XVIIth Congress of the International Society of Electrophysiology and Kinesiology

MUSCLES IN MOTION: Moving Research into Clinical Practice

18–21 June 2008 Niagara Falls, Ontario, Canada
***MUSCLES IN MOTION:* Moving Research into Clinical Practice**

NOTE TO READER: The full proceedings of the XVII Congress of the International Society of Electrophysiology and Kinesiology (ISEK), hosted in Niagara Falls, Ontario, Canada from 18 June 2008 to 21 June 2008, were constructed as a set of web pages delivered on a CD. As such, each abstract was contained in a separate PDF file, each file being hyperlinked from a Table of Contents and also from an Author Index. We have combined those files into this one PDF file. First, we provide the Table of Contents (without hyperlinks). Second, we provide the Author Index (without hyperlinks). Third, we provide all of the presented abstract papers, concatenated. Readers are encouraged to peruse the Table of Contents and Author Index in order to identify abstracts of interest to them. To find the full abstract within the abstract file, readers can search within the PDF file by the identified title or author. Note that some author last names are not unique and many authors contributed multiple papers.

Edited by Linda McLean and Paolo Bonato, Scientific Program Chairs

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Roberto Merletti[1], [1]*Department of Electronics, Politecnico di Torino, Italy*

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Denmark

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MPO1.2 NEUROMUSCULAR CONTROL DURING ISOKINETIC KNEE EXTENSION IN TOP LEVEL KARATEKA
Paola Sbriccoli[1], Francesco Felici[1], Alberto Di Mario[2], *[1]IUSM, Italy, [2]FIJLKAM, Italy*

MPO1.3 THE RELATIONSHIP BETWEEN OPEN AND CLOSED KINETIC CHAIN STRENGTH OF THE LOWER LIMB AND JUMPING PERFORMANCE IN THE ANTERIOR CRUCIATE LIGAMENT DEFICIENT SUBJECTS
Harukazu Tohyama[1], Katsunori Ikoma[1], Masayuki Ueda[1], Makoto Yuri[1], Takeshi Chiba[1], Ryohei Urayama[1], Kyoichi Hori[1], Kazunori Yasuda[1], *[1]Hokkaido University School of Medicine, Japan*

MPO1.4 INCREASED CO-ACTIVATION OF ANTAGONIST MUSCLES BY FORCE MODULATION TRAINING
Massimo Mischi[1], *[1]Eindhoven University of Technology, Netherlands*

MPO1.5 THE EFFECT OF PATELLAR TAPING ON VMO ACTIVATION DURING FUNCTIONAL ACTIVIES IN HEALTHY SUBJECTS
Robbie Trachter[1], Lisa Sheehy[1], Linda McLean[1], *[1]Queen's University, Canada*

MOTOR PERFORMANCE AND SPORT MPO2
Thursday June 19 15:00-16:30 Strategy Room 2

MPO2.1 NEUROMUSCULAR RESPONSES TO CONTINUOUS AND INTERMITTENT VOLUNTARY CONTRACTIONS
Alberto Rainoldi[1], Marco Gazzoni[2], Massimiliano Gollin[1], Marco Alessandro Minetto[3], *[1]Motor Science Research Center, SUISM, University of Turin, Italy, [2]Laboratory for Engineering of the Neuromuscular System, Department of Electronics, Polytechnic of Turin, Italy, [3]Division of Endocrinology and Metabolism, Department of Internal Medicine, University of Turin, Italy*

MPO2.2 EFFECT OF POWER-ASSIST RATIOS ON PHYSICAL ACTIVITIES DURING CYCLING WITH A TORQUE-ASSISTED BICYCLE
Tohru Kiryu[1], Masakazu Sakaguchi[1], [1]*Niigata University, Japan*

MPO2.3 EFFECTS OF EXERCISE-INDUCED MUSCLE DAMAGE ON FATIGUE-RELATED VOLITIONAL AND MAGNETICALLY-EVOKED NEUROMUSCULAR PERFORMANCE
Claire Minshull[1], Nigel Gleeson[1], Andrea Bailey[2], [1]*Nottingham Trent University, United Kingdom,* [2]*R.J.A.H. Orthopaedic Hospital, United Kingdom*

MPO2.4 EFFECTS OF MODE OF FLEXIBILITY CONDITIONING ON NEUROMECHANICAL PERFORMANCE IN MALES
Nigel Gleeson[1], Claire Minshull[1], Andrea Bailey[2], [1]*Nottingham Trent University, United Kingdom,* [2]*RJAH Orthopaedic and District NHS Trust, United Kingdom*

MPO2.5 DROP LANDING BIOMECHANICS – DOES SUBSEQUENT TASK AFFECT NEUROMUSCULAR MOTOR (PRE) CONTROL STRATEGY?
Paulo Mourão[1], Rui Ângelo[1], Alberto Carvalho[1], [1]*Higher Education Institute of Maia, Portugal*

MPO2.6 MUSCLE ACTIVITY AND ENERGETICS DURING NORDIC WALKING
Nicolas Nielsen[1], Pia Melcher[1], Hanne Lauridsen[1], Bente Jensen[1], [1]*University of Copenhagen, Denmark*

MOTOR PERFORMANCE AND SPORT MPO3
Friday June 20 9:00-10:30 Strategy Room 1

MPO3.1 WAVELET-EMG-ANALYSIS OF THE LEG MUSCLES IN FENCING DURING A FLÈCHE ATTACK
Corina Nüesch[1], Beat Göpfert[1], Marcel Fischer[1], Julien Frere[1], Dieter Wirz[1], Niklaus Friederich[2], [1]*Lab. for Orthopaedic Biomechanics, University Basel, Switzerland,* [2]*Dept. for Orthopaedic Surgery, Kantonsspital Bruderholz, Switzerland,* [3]*Institute for Biomechanics, ETH Zurich, Switzerland,* [4]*Centre d'Etude des Transformations des APS (CETAPS), Université de Rouen, France*

MPO3.2 SURFACE EMG AND HEART RATE RESPONSES IN PATIENTS WITH FIBROMYALGIA SYNDROME
Paul Jarle Mork[1], Håvard Wuttudal Lorås[1], Magne Rø[1], Ulf Lundberg[2], Rolf Westgaard[1], [1]*Norwegian University of Science and Technology, Norway,* [2]*Stockholm University, Sweden*

MPO3.3 COMPARISON OF HEALTH-RELATED FITNESS AMONG ADOLESCENTS OF CONTRASTING MATURITY STATUS
Masoumeh Shojaei[1], Fatemeh Akbari[1], Abbas Ali Gaeini[1], [1]*Al-Zahra University, Iran, Islamic Republic of,* [2]*Tehran University, Iran, Islamic Republic of*

MPO3.4 CHARACTERISTICS OF MUSCLE ACTIVITIES IN YOUNG AND ELDERLY TAI CHI PRACTITIONERS DURING TAI CHI GAIT
Xiaolin Ren[1], Ge Wu[1], [1]*University of Vermont, United States*

MPO3.5 IS NEURO-MUSCULAR FUNCTION INFLUENCED BY CREATINE SUPPLEMENTATION?
Ilenia Bazzucchi[1], Francesco Felici[1], Massimo Sacchetti[1], [1]*University Institute of Movement Sciences, Italy*

MPO3.6 THE SPATIAL AND TEMPORAL RELATIONS BETWEEN EMG ACTIVITY AND JOINT DISPLACEMENT DURING THE LAT PULL-DOWN EXERCISE

Yasushi Koyama[1], Hirofumi Kobayashi[1], Shiji Suzuki[1], Roger M. Enoka[2], [1]*WASEDA university, Japan,* [2]*University of Colorado, United States*

MODELING AND SIGNAL PROCESSING MSO1
Friday June 20 11:30-13:00 Strategy Room 1

MSO1.1 DEVELOPMENT OF AN SVM CLASSIFIER FOR DETECTING MERGED MOTOR UNIT POTENTIAL TRAINS
Hossein Parsaei[1], Faezeh Jahanmiri Nezhad[1], Daniel W. Stashuk1[1], Andrew Hamilton-Wright[2], [1]*University of Waterloo, Canada,* [2]*University of Guelph, Canada*

MSO1.2 CRITICAL EVALUATION OF MOTOR UNIT ARCHITECTURE MODELS
Javier Navallas[1], Armando Malanda[1], Luis Gila[2], Javier Rodriguez[1], Ignacio Rodriguez[1], [1]*Universidad Publica de Navarra, Spain,* [2]*Hospital Virgen del Camino, Spain*

MSO1.3 MODEL BASED ESTIMATON OF MOTOR UNIT PROPERTIES
Ariane Schad[1], Robert Oostenveld[2], Johannes P van Dijk[3], Machiel J Zwarts[3], Dick F Stegeman[3], Jens Timmer[1], Bernd G Lapatki[1], [1]*University of Freiburg, Germany,* [2]*Radboud University of Nijmegen, Netherlands,* [3]*Radboud University Medical Centre Nijmegen, Netherlands*

MSO1.4 CONTROLLING CLASSIFIER ENSEMBLE MEMBERS FOR MOTOR UNIT POTENTIAL CLASSIFICATION THROUGH DIVERSITY MEASURE
Sarbast Rasheed[1], Daniel Stashuk[1], Mohamed Kamel[1], [1]*University of Waterloo, Canada*

MSO1.5 FINITE ELEMENT SIMULATION OF SURFACE POTENTIAL DISTRIBUTIONS AT VARIABLE LIMB JOINT ANGLE
Mogens Nielsen[1], Thomas Graven-Nielsen[1], Jens Brøndum Frøkjær[1], Dario Farina[1], [1]*Aalborg University, Denmark,* [2]*Aalborg Hospital, Denmark*

MSO1.6 USE OF HILL-TYPE MUSCLE MODELS IN THE FAST ORTHOGONAL SEARCH ALGORITHM FOR WRIST FORCE ESTIMATION
Katherine Mountjoy[1], Evelyn Morin[1], Maryam Moradi[1], Keyvan Hashtrudi-Zaad[1], [1]*Queen's University, Canada*

MODELING AND SIGNAL PROCESSING MSO2
Saturday June 21 9:00-10:30 Strategy Room 3

MSO2.1 DIFFERENTIATION BETWEEN CROSSTALK AND CO-ACTIVATION: A FUZZY-INFERENCE BASED APPROACH
Catherine Disselhorst-Klug[1], Lars Meinecke[1], Thomas Schmitz-Rode[1], Günter Rau[1], [1]*Chair of Applied Medical Engineering, Helmholtz-Institute, RWTHAachen University, Germany*

MSO2.2 NOVEL POWER SPECTRUM-BASED FRACTAL INDICATORS FOR MYOELECTRIC PARAMETERS
Mehran Talebinejad[1], Adrian Chan[2], Ali Miri[1], [1]*University of Ottawa, Canada,* [2]*Carleton University, Canada*

MSO2.3 EFFECT OF RECTIFICATION ON SURFACE EMG COHERENCE
Madeleine Lowery[1], [1]*University College Dublin, Ireland*

MSO2.4 FEATURE SPACE TRAJECTORIES FOR USE WITH MYOELECTRIC CONTROL

OF A PROSTHETIC LIMB
Levi Hargrove[1], Yves Losier[1], Kevin Englehart[1], Bernard Hudgins[1], [1]*University of New Brunswick, Canada*

MSO2.5 SPECTRAL MULTIDIP CONDUCTION VELOCITY ESTIMATION WITH HIGH-ORDER SPATIAL FILTERS
Francesco Negro[1], Olivier Rossell[1], Francesco Sapia[1], Dario Farina[1], [1]*Aalborg University, Denmark*

MSO2.6 IMPROVING THE REPRESENTATION OF A MUSCLE IN SURFACE EMG USING “EXTREME” HIGH PASS FILTERING
Dick Stegeman[1], Bram Bielen[1], Johannes Van Dijk[1], Bert Kleine[1], Didier Staudenmann[2], Jaap Van Dieen[2], [1]*Dept. of Clin. Neurophysiology, Radboud University Nijmegen Medical Centre, Netherlands*, [2]*Fac. of Human Movement Sciences, VU University, Netherlands*

MOTOR UNITS MUO1
Thursday June 19 9:00-10:30 Strategy Room 3

MUO1.1 QUANTITATIVE ELECTROMYOGRAPHIC EVIDENCE OF CHANGES IN MUP MORPHOLOGY WITH REPETITIVE STRAIN INJURIES
Kristina Calder[1], Linda McLean[1], Dan Stashuk[2], [1]*Queen's University, Canada*, [2]*University of Waterloo, Canada*

MUO1.2 BAYESIAN AGGREGATION FOR NON-SPECIFIC ARM PAIN CLASSIFICATION
Andrew Hamilton-Wright[2], Linda McLean[1], Daniel Stashuk[3], Kristina Calder[1], [1]*Queen's University, Canada*, [2]*Mount Allison University, Canada*, [3]*University of Waterloo, Canada*

MUO1.3 QUANTITATIVE ELECTROMYOGRAPHIC CHANGES ASSOCIATED WITH MOTOR DEFICITS IN CARPAL TUNNEL SYNDROME
Joseph Nashed[1], Andrew Hamilton-Wright[1], Linda McLean[1], [1]*Queen's University, Canada*

MUO1.4 RELATIONSHIP BETWEEN BICEPS BRACHII SURFACE EMG CHARACTERISTICS AND MOTOR UNIT FIRING BEHAVIOR
Anita Christie[1], Gary Kamen[1], J. Greig Inglis[2], David Gabriel[2], [1]*University of Massachusetts, United States*, [2]*Brock University, Canada*

MUO1.5 COORDINATION OF MOTOR-UNIT FIRING PATTERNS IN ELBOW FLEXORS
Zoia Lateva[1], Kevin McGill[1], M. Elise Johanson[1], [1]*VA Palo Alto Health Care System, United States*

MUO1.6 SHORT TERM BED REST REDUCES CONDUCTION VELOCITY OF INDIVIDUAL MOTOR UNITS IN LEG MUSCLES
Corrado Cescon[1], Marco Gazzoni[1], Roberto Merletti[1], [1]*LISiN, Dip. di Elettronica, Politecnico di Torino, Italy*

MOTOR UNITS MUO2
Thursday June 19 11:30-13:00 Strategy Room 3

MUO2.1 BAYESIAN MUSCLE CHARACTERIZATION USING QUANTITATIVE ELECTROMYOGRAPHY
Lou Pino[1], Dan Stashuk[1], Shaun Boe[2], Tim Doherty[2], [1]*University of Waterloo,*

Canada, [2]London Health Sciences Centre, Canada

MUO2.2 RESOLVING SUPERIMPOSED MUAPS BY PARTICLE SWARM OPTIMIZATION
Hamid Reza Marateb[1], Kevin C. McGill[2], *[1]Laboratory of Engineering of Neuromuscular System (LISiN) , Politecnico di Torino, Italy, [2]VA Palo Alto Health Care System, United States*

MUO2.3 MOTOR UNIT TRACKING WITH HIGH-DENSITY SURFACE EMG
Joleen Blok[1], Ellen Maathuis[1], Gerhard Visser[1], *[1]Erasmus MC, University Medical Center Rotterdam, Netherlands*

MUO2.4 DECOMPOSITION OF SURFACE EMG FROM EXTERNAL ANAL SPHINCTER
Ales Holobar[1], Paul Enck[2], Heidemarie Hinninghofen[2], Roberto Merletti[1], *[1]Polytechnic of Torino, Italy, [2]University of Tübingen, Germany*

MUO2.5 DEMUSETOOL - A TOOL FOR DECOMPOSITION OF MULTICHANNEL SURFACE ELECTROMYOGRAMS
Ales Holobar[1], Damjan Zazula[2], Roberto Merletti[1], *[1]Polytechnic of Torino, Italy, [2]University of Maribor, Slovenia*

MUO2.6 MOTOR UNIT RATE CODING AND NEUROMUSCULAR PHASE ADVANCE IN SINUSOIDAL ISOMETRIC CONTRACTIONS
Christopher Knight[1], Maria Bellumori[1], *[1]University of Delaware, United States*

MOTOR UNITS MUO3
Saturday June 21 11:30-13:00 Strategy Room 2

MUO3.1 EFFECT OF IDENTIFICATION ERRORS ON SYNCHRONIZATION AND COMMON DRIVE PARAMETERS
Jayaraj Nair[1], Jose Gonzalez-Cueto[1], Zeynep Erim[2], *[1]Dalhousie University, Canada, [2]National Institute of Biomedical Imaging and Bioengineering, United States*

MUO3.2 DESIGN OF THIN-FILM ELECTRODES FOR MULTI-CHANNEL INTRAMUSCULAR RECORDINGS
Dario Farina[1], Sascha Kammer[1], Ken Yoshida[1], *[1]Aalborg University, Denmark, [2]Fraunhofer Institute for Biomedical Engineering, Germany, [3]Indiana University-Purdue University Indianapolis, United States*

MUO3.3 MOTOR CONTROL OF LOW THRESHOLD TRAPEZIUS MOTOR UNITS IN SUSTAINED CONTRACTIONS OF VARIABLE STRENGTH
Rolf H. Westgaard[1], Carlo J. De Luca[2], *[1]Norwegian University of Science and Technology, Norway, [2]Boston University, United States*

MUO3.4 MUSCLE ACTIVITY DURING LINEARLY VARYING ISOMETRIC CONTRACTION: ELECTROMYOGRAPHICAL ASPECTS
Claudio Orizio[1], Elena Baruzzi[1], Valentina Maruggio[1], Bertrand Diemont[1], Paolo Gaffurini[1], Massimiliano Gobbo[1], *[1]University of Brescia, Brescia, Italy, Italy*

MUO3.5 EFFECT OF SMALL MOTOR UNIT POTENTIALS ON THE MOTOR UNIT NUMBER ESTIMATE
Hans van Dijk[1], Machiel Zwarts[1], Jurgen Schelhaas[1], Dick Stegeman[1], *[1]Radboud University Nijmegen Medical Centre, Netherlands*

MUO3.6 STRATEGIES FOR MOTOR UNIT OPTIMISATION DURING PAIN
Kylie Tucker[1], Paul Hodges[1], *[1]University of Queensland, Australia*

NEUROPHYSIOLOGY NPO1
Friday June 20 11:30-13:00 Strategy Room 2

NPO1.1 THE EFFECT OF A PRE-CONDITIONING ACTIVATION ON MUSCLE FIBER CONDUCTION VELOCITY AND TWITCH FORCE DURING DOUBLET SUPRAMAXIMAL STIMULATION
Ernest Nlandu Kamavuako[1], Dario Farina[1], Kristian Hennings[1], [1]*Aalborg University, Denmark*

NPO1.2 SELECTIVE INCREASE IN CORTICOSPINAL EXCITABILITY WITH TACTILE EXPLORATION
Patricia Oliver[1], François Tremblay[2], [1]*University of Ottawa, Canada,* [2]*Elizabeth Bruyere Research Institute, Canada*

NPO1.3 H-REFLEX VARIABILITY UNDER PRESYNAPTIC INHIBITION
Rinado Mezzarane[1], André Kohn[1], [1]*University of Sao Paulo, Brazil*

NPO1.4 CHANGES IN REFLEX EXCITABILITY CAN SUGGEST PREPARATORY ACTIVITY IN THE SPINAL CORD FOR MUSCLE CONTRATION
Emerson Fachin Martins[1], André Fabio Kohn[2], [1]*University Municipal of São Caetano do Sul, Brazil,* [2]*University of São Paulo, Brazil*

NPO1.5 INTERNEURON MEDIATION OF MUSCLE CO-CONTRACTION: FCU LATE RESPONSES AFTER DISTAL ULNAR STIMULATION
Niles Roberts[1], Jacqueline Wertsch[2], [1]*Zablocki VA Medical Center, United States,* [2]*Medical College of Wisconsin, United States*

OCCUPATIONAL BIOMECHANICS KEYNOTE OBK1
Thursday June 19 14:00-14:50 Great Room C

OBK1.1 OCCUPATIONAL BIOMECHANICS OF THE SPINE
Jim Potvin[1], [1]*McMaster University, Canada*

OCCUPATIONAL BIOMECHANICS SYMPOSIUM OBS1
Thursday June 19 15:00-16:30 Great Room C

OBS1.1 CAN LOW LEVEL EMG BE USED TO PREDICT SUBJECTIVE DISCOMFORT SCORES
Jack Callaghan[1], Jennifer Durkin[1], Diane Gregory[1], Erika Nelson-Wong[1], [1]*University of Waterloo, Canada*

OBS1.2 TRUNK EXTENSOR ACTIVITY IN LOW BACK PAIN PATIENTS
Jaap van Dieën[1], Wolbert van den Hoorn[1], Onno Meijer[1], [1]*VU University, the Netherlands*

OBS1.3 REHABILITATING LOW BACK DISORDERS: CONSIDERING CORRUPTED MOTOR PATTERNS
Stuart McGill[1], [1]*University of Waterloo, Canada*

POSTURE AND BALANCE PBO1
Friday June 20 15:00-16:30 Strategy Room 2

PBO1.1 EXERCISE-INDUCED REMODELING OF MUSCLE ACTIVATION PATTERNS

FOLLOWING STROKE
Vicki Gray[1], Tanya Ivanova[1], S. Jayne Garland[1], [1]*University of Western Ontario, Canada*

PBO1.2 CORRELATION BETWEEN HEAD AND SHOULDERS POSTURE ALIGNMENT AND MASTICATORY MUSCLES SURFACE ELECTROMYOGRAPHIC ACTIVITY IN TEMPOROMANDIBULAR DISORDER SUBJECTS.
Cláudia Lopes Duarte[1], Fausto Bérzin[1], [1]*State University of Campinas, Brazil*

PBO1.3 ACTIVITY OF COMPONENTS OF THE PARASPINAL AND ABDOMINAL MUSCLE GROUPS DIFFERS WITH SPINAL CURVES IN SITTING
Andrew Claus[1], Julie Hides[1], Lorimer Moseley[2], Paul Hodges[1], [1]*The University of Queensland, Australia,* [2]*Oxford University, United Kingdom*

PBO1.4 THE EFFECT OF FATIGUE ON POSTURAL CONTROL DURING ONE-LEG STANCE IN FEMALE ELITE HANDBALL PLAYERS
Mette K. Zebis[1], Jørgen Skotte[1], Lars L. Andersen[1], Per Aagaard[3], [1]*National Research Centre for the Working Environment, Denmark,* [2]*Institute of Sports Medicine Copenhagen, Bispebjerg Hospital, Denmark,* [3]*University of Southern Denmark, Denmark*

POSTURE AND BALANCE PBO2
Saturday June 21 9:00-10:30 Strategy Room 2

PBO2.1 EVIDENCE OF PLATEAU POTENTIAL IN MOTONEURONS DRIVING FOOT MUSCLES DURING QUIET STANCE
Liria Okai[1], André Kohn[2], [1]*Physical Therapy School - Universidade de Santo Amaro, Brazil,* [2]*Universidade de São Paulo, Brazil*

PBO2.2 A STUDY ON THE STRIDE RATE VARIABILITY USING TREADMILL ON DEMAND
JinSeung Choi[1], DongWon Kang[1], Gyerae Tack[1], [1]*Konkuk University, Korea, Republic of*

PBO2.3 CIRCUMVENTION OF A SUDDENLY APPEARING OBSTACLE
Jaap van Dieën[1], Mendel Douma[1], Leen Nugteren[1], Mirjam Pijnappels[1], Idsart Kingma[1], [1]*VU University Amsterdam, Netherlands*

PBO2.4 TRADE-OFF AND COACTIVATION BETWEEN GASTROCNEMII DURING A QUIET STANDING TEST: PRELIMINARY RESULTS
Taian Vieira[1], Roberto Merletti[1], [1]*Polytechnic of Turin, Italy,* [2]*Federal University of Rio de Janeiro, Brazil*

PBO2.5 ASSESSMENT OF GASTROCNEMIUS HETEROGENEITY USING A HIGH DENSITY SEMG SYSTEM
Taian Vieira[1], Francesco Mastrangelo[1], Roberto Merletti[1], [1]*Polytechnic of Turin, Italy,* [2]*Federal University of Rio de Janeiro, Brazil*

POSTURE AND BALANCE PBO3
Saturday June 21 11:30-13:00 Great Room C

PBO3.1 ASSESSMENT OF POSTURAL SWAY DURING MULTIPLE LOAD AND VISUAL CONDITIONS
Pilwon Hur[1], Karl Rosengren[1], Gavin Horn[1], Tad Schroeder[1], Sarah Ashton-Szabo[1], Elizabeth Hsiao-Wecksler[1], [1]*University of Illinois at Urbana-Champaign, United*

States

PBO3.2 HUMAN ANKLE ANGLE EXCURSIONS LINKED TO DETECTION OF SMALL HORIZONTAL SINUSOIDAL PLATFORM TRANSLATIONS
Rakesh Pilkar[1], Viprali Bhatkar[1], Christopher Storey[2], Charles Robinson[1], [1]*Clarkson University, United States,* [2]*Louisiana State Univ HSC, United States,* [3]*VA Medical Center, United States*

PBO3.3 TA ACTIVATION OCCURS ONLY WITH CORRECTLY DETECTED, SHORT PLATFORM TRANSLATIONS MADE AT AND ABOVE DETECTION THRESHOLD
Xiaoxi Dong[1], Charles Robinson[2], Charles Robinson[1], [1]*Clarkson University, United States,* [2]*VA Medical Center, United States*

PBO3.4 SHOULDER MUSCLE CONTRIBUTION TO RECOVERY FROM A TRIP
Idsart Kingma[1], Mirjam Pijnappels[1], Jaap van Dieën[1], [1]*VU University Amsterdam, Netherlands*

PBO3.5 POSTURE-MOVEMENT STRATEGIES TO ADAPT TO REPETITIVE MOTION-INDUCED ARM FATIGUE
Jason Fuller[1], Karen Lomond[1], Julie Côté[1], [1]*McGill University, Canada*

PBO3.6 ASSOCIATIONS AMONG BALANCE, ELECTROMYOGRAPHY AND LOWER EXTREMITY INJURY IN BALLET DANCERS: A PRELIMINARY STUDY
Jung-Hsien Liao[1], Cheng-Feng Lin[1], Fong-Chin Su[1], Jin-Yang Chen[1], [1]*National Cheng Kung University, Taiwan*

PHYSICAL MEDICINE AND REHABILITATION PMO1
Thursday June 19 11:30-13:00 Great Room C

PMO1.1 MEASURING THE EFFICACY OF A NOVEL BIOFEEDBACK DEVICE FOR POST-STROKE REHABILITATION OF THE UPPER EXTREMITY
Don Yungher[1], William Craelius[1], [1]*Rutgers University, United States*

PMO1.2 MOTOR UNIT PROPERTIES OF THE BICEPS BRACHII OF STROKE PATIENTS ASSESSED WITH SURFACE ARRAY EMG
Laura Kallenberg[1], Hermie Hermens[2], [1]*Roessingh Research and Development, Netherlands,* [2]*University of Twente, Netherlands*

PMO1.3 FUNCTIONAL IMPLICATION OF GAIT AFTER LEFT OR RIGHT-SIDED STROKE
Joao Correa[1], Carolina Rocco[1], Daniel Andrade[1], Claudia Oliveira[1], Fernanda Correa[1], [1]*UNINOVE, Brazil*

PMO1.4 THE EFFECTS OF CIRCUMFERENTIAL AIR-SPLINT PRESSURE ON THE SOLEUS STRETCH REFLEX IN SUBJECTS WITH AND WITHOUT CEREBRAL VASCULAR ACCIDENT
James Agostinucci[1], [1]*University of Rhode Island, United States*

PMO1.5 THE NEUROMUSCULAR DEMANDS OF ALTERING FOOT PROGRESSION ANGLE DURING GAIT
Derek Rutherford[1], Cheryl Hubley-Kozey[1], [1]*Dalhousie University, Canada*

PMO1.6 TIME-FREQUENCY ANALYSIS OF LEG MUSCLES DURING GAIT IN PATIENTS WITH DIPLEGIC CEREBRAL PALSY
Jacqueline Romkes[1], Reinald Brunner[1], [1]*University Children's Hospital Basel, Switzerland*

PHYSICAL MEDICINE AND REHABILITATION PMO2
Thursday June 19 15:00-16:30 Strategy Room 1

PMO2.1 THE EFFECT OF CIRCUMFERENTIAL PRESSURE ON F-WAVE RESPONSE IN THE TIBIALIS ANTERIOR MUSCLE
James Agostinucci[1], Amy Woodard[1], [1]*University of Rhode Island, United States*

PMO2.2 THE EFFECTS OF TIZANIDINE ON THE MEDIUM LATENCY RESPONSE IN THE M. FLEXOR CARPI RADIALIS.
Carel Meskers[1], Alfred Schouten[2], Marieke Rich[1], Jasper Schuurmans[2], Jurriaan de Groot[1], Hans Arendzen[1], [1]*Leiden University Medical Center, Netherlands,* [2]*Delft University of Technology, Netherlands*

PMO2.3 EXPLORATION OF THE PHYSIOLOGICAL BASIS OF MUSCLE ACTIVITY WITH NERVE ELONGATION IN A CLINICAL SETTING
Michel Coppieters[1], Andrew McLean[1], Paul Hodges[1], [1]*The University of Queensland, Australia*

PMO2.4 ROBOTIC GAIT TRAINING IN CHILDREN WITH CEREBRAL PALSY
Ben Patritti[1], Anat Mirelman[1], Marlena Pelliccio[1], Lynn Deming[1], Donna Nimec[1], Paolo Bonato[2], [1]*Harvard Medical School, United States,* [2]*Harvard-MIT Division of Health Sciences and Technology, United States*

PMO2.5 INFLUENCE OF PERIPHERAL AFFERENT STIMULATION ON CORTICOMOTOR EXCITABILITY: A LOWER LIMB STUDY
Charles Tardif[1], Guillaume Léonard[2], Martin Héroux[3], Louis E. Tremblay[1], [1]*University of Ottawa, Canada,* [2]*Université de Sherbrooke, Canada,* [3]*Queen's University, Canada*

PMO2.6 MODULATION OF NEURAL ACTIVITY DURING MENTAL SIMULATION OF KNEE EXTENSION : AGE AND TYPE OF MUSCLE CONTRACTION EFFECTS
Louis E. Tremblay[1], Cédric Tremblay[1], Charles M. Tardif[1], [1]*University of Ottawa, Canada*

PHYSICAL MEDICINE AND REHABILITATION PMO3
Thursday June 19 17:30-19:00 Strategy Room 1

PMO3.1 DIFFERENCES IN MUSCLE ACTIVATION PATTERNS BETWEEN SUBJECTS WITH CHRONIC LOW BACK PAIN AND HEALTHY CONTROLS DURING WALKING
Marije van der Hulst[1], Miriam Vollenbroek-Hutten[1], Johan Rietman[1], Leendert Schaake[1], Hermie Hermens[1], [1]*Roessingh Research and Development, Netherlands,* [2]*Rehabilitation Centre het Roessingh, Netherlands,* [3]*Faculties of Electrical Engineering, Mathematics & Informatics, University of Twente, Netherlands*

PMO3.2 SEGMENTAL KINEMATICS DURING MANUAL ROTATION MOBILIZATION OF THE ATLANTO-AXIAL JOINT: MORPHOLOGY OR INTERVENTION DETERMINED?
Erik Cattrysse[1], Patrick Kool[1], Steven Provyn[1], Jean-Pierre Baeyens[1], Jan-Pieter Clarys[1], Van Roy Peter[1], [1]*Vrije Universiteit Brussel, Belgium,* [2]*University College Antwerp, Belgium*

PMO3.3 A NOVEL METHOD FOR NECK-COORDINATION EXERCISE – A PILOT STUDY ON PERSONS WITH NON-SPECIFIC CHRONIC NECK PAIN
Ulrik Röijezon[1], Martin Björklund[1], Mats Djupsjöbacka[1], [1]*Centre for Musculoskeletal*

Research, University of Gavle, Sweden, [2]Alfta Research Fundation, Sweden, [3]Dept of Community Medicine and Rehabilitation, Umea University, Sweden

PMO3.4 EFFECT OF TIBIAL ROTATION IN OPEN AND CLOSED CHAIN EXERCISES ON VASTUS MEDIALIS OBLIQUUS ACTIVATION
Lisa Sheehy[1], Shannon Gallagher[1], Linda McLean[1], [1]*Queen's University, Canada*

PMO3.5 A METHOD TO MONITOR THE STATUS OF FUNCTIONAL RECOVERY IN TENDINOPATHIES: A COMBINATION OF EMG AND STRENGTH MEASUREMENTS
Omid Alizadehkhaiyat[1], Anthony Fisher[2], Graham Kemp[1], Simon Frostick[1], [1]*University of Liverpool, United Kingdom,* [2]*Royal Liverpool and Broadgreen University Hospitals, United Kingdom*

PMO3.6 MYOTEL: ADRESSING NECK SHOULDER PAIN AND ITS RELATED DISABILITIES BY ASSESSING AND FEEDBACK SEMG IN THE DAILY (WORK) ENVIRONMENT
Miriam Vollenbroek-Hutten[1], Rianne Huis in 't Veld[1], Hermie Hermens[2], [1]*Roessingh Research and Development, Netherlands,* [2]*University of Twente, Faculty Electrical Engineering, Mathematics and Computer Science, Netherlands*

PHYSICAL MEDICINE AND REHABILITATION PMO4
Friday June 20 11:30-13:00 Great Room C

PMO4.1 NEUROMUSCULAR PATTERN OF THE PELVIC FLOOR MUSCLES DURING LIMB MOVEMENTS
Jenny Sjödahl[1], Annelie Gutke[1], Joanna Kvist[1], Birgitta Öberg[1], [1]*Department of Medical and Health Sciences, Linköping University, Sweden*

PMO4.2 MAXIMUM PELVIC FLOOR MUSCLE ACTIVATION AND INTRAVAGINAL PRESSURE ACHIEVED DURING COUGHING IN WOMEN WITH AND WITHOUT INCONTINENCE
Stéphanie Madill[1], Marie-Andrée Harvey[2], Linda McLean[1], [1]*Queen's University, Canada,* [2]*Kingston General Hospital, Canada*

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Cindy Brown[1], Linda McLean[1], [1]*Queen's University, Canada*

PMO4.4 INTRAVAGINAL PRESSURE AND PELVIC FLOOR MUSCLE ACTIVATION DURING COUGHING IN WOMEN WITH AND WITHOUT STRESS URINARY INCONTINENCE
Stéphanie Madill[1], Marie-Andrée Harvey[2], Linda McLean[1], [1]*Queen's University, Canada,* [2]*Kingston General Hospital, Canada*

PMO4.5 PELVIC FLOOR MUSCLE RESPONSE TO A PRESSURE PAIN STIMULUS IN WOMEN WITH PROVOKED VESTIBULODYNIA
Evelyne Gentilcore-Saulnier[1], Corrie Goldfinger[1], Linda McLean[1], Caroline Pukall[1], Susan Chamberlain[1], [1]*Queen's University, Canada,* [2]*Kingston General Hospital, Canada*

PMO4.6 EMG ONSET DETECTION IN PELVIC FLOOR MUSCLES OF CONTINENT AND INCONTINENT WOMEN
Luc Janssens[1], Filip Staes[2], F Penninckx[4], R Vereecken[5], W De Weerdt[3], A Devreese[2], [1]*GroupT Leuven Engineering College, Belgium,* [2]*Department of Physiotherapy and Rehabilitation, University Hospital Leuven, K.U. Leuven, Belgium,*

[3]Department of Rehabilitation Sciences, K.U. Leuven, Belgium, [4]Department of Abdominal Surgery, University Hospital Leuven, K.U. Leuven, Belgium, [5]Department of Urology, University Hospital Leuven, K.U. Leuven, Belgium

PHYSICAL MEDICINE AND REHABILITATION PMO5
Friday June 20 15:00-16:30 Strategy Room 1

PMO5.1 TBA

TBA

PMO5.2 EVALUATION OF THE FUNCTIONS OF THE UPPER LIMB AND THE COGNITIVE FUNCTION WITH THE IMPROVED MODEL OF A HAPTIC DEVICE SYSYTEM IN HEALTHY CHILDREN

Yumi Ikeda[1], Kaoru Inoue[1], Yuko Ito[1], Takafumi Terada[1], Hitoshi Takei[1], Osamu Nitta[1], Yorimitsu Furukawa[1], Shu Watanabe[1], Seiki Kaneko[1], *[1]Arakawa-ku,7-2-10, Japan*

PMO5.3 KINEMATIC AND ELECTROMYOGRAPHIC CHARACTERIZATION OF PHANTOM LIMB MOVEMENTS IN ABOVE-ELBOW AMPUTEES

Catherine Mercier[1], Martin Gagné[1], Sébastien Hétu[1], Joëlle Dubé[1], Pierre-Olivier Lauzon[1], *[1]Université Laval, Canada*

PMO5.4 AUGMENTED REALITY: A TOOL FOR MYOELECTRIC PROSTHESES

Alcimar Soares[1], Edgard Lamounier[1], Kenedy Nogueira[1], Adriano Andrade[1], *[1]Federal University of Uberlandia, Brazil*

PMO5.5 EVALUATING THE USE OF A REHABILITATION DOG AS A WALKING AID: KINEMATIC ANALYSIS

Anuja Darekar[1], Alison Oates[1], Claire Perez[1], Lynda Rondeau[2], Joyce Fung[1], *[1]Mcgill University, Canada, [2]Centre de Réadaptation Estrie, Canada*

PMO5.6 EVALUATING THE USE OF REHABILITATION DOG AS A WALKING AID: EMG ANALYSIS

Alison Oates[1], Anuja Darekar[1], Adriana Venturini[1], Lynda Rondeau[1], Joyce Fung[1], *[1]School of Physical & Occupational Therapy, McGill University and Jewish Rehabilitation Hospital Research Site of the Montreal Interdisciplinary Research Centre in Rehabilitation (CRIR), Canada, [2]Centre de Readaptation Estrie Research Site of CRIR, Canada*

PHYSICAL MEDICINE AND REHABILITATION PMO6
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PMO6.1 THE EFFECT OF STRENGTH TRAINING ON HAND FUNCTION IN CHILDREN WITH CEREBRAL PALSY

Karin Roeleveld[1], Siri Brændvik[2], Ann-Kristin Elvrum[2], Randi Sæther[2], Torarin Lamvik[2], Beatrix Vereijken[1], *[1]Human Movement Science Programme, NTNU, Norway, [2]St. Olavs University Hospital, Norway*

PMO6.2 REHABILITATION OF ATROPHIC MUSCLES AFTER TOTAL ANKLE REPLACEMENT, TAR

Vinzenz von Tscharner[1], Victor Valderrabano[1], *[1]University of Calgary, Human Performance Laboratory. Orthopaedic Department, University Hospital of Basel, 4031 Basel, Switzerland, Canada*

PMO6.3 EFFECT OF MULTILEVEL BOTULINUM TOXIN A INJECTIONS ON SURFACE EMG PATTERNS DURING GAIT IN CHILDREN WITH CEREBRAL PALSY
Lisette van der Houwen[1], Vanessa Scholtes[1], Jules Becher[1], Jaap Harlaar[1], [1]*VU University Hospital, Netherlands*

PMO6.4 WALKING IN PEOPLE WITH MYELOMENINGOCELE: MUSCLE ACTIVITY MECHANISM
Chia-Lin Chang[1], Beverly Ulrich[2], [1]*University of Pittsburgh Medical Center, United States,* [2]*University of Michigan, United States*

PMO6.5 ABNORMAL ELECTROMYOGRAPHIC SIGNALS IN THE ADULT CEREBRAL PALSY POPULATION
Melany Westwell[1], Sylvia Ounpuu[1], Katharine Bell[1], [1]*Connecticut Children's Medical Center, United States*

PMO6.6 THE INFLUENCE OF DIABETIC NEUROPATHY AND THE USE OF HABITUAL SHOES IN LOWER LIMB EMG AND IN VERTICAL FORCES DURING GAIT
Paula Akashi[1], Isabel Sacco[1], [1]*University of Sao Paulo, Brazil*

ROBOTICS KEYNOTE RBK1
Saturday June 21 8:00-8:50 Great Room C

RBK1.1 INNOVATIVE TECHNOLOGIES FOR QUANTIFYING AND TREATING LOWER EXTREMITY IMPAIRMENTS FOLLOWING STROKE
Joe Hidler[1,2], [1]*National Rehabilitation Hospital, United States,* [2]*Catholic University, United States*

ROBOTICS SYMPOSIUM RBS1
Saturday June 21 9:00-10:30 Great Room C

RBS1.1 MOTOR RECOVERY AFTER STROKE – OPTIMIZING RESULTS
Joel Stein[1,2], [1]*Spaulding Rehabilitation Hospital, United States,* [2]*Harvard Medical School, United States*

RBS1.2 CONTACT ROBOTICS: TOOLS TO STUDY AND TREAT NEUROLOGICAL DISORDERS
Neville Hogan[1], [1]*Massachusetts Institute of Technology, Unites States*

RBS1.3 ROBOTICS FOR LOCOMOTOR TRAINING AFTER SPINAL CORD INJURY
Patricia Winchester[1], Ross Querry[1], Keith Tansey[1], [1]*University of Texas, United States*

SPACTICITY ASSESSMENT KEYNOTE SAK1
Friday June 20 14:00-14:50 Great Room C/B

SAK1.1 CLINICAL ASSESSMENT OF MOTOR DEFICITS IN SPASTIC CEREBRAL PALSY
Kenton Kaufman[1], [1]*Biomechanics/Motion Analysis Laboratory, Mayo Clinic, United States*

SPACTICITY ASSESSMENT SYMPOSIUM SAS1
Friday June 20 15:00-16:30 Great Room C

SAS1.1 NEUROMUSCULAR DEFICITS IN SPASTIC CEREBRAL PALSY
Jessica Rose[1], [1]*Stanford University School of Medicine, United States*

SAS1.2 BIOMECHANICAL CHANGES IN SPASTIC HUMAN SKELETAL MUSCLE
Richard Lieber[1], [1]*University of California, United States*

TELEREHABILITATION KEYNOTE TRK1
Thursday June 19 8:00-8:50 Great Room C

TRK1.1 TELEREHABILITATION; ENABLING REMOTE MONITORING AND REMOTELY SUPERVISED TRAINING
Hermie Hermens[1,2], Miriam Vollenbroek-Hutten[1], [1]*Roessingh Research and Development, The Netherlands,* [2]*University of Twente, The Netherlands*

TELEREHABILITATION SYMPOSIUM TRS1
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Paolo Bonato[1,2], Fabrizio Cutolo[1,3], Danilo De Rossi[3], Richard Hughes[1], Shyamel Patel[1], Maurizio Schmid[1,4], Joel Stein[1], Alessandro Tognetti[3], [1]*Harvard Medical School, United States,* [2]*Harvard-MIT Division of Health Sciences and Technology, Unites States,* [3]*University of Pisa, Italy,* [4]*University Roma, Italy*

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Roberto Colombo[1], Fabrizio Pisano[2], Alessandra Mazzone[1], Carmen Delconte[2], Giuseppe Minuco[1], [1]*Bioengineering Service, Fondazione Salvatore Maugeri, IRCCS, Italy,* [2]*Neurology Dept., Fondazione Salvatore Maugeri, IRCCS, Italy*

TRS1.3 TECHNOLOGY SUPPORTED FEEDBACK STRATEGIES TO TREAT PATIENTS WITH CHRONIC DISORDERS
Miriam Vollenbroek-Hutten[1], Hermie Hermens[1,2] , [1]*Roessingh Research and Development, The Netherlands,* [2]*University of Twente, Netherlands*

VIRTUAL REALITY KEYNOTE VRK1
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Judith Deutsch[1], [1]*Department of Rehabilitation and Movements Science, University of Medicine and Dentistry of New Jersey, Unites States*

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Heidi Sveistrup[1], [1]*University of Ottawa, Canada*

VRS1.2 USE OF VIRTUAL REALITY TO FACILITATE RECOVERY OF HAND FUNCTION
Sergei Adamovich[1], Alma Merians[2], [1]*New Jersey Institute of Technology, United States,* [2]*University of Medicine and Dentistry of New Jersey, United States*

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Susan Whitney[1], Patrick Sparto[1], Khalid Alahmari[1], Mark Redfern[1], Joseph Furman[1],
[1]*University of Pittsburgh, United States*

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Vinzenz von Tscharner[1], Aurel Coza[1], Benno M. Nigg[1], [1]*University of Calgary, Human Performance Laboratory, Canada*

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MOTOR CONTROL MCP1
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MOVEMENT DISORDERS MDP1
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School of Dentistry – University of São Paulo, Brazil, [3]University of Ribeirão Preto – UNAERP, Brazil, [4]Ribeirão Preto School of Medicine – University of São Paulo, Brazil

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MDP1.6 "PATTERNIZING" STANDARDS OF SIT-TO-STAND MOVEMENT IN CEREBRAL PALSY WITH SPASTIC DIPLEGIA
Yonetsu Ryou[1], Nitta Osamu[1], *[1]Kanagawa University of Human Services, Japan, [2]Tokyo Metropolitan University, Japan*

MDP1.7 THE EFFECT OF MANDIBULAR OSTEOPOROSIS ON THE SOMATOGNATHIC SYSTEM: AN ELECTROMYOGRAPHIC ANALYSIS
Selma Siéssere[1], Luiz Gustavo Sousa[1], Mathias Vitti[1], Naira Albuquerque Lima[1], Paulo Batista Vasconcelos[1], Simone Cecilio Hallak Regalo[1], Richard Honorato Oliveira[1], Marisa Semprini[1], *[1]University of São Paulo, Brazil*

MDP1.8 VARIATIONS IN ANKE PLANTAR FLEXOR FUNCTION IN CHILDREN WITH CEREBRAL PALSY: POSSIBLE IMPLICATIONS FOR TREATMENT OPTIONS
Sylvia Ounpuu[1], Melany Westwell[1], Peter DeLuca[1], *[1]Connecticut Children's Medical Center, United States*

MDP1.9 SURFACE ELECTROMYOGRAPHIC STUDY OF MEIGE SYNDROME PATIENT: A CASE REPORT.
Cláudia Lopes Duarte[1], Cynthia Bicalho Borini[1], Lucielma S. S. Pinto[1], Jacks Jorge[1], Fausto Bérzin[1], *[1]State University of Campinas, Brazil*

MDP1.10 PAIN REDUCTION IMPROVES BALANCE CONFIDENCE IN PEOPLE WITH PERIPHERAL NEUROPATHY
Emily Bourgeois[1], Sarah Lundmark[1], Brad Manor[1], Li Li[1], *[1]Louisiana State University, United States*

MDP1.11 ELIMINATION OF PAIN BY PASTING METAL PARTICLES ON DOMAIN OF SKIN DETECTED PAIN FOR DISEASE OF HIPBONE AND KNEE
Kazuyoshi Sakamoto[1], Norio Saito[1], *[1]The University of Electro-Communications, Japan, [2]Super Medical Laboratory, Japan*

MDP1.12 ELECTROMYOGRAPHY AS IT PERTAINS TO THE STUDY OF MUSCULAR MOVEMENT WITHIN OCCUPATIONAL THERAPY
César F. Amorim[1], Lucimara J. Amorim[2], Ligia de Castro Pereira[3], Jamilson Brasileiro[4], Tamotsu Hirata[1], *[1]São Paulo State University – Unesp-Feg, Brazil, [2]Paulista University, Brazil, [3]Toronto – Ontario, Canada, [4]UFRN, Brazil*

MDP1.13 NEUROPATHIC PAIN EFFECTS PHYSICAL FUNCTION
Sarah Lundmark[1], Emily Bourgeois[1], Brad Manor[1], Li Li[1], *[1]Louisiana State University, United States*

MDP1.14 GAIT CO-ACTIVATION PATTERNS IN TERMINAL OSTEOARTHRITIS KNEE PATIENTS
Vassilios Vardaxis[1], Dane Hansen[1], Becca Meier[1], Craig Mahoney[2], *[1]Des Moines University, United States, [2]Mercy Medical Center, United States*

MUSCLE FATIGUE MFP1
Thursday June 19, 10:30-11:30 and 16:30-17:30 Great Room A

MFP1.1 EFFECTS OF FATIGUE ON SIGNAL-DEPENDENT NOISE DURING ISOMETRIC FORCE PRODUCTION
Olivier Missenard[1], Denis Mottet[1], Stéphane Perrey[1], *[1]EA 2991, France*

MFP1.2 SPATIAL DEPENDENCY IN SURFACE ELECTROMYOGRAPHY AFTER ECCENTRIC ARM EXERCISE
Harri Piitulainen[1], Paavo Komi[1], Janne Avela[1], *[1]Neuromuscular Research Center, Department of Biology of Physical Activity, University of Jyväskylä, Finland*

MFP1.3 COMPARATIVE STUDY BETWEEN EMG FATIGUE ESTIMATORS
Miriam González[2], Armando Malanda[1], Mikel Izquierdo[2], *[1]Universidad Pública de Navarra, Spain, [2]Centro de Estudios e Investigación en Medicina del Deporte, Spain*

MFP1.4 ANALYSIS OF MAXIMAL GRIP FORCE USING EMPIRICAL MODE DECOMPOSITION
Ke Li[1], David Hewson[1], Jacques Duchêne[1], Jean-Yves Hogrel[2], *[1]Université de Technologie de Troyes, France, [2]Institut de Myologie, France*

MFP1.5 ELECTROMYOGRAPHY FATIGUE THRESHOLD OF DELTOID AND UPPER TRAPEZIUS DURING SCAPULAR PLANE ARM ABDUCTION
Gleyson R. V. Stabile[1], Ligia M. Pereira[1], Maryela O. Menacho[1], Marcio M. Kawano[1], Hugo M. Pereira[1], Alan L. Z. Eis[1], Fabio Y. Nakamura[1], Jefferson R. Cardoso[1], *[1]Universidade Estadual de Londrina, Brazil*

MFP1.6 EFFECT OF ECCENTRIC STRENGTH TRAINING ON MEDIAN FREQUENCY AND TIME OF FATIGUE IN DIFFERENT LEVELS OF ISOMETRIC CONTRACTION.
Anielle C. M. Takahashi[1], Ana Beatriz de Oliveira[1], Ruth C. de Melo[1], Robison J. Quitério[2], Patricia S. Porto[1], Aparecida M. Catai[1], *[1]Universidade Federal de São Carlos - UFSCar, Brazil, [2]Universidade Estadual Paulista - UNESP, Brazil*

MFP1.7 EMG SIGNAL ANALYSIS DURING TREADMILL RUNNING AT ELECTROMYOGRAPHIC FATIGUE THRESHOLD INTENSITY.
Mauro Gonçalves[1], Sarah Regina Dias da Silva[1], Carina Helena Wasem Fraga[1], Priscila de Brito Silva[1], *[1]São Paulo State University, Brazil*

MFP1.8 NEUROMUSCULAR FATIGUE OF THE LUMBAR EXTENSORS INFLUENCES WALKING GAIT PARAMETERS IN HEALTHY WOMEN
Michael Olson[1], *[1]Southern Illinois University Carbondale, United States*

MFP1.9 MYOELECTRIC FATIGUE PROFILES DURING ELECTRICALLY-ELICITED CONTRACTIONS OF VASTUS LATERALIS, VASTUS MEDIALIS OBLIQUUS, AND VASTUS MEDIALIS LONGUS MUSCLES
Alberto Botter[1], Roberto Merletti[1], Marco Alessandro Minetto[2], *[1]Laboratory for Engineering of the Neuromuscular System, Department of Electronics, Polytechnic of Turin, Italy, [2]Division of Endocrinology and Metabolism, Department of Internal Medicine, University of Turin, Italy*

MFP1.10 ELECTROMYOGRAPHIC ASSESMENT PRE TO POST SURGERY OF THE ERECTOR SPINAE IN PATIENTS WITH HERNIATION OF NUCLEUS PULPOSUS
Rony Silvestre[1], Edith Elgueta[1], Alex Araneda[1], *[1]hospital Del Trabajador, Chile, [2]universidad De Talca, Chile, [3]universidad Santo Tomas, Chile*

MFP1.11 A FATIGUE STUDY IN AUXILIARY NURSES
Mar Marcilla[1], Irene Rubio[1], Martin Caicoya[1], *[1]Direccion General De Funcion Publica, Servicio De Prevencion De Riesgos Laborales, Spain*

MFP1.12 THE EFFECT OF HYPERTHERMIA AND PROLONGED SUBMAXIMAL CYCLING ON NEUROMUSCULAR CONTROL OF MAXIMAL VOLUNTARY CONTRACTION OF THE KNEE

EXTENSORS
Angus Hunter[1], Yumna Albertus-Kajee[2], Alan St Clair Gibson[3], *[1]University of Stirling, United Kingdom, [2]University of Cape Town, South Africa, [3]University of Northumbria, United Kingdom*

MECHANOMYOGRAPHY AND MUSCLE-JOINT MECHANICS MMP1
Saturday June 21, 10:30-11:30 and 15:00-16:30 Great Room A

MMP1.1 MRI ANALYSIS OF THE FORCED FLEXION MOVEMENTS OF THE UNILATERAL HIP JOINT IN SUPINE POSITION
Hitoshi Takei[1], Hideyuki Usa[2], Atsushi Senoo[1], Yumi Ikeda[1], Yorimitsu Furukawa[1], Shu Watanabe[1], Ken Yanagisawa[1], *[1]Tokyo Metropolitan University, Japan, [2]Senkawa Shinoda Orthopaedic Clinic, Japan*

MMP1.2 INFLUENCE OF ANGULAR VELOCITY OF A DYNAMOMETER IN JOINT POSITION SENSE OF KNEE IN NORMAL SUBJECTS
Fabio Cyrillo[1], Cristina Cabral[1], *[1]UNICID, Brazil*

MMP1.3 EVALUATION OF THE MUSCULAR RELATION OF VASTUS MEDIALIS AND VASTUS LATERALIS DURING THE LANDING OF CABRIOLE IN A BALLET DANCER – CASE STUDY
Fabio Cyrillo[1], Fabiana Crovador[1], Ligia Niero[1], Francine Gondo[1], *[1]UNICID, Brazil*

MMP1.4 ANALYSIS OF FORCES DURING ISOMETRIC MOVEMENT OF THE DELTOID MUSCLE: ELECTROMYOGRAPHIC STUDY
César F Amorim[1], Fabiano Politti[2], Jamilson Brasileiro[3], Ligia C. Pereira[4], Tamotsu Hirata[1], *[1]São Paulo State University – UNESP-FEG, Brazil, [2]University of Campinas – UNICAMP, Brazil, [3]Federal University of Rio Grande do Norte – UFRN, Brazil, [4]Toronto – Ontario - Canada., Canada*

MMP1.5 THE USE OF MECHANOMYOGRAPHY SIGNAL IN THE ASSESSMENT OF MUSCLE FIBER TYPE
Bruno Jotta[1], Rafael Rühl[2], Alexandre Pino[1], Marco Garcia[1], Márcio Souza[1], *[1]Biomedical Engineering Program – COPPE – Federal University of Brazil, Brazil, [2]Laboratory of Biomechanics – School of Physical Education and Sports – Federal University of Brazil, Brazil*

MMP1.6 REPEATABILITY OF THE MECHANOMYOGRAPHIC AMPLITUDE VERSUS ISOMETRIC TORQUE PATTERNS OF RESPONSES
Eric Ryan[1], Travis Beck[1], Trent Herda[1], Pablo Costa[1], Jason DeFreitas[1], Joel Cramer[1], *[1]University of Oklahoma, United States*

MOTOR PERFORMANCE AND SPORT MPP1
Saturday June 21, 10:30-11:30 and 15:00-16:30 Great Room A

MPP1.1 CHANGES IN ISOKINETIC TORQUE RATIO AFTER A AEROBIC EXERCISE IS DEPENDENT ON THE VELOCITY OF TESTING AND METHOD OF CALCULATION
Benedito Denadai[1], Anderson Oliveira[1], Rogério Corvino[1], Fabrizio Caputo[1], *[1]São Paulo State, Brazil*

MPP1.2 EVALUATION OF MUSCULAR FATIGUE DURING REPETITVE SKIING TURNS BY REFERRING TO SQUAT EXERCISE
Tohru Kiryu[1], Toshihiko Akutsu[1], Kazuya Motohashi[1], Yukihiko Ushiyama[1], *[1]Niigata University, Japan*

MPP1.3 ADDUCTOR MAGNUS, RECTUS ABDOMINIS, SERRATUS ANTERIOR AND TRICEPS ACTIVITY DURING REVERSE PUNCH (GYAKU ZUKI) EXECUTION
Jonas Gurgel[1], Ernesto Yee[1], Flávia Porto[1], Gustavo Sepúlveda Silva[1], Ismael Baldissera[1],

Fabiano Souza Gonçalves[1], Gabriel Espinosa da Silva[1], Pedro Osório[3], *[1]Aerospace Biomechanics Research Laboratory / PUCRS, Brazil, [2]Physical Activity Assessments and Research Laboratory / PUCRS, Brazil, [3]Universidade Luterana do Brasil, Brazil*

MPP1.4 DROP LANDING BIOMECHANICS – DOES FALLING HEIGHT IMPLY NEUROMUSCULAR SYSTEM (PRE) ADJUSTMENT?
Rui Ângelo[1], Paulo Mourão[1], Alberto Carvalho[1], *[1]Higher Education Institute of Maia, Portugal*

MPP1.5 ELECTROMYOGRAPHY ANALYSIS OF TWO DIFFERENT SQUAT EXERCISES – RELATIONSHIP OF FOOT POSITIONS
Flávia Porto[1], Jonas Gurgel[1], Fernando Luchetta[2], Ismael Baldissera[1], Fabiano Gonçalves[1], Gustavo Sepúlveda[1], Felipe Flores[1], Yuri Shoereder[1], *[1]Aerospace Biomechanics Research Laboratory / PUCRS, Brazil, [2]Physical Activity Assessments and Research Laboratory / PUCRS, Brazil*

MPP1.6 EFFECTS OF DISTANCE OF EXTERNAL FOCUS OF ATTENTION ON THE LEARNING OF BASKETBALL SET SHOT IN ADOLESCENT FEMALES
Masoumeh Shojaei[1], Marina Daneghian[1], Shahla Hojjat[1], *[1]AlZahra University, Iran, Islamic Republic of, [2]Islamic Azad University, Iran, Islamic Republic of*

MPP1.7 MUSCLE FUNCTION INVESTIGATED WITH SURFACE ELECTROMYOGRAPHY IN LATRAL ELBOW TENDINOPATHY
Omid Alizadehkhaiyat[1], Anthony Fisher[2], Graham Kemp[1], Simon Frostick[1], *[1]University of Liverpool, United Kingdom, [2]Royal Liverpool and Broadgreen University Hospitals, United Kingdom*

MPP1.8 MUSCLES THAT CROSS THE KNEE JOINT HAVE A PREFERRED ORIENTATION TO RESIST A FORCE APPLIED FROM OUTSIDE EXCEPT FOR EXTENSION AND FLEXION
Fuminari Kaneko[1], Nobuhiro Aoki[1], Tatsuya Hayami[1], Akihiro Kanamori[1], *[1]National Institute of Advanced Industrial Science and Technology, Japan*

MPP1.9 TEMPOROMANDIBULAR DYSFUNCTION AND SPORT-RELATED FACIAL FRACTURES: A CASE REPORT
Lilian Ries[1], Fausto Bérzin[2], *[1]Santa Catarina State Development University-UDESC, Brazil, [2]Piracicaba Odontology Faculty-UNICAMP, Brazil*

MPP1.10 ELECTROMYOGRAPHY ANALYSIS OF THREE DIFFERENT TYPES OF CHEST EXERCISES: INCLINED BENCH PRESS, INCLINED DUMBBELL PRESS AND INCLINED DUMBBELL FLY.
Jonas Gurgel[1], Rafael Engrazia[1], Flávia Porto[1], Gustavo Sepúlveda[1], Ismael Baldissera[1], Renata Vivian[1], Gabriel Espinosa[1], Ernesto Yee[1], Felipe Flores[1], *[1]Aerospace Biomechanics Research Laboratory / PUCRS, Brazil, [2]Physical Activity Assessments and Research Laboratory / PUCRS, Brazil*

MPP1.11 EFFECTS OF A HIGH-REPETITION RESISTANCE TRAINING PROGRAM ON METABOLIC AND NEUROMUSCULAR VARIABLES
Camila Greco[1], Anderson Oliveira[2], Marcelo Pereira[2], Tiago Figueira[1], Jailton Pelarigo[1], Vinícius Ruas[1], *[1]Human Performance Laboratory, Brazil, [2]Laboratory of Biomechanics, Brazil*

MPP1.12 ELECTROMYOGRAPHY ANALYSIS OF FOUR DIFFERENT TYPES OF ABDOMINALS EXERCISES
Jonas Gurgel[1], Ismael Baldissera[1], Flávia Porto[1], Ernesto Yee[1], Felipe Flores[1], Gustavo Sepúlveda[1], Renata Vivian[1], Gabriel Espinosa[1], Simone Padilha[2], Vinicius Duval Leite[2], *[1]Aerospace Biomechanics Research Laboratory, Brazil, [2]Physical Activity Assessments and Research Laboratory, Brazil*

MODELING AND SIGNAL PROCESSING MSP1

Saturday June 21, 10:30-11:30 and 15:00-16:30 Great Room A

MSP1.1 VARIANCE-BASED SIGNAL CONDITIONING IMPROVES SPIKE DETECTION IN MULTI-UNIT INTRA-FASCICULAR RECORDINGS
Ernest Nlandu Kamavuako[1], Ken Yoshida[1], Winnie Jensen[1], [1]*Aalborg University, Denmark,* [2]*IUPUI, United States*

MSP1.2 USE OF SPATIAL INFORMATION IN 2D ARRAY SEMG DECOMPOSITION
Casper T. Smit[1], Laura A.C. Kallenberg[1], Hermie J. Hermens[1], Hermie J. Hermens[2], [1]*Roessingh Research and Development, Netherlands,* [2]*University of Twente, Netherlands*

MSP1.3 EXTRACTION OF REPRESENTATIVE POTENTIALS IN MUAP SETS
Armando Malanda[1], Javier Navallas[1], Luis Gila[2], Javier Rodríguez[1], Ignacio Rodríguez[1], [1]*Universidad Pública de Navarra, Spain,* [2]*Hospital de Navarra, Spain*

MSP1.4 TEAGER-KAISER ENERGY OPERATOR IMPROVES SIGNAL-TO-NOISE RATIO OF EMG SIGNAL IN YOUNG AND OLD ADULTS
Stanislaw Solnik[1], Allison Gruber[1], Patrick Rider[1], Ken Steinweg[2], Joseph Helseth[1], Paul DeVita[1], Tibor Hortobágyi[1], [1]*Biomechanics Laboratory, East Carolina University, United States,* [2]*The Brody School of Medicine, East Carolina University, United States*

MSP1.5 REAL-TIME REDUCTION OF POWER LINE INTERFERENCE IN MULTI-CHANNEL SURFACE EMG
Luca Mesin[1], Andreas Boye[2], Amedeo Troiano[1], Roberto Merletti[1], Dario Farina[2], [1]*Laboratory for Engineering of the Neuromuscular System (LISiN), Department of Electronics, Italy,* [2]*Center for Sensory-Motor Interaction (SMI), Department of Health Science and Technology, Aalborg University, Denmark*

MSP1.6 A MAXIMUM LIKELIHOOD APPROACH FOR THE DETECTION OF MUSCULAR ACTIVATION TIMING
Silvia Conforto[1], Daniele Bibbo[1], Tommaso D'Alessio[1], [1]*Dipartimento di Elettronica Applicata, Università degli Studi di Roma TRE, Italy*

MSP1.7 THE EFFECT OF BACKGROUND MUSCLE ACTIVITY ON COMPUTERIZED DETECTION OF EMG ONSET AND OFFSET
Angela Lee[1], Jacek Cholewicki[1], Peter Reeves[1], [1]*Michigan State University College of Osteopathic Medicine, United States*

MSP1.8 APPLICATION OF THE HILBERT SPECTRUM FOR FEATURE EXTRACTION FROM ELECTROMYOGRAPHIC SIGNALS
Adriano Andrade[1], Alcimar Soares[1], [1]*Federal University of Uberlandia, Brazil*

MSP1.9 A NOTE ON THE SIGNAL-TO-NOISE RATIO OF MYO-ELECTRIC CHANNEL
Ning Jiang[1], Philip Parker[1], Kevin Englehart[1], [1]*University of New Brunswick, Fredericton, Canada*

MSP1.10 ASSESSMENT OF SURFACE EMG RMS AMPLITUDE DURING MAXIMUM VOLUNTARY CONTRACTION
Taian Vieira[1], Belmiro Salles[2], Liliam Oliveira[2], [1]*Polytechnic of Turin, Italy,* [2]*Federal University of Rio de Janeiro, Brazil*

MSP1.11 RELIABILITY OF THE MECHANOMYOGRAM DURING INCREMENTAL ISOMETRIC MUSCLE ACTIONS
Trent Herda[1], Eric Ryan[1], Pablo Costa[1], Jason DeFreitas[1], Travis Beck[1], Joel Cramer[1], [1]*University of Oklahoma, United States*

MSP1.12 IN VIVO PASSIVE FORCES OF THE INDEX FINGER MUSCLES: VALIDATING CURRENT MODELS

Jin Qin[1], David Lee[1], Jack Dennerlein[1], [1]*Harvard University, United States*

MSP1.13 A PHYSIOLOGICALLY ACCURATE MOTOR UNIT ARCHITECTURE MODEL
Javier Navallas[1], Armando Malanda[1], Luis Gila[2], Javier Rodriguez[1], Ignacio Rodriguez[1], [1]*Universidad Publica de Navarra, Spain,* [2]*Hospital Virgen del Camino, Spain*

MSP1.14 ESTIMATION OF MONOPOLAR SIGNALS FROM SPHINCTER MUSCLES
Luca Mesin[1], [1]*Laboratory for Engineering of the Neuromuscular System (LISiN), Department of Electronics, Polytechnic of Turin, Italy*

MSP1.15 EMG-TO-TORQUE MODELING: INTERPRETING THE ROLE OF MUSCLES CROSSING THE KNEE DURING AN IMPACT LANDING
Jeffery Podraza[1], Scott White[1], [1]*University at Buffalo, United States*

MOTOR UNITS MUP1
Friday June 20, 10:30-11:30 and 16:30-17:30 Great Room A

MUP1.1 RELATIONSHIPS BETWEEN MOTOR UNIT SIZE AND RECRUITMENT THRESHOLD IN YOUNG AND OLDER ADULTS
Brett W Fling[2], Kim Johnson[1], Christopher A Knight[3], Anita Christie[1], Gary Kamen[1], [1]*University of Massachusetts, United States,* [2]*University of Michigan, United States,* [3]*University of Delaware, United States*

MUP1.2 EMGLAB WEBSITE FOR EMG DECOMPOSITION
Kevin McGill[1], Edward Clancy[2], [1]*VA Palo Alto Health Care System, United States,* [2]*Worcester Polytechnic Institute, United States*

MUP1.3 MOTOR UNIT DISCHARGE IN SOLEUS MUSCLE AND CORRELOGRAM WITH CENTER OF PRESSURE PARAMETERS
Germán Vásquez[1], David Arriagada[1], Luis Campos[1], Susana Vargas[1], Rony Silvestre[1], [1]*Universidad Mayor, Chile*

MUP1.4 SPATIAL VARIABILITY OF CORTICOMUSCULAR COHERENCE
Francesco Negro[1], Kristian Hennings[1], Dario Farina[1], [1]*Aalborg Universitet, Denmark*

NEUROPHYSIOLOGY NPP1
Saturday June 21, 10:30-11:30 and 15:00-16:30 Great Room A

NPP1.1 MUSCULAR RESPONSE TO QUASI-STATIC TENSILE LOADING OF THE CERVICAL FACET JOINT CAPSULE
Nadia Azar[1], Srinivasu Kallakuri[2], Chaoyang Chen[2], John Cavanaugh[2], [1]*University of Windsor, Canada,* [2]*Wayne State University, United States*

NPP1.2 COMPARISON OF TWO METHODS OF DETERMINING THE LATENCY OF MOTOR EVOKED POTENTIALS TO THE INFRASPINATUS MUSCLE
Lisa Sheehy[1], Linda McLean[1], [1]*Queen's University, Canada*

NPP1.3 NEURAL MECHANISM IN BRAIN UPON PRE-DETERMINED RESPONSE TO PERIODIC AND APERIODIC STIMULI - ANALYSIS OF VISUAL STIMULATION OF MOTOR MOVEMENT
Matsuda Tadamitsu[1], Watanabe Shu[2], Kuruma Hironobu[2], Murakami Yoshiyuki[3], Watanabe Rui[4], Ikeda Yumi[2], Atushi Senoo[2], Hosoda Masataka[1], Yonemoto Kyozou[1], [1]*Ryotokuji University, Japan,* [2]*Tokyo Metropolitan University, Japan,* [3]*Edogawa Medical Special School, Japan,* [4]*Kiyose Rehabilitation Hospital, Japan*

NPP1.4 ElECTROPHYSIOLOGICAL STUDIES IN PATIENTS WITH SUBACUTE MYELO-OPTICO-

NEUROPATHY BY MAGNETIC STIMULATION
Akihisa Matsumoto[1], Yasutaka Tajima[1], Hidenao sasaki[1], *[1]Department of Neuroligy,Sapporo City Hospital, Japan*

NPP1.5 ELECTROMYOGRAPHY OF MASTICATORY MUSCLES IN PARTIALLY EDENTULOUS CHILDREN AFTER ORAL REHABILITATION
Maria Beatriz Gavião[1], Flávia Gambareli[1], Marcia Serra-Vicentin[1], Camila Rocha[1], *[1]State University of Campinas, Piracicaba Dental School, Brazil*

NPP1.6 A MODEL OF THE MUSCLE-FIBER CONDUCTION-VELOCITY RECOVERY FUNCTION
Kevin McGill[1], Zoia Lateva[1], *[1]VA Palo Alto Health Care System, United States*

NPP1.7 DO SEX, PAIN AND THE OVARIAN CYCLE INFLUENCE MASTICATORY MUSCLE ACTIVITY IN SUBJECTS WITH TEMPOROMANDIBULAR DISORDERS?
Mariana Trevisani Arthuri[1], Gustavo Hauber Gameiro[1], Tatiane de Freitas Salvador[1], Frederico Andrade e Silva[1], Fausto Bérzin[1], Maria Cecília Ferraz de Arruda Veiga[1], *[1]Faculty of Dentistry of Piracicaba - University of Campinas – Unicamp, Brazil*

NPP1.8 SCALENE MUSCLE ACTIVITY DURING PROGRESSIVE INSPIRATORY LOADING UNDER PRESSURE SUPPORT VENTILATION IN NORMAL HUMANS
Linda Chiti[1], Giuseppina Biondi[1], Capucine Morelot-Panzini[1], Mathieux Raux[1], Thomas Similowski[1], Francois Hug[3], *[1]university Paris6, France, [2]groupe Hospitalier Pitié Salpêtrière, France, [3]university Of Nantes, France*

NPP1.9 EFFECT OF PERIPHERAL SENSORY INPUTS ON SOLEUS H-REFLEX DURING ROBOT-INDUCED PASSIVE STEPPING IN HUMAN
Kimitaka Nakazawa[1], Kiyotaka Kamibayashi[1], Tsuyoshi Nakajima[1], Makoto Takahashi[1], Masako Fujita[1], Tetsuya Ogawa[1], Masami Akai[1], *[1]National rehabilitation center for persons with disabilities, Japan*

NPP1.10 NOVEL INSIGHTS INTO CRAMP PATHOPHYSIOLOGY FROM M-WAVE ANALYSIS DURING CRAMP DISCHARGE
Marco Alessandro Minetto[1], Alberto Botter[2], Domenico De Grandis[3], Roberto Merletti[2], *[1]Division of Endocrinology and Metabolism, Department of Internal Medicine, University of Turin, Italy, [2]Laboratory for Engineering of the Neuromuscular System, Department of Electronics, Polytechnic of Turin, Italy, [3]Division of Neurology, Department of Neuroscience, Civile Hospital Santa Maria della Misericordia, Italy*

NPP1.11 MOTOR RESPONSE EVOKED DURING PROLONGED STIMULATION TACTILE: ELECTROMIOGRAPHIC AND KINEMATIC EVIDENCE.
Fresia Vargas[2], Consuelo Calderón[2], German Vásquez[1], David Arriagada[1], Luis Campos[1], Susana Matzner[1], Rony Silvestre[1], Antoni Valero-Cabre[3], *[1]Universidad Mayor, Chile, [2]TEDES, Chile, [3]Boston University, United States*

POSTURE AND BALANCE PBP1
Friday June 20, 10:30-11:30 and 16:30-17:30 Great Room A

PBP1.1 EFECTS OF SENSORY STIMULI ON POSTURAL CONTROL PROSPECTIVE CASE STUDY.
Edith Elgueta[1], Valeska Gatica[1], *[1]university Of Talcaca, Chile*

PBP1.2 OCCLUSION STATUS EVALUATED DURING MEASUREMENTS OF DYNAMIC BALANCE BY MEASURING MASSETER ACTIVITY USING EMG SYSTEMS.
Masataka Hosoda[1], Tadashi Masuda[2], Koji Isozaki[1], Kiyomi Takayanagi[3], Hideki Moriyama[3], Ken Nishihara[3], Noboru Sakanoue[1], Shigeki Miyajima[1], Tadamitsu Matsuda[1], Akira Takanashi[1], Kotomi Shiota[1], Sadao Morita[4], *[1]department Of Physical Therapy, Ryotokuji University, Japan, [2]laboratory Of Biosystem Modeling, School Of Biomedical Science, Tokyo Medical And Dental*

University, Japan, [3]department Of Physical Therapy, Saitama Prefectural University, Japan, [4]department Of Rehabilitation Medicine, Tokyo Medical And Dental University Graduate School, Japan

PBP1.3 ELECTROMYOGRAPHIC ANALYSIS OF THE ANTERIOR, MEDIAL AND POSTERIOR PARTS OF THE TEMPORAL MUSCLE DURING MASTICATION
Fausto Bérzin[1], Mirian Nagae[1], Marcelo Alves[1], Maria Bérzin[1], *[1]State University of Campinas, Brazil*

PBP1.4 ELECTROMYOGRAPHIC AND CEPHALOMETRIC CORRELATION OF MANDIBULAR BIOMECHAMICAL WITH THE PREDOMINANT MASTICATORY MOVEMENT
Maria Coelho-Ferraz[1], Cesar Amorim[1], Mário Vedovello Filho[1], Fausto Bérzin[1], *[1]Araras Dental College, Brazil*

PBP1.5 THE CORRELATION OF MOBILE TRACING OF SOMATIC GRAVITY CENTER AT THE TIME OF STANDING UP WITH ACTING SPEED
Osamu Nitta[1], Yorimitsu Furukawa[1], Yumi Ikeda[1], Hironobu Kuruma[1], Ken Yanagisawa[1], Johon Surya[1], *[1]Tokoyo Metropolitan University, Japan*

PBP1.6 DOES KNEE ANGLE ALIGNMENT AFFECTS POSTURAL CONTROL?
Cassio Siqueira[1], Gabriel Moya[1], Rene Caffaro[1], Isabel Sacco[1], André Kohn[1], Clarice Tanaka[1], *[1]University of São Paulo, Brazil*

PBP1.7 COORDINATION TRAINING OF POSTURAL MUSCLES - DOES IT IMPROVE POSTURAL CONTROL?
Marie Birk Jørgensen[1], Jørgen Skotte[1], Mette Kreutzfeld Zebis[1], Karen Søgaard[1], Gisela Sjøgaard[1], *[1]National Research Centre for the Working Environment, Denmark*

PBP1.8 CENTER OF PRESSURE EXCURSION DURING VOLUNTARY TRUNK MOVEMENT IN SITTING
Richard Preuss[1], Milos Popovic[2], *[1]Toronto Rehabilitation Institute Lyndhurst Centre, Canada, [2]University of Toronto, Institute of Biomaterials and Biomedical Engineering, Canada*

PBP1.9 THE RELATIONSHIP BETWEEN MUSCLE POWER AND BALANCE IN POSTURAL CONTROL
Kotomi Shiota[1], Akira Takanashi[1], Shigeki Miyajima[1], Tadamitsu Matsuda[1], Makoto Ikeda[2], *[1]Ryotokuji University, Japan, [2]Tokyo Metropolitan University, Japan*

PBP1.10 ELECTROMYOGRAPHIC ANALYSIS OF THE STERNOCLEIDOMASTOIDEUS MUSCLE DURING HEAD DYNAMIC MOVEMENTS
Fausto Bérzin[1], Maria Aranha[1], Fabiana Forti[1], Daniela Silva[1], *[1]State University of Campinas, Brazil*

PBP1.11 FUNCTIONAL ANALYSIS OF THE ANTERIOR, MEDIAL AND POSTERIOR PARTS OF TEMPORAL MUSCLE DURING MOVEMENTS OF THE MANDIBLE – ELECTROMYOGRAPHIC STUDY
Fausto Bérzin[1], Mirian Nagae[1], Marcelo Alves[1], Maria Bérzin[1], *[1]State University of Campinas, Brazil*

POSTURE AND BALANCE PBP2
Saturday June 21, 10:30-11:30 and 15:00-16:30 Great Room A

PBP2.1 EFFECT OF THE APPLICATION OF AN EMBEDDED ENVIRONMENT ON THE MIGRATION OF THE COP DURING THE MAINTENANCE OF THE BIPED POSITION IN YOUNG WOMEN
Alex Araneda[1], Rodrigo Guzmán[1], *[1]Santo Tomás University, Chile*

PBP2.2 THE VELOCITY OF THE CENTER OF PRESSURE FLUCTUATION DURING QUIET STANDING

REFLECTS THE NEURAL MODULATION OF THE ANKLE TORQUE
Kei Masani[1], Albert Vette[1], Milos Popovic[1], [1]*University of Toronto, Canada,* [2]*Toronto Rehab, Canada*

PBP2.3 MODELING THE NEUROMUSCULAR CONTROL FOR HUMAN ERECT POSTURE
Eduardo Naves[1], Adriano Pereira[1], Adriano Andrade[1], Alcimar Soares[1], [1]*Federal University of Uberlandia, Brazil*

PBP2.4 INFLUENCE OF VISUAL INPUT IN MOTOR STRATEGIES IN YOUNGS ADULTS
Rony Silvestre[1], Angelo Bartsch[2], Cristian Cuadra[2], Pablo Ortega[2], Marinella Razeto[2], [1]*Universidad Mayor, Chile,* [2]*Universidad de Playa Ancha, Chile*

PBP2.5 INFLUENCE OF THE COMBINED USE OF IMPROVING STRENGTH OF THE MAXIMAL GLUTEAL MUSCLE AND STRETCHING OF THE GREATER PSOAS MUSCLE ON AN ANGLE OF INCLINATION OF PELVIS AND SWAY OF GRAVITY CENTER
Nanami Ikeda[1], Yuuko Ishizaki[1], Yukio Kubota[1], Hitoshi Takei[2], [1]*Kawakita General Hospital, Japan,* [2]*Tokyo Metropolitan University, Japan*

PBP2.6 QUANTIFICATION OF FORWARD HEAD POSTURE AND ANALYSIS OF STERNOCLEIDOMASTOID AND UPPER TRAPEZIUS PROPERTIES IN ASYMPTOMATIC YOUNG ADULTS
Hsin-Min Lee[1], Jia-Yuan You[1], [1]*Department of Physical Therapy, I-Shou University, Taiwan*

PBP2.7 INTENSIVE RETURN TO WORK REHABILITATION IMPROVES STABILIZATION PATTERNS IN WHIPLASH-ASSOCIATED DISORDER INDIVIDUALS.
Nancy St-Onge[1], Julie N. Côté[2], Isabelle Patenaude[2], Joyce Fung[2], [1]*Concordia University, Canada,* [2]*McGill University, Canada,* [3]*Constance-Lethbridge Rehabilitation Center, Canada,* [4]*Jewish Rehabilitation Hospital, Canada*

PBP2.8 EFFECT OF FATIGUE AND PROTECTIVE CLOTHING ON FUNCTIONAL BALANCE OF FIREFIGHTERS
Pilwon Hur[1], Karl Rosengren[1], Gavin Horn[1], Denise Smith[1], Elizabeth Hsiao-Wecksler[1], [1]*University of Illinois at Urbana-Champaign, United States*

PBP2.9 ELECTROMYOGRAPHY OF ORBICULARIS ORIS MUSCLE IN CHILDREN
Camila Rocha[1], Paulo Cária[1], Fausto Bérzin[1], Maria Beatriz Gavião[1], [1]*State University of Campinas, Piracicaba Dental School, Brazil*

PBP2.10 RATE OF FORCE DEVELOPMENT AND CO-ACTIVATION AROUND THE KNEE IN ELDERLY FALLERS
Mauro Gonçalves[1], Marcelo Pereira[1], Patricia Aguiar[1], [1]*SAO PAULO STATE UNIVERSITY, Brazil*

PHYSICAL MEDICINE AND REHABILITATION PMP1
Thursday June 19, 10:30-11:30 and 16:30-17:30 Great Room A

PMP1.1 HOFFMANN REFLEX FROM THE VASTUS MEDIALIS OBLIQUE MUSCLE DURING TRACTION OF THE LEG
Yoshitsugu Tanino[1], Kyosuke Takasaki[1], Shinichi Daikuya[2], Tetsuei Kawano[3], Toshiaki Suzuki[1], [1]*Kansai University of Health Sciences, Japan,* [2]*Kishiwada Eishinkai Hospital, Japan,* [3]*Hachisuba Ckinic, Japan*

PMP1.2 ANALYSIS OF THE ELECTROMYOGRAPHIC ACTIVITY AND THE FORCE OF KNEE EXTENSOR MUSCLES BEFORE AND AFTER TWO DIFFERENT ELECTROSTIMULATION TRAINING PROGRAMS.
Fabiana Forti[1], Janaína Moraes[2], Rinaldo Guirro[3], Daniel Sakabe[2], Fausto Bérzin[1], [1]*PPG em*

Biologia Buco Dental, Faculdade de Odontologia de Piracicaba, Universidade Estadual de Campinas, Brazil, [2]Faculdades Integradas Einstein de Limeira, Brazil, [3]PPG em Fisioterapia, Universidade Metodista de Piracicaba, Brazil

PMP1.3 EFFECT OF INDUCED ISCHEMIA ON ELECTROMYOGRAPHIC FREQUENCY
Clívea Bandeira[1], Kelly Berni[1], Delaine Bigaton[1], Cristiane Pedroni[2], *[1]Methodist University of Piracicaba, Brazil, [2]Piracicaba Dental College – Campinas State University, Brazil*

PMP1.4 CONTROLLER OF ULTRASONIC MOTOR FOR REHABILITATION ROBOT
Toshihiro Kawase[1], Kazuo Kurashige[1], Masato Watanabe[1], Hiroyuki Kambara[1], Yasuharu Koike[1], *[1]Tokyo Institute of Technology, Japan*

PMP1.5 EVALUATION PROCESS FOR AGONIST AND ANTAGONIST MUSCLE ACTIVITIES DURING ACTIVE AND PASSIVE WALKING WITH LOKOMAT
Yosuke Ichishima[1], Tsuyoshi Nakajima[2], Kiyotaka Kamibayashi[2], Kimitaka Nakazawa[2], Tohru Kiryu[1], *[1]Niigata University, Japan, [2]Research Institute of National Rehabilitation Center, Japan*

PMP1.6 REDUCTION OF FORCE POTENTIATION INDUCED BY RANDOMIZED FREQUENCY ELECTRICAL STIMULATION: POTENTIAL METHOD FOR THE PREVENTION OF MUSCLE FATIGUE
Noritaka Kawashima[1], Sureshkumar Ahthavan[2], Kei Masani[1], Milos Popovic[1], *[1]Toronto Rehabilitation Institute, Canada, [2]University of Toronto, Canada*

PMP1.7 DEVELOPMENT OF A ROBOTIC ARM FOR UPPER LIMB CONTROLED BY ELECTROMIOGRAPHY IN REAL TIME COMBINED WITH FES, FOR THE RESTARUATION OF THE FUNCTION TO THE UPPER LIMB IN PATIENT WITH SPINAL CORD DAMAGE AND STROKE
Luis Campos[1], Karla Lever[1], Rony Silvestre[2], *[1]Universidad Mayor, Chile, [2]Hospital del Trabajador, Chile*

PMP1.8 PATHOLOGIC NORMOREFLEXIA DUE TO COEXISTING UPPER AND LOWER MOTOR NEURON CONDITIONS; TWO CASE REPORTS.
Waqaas Quraishi[1], Barry Root[1], *[1]Long Island Jewish Medical Center, United States*

PMP1.9 THE EFFECT OF ELECTRODE DISLOCATION ON SURFACE ELECTROMYOGRAPHY AMPLITUDE: IMPLICATIONS FOR AMBULATORY MEASUREMENTS
Ruud de Nooij[1], Laura A.C. Kallenberg[1], Hermie J. Hermens[1], Hermie J. Hermens[2], *[1]Roessingh Research and Development, Netherlands, [2]University of Twente, Faculty of Electrical Engineering, Mathematics and Computer Science, Netherlands*

PMP1.10 ON THE USE OF ELASTIC STRAPS TO IMPROVE GONIOMETER MEASUREMENTS
Tatiana de Oliveira Sato[1], Heleodório Honorato dos Santos[1], Ana Beatriz de Oliveira[1], Ana Maria Forti Barela[2], Helenice Jane Cote Gil Coury[1], Tania de Fátima Salvini[1], *[1]Universidade Federal de São Carlos, Brazil, [2]Universidade Cruzeiro do Sul, Brazil*

PMP1.11 ACTIVATION OF THE PREFRONTAL CORTEX DURING LEARNING BY MOBILE SOFTWARE
Shu Watanabe[1], Kanji Akahori[2], Hironobu Kuruma[1], Masahiro Matsuda[1], Yoshiyuki Murakami[1], Kyozo Yonemoto[1], Masao Inoue[3], *[1]Tokyo Metropolitan University, Japan, [2]Tokyo Institute of Technology, Japan, [3]Shimadzu Corporation, Japan*

PMP1.12 A METHOD FOR BETTER POSITIONING BIPOLAR ELECTRODES FOR LOWER LIMB EMG RECORDINGS
Aline Gomes[1], Andrea Onodera[1], Mitie Otuzi[1], Denise Pripas[1], Isabel Sacco[1], *[1]University of Sao Paulo, Brazil*

PHYSICAL MEDICINE AND REHABILITATION PMP2

Thursday June 19, 10:30-11:30 and 16:30-17:30 Great Room A

PMP2.1 MUSCULAR EFFECTS OF POSTERIOR CROSSBITE TREATMENT
Moara De-Rossi[1], Maria Beatriz Duarte Gavião[1], Janaina De-Rossi[1], Matias Vitti[1], Simone Cecilio Hallak Regalo[1], [1]*State University of Campinas, Brazil*

PMP2.2 ON THE USE OF EMG AMPLITUDE RATIOS TO ASSESS THE COORDINATION OF BACK MUSCLES
Christian Larivière[1], A. Bertrand Arsenault[2], [1]*Occupational Health and Safety Research Institute Robert-Sauvé (IRSST), Canada,* [2]*School of rehabilitation, University of Montreal, Canada*

PMP2.3 EVALUATION OF THE FATIGABILITY OF THE CERVICAL EXTENSOR MUSCLES WHILE DOING THE NECK EXTENSOR ENDURANCE TEST IN PATIENTS WITH MYOGENOUS TEMPOROMANDIBULAR DISORDERS AND HEALTHY SUBJECTS: PRELIMINARY RESULTS
Susan Armijo Olivo[1], Rony Silvestre[2], Jorge Fuentes[1], Bruno Da Costa[1], Inae Gadotti[1], David Magee[1], [1]*University of Alberta, Canada,* [2]*Mayor University, Chile*

PMP2.4 ELECTROMYOGRAPHIC EVALUATION OF THE PERFORMANCE OF FLEXOR CERVICAL MUSCLES IN PATIENTS WITH TEMPOROMANDIBULAR DISORDERS WHEN EXECUTING THE CRANIOCERVICAL FLEXION TEST (CCFT): PRELIMINARY RESULTS
Susan Armijo Olivo[1], Rony Silvestre[2], Jorge Fuentes[1], Bruno Da Costa[1], Inae Gadotti[1], David Magee[1], [1]*University of Alberta, Canada,* [2]*Mayor University, Chile*

PMP2.5 ELECTROMYOGRAPHIC STUDY OF PATIENTS WITH MASTICATORY MUSCLES DISORDERS, PHYSIOTHERAPEUTIC AND ODONTOLOGIC TREATMENT: RANDOMIZED CONTROLLED TRIALS
Daniela Biasotto-Gonzalez[1], João Corrêa[1], Manoela Martins[1], Daniel Andrade[1], Luciane Jesus[1], [1]*UNINOVE, Brazil*

PMP2.6 OCCLUSAL ADJUSTMENT INFLUENCE ON THE MASTICATORY MUSCLES EMG IN A PACIENT TREATED BY FUNCTIONAL MAXILLARY ORTHOPEDICS – A STUDY OF CASE
Stela Wilhelmsen[1], Liege Ferreira[1], Cristiane Pedroni[1], José Santos[1], Fausto Bérzin[1], [1]*Piracicaba Dental College – Campinas State University, Brazil*

PMP2.7 EFFECT OF HVES IN PATIENTS WITH TMD – ELECTROMYOGRAPHIC EVALUATION
Ana Flávia Almeida[1], Kelly Berni[1], Delaine Bigaton[1], Cristiane Pedroni[2], [1]*1Methodist University of Piracicaba, Brazil,* [2]*Piracicaba Dental College – Campinas State University, Brazil*

PMP2.8 DIAGNOSIS CONTRIBUITION OF SURFACE ELECTROMYOGRAPHY FOR TEMPOROMANDIBULAR DISORDERS
Cristiane Pedroni[1], Anamaria Oliveira[2], Eduardo Sakai[3], Delaine Bigaton[4], Ana Flavia Almeida[4], Clívea Bandeira[4], Marcelo Alves[1], Fausto Bérzin[1], [1]*Piracicaba Dental College – Campinas State University, Brazil,* [2]*Ribeirão Preto Medical School, University of São Paulo, Brazil,* [3]*3Araras Dental College, UNIARARAS, Brazil,* [4]*Methodist University of Piracicaba, Brazil*

PMP2.9 IMMEDIATE ALTERATIONS IN SUPRAHYOIDS MUSCLES KINETICS AFTER FUNCTIONAL MAXILLARY ORTHOPEDICS APPLIANCE INSTALLATION – AN ELECTROMYOGRAPHIC STUDY
Eduardo Sakai[1], Fausto Bérzin[1], Cristiane Pedroni[1], Denise Santos[1], Marcelo Alves[1], [1]*Araras Dental College, Brazil*

PMP2.10 COMPARISON OF WATER DEGLUTITION DATA WITH AND WITHOUT FUNCTIONAL MAXILLARY ORTHOPEDIC APPLIANCE , REGISTERED WITH SURFACE ELECTRODES
Eduardo Sakai[1], Cristiane Pedroni[1], Cristina Fiuza[1], Luciano Ribeiro[1], [1]*Araras Dental College, Brazil*

PMP2.11 EMG PROFILE OF BRAZILIAN WOMEN WITH CHRONIC STRESS AND

TEMPOROMANDIBULAR DISFUNCTION
Maria Bérzin[1], Fausto Bérzin[1], Marcelo Alves[1], [1]*State University of Campinas, Brazil*

PMP2.12 EFFECTS ON EXCITABILITY CHANGES IN HUMAN PRIMARY MOTOR CORTEX (M1) ARE VARIABLE DEPENDENT ON CHARACTERISTICS OF FUNCTIONAL ELECTRICAL STIMULATION (FES)
Kenichi Sugawara[1], Shigeo Tanabe[2], Toshio Higashi[1], Takamasa Tsurumi[1], Tatsuya Kasai[3], [1]*Kanagawa University of Human Services, Japan,* [2]*Kinjo University, Japan,* [3]*Hiroshima University, Japan*

PHYSICAL MEDICINE AND REHABILITATION PMP3
Friday June 20, 10:30-11:30 and 16:30-17:30 Great Room A

PMP3.1 THE EFFECTS OF JOINT TRACTION AND POSITION OF UPPER LIMB ON REACTION TIME OF QUADRICEPS FEMORIS
Takayuki Koyama[1], Ken Yanagisawa[2], Osamu Nitta[2], Jun-ya Aizawa[3], [1]*Surugadai Nihon University Hospital, Japan,* [2]*Tokyo Metropolitan University, Japan,* [3]*Ryotokuji University, Japan*

PMP3.2 THE EFFECT OF INCREASING THE INSTABILITY OF WEIGHT BEARING SURFACE ON THE ACTIVITY OF THE PERIARTHICULAR SHOULDER MUSCLES DURING CLOSE CHAIN EXERCICES
khosro Kalantari[1], Mehri Ghasemi[1], Simin Berenji[1], [1]*Shaheed Beheshti Medical University, Iran, Islamic Republic of*

PMP3.3 A CASE REPORT OF A PATIENT WITH MYOFASCIAL PAIN SYNDROME: AN ELECTROMYOGRAPHIC STUDY.
Maria Fernanda M. Aranha[1], Cláudia Lopes Duarte[1], Sílvio Martins[2], Fausto Bérzin[1], [1]*State University of Campinas, Brazil,* [2]*Sugar Cane Suppliers Hospital, Brazil*

PMP3.4 INFLUENCE OF ANKLE POSITION IN THE BIOMECHANICS OF PILATES' FOOTWORK EXERCISES
Bergson Queiroz[1], Daniella Gomes[1], Cristina Abrami[2], Isabel Sacco[1], [1]*University of Sao Paulo, Brazil,* [2]*Centro de Ginástica Postural Angélica, Brazil*

PMP3.5 INFLUENCE OF FUNCTIONAL INSTABILITY IN ANKLE MUSCULAR ACTIVITY DURING A CUTTING MOVEMENT IN VOLLEYBALL PLAYERS
Eneida Yuri Suda[1], Paula M H Akashi[1], Isabel C N Sacco[1], [1]*University of São Paulo, Brazil,* [2]*Centro Universitário Capital, Brazil,* [3]*Universidade do Grande ABC, Brazil*

PMP3.6 EMG ONSET TIMING DURING A CUTTING MANEUVER IN VOLLEYBALL PLAYERS WITH FUNCTIONAL ANKLE INSTABILITY
Eneida Yuri Suda[1], Isabel C N Sacco[1], [1]*University of São Paulo, Brazil,* [2]*Centro Universitário Capital, Brazil,* [3]*Universidade do Grande ABC, Brazil*

PMP3.7 CONTRALATERAL EFFECTS FOLLOWING UNILATERAL SUBMAXIMAL ECCENTRIC AND CONCENTRIC CONTRACTIONS TRAINING
Azusa Uematsu[1], Hirofumi Kobayashi[1], Shuji Suzuki[1], [1]*Waseda University, Japan*

PMP3.8 EMG DATA OF THE KNEE MUSCLES CHANGES DEPENDING ON THE DIRECTION OF PASSIVE RESISTANCE DURING THE CLOSED KINETIC CHAIN TASK
Nobuhiro Aoki[1], Fuminari Kaneko[2], Tatsuya Hayami[3], Akihiro Kanamori[3], Yasuyoshi Wadano[1], [1]*Ibaraki Prefectural University of Health Sciences, Japan,* [2]*National Institute of Advanced Industrial Science and Technology, Japan,* [3]*Tsukuba University, Japan*

PMP3.9 EFFECT OF ACUPUNCTURE TREATMENT ON MASTICATORY MUSCLES IN INDIVIDUALS WITH TEMPOROMANDIBULAR DISORDERS INDIVIDUALS

Simone Cecilio Hallak Regalo[1], Sandra Valéria Rancan[1], Jaime Eduardo Cecilio Hallak[1], Solange Aparecida Bataglion[1], Adriana Pedersoli[1], Marisa Semprini[1], Patrícia Tiemy Hotta[1], César Bataglion[1], [1]*University of São Paulo, Brazil*

PMP3.10 HAMSTRING MUSCLE CONTRACTION AND TIBIAL POSTERIOR TRANSLATION UNDER ANTERIOR DRAWER TRACTION USING SLING FOR THE PATIENTS WITH PCL INSUFFICIENCY

Takahiro Sakai[1], Maki Koyanagi[2], Masaki Yoshida[2], Kumi Akataki[2], Hirohisa Fujisaki[2], Tomohisa Yamagata[2], Mutsumi Sato[3], Kuniyuki Hidaka[3], Ken Nakata[4], [1]*Shijonawate Gakuen University, Japan,* [2]*Osaka Electro-Communication University, Japan,* [3]*Osaka University Hospital, Japan,* [4]*Osaka University Graduate School of Medicine, Japan*

PMP3.11 EMG ACTIVITY OF THE MASTICATORY MUSCLES IN COMPLETE DENTURE WEARING INDIVIDUALS

Patricia Hotta[1], Takami Hotta[1], Mathias Vitti[1], Elaine Coronatto[1], Cesar Bataglion[1], Paulo Vasconcelos[1], Rogerio Pavão[1], Simone Regalo[1], [1]*University of São Paulo, Brazil*

PMP3.12 ELECTROMYOGRAPHIC AND BITE FORCE MODIFICATIONS CAUSED IN MASTICATORY MUSCULATURE OF BRAZILIAN WHITE CIVILIZATION WHEN COMPARED TO BRAZILIAN INDIGENOUS CIVILIZATION

Carla Santos[1], Mathias Vitti[1], Wilson Mestriner Junior[1], Paulo Vasconcelos[1], Jaime Hallak[2], Fernando Dias[1], Carlos Regalo[1], Simone Regalo[1], [1]*Ribeirão Preto School of Dentistry – University of São Paulo, Brazil,* [2]*Ribeirão Preto School of Medicine – University of São Paulo, Brazil*

PHYSICAL MEDICINE AND REHABILITATION PMP4

Friday June 20, 10:30-11:30 and 16:30-17:30 Great Room A

PMP4.1 IMMEDIATE EFFECT OF THE SELECTIVE NEUROMUSCULAR ELECTRICAL STIMULATION OF THE VASTUS MEDIALIS OBLIQUE MUSCLE

Jamilson Brasileiro[1], Cesar Amorim[1], Denise Augusto[1], Paula Ventura[1], João Nogueira[1], [1]*Universiade Federal do Rio Grande do Norte, Brazil,* [2]*Universidade Estadual Paulista, Brazil*

PMP4.2 LOCAL HEAT COMBINED WITH ELECTRICAL STIMULATION TO INCREASE SKIN BLOOD FLOW: A POTENTIAL MODALITY FOR WOUND HEALING

Abdul-Majeed Al-Malty[1], Jerrold Petrofsky[1], [1]*The Hashemite University, Jordan,* [2]*Loma Linda University, United States*

PMP4.3 ANALYSIS OF HEART RATE RECOVERY TIME IN EXERCISE OF DEEP TRUNK MUSCLES USING TWO DEVICES

Yorimitsu Furukawa[1], Osamu Nitta[1], Hitoshi Takei[1], Hironobu Kuruma[1], Osamu Nkamata[2], Yumi Ikeda[1], Nami Shida[1], Makoto Ikeda[1], Ken Yanagisawa[1], [1]*Tokyo Metropolitan University, Japan,* [2]*Bukyo Gakuin University, Japan*

PMP4.4 DIABETIC NEUROPATHY PROGRESSION EFFECTS IN GROUND REACTION FORCES AND IN LOWER LIMB EMG DURING BAREFOOT GAIT

Paula Akashi[1], Isabel Sacco[1], [1]*University of Sao Paulo, Brazil*

PMP4.5 ALTERATION IN ACTIVITY PATTERN OF MASTICATORY MUSCLES OF CHILDREN WITH CEREBRAL PALSY

Lilian Ries[1], Fausto Bérzin[2], [1]*Santa Catarina State Development University-UDESC, Brazil,* [2]*Piracicaba Odontology Faculty-UNICAMP, Brazil*

PMP4.6 THE CHARACTERISTICS OF MOTION OF HEMIPLEGIC PATIENTS WHILE EATING WITH NON- DOMINANT HAND

Asako Matsubara[1], Tsuneji Murakami[1], Tatsunori Sawada[2], [1]*Hiroshima General Rehabilitation*

Center, Japan, [2]Seirei Christopher University, Japan

PMP4.7 MUSCULAR ACTIVITY OF SPINAL ERECTORS AND RECTUS ABDOMINAL IN PATIENTS WITH ENCEPHALOPATHY DURING THERAPEUTIC RIDING
Fabio Cyrillo[1], Rebeca Santos[2], Fernanda Borges[2], *[1]UNICID, Brazil, [2]HIPICA STO AMARO - FUND. SELMA, Brazil*

PMP4.8 ASYMMETRY INDEX OF MASTICATORY MUSCLE AS A PARAMETER FOR ANALYSIS OF THE TMD TREATMENT OUTCOMES
Cláudia Maria de Felício[1], Melissa O. Melchior[1], Cláudia L.P Ferreira[1], Marco A.M. Rodrigues da Silva[2], *[1]Faculdade de Medicina de Ribeirão Preto-Universidade de São Paulo, Brazil, [2]Faculdade de Odontotologia de Ribeirão Preto-Universidade de São Paulo, Brazil*

PMP4.9 COMPARATIVE SUDY OF ELECTROMYOGRAPHIC ACTIVITY OF VENTILATORY MUSCLES DURING MAXIMAL SUSTAINED INSPIRATION EXERCISE, FLOW AND VOLUME RESPIRATORY STIMULATOR
Fabio Cyrillo[1], Roberta Lima[2], Yara Teixeira[2], Camila Torriani[2], Rodrigo Raimundo[2], *[1]UNICID, Brazil, [2]FMU, Brazil*

PMP4.10 LATISSIMUS DORSI AND ITS ROLE IN BREATHING PROCESS
Fabio Cyrillo[1], Tais Brasil[2], Camila Torriani[2], *[1]UNICID, Brazil, [2]FMU, Brazil*

PMP4.11 INFLUENCE OF POSITIONAL RELEASING THERAPY (PRT) ON TRICEPS SURAE MUSCLE DURING GAIT
Fabio Cyrillo[2], Juliana Sardinha[1], Marcio Conte[1], Camila Torriani[1], Fabiane Costa[1], *[1]FMU, Brazil, [2]UNICID, Brazil*

PMP4.12 FORCE AND SURFACE ELECTROMYOGRAPHY PATTERN OF SHOULDER MUSCLES IN WOMEN SUBMITTED TO PATEY MODIFIED RADICAL MASTECTOMY
Juliana Resende[1], Thiago Pereira[2], Ana Ribeiro[2], Anke Bergmann[2], Marco Garcia[1], *[1]Laboratory of Biomechanics - School of Physical Education and Sports - Federal University of Rio de Janeiro, Brazil, [2]HCIII - National Cancer Institute, Brazil*

PHYSICAL MEDICINE AND REHABILITATION PMP5
Saturday June 21, 10:30-11:30 and 15:00-16:30 Great Room A

PMP5.1 EMG OF SPINOCEREBELLAR DEGENERATION PATIENT AND NORMAL SUBJECTS DURING WOLKING
Kimito Momose[1], Masanori Onizaki[2], *[1]Shinshu University, Japan, [2]Fujimi Kogen Hospital, Japan*

PMP5.2 ANALYSIS OF DROP FOOT GAIT PATTERNS IN CHILDREN WITH HEMIPLEGIC CEREBRAL PALSY FOLLOWING AN ANTERIOR TIBIALIS HOME BASED NEUROMUSCULAR ELECTRICAL STIMULATION (NMES) STRENGTHEING PROGRAM
Melany Westwell[1], Matthew Luginbuhl[1], Sapna Parikh[1], *[1]Connecticut Children's Medical Center, United States*

PMP5.3 ESTIMATION OF CONTROL STRATEGIES ADOPTED BY ELITE FES ROWERS
Matthew Green[1], Brian Andrews[1], Adrian Poulton[2], Simon Goodey[3], Robin Gibbons[4], *[1]Oxford Brookes University, United Kingdom, [2]The Open University, United Kingdom, [3]London Regatta Centre, United Kingdom, [4]ASPIRE National Training Centre, United Kingdom*

PMP5.4 ACUTE EFFECTS OF FOOT TAPING ON MUSCULAR RECRUITMENT IN PATIENTS WITH FOOT DEFORMITIES
Licia Cacciari[1], Ricky Watari[1], Isabel Sacco[1], Clarice Tanaka[1], *[1]University of Sao Paulo, Brazil*

PMP5.5 DRIVING EFFICIENCY OF WHEELCHAIR AS HEMIPLEGIC-COMPARE WITH SITTING

POSTURE
Makoto Miwa[1], Utari Moriyama[1], Kohji Ihashi[1], Kimito Momose[1], [1]*Yamagata Prectural University, Japan*

PMP5.6 VALIDITY OF MUSCLE FUNCTION TESTS IN THE REHABILITATION MANAGEMENT OF CHRONIC LBP
Gerold Ebenbichler[1], Wolfgang Gruther[1], Franziska Wick[1], Christoph Leitner[1], Birgit Paul[2], Martin Posch[3], [1]*Department of Physical Medicine and Rehabilitation, MUV; VGH, Austria,* [2]*Psychosoziales Krankenhaus Eggenburg, Austria,* [3]*Section of Medical Statistics, MUV; VGH, Austria*

PMP5.7 AN EVALUATION METHOD FOR CONTROLLING SKILL OF MYOELECTRIC CONTROL HAND
Yoshiaki Hara[1], Yoshihiro Fukazawa[2], Takaaki Chin[2], Kotaro Minato[3], Masaki Yoshida[4], [1]*Hyogo Assistive Technology Research and Design Institute, Japan,* [2]*Hyogo Rehabilitation Center, Japan,* [3]*Nara Institute of Science and Technology, Japan,* [4]*Osaka Electro-Communication University, Japan*

PMP5.8 ELECTROMYOGRAPHIC PEAK ANALYSIS IN SHOULDER STABILIZER MUSCLES DURING PENDULAR EXERCISES
Michel Brentano[1], Tiago Noer[1], Tiago Arrial[1], Luiz Kruel[1], [1]*Federal University of Rio Grande do Sul, Brazil*

PMP5.9 COMPARISON OF THE BRACHIAL BICEPS AND TRICEPS ACTIVATION DURING PEC DECK AND BENCH PRESS EXERCISES WITH THE VALUES OBTAINED DURING THE 1RM TEST IN THE BICEPS CURL AND TRICEPS PRESSDOWN
Michel Brentano[1], Rafael Spinelli[1], Tiago Arrial[1], Luiz Kruel[1], [1]*Federal University of Rio Grande do Sul, Brazil*

PMP5.10 A PRELIMINARY STUDY ON USEFULNESS OF CYCLING WHEEL CHAIR TRAINING FOR THE PATIENTS WITH LOWER EXTREMITIES IMPAIRMENT
Takaaki Sekiya[1], Kazunori Seki[1], Kazuko Tsuji[2], Yasunobu Handa[1], [1]*tohoku University, Japan,* [2]*wakuya Town Hospital, Japan*

PMP5.11 GAIT ANALYSIS BEFORE AND AFTER ELECTRICAL STIMULATION FOR THE SPASTIC PARAPLEGIC PATIENTS AND HEMIPARETIC STROKE PATIENTS
Kayoko Suzuki[1], Kazunori Seki[1], Yasunobu Handa[1], [1]*Tohoku University, Japan*

PMP5.12 SURFACE ELECTROMYOGRAPHIC STUDY OF RAMSAY HUNT SYNDROME: A CASE REPORT
Eduardo Grossmann[1], Iliana Lima[1], Danilo Dressano[1], Maria Bérzin[1], Fausto Bérzin[1], [1]*Federal University of Rio Grande do Sul, Brazil*

PHYSICAL MEDICINE AND REHABILITATION PMP6
Saturday June 21, 10:30-11:30 and 15:00-16:30 Great Room A

PMP6.1 NEUROMUSCULAR RESPONSES TO STATIONARY RUNNING AT DIFFERENT CADENCES IN AQUATIC AND DRY LAND ENVIRONMENTS
Cristine Lima Alberton[1], Marcus Peikriswili Tartaruga[1], Eduardo Marczwski da Silva[1], Eduardo Lusa Cadore[1], Stéphanie Santana Pinto[1], Tiago de Menezes Arrial[1], Luiz Fernando Martins Kruel[1], [1]*Federal University of Rio Grande do Sul, Brazil*

PMP6.2 PARASPINAL RECRUITMENT PATTERNS DURING ECCENTRIC AND CONCENTRIC CONTRACTIONS IN CHRONIC LOW BACK PAIN
Germán Vásquez[1], David Arriagada[1], Francisca Arroyo[2], Rony Silvestre[1], [1]*Universidad Mayor, Chile,* [2]*Clínica Santa María, Chile*

PMP6.3 EFFECT OF THE BASE ON THE ELECTROMYOGRAPHIC AMPLITUDE DURING CLOSED KINECTIC CHAIN EXERCISES FOR UPPER LIMB AND SHOULDER GIRDLE

Anamaria Siriani de Oliveira[1], Helga Tatiana Tucci[1], Rodrigo Cappato de Araújo[1], Rodrido de Andrade[1], Jaqueline Martins[1], [1]*University of São Paulo, Brazil*

ISEK 2008 Program

Power point, Projector, Podium Microphone, Computer | Power point, Projector, Computer | NO AV

June 18, 2008 (Wednesday)

SPEAKER READY ROOM (Strategy Room 6)

Time						
8:00						
:30						
9:00	ISEK Council Meeting 9:00-12:30 Executive Board Room (3rd floor)					
:30						
10:00						
:30						
11:00						
:30						
12:00		Registration Desk Open 12:00-21:00 Great Room pre-function space in front of Great Room A (3rd floor)				
:30	ISEK Council Lunch 12:30-13:30 Executive Board Room (3rd floor)					
13:00						
:30						
14:00			Workshop Motor Unit Decomposition McGill & Clancy 14:00-18:00 (Internet) Strategy Room 7 (5th floor) CLSRM	Workshop FES in Spinal Cord Injury Popovic & Ditor 14:00-18:00 (Internet) Upper Fallsview Studio A (5th floor) CLSRM	Workshop Neurophysiology of Aging Doherty & Rice 14:00-18:00 Upper Fallsview Studio B (5th floor) CLSRM	Workshop Pelvic Floor Muscle Function Hodges 14:00-18:00 Strategy Room 5 (5th floor) CLSRM
:30						
15:00						
:30						
16:00						
:30						
17:00						
:30						
18:00						
:30			HIGH DENSITY SURFACE EMG:TECHNOLOGY AND APPLICATIONS Basmajian Lecture 18:30-19:30 Roberto Merletti Great Rooms A/B (3rd floor)			
19:00						
:30			Welcome Reception 19:30-21:30 Great Rooms A/B (3rd floor)			
20:00						
:30						
21:00						
:30						

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June 19, 2008 (Thursday)

SPEAKER READY ROOM (Strategy Room 6)

Time					
7:00	Registration Desk Open 7:00-15:00 **Great Room pre-function space in front of Great Room A (3rd floor)**				
:30					
8:00		TELERABILITATION; ENABLING REMOTE MONITORING AND REMOTELY SUPERVISED TRAINGING Keynote Lecture 8:00-8:50 Hermie Hermens **Great Room C (3rd floor)**			
:30					
9:00		Symposium on Telerehabilitation 9:00-10:30 **Great Room C (3rd floor)**	Motor Performance - MP01 9:00-10:30 **Strategy Room 1 (5th floor)**	Mechanomyogr - MM01 9:00-10:30 **Strategy Room 2 (5th floor)**	Motor Units - MU01 9:00-10:30 **Strategy Room 3 (5th floor)**
:30					
10:00					
:30		Coffee Break and Poster Session (Electrode Arrays, Ergonomics, Motor Control, Muscle Fatigue, Physical Medicine) 10:30-11:30 **Great Room A**			
11:00					
:30		Physical Medicine - PM01 11:30-13:00 **Great Room C**	Muscle Fatigue - MF01 11:30-13:00 **Strategy Room 1 (5th floor)**	Movement Dis - MD01 11:30-13:00 **Strategy Room 2 (5th floor)**	Motor Units - MU02 11:30-13:00 **Strategy Room 3 (5th floor)**
12:00					
:30					
13:00		Lunch 13:00-14:00 **Great Room A (3rd floor)**		JEK Editorial Board Meeting 13:00-14:00 (Lunch Buffet: The "Traditional Buffet") **Strategy Room 7 (5th floor)**	
:30					
14:00		OCCUPATIONAL BIOMECHANICS OF THE SPINE Keynote Lecture 14:00-14:50 Jim Potvin **Great Room C (3rd floor)**			
:30					
15:00		Symposium on Occupational Biomech 15:00-16:30 **Great Room C (3rd floor)**	Physical Medicine - PM02 15:00-16:30 **Strategy Room 1 (5th floor)**	Motor Performance - MP02 15:00-16:30 **Strategy Room 2 (5th floor)**	Electrode Arrays - EA01 15:00-16:30 **Strategy Room 3 (5th floor)**
:30					
16:00					
:30		Coffee Break and Poster Session (Electrode Arrays, Ergonomics, Motor Control, Muscle Fatigue, Physical Medicine) 16:30-17:30 **Great Room A**			
17:00					
:30		Ergonomics - ER01 17:30-19:00 **Great Room C (3rd floor)**	Physical Medicine - PM03 17:30-19:00 **Strategy Room 1 (5th floor)**	Muscle Fatigue - MF02 17:30-19:00 **Strategy Room 2 (5th floor)**	Movement Dis - MD02 17:30-19:00 **Strategy Room 3 (5th floor)**
18:00					
:30					
19:00					

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June 20, 2008 (Friday)

SPEAKER READY ROOM (Strategy Room 6)

Time					
7:00					
:30	Registration Desk Open 7:30-12:00 **Great Room pre-function space in front of Great Room A (3rd floor)**				
8:00 – :30		RECENT ADVANCES: FUNCTIONAL ELECTRICAL STIMULATION IN SPINAL CORD INJURY Keynote Lecture 8:00-8:50 Milos Popovic **Great Room C (3rd floor)**			
9:00 – 10:00		Symposium on FES 9:00-10:30 **Great Room C (3rd floor)**	Motor Performance - MP03 9:00-10:30 **Strategy Room 1 (5th floor)**	Mechanomyogr - MM02 9:00-10:30 **Strategy Room 2 (5th floor)**	Motor Control - MC01 9:00-10:30 **Strategy Room 3 (5th floor)**
:30 – 11:00		Coffee Break and Poster Session (Motor Control, Movement Dis, Motor Units, Posture & Balance, Physical Medicine) 10:30-11:30 **Great Room A**			
:30 – 12:30		Physical Medicine - PM04 11:30-13:00 **Great Room C (3rd floor)**	Modeling & SP - MS01 11:30-13:00 **Strategy Room 1 (5th floor)**	Neurophysiology - NP01 11:30-13:00 **Strategy Room 2 (5th floor)**	Motor Unit - Interest Group 11:30-13:00 (Kevin McGill) **Strategy Room 3 (5th floor)**
13:00 – :30	ISEK Council **Executive Board Room (3rd floor)**	Lunch 13:00-14:00 **Great Room A**			**Student Career Strategy Room 3 (5th floor)**
14:00 – :30		CLINICAL ASSESSMENT OF MOTOR DEFICITS IN SPASTIC CEREBRAL PALSY Keynote Lecture 14:00-14:50 Ken Kaufman **Great Room C/B (3rd floor)**			
15:00 – 16:00		Symposium on Spasticity 15:00-16:30 **Great Room C (3rd floor)**	Physical Medicine - PM05 15:00-16:30 **Strategy Room 1 (5th floor)**	Posture & Balance - PB01 15:00-16:30 **Strategy Room 2 (5th floor)**	Motor Control - MC02 15:00-16:30 **Strategy Room 3 (5th floor)**
:30 – 17:00		Poster Session (Motor Control, Movement Dis, Motor Units, Posture & Balance, Physical Medicine) 16:30-17:30 **Great Room A**			
:30 – 18:00					
:30 – 22:00		**18:30 – 19:00 Cocktails** **19:00 – 21:30 Banquet** **22:00 – Fireworks over the Falls** **Great Room A**			

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June 21, 2008 (Saturday)

SPEAKER READY ROOM (Strategy Room 6)

Time				
7:00 :30				
8:00 :30	INNOVATIVE TECHNOLOGIES FOR QUANTIFYING AND TREATING LOWER EXTREMITY IMPAIREMENTS FOLLOWING STROKE Keynote Lecture 8:00-8:50 Joe Hidler **Great Room C (3rd floor)**			
9:00 :30 **10**:00	Symposium on Robotics 9:00-10:30 **Great Room C (3rd floor)**	Motor Control - MC03 9:00-10:30 **Strategy Room 1 (5th floor)**	Posture & Balance - PB02 9:00-10:30 **Strategy Room 2 (5th floor)**	Modeling & SP - MS02 9:00-10:30 **Strategy Room 3 (5th floor)**
:30 **11**:00	Coffee Break and Poster Session (Mechanomyogr, Motor Performance, Modeling & SP, Neurophysiology, Posture & Balance, Physical Medicine) 10:30-11:30 **Great Room A (3rd floor)**			
:30 **12**:00 :30	Posture & Balance - PB03 11:30-13:00 **Great Room C (3rd floor)**	Motor Control - MC04 11:30-13:00 **Strategy Room 1 (5th floor)**	Motor Units - MU03 11:30-13:00 **Strategy Room 2 (5th floor)**	Physical Medicine - PM06 11:30-13:00 **Strategy Room 3 (5th floor)**
13:00 :30	Lunch, ISEK General Assembly 13:00-14:00 **Great Room C (3rd floor)**			
14:00 :30	VIRTUAL REALITY AND GAMING APPROACHES TO PROMOTE WALKING AND MOBILITY POST-STROKE: FROM THE LAB TO THE CLINIC Keynote Lecture 14:00-14:50 Judy Deutsch **Great Room C/B (3rd floor)**			
15:00 :30 **16**:00	Symposium on Virtual Reality 15:00-16:30 **Great Room C (3rd floor)**	Poster Session (Mechanomyogr, Motor Performance, Modeling & SP, Neurophysiology, Posture & Balance, Physical Medicine) 15:00-16:30 **Great Room A**		
:30 **17**:00	Award Ceremony 16:30-17:00 **Great Room C (3rd floor)**			
:30 **18**:00				

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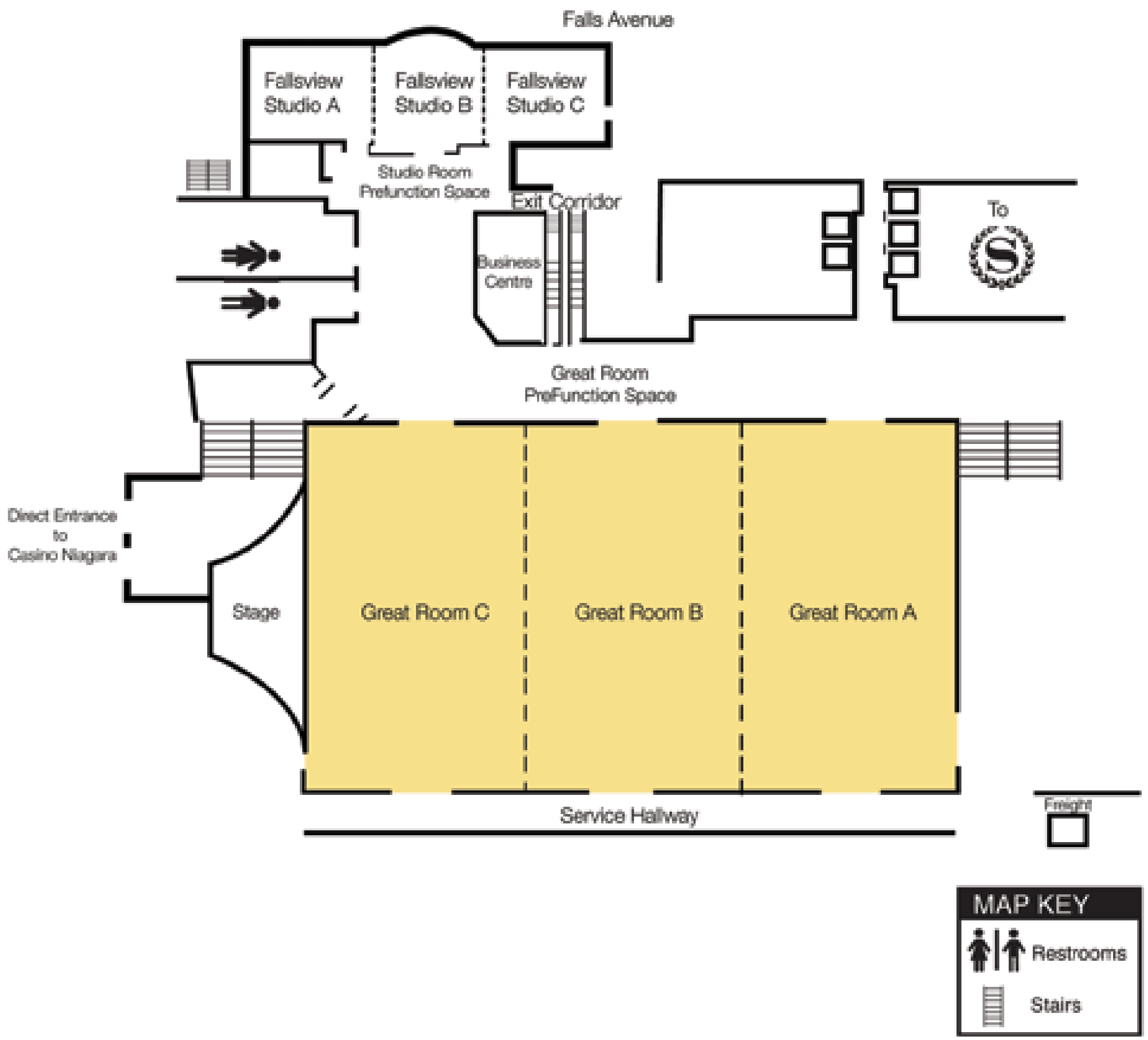
3rd FLOOR



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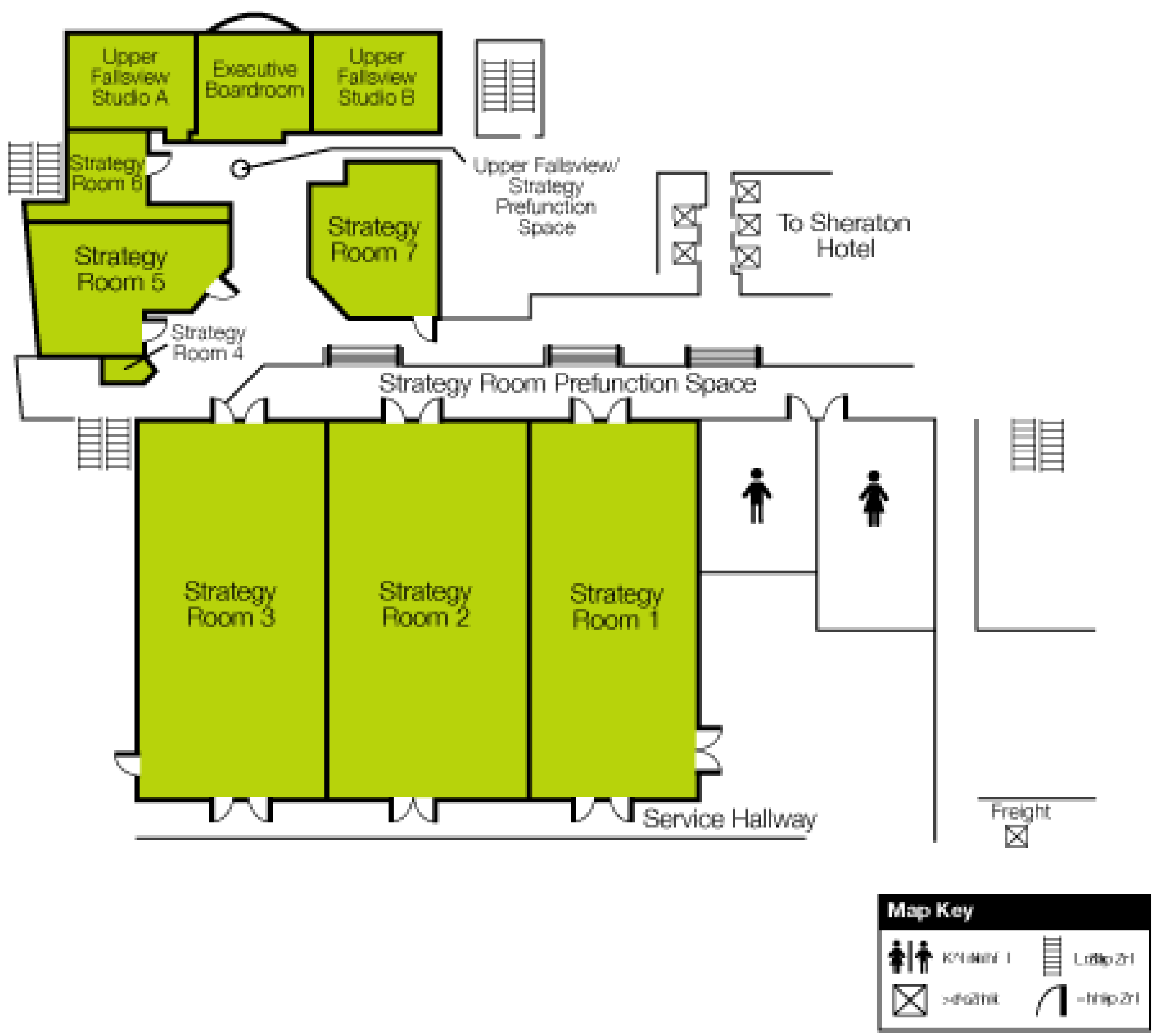
5th Floor



XVIIth
Congress of the International Society of Electrophysiology and Kinesiology

MUSCLES IN MOTION: Moving Research into Clinical Practice

18–21 June 2008 Niagara Falls, Ontario, Canada
***MUSCLES IN MOTION:* Moving Research into Clinical Practice**

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Abstracts

Basmajian Lecture

Dr. Roberto Merletti

TWO DIMENSIONAL HIGH DENSITY SURFACE EMG (HD-EMG) TECHNOLOGY AND APPLICATIONS

Roberto Merletti[1], Aleš Holobar[1], Dario Farina[2]

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INTRODUCTION

The motor unit action potentials (MUAPs) sum to produce a spatial distribution of voltage on the skin surface above the muscle. This time varying “electrical image” may be acquired with an electrode grid that provides two dimensional (2D) sampling in space. The time evolution of the resulting “image” may be tracked by sampling in time. This methodology is referred to as High Density EMG (HD-EMG) (Zwarts et al. 2003).
The structure of the electrical image (potential map), its evolution in time and its time integrals (RMS or ARV) contain information of anatomical and physiological relevance related to force (Staudeman et al.2007) and fatigue. The problem of extracting this information from HD-EMG has been addressed by many research groups and represents a challenge in surface EMG technology (Kleine et al. 2000).

EMG TOPOGRAPHY

The distribution of EMG amplitude (ARV or RMS) provides a topographical representation of muscle activity. Fig 1 depicts the distribution of single differential (SD) EMG RMS at two different levels of isometric contraction of the upper trapezius and demonstrates that a single sampling point is not representative of the spatially heterogeneous muscle activity. Fig. 2 shows the change of EMG distribution above a portion of the trapezius muscle during a contraction sustained to endurance and a shift of the centroid of the map during the contraction time. Subjects with greater activity shift show longer endurance time (Farina et al 2008, Kleine et al 2000).

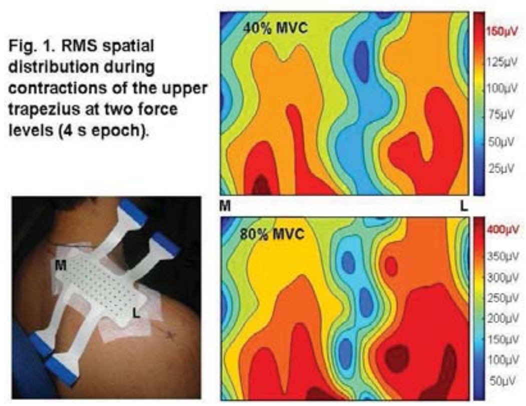

Fig. 1. RMS spatial distribution during contractions of the upper trapezius at two force levels (4 s epoch).

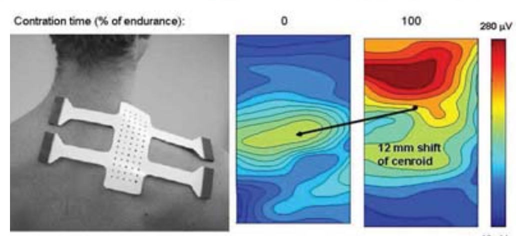

Fig. 2. Change of RMS map under the 2D array during a low level contraction sustained to endurance. The 2D array is placed orthogonally to the position indicated in Fig. 1.The black line shows the shift of the centroid of the SD RMS distribution between the initial and the final map (t=0 and t = endurance time).

1

The most relevant contributions to surface EMG analysis over the past decade are related to the decomposition of the signal into the constituent MUAP trains (Fig. 3). Methods for increasing the spatial selectivity of the recording (Disselhorst-Klug et al. 1997; Ostlund et al. 2006) and for identifying individual sources have been developed (Roeleveld and Stegeman 2002). One algorithm developed for the decomposition of the surface EMG is the Correlation Kernel Compensation (CKC) method (Holobar et al. 2004). This algorithm is fully automatic and resolves superimpositions of MUAPs.
The interpolated surface spatial distribution of a MUAP extracted from HD-EMG recordings with this algorithm is depicted in Fig. 4 both as an instantaneous color map and as a 3D voltage plot. The surface approach may provide information on a larger number of motor units than it is currently possible with selective intramuscular recordings and provides information about the motor control strategies implemented by the CNS.
Fig 5 shows the interpolated discharge rate of 18 motor units identified by the CKC decomposition algorithm (Holobar and Zazula 2004) during a ramp-up-ramp-down isometric contraction of the abductor pollicis brevis. The “common drive” phenomenon (De Luca et al. 1982) is evident in the plot.

CONDUCTION VELOCITY AND OTHER MOTOR UNIT PROPERTIES

The spatio-temporal distribution of surface electrical potential contains information on the velocity, direction and origin of propagation of action potentials along the muscle fibers and about neural control strategies (Farina et al, 2004; Grönlund et al. 2005). Knowledge of fiber direction and innervation zone location is relevant, for instance, for defining “optimal” locations for estimating EMG variables, for injection of botulinum toxin or for safer performance of episiotomy surgery (Enck et al. 2004).
The change of velocity of propagation of MUAPs is an indicator of muscle fatigue (Merletti et al. 1990).

APPLICATIONS

The diagnostic contribution of HD-EMG is not yet fully accepted in clinical neurophysiology (Drost et al. 2006). However, the applications of this technique in ergonomics, sport and rehabilitation medicine, and gynecology are rapidly growing. For example, HD-EMG recordings from the external anal sphincter (EAS) muscle aims at reducing the risk of partial denervation following episiotomy (Fig. 6 and 7). This technique is being clinically applied with the development of disposable probes used to support clinical decisions about episiotomy and tumor resection (Enck et al. 2004).
Other applications of HD-EMG concern motor unit counting (Blok et al. 2005), monitoring of activities at workstations, prevention of work related neuromuscular disorders, and the study of reinnervation (Lanzetta et al. 2002).

CONCLUSIONS

Surface EMG is becoming an imaging technique that can be integrated with ultrasonic and MR approaches. While the traditional one-electrode-pair technique remains the one of choice for movement analysis, the HD-EMG technology is opening new windows to look into the mechanisms associated to movement.

REFERENCES

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De Luca CJ et al. Control scheme governing concurrently active human motor units during voluntary contractions. J. Physiol. 1982; 329: 129-142.

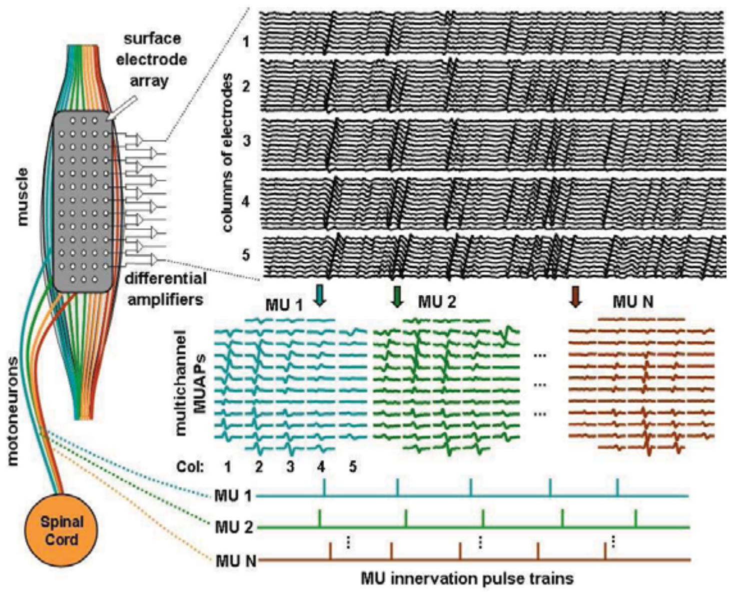

Fig. 3. Surface EMG is detected using a 2D array and the 2D "signatures" of the individual motor units (MU) are extracted using a decomposition algorithm. Available algorithms are based either on template recognition and classification or on blind identification of repeating MUAPs. Individual MU discharge rates are then estimated and MUAPs are extracted by spike triggering averaging of the surface EMG (Holobar and Zazula, 2004). The algorithm for blind MUAP identification automatically resolves superimpositions.

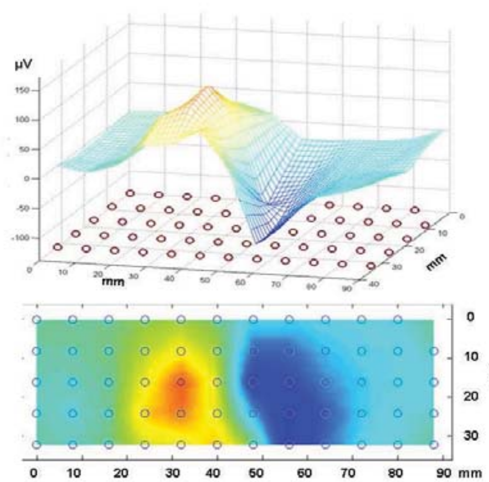

Fig. 4. 3D and color map representation in space (skin surface) of a single differential MUAP at a specific time sample. The 2D electrode array is placed on one side of the innervation zone. The circles represent the electrodes of the array. Propagation direction is to the left.

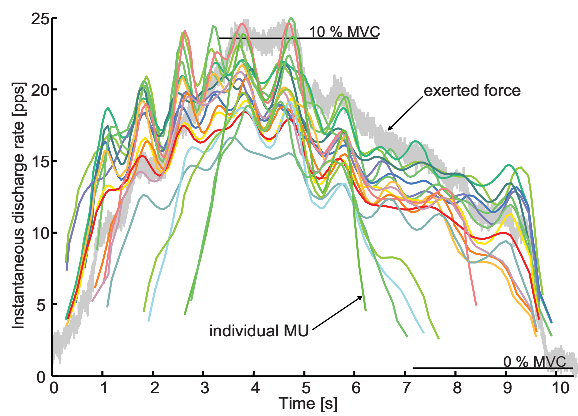

Fig. 5. Isometric force (grey) and instantaneous discharge rates of 18 motor units of the abductor pollicis brevis muscle during its ramp-up and ramp-down contraction. The phenomenon of common drive, previously observed with needle EMG (De Luca et al. 1982), is evident. MU discharge patterns were identified by the Convolution Kernel Compensation algorithm applied to 64-channel surface EMG (Holobar and Zazula 2004).

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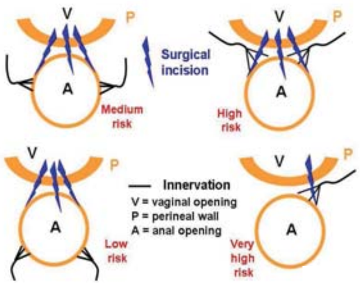

Fig. 6. Section of the recto-vaginal septum (episiotomy) may imply partial denervation and later loss of control of the EAS.

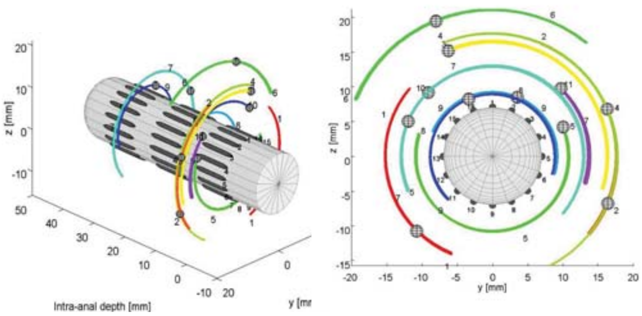

Fig. 7. Identification of the innervation zone (IZ) and of the motor unit arrangement in the EAS muscle by means of an intra-anal probe with three circumferential electrode arrays. The IZs (grey dots) and the motor units (colored arcs) are located differently in different individuals. A CV of 4m/s is assumed for all MUs. The MUs with larger radius are those with lower angular conduction velocity.

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Staudenmann D. et al. Independent component analysis of high-density electromyography in muscle force estimation. IEEE Trans. Biomed. Eng. 2007; 54: 751-754.

Zwarts MJ, Stegeman DF. Multichannel surface EMG: basic aspects and clinical utility. Muscle Nerve 2003; 28: 1-17.

ACKNOWLEDGEMENTS
This work was supported by the EU projects OASIS and CyberManS, by the EU Marie Curie EIF fellowship, by the projects MESM and OSMA of the European and Italian Space Agencies, and by the Bank Foundations Compagnia di San Paolo and Fondazione Cassa di Risparmio di Torino.

Symposium

Applications of Functional Electrical Stimulation (FE)

RECENT ADVANCES: FUNCTIONAL ELECTRICAL STIMULATION IN SPINAL CORD INJURY

Milos R. Popovic[1,2]

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[2] Toronto Rehab, ON, Canada
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ABSTRACT

Functional electrical stimulation (FES) is a technique developed in the early 1960s. Presently, FES is used primarily to develop neuroprosthetic systems for individuals with some form of paralysis. In this context, FES systems in these individuals are used as a substitute for functions that have been impaired. For example, individuals who are unable to breathe on their own may have a neuroprosthesis implanted to allow them to breathe without the help of a ventilator. Similarly, individuals who are unable to grasp receive a neuroprosthesis for grasping to help them grasp and release objects they are unable to grasp and release voluntarily.

Since 2000-2001, there has been a major shift in thinking with respect to FES technology and its applications. A new application has emerged; FES is being used as a short-term therapeutic intervention to facilitate recovery of voluntary function, instead of using FES as a permanent neuroprosthetic system patients have to use/wear all the time to perform a task they are unable to perform voluntarily. This new and emerging application of FES has been termed, FES therapy, or simply FET.

In this lecture, there will be discussion of an application of FES therapy that can be used to restore voluntary grasping function in individuals with spinal cord injury.

The second part of the talk will focus on another little known application of FES. We will examine the use of FES to prevent syncope in individuals with SCI. Yet unpublished results pertaining to this application of FES will be discussed.

The Effects of Functional Electrically Stimulated (FES) Exercise on the Risk of Deep Vein Thrombosis and Pressure Sore Development After Spinal Cord Injury

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Spinal Cord Injury Laboratory, Brock University,
St. Catharines, Ontario, Canada
E-mail: dditor@brocku.ca

INTRODUCTION

There is considerable interest in FES-assisted cycling as a means of enhancing health in persons with spinal cord injury (SCI). There are many purported benefits to the neuromusculoskeletal system that have been attributed to FES-assisted exercise and in particular to the most widespread applicaton, FES-assisted cycling. These include benefits associated with muscle (i.e., reducing atrophy, spasticity or enhancing size, strength or function) and the cardiovascular system (i.e., enhancing oxygen uptake, increasing cardiac output or reducing sub-maximal heart rate) as well as more putative effects on bone (i.e., minimizing loss of bone mineral density). However, there is little evidence that demonstrates if these mostly physiological benefits actually translate to reductions in actual health complications. The primary purpose of this study was to determine the effects of FES-cycling on the risk of some common secondary health complications frequently encountered in persons with SCI, specifically cardiovascular complications such as deep vein thromboses (DVT) and pressure ulcers (PU). The present report reflects a preliminary analysis for 11 of the 19 subjects currently recruited to an ongoing study in which up to 48 participants are expected to take part in a 12 week program of FES-assisted cycling exercise.

OBJECTIVE

Although FES-cycling enhances lower limb blood flow and muscle mass in individuals with spinal cord injury (SCI), it is unknown if these physiological benefits translate to reductions in actual health complications. The primary purpose of this study was to determine the effects of FES-cycling on the risk of deep vein thromboses (DVT) and pressure ulcers (PU) in individuals with SCI. A secondary purpose was to investigate changes in common femoral artery blood flow (CFA-BF) after FES-training, and the perceived therapeutic effects.

METHODS

To date, pre-and-post exercise measures only, have been obtained for 11 of 19 individuals recruited into a 12-week, thrice-weekly, double-crossover, FES-cycling trial. The risk of DVT was determined by plasma d-dimer concentrations, while PU risk was determined via the half-time to recovery in skin temperature ($t_{½}$) following a superficial cold stimulus (17°C, 2 minutes). Doppler ultrasound was used to determine CFA-BF, while a 7-point patient-global scale (1=terrible, 7=delighted) was used to determine the perceived effects of FES-exercise compared to typical physical activities.

OUTCOME MEASURES

DVT risk: Plasma d-dimer concentrations,
PU risk: Half-time to recovery in skin temperature ($t½$) following a superficial cold stimulus (17°C, 2 minutes).
CFA-BF: Doppler ultrasound
Perceived effects of FES-cycling: 7-point patient global scale:
(1=terrible, 7=delighted).

RESULTS

There were no significant changes in d-dimer concentrations (pre=458.7±612 vs. post=336.2±185 µg/L, p=0.42, n=11) or $t_{½}$ (pre=245.9±52.3 vs. post=240.2±48.9 sec, p=0.41, n=10) following FES-cycling. There was a trend for increased CFA-BF (pre=355.6±131.2 vs. post=465.4±168.5 mL/min, p=0.15, n=6) after training and a significantly greater perceived therapeutic effect of FES-cycling compared to typical physical activities (6.5±0.6 vs. 5.0±1.3, respectively, p=0.03, n=6).

SUMMARY/CONCLUSIONS

Preliminary data suggest that d-dimer (DVT risk) and $t½$ (pressure sore risk / skin blood flow) may not be sensitive to change after a 12-week FES-cycling program despite apparent changes in CFA-BF (peripheral circulation) and perceived overall therapeutic effect.

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FES IN LOCOMOTOR TRAINING FOR INDIVIDUALS WITH SCI: PROMOTING FUNCTION AND NEUROPLASTICITY

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INTRODUCTION

In individuals with motor-incomplete spinal cord injury (SCI) body weight supported (BWS) locomotor training improves overground walking ability, as does locomotor training with functional electrical stimulation (FES). Using FES as a component of BWS locomotor training is a logical combination. FES can be used as a neuroprosthesis to assist with dorsiflexion, or higher levels of stimulation can be used to evoke a withdrawal reflex for stepping. Other possibilities for locomotor training include robotic gait orthoses or the assistance of a therapist. Among these various options there have been no comparisons to determine whether one form of training offers benefits that are greater than the alternatives.

The purpose of this study was to compare functional and neurophysiologic changes associated with each of three different forms of locomotor training.

METHODS

We compared measures of walking function (walking speed), coordination (intralimb hi-knee angle consistency, step symmetry), functional balance scores (Berg Balance score), and changes in spinal reflex activity (H/M ratio, reciprocal inhibition, flexor reflex sensitivity, quadriceps excitability to stretch). Fifty-four (54) subjects with chronic (> 1 year) motor-incomplete SCI participated. Subjects were pseudo-randomly stratified into 1 of 4 BWS assisted-stepping groups with stratification based on initial lower extremity motor scores.

Subjects participated in training 5 days/week in 1-hour sessions for 3 months. All training groups utilized BWS, but different forms of assistance were provided to aid in stepping. The four training groups are: 1) treadmill training with manual assistance (TM; trainers provided manual assistance to the stepping limb), 2) treadmill training with stimulation (TS; stimulation to the common peroneal nerve was used to elicit a flexion withdrawal response at the onset of each step) , 3) overground training with stimulation (OG; foot-drop stimulators provided assistance with dorsiflexion), or 4) treadmill training with robotic assistance (LR; passive mechanical guidance was provided by a powered gait orthosis.).

RESULTS

Following training, between-group differences in walking speed were significant only for the OG versus LR group, with the OG group having greater improvement in walking speed. However, the mean change in walking speed on the OG group was different than that of the TM and TS groups as well. Intralimb coordination and step symmetry improved across training groups. Scores of functional standing balance increased in the OG group. There were changes in some spinal reflexes in association with training. When data from the from the TS and OG groups were pooled there was a significant improvement in reciprocal inhibition, however there was great variability in the response of this spinal circuit in those assigned to the TS group with some demonstrating reciprocal facilitation (rather than inhibition) following training. The TS group demonstrated the greatest changes in spinal reflex activity as measured by response to muscle stretch.

SUMMARY/CONCLUSIONS

These results suggest that training in the overground environment in combination with electrical stimulation as a neuroprosthetic to assist with foot clearance may offer benefits above those of training on the treadmill. However, from a clinical perspective it appears that all forms of training are associated with some improvements in function, suggesting that the important element is that the individual have the opportunity to practice the task of walking.

The training group that received the greatest intensity of stimulation (the TS group) showed the greatest change in both electrophysiologic and mechanical measures of spinal reflex excitability. In some cases, it is likely that these reflex changes are advantageous for motor function (i.e., adaptive). However, it is possible that some of the changes may be detrimental to function (i.e., maladaptive). For example from a functional perspective, proper functioning of the reciprocal inhibition circuit is thought to be important for limiting inappropriate muscle co-contraction. Therefore, improvements in inhibition are adaptive, while decreases in inhibition (as observed in some individuals in the TS group) are maladaptive.

These data demonstrate that electrical stimulation can be an important adjunct to locomotor training, with effects at both the functional and neurophysiologic levels. Further, the combined use of electrical stimulation and training are effective even in individuals with chronic motor impairment due to spinal cord injury.

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PROVOKING NEW LOCOMOTOR PATTERNS USING FUNCTIONAL ELECTRICAL STIMULATION IN INCOMPLETE SPINAL CORD INJURY

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Introduction
Many patients recover some walking function following incomplete Spinal Cord Injury (SCI). However, their gait remains impaired such that they must use aids such as walkers, canes or crutches, and they can only ambulate at slow speeds for short distances. Progress in rehabilitation engineering has led to the development of interventions such as Functional Electrical Stimulation (FES)-Assisted walking therapy that can produce significant improvements in overground walking speed and gait quality [1]. It is not know if these improvements are due to motor learning, changes in muscle strength and fiber type, cardiovascular conditioning, personal confidence or a combination of all the above. The purpose of this study was to identify the amount of motor learning that took place following a 16-week regimen of intense FES-assisted walking. We evaluated differences in locomotor coordination by analyzing the EMG patterns during overground walking before and after treatment.

Methods
9 subjects with chronic, incomplete SCI (ASIA C or D) participated in this study. In a gait laboratory, each subject performed several cycles of reciprocal walking overground using their preferred gait aids while EMG of seven major muscle of the lower limbs were recorded bilaterally. Four basic muscle synergies related to gait were defined a priori (left and right extensors, left and right flexors). The expression of these synergies during the gait cycle were computed from the time-normalized EMG data of all subjects [2]. Following a baseline EMG assessment, each subject was fitted with a multi-channel surface FES system for walking that produced patterned contractions of the four muscle synergies. Subjects underwent a training regimen in which they walked on a treadmill using the neuroprosthesis three times per week for 16 weeks. At the end of treatment, each subject repeated the EMG assessment. The EMG signals were decomposed to the four synergies and then reconstructed. The amount of variance explained by the synergies was thus calculated as a percentage of the total variance.

Results
All subjects demonstrated high levels of variance in their kinematics and EMG patterns before and after treatment. The four basic muscle synergies accounted for 64% of the total variance before treatment and 78% of the total variance after treatment. This

difference was significant ($p<0.05$). Following treatment, it was generally found that the representation of the two extensor synergies was increased while the flexor synergies decreased or stayed the same.

Conclusions

The results suggest that FES-assisted walking therapy has a significant neurological training effect as seen in the locomotor patterns of muscle activation. The changes observed represent an increase in the representation of certain components of gait. In one sense, subjects developed more simplified patterns. These findings suggest that FES-assisted training is neurally potent.

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Symposium

Occupational Biomechanics of the Spine (OB)

OCCUPATIONAL BIOMECHANICS OF THE SPINE

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INTRODUCTION

It is well known that spine injuries, most often in the lower back, are prevalent in the workplace. This paper will provide some examples of previous occupational spine biomechanics research, to give some historical perspective on the development of this field. Where relevant, some of my own research will be integrated into this paper, but this certainly does not presuppose that these studies have had the same impact as the other seminal studies also cited. What follows is a very brief and incomplete summary of the progression of this field.

TISSUE TOLERANCE

For decades, lumbar intervertebral disc compression force was the main variable of interest, when assessing LBI risk in the workplace. NIOSH (1981) integrated biomechanical and epidemiological data and recommended a compression force Action Limit of 3400 N. Jager and Luttman (1991) and Genaidy et al (1993) have presented predictive models for further delineating tolerance based on variables like gender, age, percentile, lumbar level etc. Brinkman et al (1988) found that the number of cycles to failure decreased with increasing repeated load magnitude. Kumar (1990) and Norman et al (1998) have identified cumulative loading as an important risk factor for LBI. Parkinson and Callaghan (2007) provide further evidence to support the use of a non-linear weighting of forces before the integral is calculated. More recently, other variables like lumbar shear force, spine flexion, spine twisting and joint stability/buckling (see below) have also been implicated as LBI risk factors.

BIOMECHANICAL SPINE MODELS

With the focus on spine compression force, methods were needed to estimate exposure. Direct measures are not feasible, especially in the workplace; so biomechanical models have been developed for this purpose. Morris et al (1961) provided one of the first attempts to model the physiological mechanisms involved to develop the forces required for static lifting. The spine was conceptualized as an elastic rod, which is highly dependent on extrinsic support by structures such as the paraspinal muscles. The model was an attempt to calculate the actual forces on the sine and included consideration of: 1) moments created by the body weight and load, 2) muscle tension from trunk extensor and abdominal muscles, 3) trunk pressures and, 4) forces transmitted by the spine. A single-equivalent model of the erector spinae was used to calculate internal muscle contributions to compression force. More detail was provided in subsequent models (eg. Chaffin, 1969, Anderson et al, 1985), but these were also limited to very simplified muscular representations and static, sagittal plane loading.

McGill and Norman (1986) developed a much more complex model, incorporating 3D dynamics, 7 ligaments and 48 muscles driven by surface EMG. Since that time, McGill has continued to enhance the model and a number of other researchers have developed models with impressive complexity (eg. Marras & Sommerich, 1991). While these models do not make direct measurements of spinal load, they do provide our best estimate of low back tissue loading and are sensitive to differences between individuals and task conditions. While, these complex methods are not practical for use by ergonomics practitioners, they can form the basis for more user-friendly tools (eg. Michigan 3DSSPP© and WatBak© software packages) and/or can be used as criteria for their partial validation. They are also valuable for determining individual muscle and passive tissue loads and their subsequent contributions to lumbar joint compression and shear forces.

SPINE BIOMECHANICS DURING LIFTING AND LOWERING

Manual materials handling has been an important focus for LBIs. We used the McGill & Norman (1986) model to study subjects lifting loads ranging from 5.8 to 32.4 kg with both the “stoop” (straight leg) and the “squat” (bent knee) technique most often recommended for safe lifting (Potvin, et al.1991). The peak lumbar moments were found to be very similar for both methods at each load. However, muscles were found to contribute almost all of this moment during squats, and an average of 86% of the moment during stoop lifts. While this difference was noticeable, it still demonstrated a somewhat surprising dominant contribution of muscles, even when the trunk was forced into flexion. This was also observed when subjects performed lifts over simulated bins, designed to more naturally limit knee flexion (McKean & Potvin, 2001). Subjects achieved the additional required trunk flexion almost entirely through increased flexion about the hips, with little additional spine flexion. This appeared to demonstrate an aversion to loading the passive tissues, even when extreme trunk flexions were required.

In an excellent review, van Dieën (1999) suggests that there is little support to recommend one lifting technique over the other, and that efforts to control LBI risk should focus on reducing spine postural asymmetries, controlling speed, optimizing load height and reducing load magnitude and horizontal reach. These variables have all been demonstrated to increase LBI risk and have been variously incorporated into a number of ergonomic assessment tools such as the NIOSH Lifting Equation (NIOSH, 1992) and Liberty Mutual Tables (Snook & Ciriello, 1991).

REPETITIVE AND/OR PROLONGED SPINE LOADING

Repetitive and/or prolonged spine loading can cause chronic damage to spine tissues, but the subsequent fatigue can also lead to changes in coordination and movement strategies that can precipitate more acute injuries. Potvin (1992) studied subjects lifting for 2 hours and observed significant increases in spine flexion, with two subjects progressing to full flexion-relaxation for every lift performed when fatigued in the last 15 minutes. As noted above, such a reliance on passive tissues is almost never observed during rested lifting.

Further, van Dieën et al (2001) studied repetitive lifting and found that the peak compression forces were much higher than the median levels. This implies that continuous lifting may increase the probability of an adherent, dangerous lift, even in the absence of fatigue. Dickey et al (2003) studied repetitive spine flexion and observed creep in the maximum flexion angle and increases in the angle eliciting flexion-relaxation. They hypothesized that these changes may reflect deleterious neuromuscular adaptations to repeated passive tissue loading.

Kumar (1990) first identified cumulative loading as a risk factor for LBI. Norman et al (1998) performed a large study in an automotive assembly environment and provided empirical evidence for this link. Since that time, Callaghan and his colleagues have been very active in developing and validating methods that will allow practitioners to quantify this exposure (see Callaghan, 2005). While much more cumbersome to quantify than peak loads, these researchers have taken on the challenging task of making these measurements accessible so that these important risk factors are not ignored when LBI risk is assessed.

Motion capture systems are starting to be used to quantify risk in virtual environments, before poor designs even exist outside a computer. We have recently explored the efficacy of using this technology to quantify cumulative spine and shoulder loads and this method does show promise, at least for the companies currently having the resources to invest in this technology (Godin et al, 2006).

SPINE STABILITY

While the early emphasis in occupational biomechanics was to decrease loading exposures, mechanical stability has become an increasingly prevalent issue in the spine biomechanics literature. Bergmark (1989) was the first to attempt to quantify this for the spine, and this was followed by a number of models, including the notable work of Cholewicki and McGill (1996) who emphasized the importance of maintaining some level of spine loading to maintain stability. Panjabi (1992) cited spinal instability as an important risk factor for LBI and identified three systems contributing to spine stability: 1) neural/ feeback, 2) passive tissue, 3) active muscle. These systems interact to maintain joint rotational stiffness (JRS) and stability.

Potvin & Brown (2005) presented a simplified equation that allows for the accurate delineation of individual muscle contributions to JRS. Kevin Granata was making substantial contributions to this area with his study of the effects of reflexes on spine stability, with direct implications for sudden, expected loading in the workplace (eg. Moorhouse & Granata, 2007). We are currently attempting to apply our equation to quantify the recruitment strategies necessary to maintain JRS during a variety of dynamic motions. In this regard we, and a number of other labs, are attempting to carry the torch from Kevin's groundbreaking work.

FUTURE DIRECTIONS

This paper has presented a very brief, and incomplete, overview of the progression of research emphasis in occupational biomechanics of the spine. Much work remains to be done in areas including, but not limited to: 1) quantifying tissue tolerance to various loading modes, especially other than compression force, 2) quantifying and understanding the role of repetitive and cumulative loading in LBI, 3) quantifying and understanding the somewhat conflicting goals of reducing tissue demands while maintaining spine stability, 4) understanding how multi-tasks combine to affect LBI risk and how this can be assessed with ergonomic tools, and a variety of other areas that will be discussed.

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CAN LOW LEVEL EMG BE USED TO PREDICT SUBJECTIVE DISCOMFORT SCORES?

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INTRODUCTION

With changes to industrial demands on workers and increasing automation, there are increased computer and rapid assembly components in a large number of jobs. Prolonged sitting or standing have become an integral part of most working environments.

In prolonged repetitive exposures, elevated activation levels with fewer rest periods have been shown to increase pain reporting (Jonsson, 1982; Veiersted et al., 1990). However, the comparison of muscle recruitment levels between standing and sitting postures has revealed little difference (Callaghan et al., 2001, Althoff et al. 1992). Agonist /antagonist muscle co-activation has been well documented in individuals with low back pain. It has often been assumed that this is an adaptive response. Since most studies have utilized intact groups (pain and healthy control), it is impossible to ascertain whether muscle co-activation is causal or adaptive.

The main purpose of this summary is to discuss the trunk muscle activation levels and whether they can be predictive of subjective discomfort developed in previously asymptomatic individuals.

METHODS

The studies used to present the relationship between low demand tasks and the development of low back discomfort (LBD) represent work done on office sitting, automotive seat studies, and prolonged standing. These studies taken collectively represent results on 74 subjects, with equal male and female representation. The studies typically required subjects to undergo 2 hours of exposure.

In general 8 to 16 pairs of disposable electromyographic (EMG) electrodes (Ag-AgCl) were used to collect from the left and right thoracic and lumbar erector spinae, left and right rectus abdominis, external and internal obliques, latissimus dorsi, gluteus medius (GM), and the multifidus. EMG data were bandpass filtered from 10 to 1000 Hz and differentially amplified (common-mode rejection ratio >90 dB at 60 Hz, input impedance >10 Mohms) to generate a maximum amplification of approximately 2V (Model AMT-8, Bortec, Calgary, AB, Canada). The EMG signal was A/D converted at 2048 samples/s using a 16-bit A/D card with a ±2.5-V range. Kinematic and kinetic data were also collected with various methods for each of the studies. Ratings of perceived discomfort were recorded at 15 minute intervals using a 100mm visual analogue scale (VAS).

RESULTS AND DISCUSSION

All of the occupational tasks studied resulted in increasing discomfort over time. In the prolonged standing work individuals were clearly separated into two groups, those who developed LBD representing clinically relevant magnitudes on VAS and those who did not (Nelson-Wong et al., 2008) (Figure 1). In all of the studies, typical measures for quantifying the EMG responses such as amplitude measures, APDF and GAPS analyses (Gregory & Callaghan, 2008), and frequency measures (Durkin et al., 2006) of fatigue did not relate to the LBD reported. In fact, all subjects exhibited clear muscle rest periods and gaps regardless of the LBD levels documented. The only muscle activity measures that have been sensitive to the perceived LBD scores have been approaches linking how different muscle groups respond, which is indicative of muscle recruitment strategies and control. Both co-activation of antagonist/agonist muscles in the trunk (Gregory et al., 2006) and the hip (Nelson-Wong et al., 2008) have shown the ability to differentiate between exposures and individuals with high LBD reporting (Figure 2a) from individuals who reported low levels of LBD (Figure 2b).

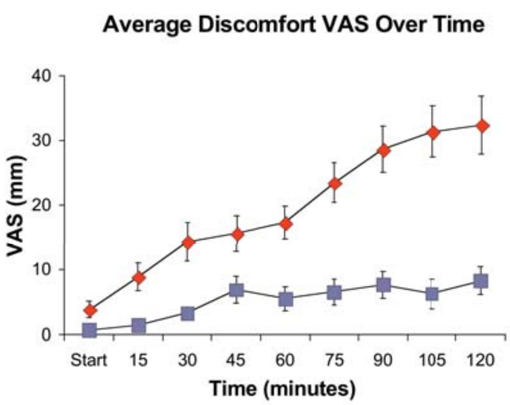

Figure 1: Discomfort developed over 2 hours of standing (red = LBD group)

SUMMARY/CONCLUSIONS

The typical occupational approaches to quantify muscle activation levels associated with pain reporting have not transferred to low level prolonged tasks with limited movement such as sitting and standing. These exposures can produce levels of discomfort that are quite high. The quantification of how different muscle groups interact in these exposures has demonstrated good ability to correlate with LBD development as well as predict individuals who will develop low back pain over time.

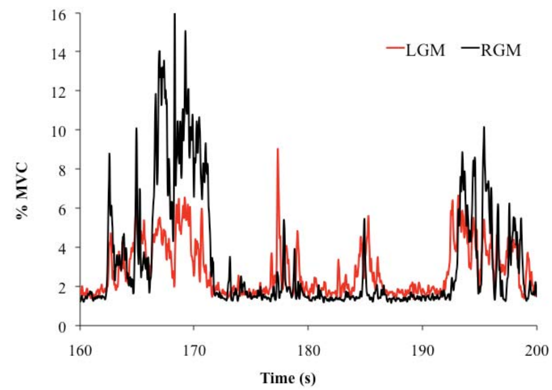

Figure 2a: High discomfort pattern with co-contraction of GM muscles.

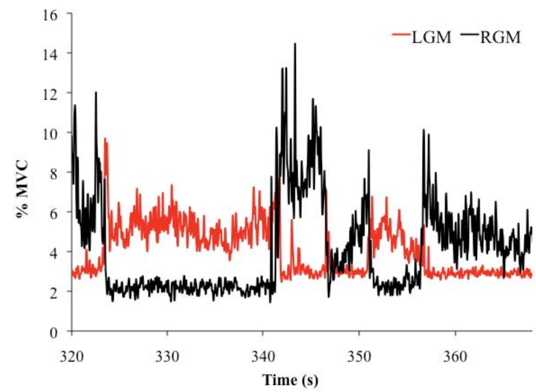

Figure 2b: Low discomfort reporting muscle activity pattern during standing

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TRUNK EXTENSOR ACTIVITY IN LOW BACK PAIN PATIENTS

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INTRODUCTION

In a previous study on trunk muscle activity in patients with low back pain (LBP), it was shown that the ratio of lumbar over thoracic extensor muscle activity was increased when compared to healthy controls (Dieën et al. 2003). The lumbar extensor fascicles originate from the lumbar vertebrae, cross one or several lumbar levels and insert on the pelvis. Their thoracic synergists originate from thoracic vertebrae and ribs, cross the whole lumbar spine and insert on the pelvis. Mechanical modeling predicted that preferential recruitment of the short lumbar fascicles relative to the thoracic fascicles would increase trunk stiffness (Dieën et al. 2003). Furthermore, it was shown that such a change in recruitment can be triggered in healthy subjects by (the threat of) mechanical perturbations to the trunk and by experimentally induced pain (Dieën et al. 2004). We therefore propose that this change in trunk muscle recruitment reflects an adaptive strategy aimed at protecting the painful body part.

The aim of this study was to test the hypothesis that LBP patients compared to healthy controls would show stronger preferential recruitment of lumbar over thoracic extensors in a functional motor task, specifically in gait. With gait velocity the phase difference between transverse pelvis and thorax rotations increases, but less so in patients. In addition, LBP patients generally prefer lower gait velocities (Lamoth et al. 2002). We therefore hypothesized that the difference in muscle recruitment between patients and controls would be larger at higher gait velocities, because high gait velocity would be perceived as more challenging by the patients.

METHODS

Fourteen chronic low back pain patients (9 female, 5 male) and twelve healthy controls (8 female, 4 male) walked on a treadmill for three minutes at 15 velocities that ranged from 0.6 to 6.2 km/h, with increments of 0.4 km/h. EMG of selected trunk muscles was measured and linear envelopes were calculated. The sum of the linear envelopes of the 2 lumbar electrode sites (3 cm paravertebral to L3) was divided by the sum of the envelopes of the two 2 thoracic sites (3 cm paravertebral to T10). Of this L/T ratio the average over each measurement period was calculated.
GEE analysis with L/T as the dependent variable, subject as a grouping variable, health status as a factor and velocity as a covariate was performed using SPSS.

RESULTS

The L/T ratio showed a periodic pattern with peaks at each heel strike, most obviously so at the higher gait velocities.

As hypothesized, the average L/T ratio was significantly higher in the patient group than in the control group ($p = 0.012$). However, substantial between subject differences were found in the patient group, with three subjects having lower ratios than all controls. In addition, significant velocity ($p = 0.002$) and velocity by health status interaction (p=0.015) effects were found. In contrast, with our hypothesis the interaction reflected a decrease of the L/T ratio with velocity in patients and a relatively constant value in the controls (Figure 1). The GEE model parameters reflected the decline in L/T ratio with velocity among the cases and the relatively constant value in the controls.

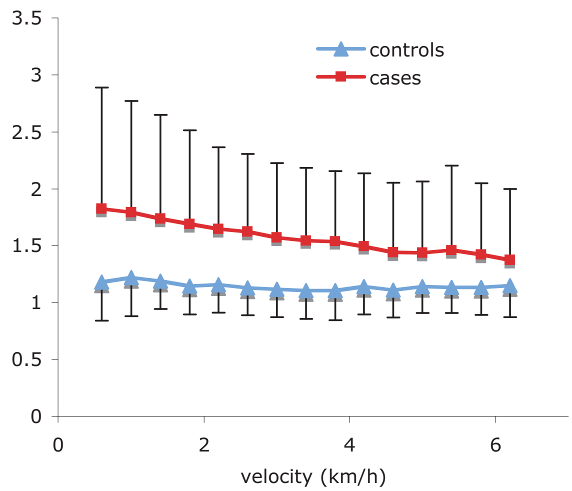

Figure 1. L/T ratio as a function of gait velocity in cases and controls. Error bars indicate one SD.

DISCUSSION

The impact force at heel strike is considered to threaten spine stability (Saunders et al., 2004). Therefore, the modulation of the L/T ratio over the stride cycle with peaks around heel contact appears in line with the assumed importance of this ratio for spine stability. As hypothesized and in line with previous findings, the L/T ratio was on average increased in patients. However, between-subject variance was large and some had ratios lower than the controls. Reeves et al. (2006) also reported higher than normal ratios in some and lower than normal ratios in other LBP patients. As mentioned above we assume that the higher ratio reflects a strategy to increase trunk stiffness. It could be that the lower ratio reflects an adaptation based on another criterion (e.g. preferential thoracic activity would reduce joint load), or it could reflect failure to adapt.

In contrast with our hypothesis, the L/T ratio decreased with increasing velocity in the patients. It could be that the higher velocity forced patients to reduce trunk stiffness and thus lower the ratio. With increasing gait velocity, transverse rotation between thorax and pelvis increases. It has been suggested that this allows pelvis rotation to contribute to step length (Bruijn et al. 2007). If trunk stiffness is maintained at a high level, this could limit rotation between thorax and pelvis and as such hamper increasing gait velocity.

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REHABILITATING LOW BACK DISORDERS: CONSIDERING CORRUPTED MOTOR PATTERNS

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OVERVIEW

Back troubles appear to be both a cause and a consequence of corrupted motor patterns. Inappropriate patterns lead to joint overload, causing tissue damage or exacerbation of existing damage. Overload may be in the form applied force, such as compression. Or, overload may cause destabilizing forces within the spinal column. Examples of corrupted patterns include not integrating major spine stabilizers such as the latissimus dorsi during certain spine extension tasks. Without latissimus forces, other extensor muscles with less mechanical advantage create the extensor torque resulting in higher joint forces. In addition, the stiffening contributions of latissimus over the entire lumbar region assist to insure sufficient stability. The corollary is that missing latissimus force and stiffness compromises spine stability. But not all perturbed patterns that affect the back involve back muscles. Many people with a history of back troubles present with "Gluteal amnesia". For example, failure to optimally integrate muscles such as gluteus medius during lifting causes over activity in the hamstrings and spine extensors. The result is compromised power generation in the hips relegating more load to the lumbar spine. Correcting gluteal amnesia reduces spine load and generally enhances spine stability.

Examples of exercises to correct these corrupted patterns are proposed. These corrective exercises precede any serious rehabilitation attempts involving enhancement of endurance and strength.

Other issues regarding muscle activation patterns will be discussed. For example, for some reason the timing of muscle onset has become popular in back pain rehabilitation discussions. Evidence will be presented to show that this is of little mechanical consequence when compared to the importance of when peak muscle forces occur, together with the rates of muscle force development and the rate at which the muscle can relax.

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Symposium

Robotics in Rehabilitation (RB)

INNOVATIVE TECHNOLOGIES FOR QUANTIFYING AND TREATING LOWER EXTREMITY IMPAIRMENTS FOLLOWING STROKE

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INTRODUCTION

The field of rehabilitation has undergone significant changes over the last decade. Over this period of time, we have seen new treatment options being tested and incorporated into the daily therapeutic programs of patients following neurological injuries such as stroke, spinal cord injury, and traumatic brain injury. One such approach has been to utilize robotic devices to deliver mass practice therapy for both upper and lower extremity therapy (Hidler et al., 2006). Robotic devices present numerous potential benefits to rehabilitation, however they also present numerous challenges. This presentation will provide an overview of some technologies that are now being used for locomotor training in individuals following neurological disorders.

Rehabilitation has also continued to adopt the concept of evidence-based practice, whereby interventions must be shown to work before they are integrated into standard of care. However there are potential pitfalls with this approach, since we often treat the system as a 'Black Box' rather than a system of components, perhaps some of which are not operating as they should. A more scientific approach would be to break down the problem from the top down. For example, a common behavior exhibited by stroke survivors is poor gait, characterized by asymmetric walking patterns poor endurance, and slow walking speeds. Breaking down the problem, we must ask what impairments are responsible for these behaviors and which interventions are best suited for the impairment? Treating spasticity, for instance, requires a much different approach than abnormal muscle synergy patterns. By breaking down the system by impairments, we can develop targeted treatments and perhaps more effective outcomes.

In this talk, the primary impairments responsible for walking disorders following stroke will be discussed. We will see how innovative technologies are being used to study these impairments with quantitative measures. With a clear understanding of the various lower extremity motor impairments in stroke, new therapeutic interventions will be discussed. Specifically, the results of a 5-year randomized clinical trial will be discussed, which compared robotic-assisted gait training with conventional gait training. The talk will conclude with an overview of a new gait training technology that is allowing patients to begin their therapy early after injury, and allows them to practice functional tasks including stairs, sit to stand, and the navigation of obstacles.

Weakness and Coordination in Acute and Chronic Stroke

It is believed that two primary contributors of lower extremity impairments in stroke are weakness and abnormal synergy patterns. In the upper limb of chronic stroke survivors, there is a strong presence of abnormal synergistic joint moments that significantly reduce limb coordination and control during both static (Beer, 1999) and dynamic (Beer, 2000; Dewald, 2001) tasks. In the context of this work, synergy patterns are defined as the set of joint moments generated throughout the limb of interest for a given task. Similar synergy patterns have not been quantified in the lower extremity of hemiparetic stroke subjects. In our previous studies, we have investigated both synergy patterns and weakness in acute (Hidler, 2007) and chronic (Neckel, 2006) stroke. Subjects were tested in a standing position in order to simulate a more functional posture. With their paretic leg attached to a 6-degrees of freedom loadcell, stroke subjects were asked to perform maximum exertions at the ankle, knee and hip joints individually. We simultaneously monitored the forces generated at adjacent joints as a measure of concurrent synergy patterns. As shown in Figure 1, we found that when compared to age-matched control subjects,

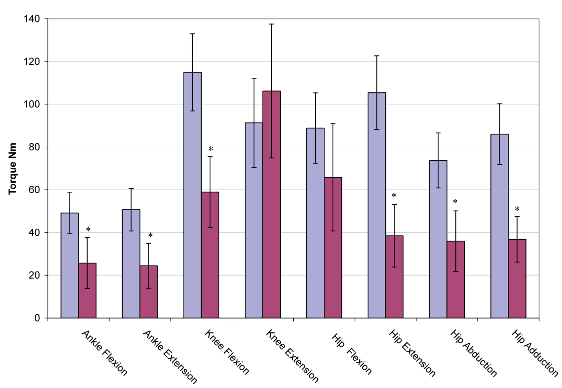

Fig 1. Maximum primary isometric torques for the control and stroke subjects (black = control; white = stroke). (* indicates p < 0.05). (Reproduced from Neckel et al., 2006)

stroke subjects demonstrate extensive weakness throughout their paretic leg. However when synergistic forces generated throughout the leg were looked at, we did not find the presence of abnormal muscle synergy patterns. Interestingly, we did find a strong presence of co-contraction in the paretic leg. Instead, stroke subjects tend to co-contract antagonistic muscles so that the net force generated at a particular joint is significantly reduced. These studies found that both acute and chronic stroke subjects experience profound weakness in their paretic leg, which is at least partially attributable to co-contraction of antagonistic muscles. Yet we did not observe the presence of abnormal synergy patterns similar to those reported in the upper limb.

Spasticity in Stroke

Another possible contributor to lower extremity impairment in stroke is reflex hyper-excitability, namely spasticity. Previous studies have investigated spasticity in the paretic limbs of stroke survivors yet have mainly focused on single-joint movements (Powers, 1988). Since gait involves multi-joint movements, our goal was to quantify spasticity during simultaneous movements of the hip, knee, and ankle joints (Black, 2007). Here, chronic stroke subjects were asked to pre-activate their paretic leg to various levels of baseline forces, after which their leg was extended at various speeds. It was found that in most lower extremity muscles, stroke subjects demonstrated enhanced reflex responses during imposed multi-joint movements, particularly at high speeds. Additionally, the long-latency reflex component (e.g. response between 40-150 ms) accounted for the majority of the differences when compared to age-matched control subjects (Figure 2).

In summary, our preliminary studies indicate that the primary contributors to lower extremity impairments in stroke are weakness and spasticity, both being deficits that account for disturbances in gait.

New Treatment Interventions

We have recently concluded a 5-year multi-center study comparing gait training with the Lokomat robotic gait-orthosis with conventional physical therapy. In the study,

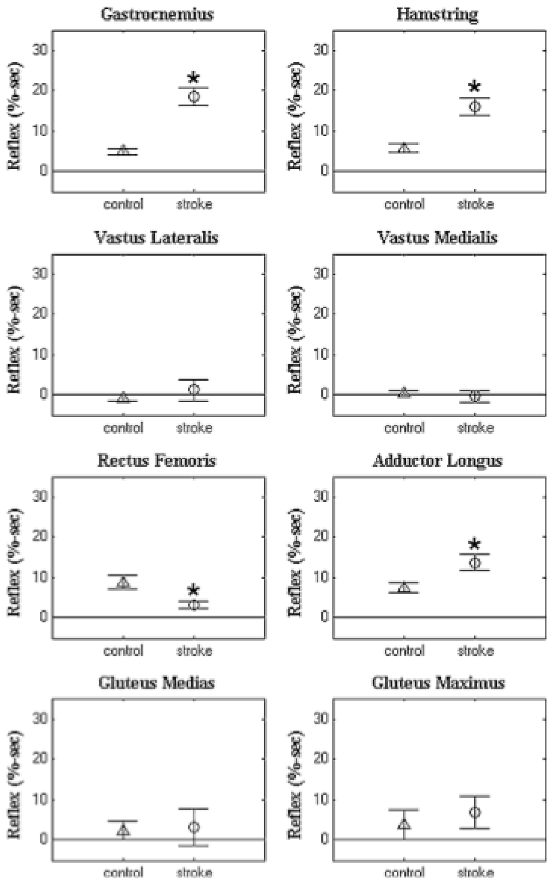

Fig 2. Average reflex responses in each muscle for each group computed using data from all trials across all speeds and pre-activation levels. Asterisk (*) indicates means are statistically different (p ≤ 0.05). (Reproduced from Black et al., 2007)

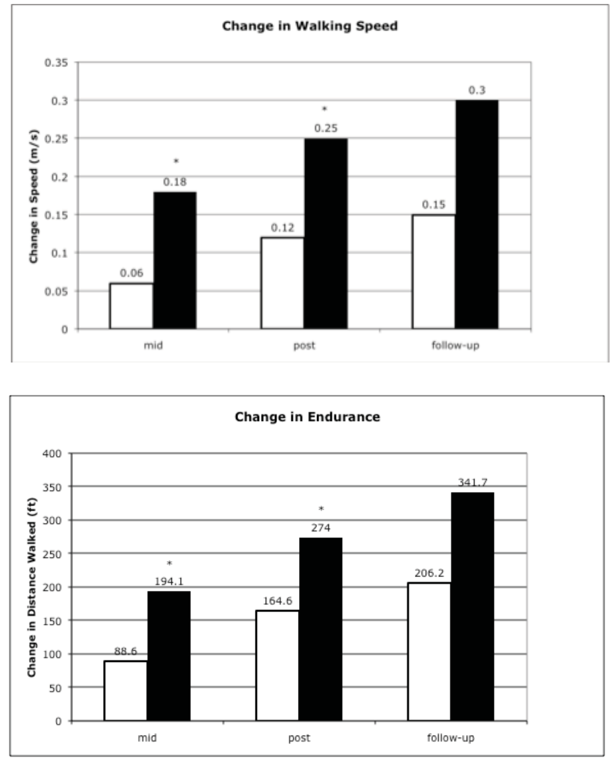

Fig 3. Changes in walking speed and endurance for Lokomat and Conventional Therapy groups. The Conventional group outperformed the Lokomat group in both improvements in walking speed and endurance.

63 subjects were trained for 8 weeks in either the Lokomat or with conventional gait training methods (33 Lokomat, 30 Conventional). Evaluation of walking ability was done at study entry, after 12 sessions, after 24 sessions, and at a 3-month follow up, which included over-ground walking speed, endurance, balance, strength, measures of disability and quality of life.

It was found that the conventional therapy group achieved significantly greater gains in both walking speed and walking endurance than the Lokomat group, while differences in measures of balance, strength, disability and quality of life were not different between the two groups.

Our findings bring into question the utility of robotic devices in treating individuals with neurological injuries, at least in their current form. That is, devices such as the Lokomat present numerous restrictions to the individuals, so that they are unable to practice key aspects of walking. For example, the device limits the amount of trunk and pelvis motion, does not allow circumduction, and dissociates the arm from walking patterns. We believe that all of these behaviors significantly limit the utility of devices such as the Lokomat and need to be addressed in future robotic systems.

In our laboratory at the National Rehabilitation Hospital in Washington DC, we have made a first pass at developing a gait training device that builds off our prior research. We believe that in order to maximize returns in walking ability after neurological injuries such as stroke, spinal cord injury and other disorders, individuals need to practice walking over-ground, including the navigation of obstacles. The system must encourage key aspects of gait (e.g. weight shifting, trunk control etc) yet must provide adequate support for the subject for safety purposes. In the end, we have developed a system called ZeroG, which is an over-ground dynamic body-weight support system.

Fig 4. ZeroG dynamic over-ground body-weight support system.

The system, shown in Figure 4, allows individuals to practice walking over-ground as well as over obstacles such as stairs and uneven terrain, in a safe controlled manner. Using devices such as ZeroG, individuals can begin practicing the gait training early after their injuries, which is important to the recovery of sable walking patterns.

SUMMARY/CONCLUSIONS

We believe the optimal rehabilitation treatments will be based on our understanding of impairments, so that interventions can properly target the primary cause of disability. Devices such as the Lokomat and ZeroG still require significant testing to determine the optimal population and treatment interventions for maximizing recovery.

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MOTOR RECOVERY AFTER STROKE –OPTIMIZING RESULTS

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STROKE THERAPIES

Enhancing motor recovery after stroke has been the focus of a broad range of research efforts in the past several years, based on the improved understanding of the plasticity of the brain and our ability to modulate this response to injury. Approaches to enhancing this recovery can be divided into several categories, including: the use of therapeutic exercise/practice, provision of sensory stimulation directly to the involved limb, direct brain stimulation, medications, and the use of stem cells and growth factors. Evidence of brain plasticity in response to exercise/motor task practice treatment is well established (Liepert, 2000).

While these approaches appear to be distinct, there is, in fact, considerable overlap. In particular, the use of therapeutic exercise/practice is a component of virtually all of these therapies. The human brain develops, functions, and remodels itself based on an incompletely understood interplay between intrinsic biological programming and interaction with the external environment. With regard to the motor system, this interplay between brain and environment occurs largely through task-oriented movements. Thus treatments are combined with task-oriented motor practice. A recent example of this approach was the Everest study sponsored by the Northstar Neuroscience company, in which low-level electrical stimulation of the brain was combined with intensive exercise therapy, and compared with exercise therapy alone (Northstar, 2008). Interestingly, while this particular study found no incremental benefit of targeted brain electrical stimulation, it did show improved motor function attributed to exercise in both the study and control groups. Therefore interventions can be divided into two fundamental parallel tracks of research: 1) finding the best exercise/practice treatment program, and 2) finding the best adjunctive method(s) of facilitating activity-dependent motor recovery in the brain to enhance the effects of practice.

Critically important issues regarding exercise/practice for motor recovery remain to be resolved. These include determining the best type of exercise therapy, the optimal time window for providing specific types of exercise, the optimal duration of such training, the intensity of treatment (i.e. session length and frequency of sessions), and a method for establishing therapeutic endpoints (i.e. when to cease therapy). Even a key study establishing the efficacy of a particular form of therapeutic exercise, such as the EXCITE trial (Wolf, 2006), leaves most of these questions largely unanswered. Attempts to ascertain the optimal intensity of physical therapy have been made, but have only answered a portion of this critical question (Kwakkel, 1999).

ROBOTICS

The use of robotic and other technologies to facilitate therapeutic exercise shows considerable promise, though some caveats exist as well. Cost remains a substantial concern, as these technologies currently represent an extra cost, rather than a replacement for other expenses (i.e. physical or occupational therapist time), and the equipment itself remains quite expensive. These issues should be addressed by improvements in technology that are expected to lead to easier to use devices that do not require direct therapist supervision, and reductions in cost as technologies mature and efficiencies of scale are achieved.

Another limitation of robotic technologies is the relatively constrained nature of the range of activities that can be performed. For example, use of a robot for gait training, such as the Lokomat® (Hocoma, Inc), does not incorporate the important activities of rising from a seated position, nor practicing turns. General purpose anthropomorphic robots remain largely in the realm of science fiction at present, and existing devices are often highly specialized. The development of mobile wearable devices for the upper limb, however, such as the Myomo e100 (Stein, 2007), creates the potential for more general purpose assistive devices to facilitate motor retraining. Considerable more research and development is needed before devices truly capable of providing robotic assistance in a broad range of functional activity training become available.

Among the advantages of robotic devices is the ability to create games or virtual reality scenarios to help make the activities more engaging and/or better simulate functional tasks. Programmable devices provide a consistent and reliable form of exercise training, allowing repeated exercises to be performed within precise parameters. Many of these devices also have measurement capabilities, allowing progress to be monitored in a quantitative fashion, and improving our understanding of the underlying changes in motor performance (Rohrer, 2004). Ultimately robotic devices may be expected to be labor-saving, and allow the delivery of more extensive therapy programs without requiring increasing the personnel available. Even with the current state of development, provision of partial body weight supported treadmill training with the use of a Lokomat® requires fewer personnel than providing this therapy with a purely manual system.

FUTURE THERAPIES

The use of progenitor/stem cells to repopulate damaged areas of the brain with replacement neurons remains an appealing long-term therapeutic goal for recovery after stroke. Based on our knowledge of brain plasticity, it is clear that any replacement neurons will need to be appropriately incorporated and entrained into functioning brain networks to provide benefit. Some form of exercise therapy involving task-oriented practice will undoubtedly be a critical component of the care of these patients. Efforts to determine the best therapeutic exercise paradigms and exercise aids will create the foundation for these therapies in the future.

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CONTACT ROBOTICS:
TOOLS TO STUDY AND TREAT NEUROLOGICAL DISORDERS

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ROBOTIC THERAPY

Since its modest beginnings over a decade ago, clinical evidence has shown that robotic therapy for stroke patients is effective (Aisen et al., 1997) and provides durable benefits (Volpe et al., 1999). The sustained rapid growth of research into this therapeutic application of robots is justified by mounting clinical evidence of its effectiveness. Recent meta-analyses of multiple clinical studies reported by Prange et al. (2006) and Kwakkel et al. (2007) have confirmed significant improvements due to robotic therapy, the latter reporting a mean effect of 7 to 8 % on motor impairment scores.

CONTACT ROBOTICS

It therefore appears that robotic therapy is well on its way to being one of the vanguard applications of *contact robotics*, applications featuring close physical contact and cooperation between robots and humans. Especially for low-functioning patients—for whom alternative approaches such as constraint-induced movement therapy are inapplicable—delivering therapy requires forceful yet sensitive physical interaction with the patient. This is a difficult engineering challenge: though haptic displays are maturing rapidly, that technology operates at very low ("fingertip") forces—reasonably so, as it is designed to interact with hands and fingers. In contrast, a therapy robot must typically operate at much higher forces. This is self-evident for locomotor therapy; the robot may need to exert forces comparable to human body weight, while maintaining sensitive control of interaction.

ENGINEERING CHALLENGES

Actuator technology is the most prominent limiting factor. While sensor technology has improved dramatically in recent decades and computational capacity continues to follow Moore's Law, actuator technology has lagged behind. Generating forces up to a human's body weight while providing a gentle, compliant behavior requires actuators with low and tunable endpoint mechanical impedance as well as high force density (the ratio of force capacity to weight), but available technologies (e.g., electromagnetic, hydraulic, pneumatic, piezoelectric) generally excel in at most one of these capacities. Newer actuator technologies (shape-memory alloys, electro-active polymers) are under investigation but have yet to approach satisfactory force magnitude or force density.

One appealing compromise solution is to use feedback control—especially force feedback—to reduce or shape the output mechanical impedance of conventional actuators. Unfortunately, substantial reduction of output mechanical impedance requires high-gain force feedback. Using typical controller design procedures, high-gain force feedback induces instability on interaction—clearly unacceptable for a robot intended to work in close physical contact with humans. Alternative design procedures (Buerger & Hogan 2007) promise to yield better performance, though much remains to be done.

BIOLOGY OF RECOVERY

Another prominent challenge of therapeutic robotics—especially acute for neurological injury, by far the most prevalent cause of sensory-motor disability—is the present state of ignorance about how robotic therapy actually works. Ultimately, recovery depends on biology, yet the details of the recovery process remain largely unknown. A deeper understanding is needed to accelerate refinement of robotic therapy or suggest new approaches. Fortunately, robots provide an excellent instrument platform from which to study recovery, especially at the behavioral level relevant to robotic treatment.

A series of studies reviewed in (Hogan et al. 2006) provide some emerging insight. Programming a robot to provide progressive resistance training to alleviate muscle weakness afforded no advantage over "assist-as-needed" training to promote sensorimotor coordination; recovery is not merely a matter of regaining strength. Using a continuous passive motion paradigm afforded no reduction of impairment, demonstrating the importance of the patient's active participation. Together with many other studies, this suggests that robotic therapy works by guiding, enhancing and perhaps even evoking activity-dependent neural plasticity. That insight may be used to design improved robotic therapy paradigms. A *performance-based progressive* algorithm that continuously assessed patient performance and progressively increased the challenge proved to be the most effective therapy we have provided to date. Remarkably (and perhaps counter-intuitively) the number of repetitions provided by intensive robotic training is not the key factor: performance-based progressive robotic therapy achieved superior outcomes with *fewer* movements (about 2/3) than our prior "assist-as-needed" therapy algorithm.

PLATEAUS AND "CHUNKING"

Robotic tools also enable completely novel observations. One striking example is the stereotypical submovements evident in the earliest actions of recovering stroke patients—remarkably similar to patterns of unimpaired movement, though substantially smaller. These submovements persist throughout recovery, becoming larger and more blended as recovery proceeds. They indicate a distinct "chunking" of motor behavior, which may have important implications for the time-course of recovery. Over a century ago, Bryan and Harter (1899) demonstrated "chunking" in the acquisition of skilled telegraphy and that it led to distinct plateaus of performance, despite continued intensive training. It does not seem unreasonable to suggest that similar plateaus of performance in recovering patients may be a consequence of "chunking" in motor behavior, though this speculation remains to be tested.

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ROBOTICS FOR LOCOMOTOR TRAINING AFTER SPINAL CORD INJURY

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INTRODUCTION

Body-weight supported treadmill training (BWSTT) has been used to enhance locomotor function in individuals following a spinal cord injury (SCI), stroke, or other neurological conditions (Barbeau et al, 1998). Locomotor training using BWSTT is based on principles that promote the movement of limbs and trunk to generate sensory information consistent with locomotion to improve the potential for the recovery of walking after neurologic injury (Harkema, 2001). To replicate a normal gait pattern multiple therapists are needed to conduct the task specific training sessions to control or assist with trunk and limb kinematics.

The success of manual BWSTT in restoring over ground locomotion has been documented in individuals following a SCI (Wernig et al, 1998, Behrman and Harkema 2000, Protas et al, 2001). Despite these improvements, the use of BWSTT has been limited due to the strenuous nature of the manual training for the therapists. As therapists fatigue, it becomes increasingly difficult to maintain symmetry between the steps; which may be an important aspect in locomotor training. Furthermore, step-to-step consistency is hard to maintain. In addition, most rehabilitative settings have limited resources to commit the time of multiple individuals to the training of one patient. The development of robotic devices to assist gait rehabilitation has been motivated by the limitations of manual locomotor training and to improve the delivery of BWSTT in the clinical setting. The goals are to reduce therapist physical demand and time, improve repeatability of step kinematics, and to increase volume of locomotor training. A popular commercially available device is the Lokomat (Hocoma AG, Volketswil, Switzerland). It uses a position controlled gait orthosis with a Lokolift body weight support system and a computer controlled interface.

EFFECTIVENESS OF ROBOTIC LOCOMOTOR TRAINING

This presentation will examine the current evidence for the efficacy of the Lokomat for locomotor training following SCI. Improvements in multiple measures of gait have been reported following Lokomat training (Hornby et al, 2005, Wirz et al, 2005). Muscle activity (EMG) was found to be similar between manual and robotic BWSTT (Colombo et al, 2001) but altered compared to treadmill walking without assistance (Hidler and Wall, 2004). Foreman et al (2006) reported attenuated EMG in the anterior tibialis with the current ankle control mechanism in the Lokomat compared to manual BWSTT.

To maximize the effectiveness in the clinical setting, robotic BWSTT can be augmented by other concurrent rehabilitation techniques. One example is coupling robotic BWSTT with FES. Synchronized FES stimulation has assisted swing limb advancement during robotic BWSTT (Querry et al, 2006). Subjects with a motor incomplete SCI who were non-ambulatory after 36 sessions of robotic BWSTT demonstrated improved overground walking outcomes with coupled FES. Biofeedback provides the subject with visual motivation to maximize active participation with robotic assistance.

NEURAL MECHANISMS OF LOCOMOTOR RECOVERY

Robotics are being utilized to understand neural mechanisms of locomotor recovery. Advantages include independent manipulation of variables, quantification of joint angles/forces, angular velocities, and synchronize control of external equipment. An example of external control by the Lokomat is the measurement of the H reflex during gait (Querry et al, 2008). Ongoing studies are examining the effect of injury severity and modulation with robotic locomotor training on H reflexes in SCI subjects.

Previously, fMRI analysis demonstrated that robotic BWSTT in individuals with motor incomplete SCI resulted in locomotor recovery and is associated with neural plasticity at supraspinal centers; sensorimotor areas and the cerebellum (Winchester et al, 2005). fMRI is limited to an isolated motor task paradigm. SPECT allows evaluation of supraspinal plasticity during actual locomotor tasks (Winchester et al, 2007). An increase in neural activity (radioactive uptake) was observed in similar areas (sensorimotor area, cerebellar vermis) during walking in the Lokomat and over ground walking compared to supine rest. Furthermore, neural activity in the sensorimotor area and cerebellar vermis were similar during walking in the Lokomat and overground walking suggesting that Lokomat training is a good training paradigm for overground walking.

ROBOTICS IMPROVE ASSESSMENTS

Robotics also provide the opportunity to standardize assessment tools. The Lokomat includes L-Force (a measure of isometric volitional torque of hip and knee flexors and extensors) and L-Stiff (a measure of resistance to passive motion at the hip and knee during three different speeds).

SUMMARY/CONCLUSIONS

Robotic BWSTT can augment gait recovery after SCI. Robots can be used as effective training tools and in conjunction with other treatments. Functional recovery of gait is associated with neural plasticity probably at both spinal and supraspinal levels. Robots can be used as testing/measuring tools and may help to determine optimal training parameters.

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Symposium

Spasticity Assessment (SA)

CLINICAL ASSESSMENT OF MOTOR DEFICITS IN SPASTIC CEREBRAL PALSY

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INTRODUCTION

This symposium will focus on motor deficits in patients with spastic cerebral palsy. Cerebral palsy (CP) is a clinical syndrome that results from a non-progressive lesion or static defect of the immature central nervous system producing a chronic motor disability. CP is the most prevalent physical disability originating in childhood with a reported incidence of 1 to 3 per 1,000 live births (Rosen and Dickinson 1992; Kuban and Leviton 1994; Blair and Stanley 1997). Spasticity is the most common physiologic manifestation of CP and the primary cause of muscle contracture in children with CP. Taft gave the frequency of spasticity as 70-80% (Taft 1995). Spasticity is caused by a lesion interrupting fibers of the tracks that control voluntary movements (pyramidal tracks). The symptoms include exaggerated, velocity sensitive, response to muscle stretching, and may include increased resistance to passive stretching (muscle tone), positive Babinski sign, and exaggerated deep tendon reflexes. Spastic diplegia is the most prevalent clinical grouping, particularly in the premature, low-birthweight infant and occurs as a result of cystic periventricular leukomalacia (Rogers, Msall et al. 1994). Physicians have grappled with classifying the types of cerebral palsy since early recognition of this disorder (Osler 1987). Cerebral palsy can be classified in a number of ways but typically is divided by tone: spastic, hypotonic, dyskinetic, or mixed. It can also be classified by body part: hemiplegia, diplegia, triplegia, double hemiplegia, or quadriplegia. The distinction between diplegia and quadriplegia is not always clear. Patient history and physical examination remain the keystones for a diagnosis and classification of cerebral palsy.

AMBULATION ABILITY

Once a diagnosis of CP has been made, a common question parents ask is "Will my child be able to walk?". There are four prognostic factors which are important to predict ambulation in CP: topography, reflex profile, developmental skills, and medical co-morbidities (Bleck 1975; Molnar and Gordon 1976; Watt, Robertson et al. 1989; da Paz Júnior, Burnet et al. 1994; Trahan and Marcoux 1994; Sala and Grant 1995; Liao, Jeng et al. 1997). The type of cerebral palsy is highly prognostic. A child with hemiplegia will walk in nearly 100% of the cases. A diagnosis of diplegia is also favorable with approximately 85-90% of children in this category achieving ambulation. Quadriplegia has a less favorable prognosis. But children with quadriplegia may have ambulatory potential as late as 7 – 8 years of age. Primitive reflex and postural reaction testing at 1 and 2 years of age can provide useful information for predicting primitive ambulatory skills. The 8 reflexes have been identified as significant with regards to ambulation: moro, asymmetric tonic neck reflex, symmetric tonic neck reflex, neck righting, tonic labyrinthine, extensor thrust (positive supporting reaction), foot placement, and parachute. The first six of these reflexes/reactions are normally absent at one year and the latter two are normally present. If reflexes/reactions normally absent by one year are present at one year, or even more significantly at two years, progress for ambulation is poor. Of all the predictor for ambulation, sitting balance is the strongest. Sitting balance is typically achieved at about six months of age. Children with CP, who are able to sit by the age of 24 months, can be expected to walk without assistive devices. Children who are unable to sit at age 4 years of age will likely be unable to walk. CP is often associated with other medical co-morbidities. Epilepsy appears to be statistically most related to ambulatory skills and is more marked for poor ambulatory outcome.

TREATMENT OPTIONS

A wide range of treatments are available for patients with CP. Treatments vary according to the patient's age, physical condition, and nature of the movement disorder. Further, one or a combination of treatment options may be used. Physical and occupational therapy is frequently used. Stretching and strengthening exercises are especially important for preventing or limiting permanent muscle shortening caused by contractures. Contractures may also be treated by serial casting. If further treatment is needed oral medications are used. Most of these medications are mildly effective. Moderate or severe spasticity is almost never completely eliminated by oral medications. Intramuscular injections with Botulinum toxin are given to patients to decrease spasticity in specific muscles. The advantage of this form of treatment is its localized effects. The medication weakens only the injected muscle and rarely causes any kind of central nervous system side effects. If this treatment is not sufficient, intrathecal baclofen therapy (ITB therapy) is considered. ITB therapy delivers an antispasmodic drug directly to the fluid surrounding the spinal cord in a small, precisely controlled dose using a pump that is placed internally. ITB therapy is effective in more than 86% of people screened for severe spasticity as a result of cerebral palsy, including many who have not had good results with oral medications. Other surgical options include selected dorsal rhizotomy and/or orthopedic surgery. Selected dorsal rhizotomy helps reduce spasticity primarily affecting the legs. Rhizotomy relives spasticity but will not improve contractures that are already present, nor is it effective treatment for dystonia, athetosis, or chorea. Orthopedic surgery can correct or prevent the muscle and bone deformities that are caused by spasticity, but does not directly change the spasticity itself. The most common orthopedic surgeries are tendon lengthening or release, tendon transfer and osteotomies.

DYNAMIC ASSESSMENT

The traditional assessment of patients with CP has focused on history and physical examination. However, physical examination provides only a passive assessment of muscle tone. Computerized motion analysis laboratories have enhanced the clinical evaluation of patients with cerebral palsy (Sutherland, Olshen et al. 1988). Currently, gait analysis techniques have emerged to aid in treatment planning for children with CP. Motion Analysis techniques make it possible to evaluate the complex gait abnormalities in these patients, often involving more than one joint, and to assess the effects of intervention. The goals for normal locomotion are 1) stability in the stance phase of gait, 2) clearance of the foot in swing, 3) proper foot preposition in swing, and 4) adequate step length (Gage 1990). Gait analysis provides a dynamic assessment of movement which is not available in any other format.

SYMPOSIUM

After providing an overall framework for the clinical assessment of children with spastic cerebral palsy, this symposium will investigate the underlying characteristics of muscles in children with cerebral palsy. First, the biomechanical changes in spastic human muscle will be discussed by Dr. Richard Lieber. A greater understanding of the underlying mechanisms, specifically the role of the motor unit in motor movement deficits in patients with cerebral palsy will then be discussed by Dr. Jessica Rose. These structural and electrophysiological changes that take place in the motor unit of patients with cerebral palsy will provide greater understanding for treatment implications.

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Neuromuscular Deficits in Spastic Cerebral Palsy

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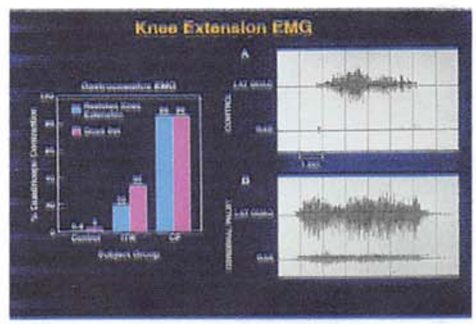
Introduction

Advances in the study of human walking have enhanced our understanding of gait disorders through quantitative analysis of kinematics, kinetics, energetics and electromyographic (EMG) measures of muscle activity during gait. However, the mechanisms underlying motor deficits in spastic cerebral palsy (CP) are not well understood. Loss of selective motor control, muscle weakness, muscle spasticity, short muscle-tendon length due to muscle: bone growth rate discrepancy, and joint contracture co-exist in spastic CP (1,2,4). The relative contribution of these motor deficits to the movement disorder varies among individuals and clinically is not well delineated. The underlying mechanisms are interrelated and arise from interruption of descending motor signals, resulting from a non-progressive brain injury around the time of birth (7). Sensory-motor deficits such as diminished proprioception and poor postural balance may also impair motor performance (3,5). Understanding the structure-function relationships that influence motor performance in CP is an essential step towards developing more effective treatment. This review summarizes a series of studies that examined neuromuscular mechanisms underlying motor deficits in spastic CP.

Selective Motor Control

Toe walking, often referred to as equinus gait, is a common gait deformity in spastic CP and exemplifies the multifaceted nature of motor deficits. Toe walking may arise from a combination of factors that lead to excessive ankle plantarflexion in stance phase of gait. It may result directly from short gastroc-soleus and joint contracture or from insufficient knee extension in late swing due to hamstrings that are short and/or slow to lengthen prior to initial contact. Loss of selective motor control also contributes to toe walking due to co-activation of gastrocnemius-soleus (GAS) during quadriceps (QUAD) contraction in late swing which extends and stabilizes the limb for initial contact (4,6). Selective motor control was assessed with EMG in 24 children, 8 with spastic diplegic CP, 8 with idiopathic toe walking and 8 controls. An obligatory co-activation between GAS and QUAD was found during voluntary knee extension in children with CP, the GAS was active for 86% of QUAD contraction compared to 0-3% for controls.

Figure. 1. Obligatory co-activation of gastrocnemius & quadriceps during voluntary knee extension in CP.

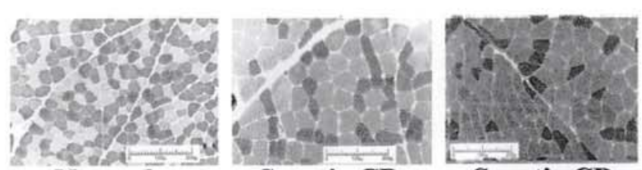


Loss of selective motor control occurred even in mild cases. The EMG test was used to differentiate mild diplegic CP from idiopathic toe walking (4,6).

Spastic Muscle Pathophysiology

Muscle structure is highly responsive to neural stimuli and reveals underlying deficits. Histologic and morphometric analysis of spastic muscle from 10 children with spastic CP was compared to controls (1). Type-1 fiber predominance was found in spastic muscle and correlated to prolonged gait EMG (r=0.77,p=.03). Increased fiber size variation

Figure 2. Spastic muscle showed increased ratio of type-1: type-2 fibers & high fiber size variability.

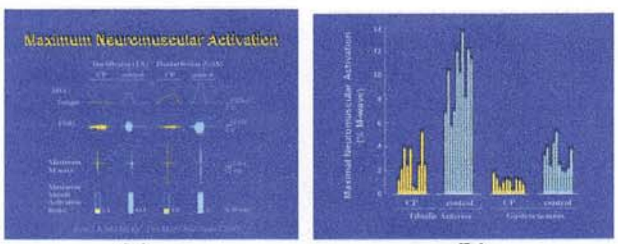


Normal **Spastic CP** **Spastic CP**

was also found in spastic muscle and correlated to energy cost during gait (r=0.69,p=.05). Findings suggest that muscle fiber changes in spastic muscle relate to degree of functional disability. Structural changes seen in spastic muscle are similar to those seen with experimental, chronic low frequency (<20 Hz) electrical stimulation, suggesting that motor-unit firing rates may be reduced in spastic CP.

Neuromuscular Activation & Motor-Unit Firing

Muscle strength, neuromuscular activation and motor-unit firing characteristics of spastic muscle were assessed during voluntary contraction of gastrocnemius (GAS) and (TA) muscles in 10 participants with spastic hemiplegic or diplegic CP and 10 controls (7). Participants with CP, produced less torque in ankle dorsiflexion and plantarflexion.. To control for antagonist co-activation and inability to fully activate muscle, maximal neuromuscular activation was assessed (voluntary EMG / M-wave amplitude) (Figure 3a and b).



(a) (b)

Figure 3. Torque (a) and maximal neuromuscular activation (a & b) in spastic CP and controls

Maximal neuromuscular activation was reduced to 34% of control values in GAS and to 25% of control values in TA. Motor-unit firing rates and recruitment were not different at submaximal levels of neuromuscular activation (Figure 4a and b). However, at maximal neuromuscular activation, predicted maximal motor-unit firing rates for TA and GAS were reduced to 50% of control values and were 16 and 13 Hz for participants with CP compared to 31 and 25 Hz for controls. Reduction of motor-unit firing rates in spastic CP may decrease muscle strength and impair muscle growth, given the number of muscle growth factors stimulated by muscle fiber activation.

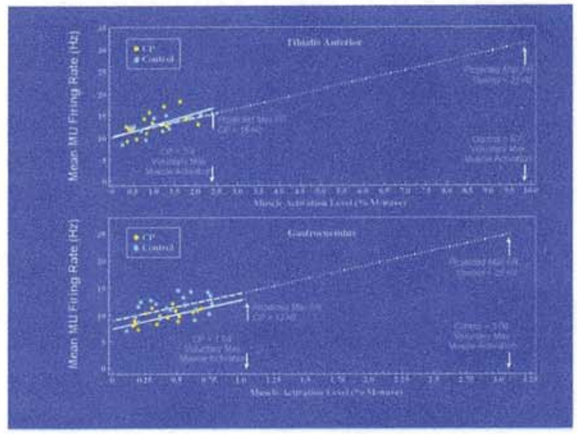
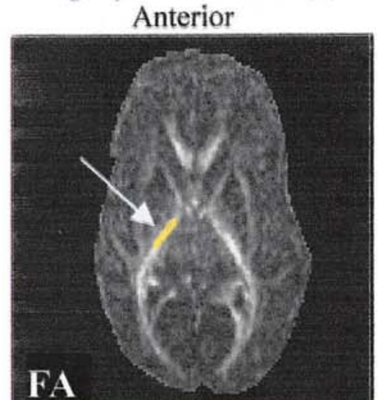


Figure 4. Motor-unit firing rates during voluntary contraction of TA and GAS in CP and controls.

Neonatal Brain Structure & Motor Deficits

Motor deficits in spastic CP, including loss of selective motor control, short muscles and decreased neuromuscular activation result from loss of descending excitatory and inhibitory signals, following injury to the developing brain. The corticospinal motor tracts descend through the internal capsule posterior limbs (PLIC). Abnormalities in the PLIC have been found to be associated with a diagnosis of CP. To better understand these structure-function relations, neonatal microstructure development of the PLIC was assessed using MRI diffusion tensor imaging (DTI) and compared to gait and motor deficits in 24 very low birthweight preterm infants (8).

Anterior



Posterior

Figure 5. Brain DTI fractional anisotropy (FA) reflects neuronal fiber tract development including integrity, number, size and/or mylenation of axons. Arrow points to internal capsule posterior limbs.

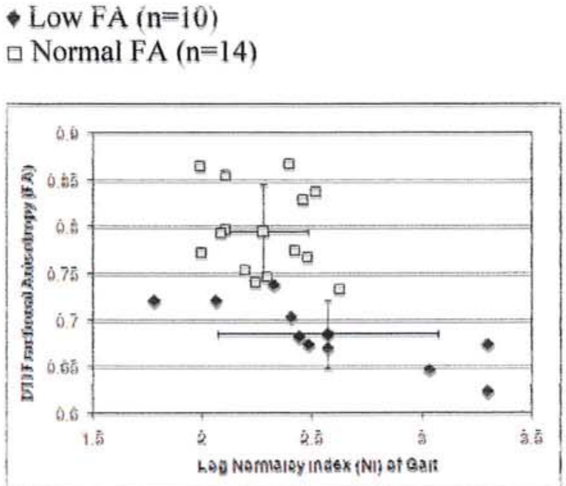


Figure 6. Scatter plot of DTI fractional anisotropy of combined right and left sides of internal capsule posterior limb and severity of gait deficits on Gillette Gait Index ($r=0.89$, $p=.001$).

Approximately 15% of very low birthweight children develop motor deficits and CP. Results indicated a strong correlation between neonatal PLIC development and later gait (Figure 6) and motor deficits (GMFCS) ($r=0.65$, $p=.04$) in preterm children with low FA in the PLIC. The data suggest that neonatal brain DTI may provide a basis for early prognosis and guidance for early intervention.

Summary

Movement disorders in spastic CP arise from loss of selective motor control, muscle weakness, short muscles, joint contracture and muscle spasticity. Sensory-motor and balance deficits may also impair function. The mechanisms underlying motor deficits are interrelated and result from interruption of descending motor signals following brain injury. Reduction in excitatory and inhibitory motor signals may result in spasticity, diminish synergist and antagonist inhibition, reduce motor drive and muscle strength, as well as impair muscle growth. These changes influence muscle structure and biomechanics, leading to short muscle-tendon length and joint contracture. This series of studies are an initial effort to examine mechanisms underlying motor deficits in CP. Further research is essential to clarify these structure-function relations and provide more effective treatment.

Acknowledgments

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BIOMECHANICAL CHANGES IN SPASTIC HUMAN SKELETAL MUSCLE

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INTRODUCTION

Skeletal muscle spasticity is a condition that occurs secondary to upper motor neuron lesions and can represent devastating events for affected individuals. Although the function of individuals with spastic muscles is severely compromised due to decreased range of motion, decreased voluntary strength, and increased joint stiffness, the basic mechanisms underlying the functional deficits that occur after the development of spasticity are not clearly understood. Additionally, while the etiology of spasticity is "central," most antispasticity therapy is directed toward the peripheral nerves and muscles. As a result, therapeutic interventions involving stretching, casting, splinting, neurectomies, intrathecal baclofen pump placement, botulinum toxin injection, and electrical stimulation of the muscles, are only marginally effective. In this review, I will summarize several current studies that define the biomechanical properties of human muscles in children with cerebral palsy (CP).

METHODS

For the intraoperative experiments, flexor carpi ulnaris (FCU) sarcomere length was measured by laser diffraction in children with CP (n=9) who were undergoing surgery for wrist flexion contractures. (Lieber and Fridén, 2002). These data were compared to "normal" FCU sarcomere length from patients (n=12) who were undergoing surgery for radial nerve palsy (Lieber *et al.* 1997). Briefly, a Helium-Neon laser beam was aligned with a specially-designed prism-mount such that the beam projected normal to one prism face and was reflected 90˚, exiting the other prism face and transilluminated an isolated muscle fiber bundle. The FCU sheath was incised 5 cm proximal to the wrist crease and a small fiber bundle directly beneath the incision was isolated using delicate blunt dissection along a natural fascicular plane within the muscle.

For the in vitro laboratory experiments, mechanical properties of single cells and small muscle fiber bundles were determined using a micromechanical testing apparatus (Fridén and Lieber, 2003; Lieber *et al.* 2003). Briefly, biopsies were excised and immediately placed in a muscle-relaxing solution so that fibers would not depolarize. Single fiber segments or small bundles were dissected from biopsies while in chilled relaxing solution under 40X magnification with epiillumination and transferred to an experimental chamber filled with relaxing solution (Fig. 1).

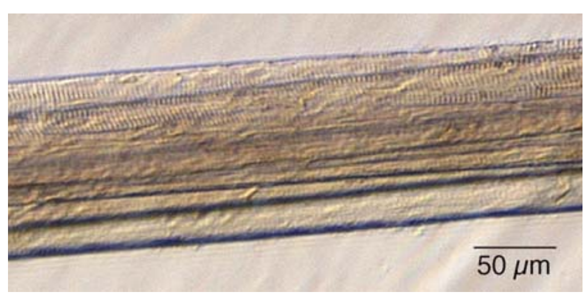

Figure 1: Micrograph of small bundle of muscle fibers for mechanical testing. The ends of this specimen are secured to a force transducer and motor.

After transfer and prior to mounting, the specimen was transilluminated with a He-Ne laser beam to define slack sarcomere length. The specimen was then secured on either side to 125 µm titanium wires using two individual 9-0 silk suture loops that were required to prevent sample slippage. One wire was secured to an ultrasensitive force and the other was secured to a micromanipulator. Structural integrity of the specimen treated in this way was excellent as evidenced by clear striation patterns and diffraction patterns. Specimens were then lengthened in 250-µm increments after which stress-relaxation was permitted for 2 minutes and sarcomere length, tension, and specimen diameter was again recorded.

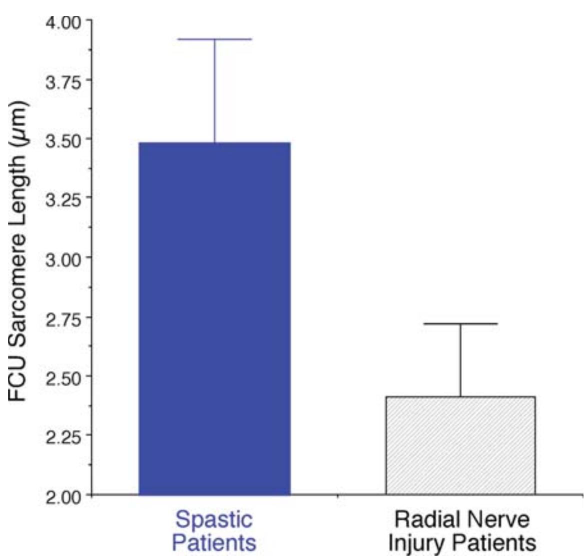

Figure 2: Sarcomere length measured from the FCU muscle of a patient measured with a spastic wrist flexion contracture, (filled bar) or radial nerve injury, which denervates wrist extensors but leaves the FCU intact (hatched bar). Note the abnormally long sarcomeres in muscles from these patients in spite of the fact that their muscles are abnormally short. Data represent mean±SEM for each group.

Specimens were elongated until mechanical failure of any of the fibers within the bundle occurred, which resulted in a precipitous loss of tension. Sarcomere length and force recorded just prior to the lengthening that resulted in failure were defined as peak sarcomere length and used to calculate ultimate stress, respectively. Fiber bundle tangent modulus was calculated as the slope of the specimen's stress-strain curve.

RESULTS AND DISCUSSION

Sarcomere length measured intraoperatively in children with severe wrist flexion contractures was dramatically longer even though their muscles were measurably shorter (Fig. 2). These data demonstrated that, while the spastic wrist flexion contractures resulted in a shortened muscle-tendon unit, the sarcomeres within the fibers were actually lengthened.

A precise mechanism that explains how a muscle could shorten due to spasticity is not currently available. Usually, muscle adaptation is explained in terms of fiber length not muscle length changes. Since fiber length remained constant but sarcomeres were consistently highly stretched, "shortening" had to occur at some other level. We hypothesize that spasticity caused muscle shortening without fiber shortening which could occur in a highly pennated muscle like the FCU if entire fibers were simply degenerated or degraded leaving a shorter muscle belly to span the entire length. Should this explanation of muscle adaptation be true, appropriate therapeutic measures would be needed to restore muscle length to normal or to permit fibers to shorten to more favorable lengths for active and passive force generation. Such procedures could include muscle-tendon lengthening, tendon grafting or some newly developed procedure to actually increase muscle length.

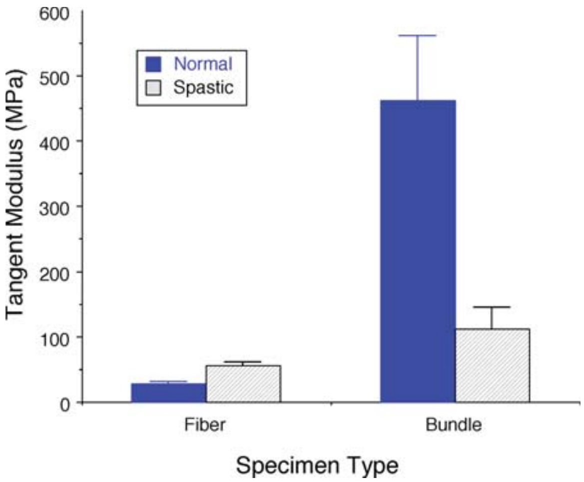

Figure 3: Tangent modulus measured from single cells (left panels) and muscle fiber bundles (right panels) of normal (open bars) and spastic (filled bars) subjects. The tangent modulus was calculated from the linear portion of the sarcomere length-stress relationship for these specimens. This indicates a severe alteration of the muscle fiber relative to the extracellular matrix. Data represent mean±SEM for each group.

Interestingly, the tensile modulus of muscle cells from spastic patients was over twice that compared to the tensile modulus from neuromuscularly intact patients (left panel of Fig. 3), demonstrating that the intrinsic passive stiffness of individual spastic muscle cells was increased. In addition, the resting sarcomere length of spastic cells (i.e., the length of the sarcomeres when the muscle cell was completely unloaded), was significantly shorter in spastic muscle cells compared to normal cells. Internal cytoskeletal structures set the resting sarcomere length in muscle cells (Wang et al. 1993; Labeit and Kolmerer, 2000). These findings demonstrate that the structures <u>within</u> the muscle cell responsible for setting resting sarcomere length and determining cellular elastic modulus are altered in spastic muscle

When comparing isolated cells to small muscle bundles, several interesting findings emerged: First, the tangent modulus measured in bundles was significantly greater compared to the same modulus measured in single cells (right panel of Fig. 3). However, the difference was much more pronounced for normal muscle bundles compared to spastic muscle bundles. Whereas spastic muscle bundle modulus was only increased 2 fold over single fiber modulus, in normal muscle, the modulus was increased over 16 times compared to the modulus of the normal isolated muscle cell. These data demonstrate a clear difference in the mechanical properties of spastic muscle tissue bundles compared to normal muscle fiber bundles. The differences are even more impressive after the structural differences between the two bundle types are considered—Only 40% of the spastic muscle bundle cross-sectional area was occupied by muscle fibers where as 95% of normal muscle bundle was occupied by muscle fibers. Morphologically, there was a huge amount of poorly organized extracellular material in spastic bundles compared to normal bundles.

One can calculate the mechanical properties of the extracellular matrix material in the bundles by subtracting single cell modulus from whole bundle modulus. When this is done, it is seen that the extracellular matrix of the spastic muscle has a modulus of ~0.2 GPa while normal muscle has a modulus of ~8 GPa—about 40 times greater. These data demonstrate that, while spastic muscle contains a larger amount of extracellular matrix material within it, the quality of that material is much lower compared to normal muscles.

SUMMARY/CONCLUSIONS

In spite of the fact that the primary lesion in CP is in the central nervous system, dramatic changes in muscle are observed. Future studies are required to study the structural basis of these observations as well as the cellular mechanisms that link neural input with muscular structure and function.

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Symposium

Telerehabilitation (TR)

TELEREHABILITATION; ENABLING REMOTE MONITORING AND REMOTELY SUPERVISED TRAINING

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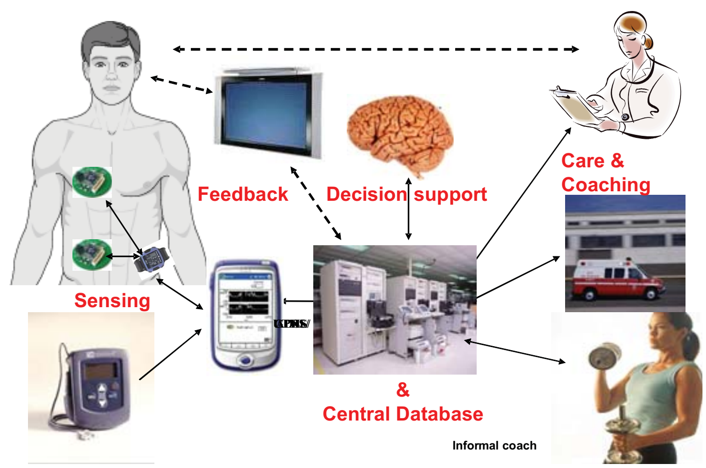
INTRODUCTION

The growing number of elderly and people with chronic disorders in our western society puts such pressure on our healthcare system that innovative approaches are required to make our health care more effective and more efficient.
One way of innovation of healthcare can be obtained by introducing new services which enable less pressure on the intramural health care and support a more independent living and self efficacy of patients. Two of such services are Remote monitoring and remotely supervised training (RMT). Remote monitoring enables freedom to the patient with the assurance that assistance is possible whenever required. Remotely supervised treatment enables efficient and effective user-centred training anywhere and anytime with an intensity not feasible in an intramural setting. It is our vision that remote monitoring and remotely supervised treatment applications will become very important for patients (safety, more in control, convenience), health care insurances (efficiency, cost reduction) and healthcare service providers (more effective, innovative).

METHODS

A schematic drawing of the main components and the information exchange pathways are shown in the figure below. RMT systems are in general quite complex, requiring dependable end-to-end systems integrating ambulant sensing of relevant biosignals and context information, an M-Health platform for a secure data transport and storage and a backend system with appropriate decisions support with respect to both technical and clinical aspects. Another key element is the feedback of the information (dotted lines), which is essential for the subject and the care providers to learn from the data and to make the appropriate decisions.



(Hermens, 2008)

RESULTS

During the past years considerable knowledge and experience is being gained with the research and development of such smart systems. In parallel, experience is gained with respect to the implementation of RMT systems in a clinical environment and how to assess its potential benefits.
Examples concern:
Activity monitoring in low back pain: the daily activity patterns of patients with

chronic low back pain have been assessed, indicating a similar amount of total daily activity compared to healthy subjects but with a different distribution over the day. The results clearly show in general a higher activity level in the morning and lower activity levels in the afternoon and evening. This provides a starting point for treatment focussed on restoration of the balance in acivity pattern by providing adequate personalised feedback

Monitoring of spasticity: continuous monitoring of muscle activity patterns by means of surface EMG measurement, in subjects with a spinal cord lesion provides insight in the magnitude and distribution of spasticity over the day. It gives direct objective and quantitative information during which activities and during which part of the day spasticity is most hindering the activities of daily living and as such it provides valuable information to optimise the drug delivery to decrease spasticity.

Myofeedback in subjects with neck shoulder and low back pain. An ambulatory myofeedback treatment was developed which involves measurement of the muscle activation of neck/shoulder muscles and providing feedback by means of vibration when relaxation in these muscles is insufficient. Clinical trials (EC project NEW) show that this treatment works well but to make the treatment more efficient, a service is being set-up to enable remote monitoring of the muscle activation patterns and remote consultation (Exozorg; EC project Myotel)

As a first step in the development of a myofeedback treatment for low back pain, low back muscle activation patterns have been monitored during the week in healthy subjects and patients with low back pain. Based on the patterns a specific ambulatory feedback program is being developed to assist in normalising the muscle activation patterns in the low back pain patients

Post rehabilitation home training: A modular system was developed to enable a wide range of upper extremity exercises at home, which can be monitored at a distance. In an international study (E.C. project Hellodoc) it was shown that the effects were very similar to the traditional intramural treatment. There was also a clear indication that subjects who train more, do improve more. This underlines the potential that with this kind of training, the patient is in the driver seat and can significantly influence the results of his training.

CONCLUSIONS

The combination of biomedical engineering with information and communication technology has opened a new area with a high potential. Regarding the present status of remotely supervised treatment one could state that until now the focus has been on the technical realisation of the sensing and transportation part of it. The development of a backend system with an appropriate decision support system is still in its infancy as well as models how to organise these services and how to make them profitable. Our experience gained in the past years in a rehabilitation setting, has learned us that this is a way forward with great potential to increase both the quality of treatment as well as its efficiency. It requires however a step-by-step approach with a strong and continuous involvement of a multidisciplinary team to guarantee a successful implementation.

WEARABLE TECHNOLOGIES TO MONITOR MOTOR RECOVERY AND FACILITATE HOME THERAPY IN INDIVIDUALS POST-STROKE

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INTRODUCTION

More than 700,000 people are affected by stroke each year in the United States [American Heart Association, 2005]. Strokes affect a person's cognitive, language, perceptual, sensory, and motor abilities [National Institute of Neurological Disorders and Stroke, 2004]. More than 1,100,000 Americans have reported difficulties with functional limitations following stroke [Center for Disease Control, 2001]. Recovery from stroke is a long process that continues beyond the hospital stay and into the home setting. The rehabilitation process is guided by clinical assessments of motor abilities, which are expected to improve over time in response to rehabilitation interventions. Telerehabilitation has the potential to facilitate extending therapy and assessment capabilities beyond what can be achieved in a clinical setting.

Accurate assessment of motor abilities is important in selecting the best therapies for stroke survivors. These assessments are based on observations of subjects' motor behavior using standardized clinical rating scales. Wearable sensors could be used to provide accurate measures of motor abilities in the home and community settings and could be leveraged upon to facilitate the implementation of telerehabilitation protocols. Herein we report results from pilot studies exploring the use of wearable sensors (accelerometers) and an e-textile glove-based system designed to monitor movement and facilitate the implementation of physical therapy based on the use of video games.

METHODS

In the studies herein reported, we investigated two technologies that have potential for application in telerehabilitation: accelerometers, which can be utilized as part of a body sensor network, and a sensorized glove, which was assembled using electro-active elastomers printed on lycra.

The first part of our study was focused on testing the hypothesis that accelerometers can be utilized to predict Wolf Functional Ability Scores (FAS). The FAS score provides a measure of the subject's quality of movement. The scores capture factors such as smoothness, speed, ease of movement, and amplitude of the compensatory movements. Twenty-three subjects who had a stroke within the previous 2 to 24 months were recruited for the study. Accelerometers were positioned on the sternum and the affected (i.e. hemiparetic) arm. Sensor data was recorded using the Vitaport 3 (Temec BV, The Netherlands) ambulatory recorder, which was worn on the waist. Subjects performed multiple repetitions of tasks requiring reaching and prehension, selected from the Wolf Motor Performance Test. The tasks included reaching to close and distant objects, placing the hand or forearm from lap to a table, pushing and pulling a weight across a table, drinking from a beverage can, lifting a pencil, flipping a card, and turning a key. Accelerometer data were processed to derive features that captured different aspects of the movement patterns and were fed to a classifier built using a Random Forest. The Random Forest approach is based on an ensemble of decision trees and is suitable for datasets with low feature-to-instance ratio.

The second part of the study herein summarized was focused on assessing feasibility of utilizing a sensorized glove to implement physical therapy protocols for motor retraining based on the use of video games. The glove was utilized to implement grasp and release of objects in the video games. This function was achieved by defining a measure of “opening of the hand” and estimating it based on processing data gathered from the sensorized glove. This measure was defined experimentally by asking individuals to hold a wooden cone-shaped object with diameter ranging from 1 cm to 11.8 cm. The output of the sensors on the glove was used to estimate the diameter of the section of the cone-shaped object corresponding to the position of the middle finger. A linear regression model was utilized to estimate the above-defined measure of “opening of the hand” (dependent variable) using the glove sensor outputs as independent variables.

RESULTS AND DISCUSSION

FAS score predictions were derived using the Random Forest approach based on feature sets derived from accelerometer data recorded during performance of motor tasks that are part of the Wolf Motor Performance Test as mentioned above. We assessed the reliability of the estimates achieved using this method by deriving the prediction error for each of the investigated motor tasks. The estimated prediction error for such motor tasks ranged between about 1 % and 13 %. This is a very encouraging result as it suggests that FAS scores could be estimated via monitoring motor tasks performed by patients in the home and community settings using accelerometers.

Encouraging results were also achieved from the study of the above-described sensorized glove. The estimation error that marked the measures of “opening of the hand” as defined above was smaller than 1.5 cm. We consider this result as satisfactory in the context of the application of interest, i.e. the implementation of video games to training grasp and release functions in individuals post-stroke.

Overall, the results herein summarized indicate that the investigated wearable technologies are suitable to implement telerehabilitation protocols.

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APPLICATION OF A SYSTEMS ARCHITECTURE FOR ROBOT-AIDED TELEREHABILITATION

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INTRODUCTION

Arm therapy is used in the neurorehabilitation of patients with upper limb paresis due to lesions of the central nervous system. Besides traditional physical therapy, task oriented repetitive movements can help patients recover motor function, improve motor coordination, learn new motor strategies and prevent secondary complications, as many studies using robot-aided therapy attest.
Telerehabilitation is the delivery of rehabilitation services through a telecommunication network and the internet. Apart from experiments in various disciplines that pertain more specifically to the field of telemedicine, the scenario of applications demonstrating the potential for remote diagnosis and treatment through robot-aided telerehabilitation is quite recent.
Telerehabilitation may well be able to optimize the therapeutic intervention, despite the fact that the patient does not directly interact with the therapist . This is so not only in the home care setting, but also in the clinical setting where it makes it possible for a therapist to monitor several patients at once, at their training stations located in different laboratories.

This paper presents a preliminary experience carried out in our Rehabilitation Institute to verify the feasibility of implementation of a telerehabilitation approach based on the application of robotic devices developed in our research laboratories.

METHODS

Rehabilitation Devices: The systems architecture implemented includes three devices developed for upper limb rehabilitation:1)a one degree of freedom (DoF) wrist manipulator specifically designed for the rehabilitation of wrist flexion and extension movement. 2) a two DoF shoulder elbow manipulator which allows robot-aided therapy by administering a sequence of reaching movements in the horizontal plane. 3) a graphic tablet-based device developed to improve the quality of movement (accuracy, efficiency and smoothness) in patients with mild impairment.
The systems architecture implemented consists of a number of rehabilitation stations (a maximum number of 16 was selected) and a supervision station, located in different laboratories. All are interconnected by means of a standard Ethernet II network. The supervision station Is also connected to the Internet in order to link up with remote rehabilitation stations in the home setting. In this preliminary presentation only the in-clinic subnet will be presented.
Rehabilitation Station: it consists of a rehabilitation device (robot or tablet) and web cam directly connected to the network (IP camera).
Supervision Station: it represents the central node of the network, where the therapist manages all the activities of each rehabilitation station. Consisting of a remote workstation, it: a) selects the patients to be supervised; b) sets up the rehabilitation

protocol and, if required, modifies the exercise parameters; c) monitors in real time the patient's performance by means of specific charts; d) transfers and archives the data and parameters acquired by the rehabilitation station; e) stores the collected data in a central data base; f) post-processes the collected data; g) reviews performance charts and allows comparison of charts; h) prints performance reports; k) allows audio/video communication with the rehabilitation station, in order to implement one-to-one and one-to-many therapist to patient interaction.

TLEREHABILITATION EXAMPLE

This systems architecture was tested during the rehabilitation of four patients after chronic stroke.

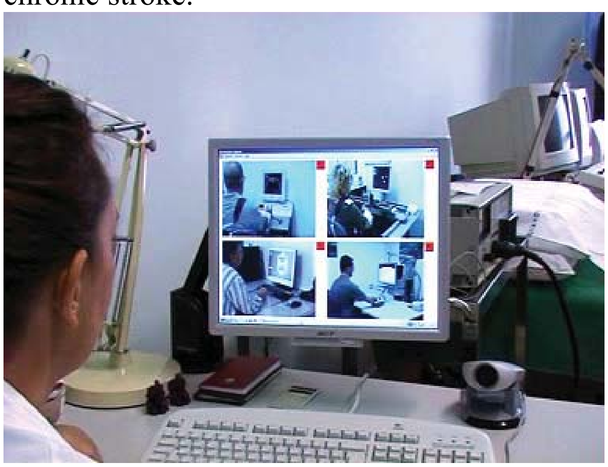

Figure 1: One-to-many therapist to patient interaction.

The testing was preceded by a learning phase in which patients were trained in order to be autonomous in connecting to the device and starting the exercise. If the patient could not attain autonomy for this task the caregiver was instructed to attach the patient to the device.

Thanks to the remote control program the therapist could take complete control of the remote device and select a new motor task when a change in difficulty level was required. The values of some exercise parameters were logged into a file of the rehabilitation station. In this way the settings of the previous session could be used as default for the following exercise session.

DISCUSSION AND CONCLUSIONS

This pilot study shows the feasibility of implementation of a telerehabilitation approach based on the application of robotic devices to increase training intensity in patients after stroke. The robot-assisted teletherapy was well accepted and tolerated by all patients. In addition, the new technological context facilitated therapy planning for the medical professionals and therapists, including the possibility to continue the rehabilitation program in the home setting. Of course, extended application in a sizeable group of patients is required to evaluate if the improvement of patients' motor ability obtained through telerehabilitation is similar to that obtained in controlled laboratory conditions.

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TECHNOLOGY SUPPORTED FEEDBACK STRATEGIES TO TREAT PATIENTS WITH CHRONIC DISORDERS

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INTRODUCTION

A lot of patients with chronic disorders experience problems with physical functioning and this is, among others, often related to deviations in motor control. As an example, development and maintenance of chronic work related neck shoulder problems seem to be related to insufficient muscle relaxation (Veiersted, Westgaard and Andersen, 1993). In addition, for patients with chronic lower back pain, evidence exist that these patients show deviation in their muscle activity patterns during walking (van der Hulst et al, 2008) as well as in their physical activity pattern over the day (van Weering et al, 2008). Besides being an explanation for a decreased level of functioning, deviations in motor control can also be used as starting point for treatment. The hypothesis is that normalization of motor control contributes positively to changes in functioning. As one of the key elements in the treatment of patients with chronic disorders is to support and empower patients to change their level of functioning themselves it would be good if treatment could be provided in the daily environment of the patient. One way of doing this is by continuously monitoring motor control parameters during normal life and give feedback to the patient on these parameters. However, insight in the most optimal feedback strategies is hardly available and needs to be researched.

Feedback strategies

Traditional care is characterized by face to face contact between healthcare professionals and patients. In most cases the patient visits the clinical practice of the care professional for diagnoses, for necessary investigations and/or for receiving treatment. The care professional uses relevant data of the patient for the decision making process and during subsequent treatment, data are used to assess treatment outcome and during the treatment sessions to provide feedback to the patient. When offering care in the home situation of the patient, face to face feedback is not as intensive as during traditional care. However using technology offers the possibility to give feedback continuously, anywhere and at anytime. Technology can not totally replace the face to face contact between the patient and professional but can take over partly the role of the face to face feedback.

METHODS

Different experiments were performed in which various feedback strategies were applied to patients with chronic pain. Voerman et al (2004) investigated the most optimal time interval of feedback on EMG patterns during computer work (Voerman et al, 2004). In a subsequent study the differences in pain and disabilities after a 4 weeks treatment, consisting of tactile feedback given on muscle relaxation at a 10 sec interval combined with a weekly visit to a professional were investigated for patients with chronic neck shoulder pain (Voerman et al, 2007). Huis in ‘t Veld et al (accepted) studied the same parameters after a 4 weeks treatment but with a visual feedback in addition to the tactile feedback and with the professional consultation partly replaced by telephone.

In addition to focus on changes after also the working mechanisms of feedback were investigated. It was investigated whether feedback does induce changes motor control and whether this is relates to changes in functioning (Voerman et al, 2007; van Weering et al, submitted).

RESULTS AND DISCUSSION

The overall results of ambulant feedback strategies to improve functioning are positive. Voerman et al (2007) showed that intensive tactile feedback (vibration) on muscle relaxation levels reduces complaints in 50% of the patients. Van Weering et al (submitted) showed that feedback on activities normalized the activity levels of patients on the days of feedback.

Concerning the modalities, Voerman et al (2004) showed that a 5, 10 and 20 second feedback interval all are effective but the 10sec time interval was most optimal to induce changes in EMG parameters. Huis in ‘t Veld (accepted) showed that this 10 sec interval tactile feedback with a continuous visual feedback on sEMG patterns in addition is possibly more effective than tactile feedback only. Concerning the working mechanism Voerman et al (2007) showed that rather than muscle activation patterns, cognitive behavioral factors underlie the working mechanisms. van Weering et al (submitted) hypothesized that an increase of awareness concerning inadequate behavior might be of importance.

SUMMARY/CONCLUSIONS

Based on the results presented in this paper it can be concluded that technology supported feedback strategies on motor control parameters positively influence the functioning of patient with chronic disorders and when provided in an ambulant way it might be an efficient way of care provision for the future. However the working mechanisms of feedback seem to be merely cognitive behavioral than a relearning of motor skill.

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Symposium

Virtual Reality in Rehabilitation (VR)

VIRTUAL REALITY AND GAMING APPROACHES TO PROMOTE WALKING AND MOBILITY POST-STROKE: FROM THE LAB TO THE CLINIC

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INTRODUCTION

Restoration of walking is identified by individuals post-stroke as one of their most important rehabilitation goals.(Bohannon et al. 1991) The standard of care for rehabilitation of ambulation includes task specific training, body weight supported treadmill training and pre-gait activities. Emerging approaches to rehabilitate walking of individuals post-stroke include motor imagery (Dunsky et al. 2006) and virtual reality (VR).

In a recent review paper, four different approaches were described, which use virtual reality technology to rehabilitate walking and mobility of individuals post-stroke.(Deutsch and Mirelman 2007) The authors found that the technology to delivery the virtual environments (VE) differed both in hardware and software. The evidence to support the use of each approach as well as the ease with which it can be applied clinically differed well. Generally though the evidence to support use of virtual reality for rehabilitation was positive, the challenge was to translate the lab findings to clinical practice.

The purpose of this paper is to elaborate on this previous publication by updating the information on use of virtual reality to improve walking and include the literature on off-the shelf gaming technology Methods of delivery, level of evidence and ease of use are reviewed. Emphasis is placed on the applications of the findings to practice.

METHODS

MEDLINE (1996- current), IEEE *Xplore* databases and selected conference publications (International VR, IWVR, Virtual Reality and Cyberpsychology) were searched using the terms virtual reality, gaming, walking, ambulation, mobility and stroke. Papers that described virtual reality and or gaming consoles for rehabilitation of walking and or mobility for persons post-stroke were selected and reviewed. Articles were included, irrespective of the level of evidence if they included a physical intervention to improve walking or mobility for individuals post-stroke. Articles that addressed deficits specific to neglect and way finding and navigations skills were excluded. Papers were reviewed for methodology and technology. Data were extracted on patient chronicity, pre-training gait speed (if provided), dosing of intervention and relevant outcome measures. Information on cost, when available was obtained from the manufacturer's website or by direct inquiry.

RESULTS AND DISCUSSION

As reported in the original review there were four methods in which virtual reality was used to rehabilitate walking post-stroke. (Deutsch et al. 2004; Fung et al. 2006; Jaffee et al. 2004; You et al. 2005) Research from some of these groups was updated since the last review. (Mirelman et al. 2007)New literature on use of gaming console for rehabilitation of mobility was found. (Flynn et al. 2007)

Tables 1: summarizes the hardware and software used to deliver the virtual reality training. Table 2 summarizes the subjects, dosing and outcome measures. Table three has information about cost, availability and level of evidence.

The VR systems currently available to rehabilitate walking for people in the chronic phase post-stroke, can be categorized as direct and indirect walking interventions. The indirect interventions have higher dosing. All groups incorporate principles of motor learning into their simulations and provide multisensory and cognitive feedback to the users. Outcomes have focused on motor performance and motor control. Cognition and self-efficacy measures which were explored in a VR way finding study (Lam et al. 2006) may be appropriate to incorporate into gait rehabilitation studies. The active ingredients in each system have not been fully explored. Transfer of the technology to practice has limitations as discussed in the next section.

SUMMARY/CONCLUSIONS

Virtual reality systems to improve walking for individuals post-stroke have been developed in a variety of ways. The early findings show transfer of training from walking in virtual environments and or physically training gait related activities to improved walking in the real world. The challenge with implementing these technologies in the clinic is that several are not commercially available and those that are in some cases cost prohibitive for most institutions. Off the shelf gaming software has only been tested for feasibility and not efficacy. Further research with gaming technology may demonstrate efficacy and therefore justify use in practice. Comparison between highly specialized virtual reality applications relative to the commercially available gaming consoles may be the object of future research.

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Table 1:

Citation	**Design**	**Hardware**	**Simulation**	**Feedback**
Jaffe et al	Randomized clinical trial With follow-up	HMD, TM with harness	Obstacle clearance	Visual, auditory, vibrotactile
You et al	RCT	Video capture	Stepping, “Sharkbait,” snowboarding	Visual, proprioceptive, faded KP and KR
Deutsch et al Mirelman et al	Double-baseline, pretest/posttest, blinded, single group single blind RCT	Desktop display with haptic robot as input device	Navigation of plane and boat through air or seascape	Visual, auditory, haptic, KR, and KR summary feedback
Fung et al	Two cases	TM on Stewart platform, Rear projected VE	Corridor walking, street crossing, park stroll	Visual, auditory, KR
Flynn et al	One case with six month follow up	Play Station II	15 games	Visual, auditory, KP and KR

Note: HMD = head-mounted device; KP = knowledge of performance; KR = knowledge of results; RCT = randomized controlled trial; TM = treadmill, VE = virtual environment
Table 2

Citation	**Sample size and acuity**	**Initial gait speed**	**Outcome measures**	**Dose**
Jaffe	*N* = 20 X = 3.7 y VR	VR Group x = .52m/s *SD* .22m/s Range .26–.82 m/s	• Balance • Gait Speed • Endurance Obstacle course	1–2 hrs, 3x/wk, for 2 weeks 120 steps/session
You	*N* = 10 X = VR 2.2 y Control 4.7 y	Not provided	FAC MMAS fMRI	1 hr, 5x/wk, for 4 weeks
Deutsch Mirelman	*N* = 6 X= 4.3 years *N = 18* X = 7.1 years	x = .64 m/s Range .14–.84 m/s x = .66 m/s Range .40–.95 m/s	Gait speed, elevations, endurance, coordination as above plus gait kinetics & kinematics community ambulation	1 hr, 3x/wk, for 4 weeks 200–500 ft movement/session As above
Fung	*n* = 2 4.5 months 2 years	 1.32 m/s 0.74 m/s	Success or failure of habituation & adaptation to VE	10- to 15-min session 39-m walks, each simulation (3) trials and increased complexity
Flynn	*N = 1*	12.73 sec on the TUG	Feasibility and Transfer to Function	1 hour, 20 sessions at home without supervision

Note: FAC = Functional Ambulation Category; MMAS = Modified Motor Assessment Scale

Table 3

	Commercially Available	Cost (in dollars)	Level of Evidence
Jafee et al	no	$$$	2
You et al	yes	$$ (10,000)	2
Deutsch et al	no	$$$	2, 3
Fung et al	yes	$$$$$$ (>500,000)	5
Flynn	yes	$ (200)	5

USING VIDEO CAPTURE VIRTUAL REALITY FOR REHABILITATION OF FUNCTIONAL MOBILITY

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Repetition is important both for motor learning and the cortical changes that create this learning. Repetition alone is not enough but must be linked to incremental success at a task. This is achieved by trial and error with feedback provided about success or lack of success through multiple sensory systems. Virtual reality (VR) systems offer the unique capability for real-time feedback to participants during practice – and importantly, the feedback is very intuitive. Practicing movements over and over as is required through trial and error however requires significant motivation. In VR, applications can be modified, tasks can be simplified and distractions minimized allowing participants to focus on key aspects of the task.

We have been using video capture VR in intervention studies for balance rehabilitation, recovery of arm function, ankle selective motor control as well as studies of learning. The platform, GestureTek IREX, uses a two-dimensional (single camera) video image of the user, processes the image with color subtraction software to remove a monochrome background and inserts the user into a virtual environment (VE). There is no need for peripherals marking body landmarks as proprietary software is used to allow the user to interact with virtual objects in a natural manner within the VE.

Applications used in various studies include: 1) a juggling task where the participant is required to reach laterally to juggle virtual balls; 2) a conveyer belt task where the participant is required to turn sideways, pick up a virtual box from a virtual conveyer belt, turn and deposit the box on a second virtual conveyer belt; and 3) a snowboard task where the user is required to lean sideways to avoid trees, rocks and other virtual objects while boarding down a hill. The applications are adjusted according to the functional abilities of the user where task difficulty is modified by changing the number of virtual objects to contact, the speed at which the objects or environment moves, or the height of the objects requiring users to reach to the ground or to step up onto a stool.

SPECIFIC APPLICATIONS

Balance and functional mobility: The position of the center of pressure (COP) is regulated during movements resulting in perturbations of balance and equilibrium. Multiple sensory inputs are used in establishing relevant frames of reference for postural control. We have studied age-related regulation of COP movement using distinct approaches to delivering VEs when visual information incongruent with vestibular and somatosensory information is provided. The lack of an exocentric frame of reference in head-mounted display virtual environments results in limiting COP movement within the base of support (BOS) in order to decrease the challenge to the postural control system.

There are limited data supporting effectiveness of exercises for balance problems even though exercise is understood to be a basic part of the management of balance problems after traumatic brain injury (TBI), in healthy aging and following a cerebrovascular accident. Our studies determined: whether early intervention with VR was feasible and safe in a sub-acute post-stroke population; whether VR would result in better community balance and functional mobility than an activity-based program with long-term TBI survivors having residual balance deficits; and whether healthy older adults who self report balance problems could benefit from a VR exercise program. All groups of participants were able to interact with the VEs safely and without secondary effects. Interventions with both the TBI survivors and healthy community dwelling older adults resulted in improvements of community level balance and functional mobility and/or improvements in reaction times during dual task activities. VR approaches to balance exercise offered a new perspective on balance retraining.

Selective motor control: Children with cerebral palsy receive physiotherapy consultation and follow a home exercise program although children are often not compliant in following these programs as they often find the exercises meaningless and uninteresting. Exercise should be challenging and meaningful to the child and clinicians must use more enjoyable methods to stimulate the child to move in ways that achieve the therapeutic goals in keeping with current principles of motor learning and motor control. The attraction of children to VR applications based on gaming and entertainment technologies is evident. We compared the VR environment to conventional exercises in order to elicit specific selective movement of ankle dorsiflexion in children with cerebral palsy. Children with CP reported higher "fun" and "interest" using VR versus conventional exercises. The children also showed greater range of active ankle dorsiflexion, longer hold times and less repetitions in the VR environment. Greater levels of co-contraction were recorded in ipsilateral gastrocnemius and contralateral tibialis anterior during the VR exercises.

Integration into clinical practice: The successful integration of VR technologies into rehabilitation has demonstrated the possibilities of practicing challenging, but safe, activities in realistic environments, while being able to control the stimulus and measure the outcome. For the therapist, there is the advantage of having full control over the level of difficulty, allowing the treatment to meet individual needs. Our current work addresses facilitators and barriers for clinical uptake of VR technologies.

SUMMARY

Our work has demonstrated positive outcome for clients with balance, functional mobility or upper extremity disorders using VR applications as an adjunct to traditional client-therapist treatment. Development of an intervention modality permitting independent practice that may permit exercise in the privacy of one's home and allow clients to selectively augment their rehabilitation has popular appeal for increasing access to appropriate intervention and decreasing wait times for activity.

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USE OF VIRTUAL REALITY TO FACILITATE RECOVERY OF HAND FUNCTION

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INTRODUCTION

Virtual reality technology may be an appropriate means to provide plasticity-mediated therapies in patient populations including stroke and cerebral palsy. Computerized systems are well suited to this and afford great precision in automatically adapting target difficulty based on individual subject's ongoing performance. Virtual environments can monitor feedback specificity and frequency, and can provide adaptive learning algorithms and graded rehabilitation activities that can be objectively manipulated to create individualized motor learning paradigms. Thus, they provide a rehabilitation tool that can be used to exploit the nervous systems' capacity for sensorimotor adaptation.

Currently there are several computerized systems under development to train upper arm movement; however, none of these systems focus on hand rehabilitation. The impact of even mild to moderate deficits in hand control effect many activities of daily living with detrimental consequences to social and work-related participation. Because of fiscal constraints, current service delivery models favor gait-training and proximal arm function. Recovery of hand function is thus is an important but difficult and challenging aspect of rehabilitation.

We have previously developed a virtual reality based training system for hand rehabilitation for patients post-stroke (Merians 2006; Adamovich 2005). We were able to track ongoing performance levels, use the data to precisely adapt the difficulty levels of the tasks to be learned and record precise kinematic and kinetic outcome measures on the patients' temporal and spatial components of hand motion. Using this system, patients improved in and retained gains made in range of motion, speed, and isolated use of the fingers. Importantly, these changes translated to improvements in functional outcome measures. As a group, subjects improved their Jebsen Test of Hand Function (JTHF) scores by 12%. However, the system was focusing on training hand alone, and could only accommodate patients with impairments in the top quartile. We have now refined and broadened the system in order to more closely model physical therapy practice and in order to accommodate patients with greater physical impairments. We present here results of pilot studies where this system was used to retrain upper extremity in patients with stroke and cerebral palsy.

METHODS

We have developed a unique exercise system that provides for haptic guidance of arm movement in three-dimensional (3D) space which is adaptive in real time as well as on a trial-by-trial basis. The system also provides for adaptive gravity and antigravity forces, global damping to aid movement stability, bilateral simulations as well as the ability to train the arm and hand together, with haptic assistance in finger extension

(Fig. 1). Virtual environment and virtual models of subject's arms and hands can be presented in a 3D workspace using stereoscopic glasses.

Four subjects (mean age=51; years post stroke =3.5) practiced approximately three hours/day for 8 days on simulations that trained the arm and hand separately. Four other subjects (mean age=59; years post stroke =4.75) practiced for the same amount of time on simulations that trained the arm and hand together. All subjects were tested pre and post training on two of our primary outcome measures, the JTHF and the Wolf Motor Function Test (WMFT).

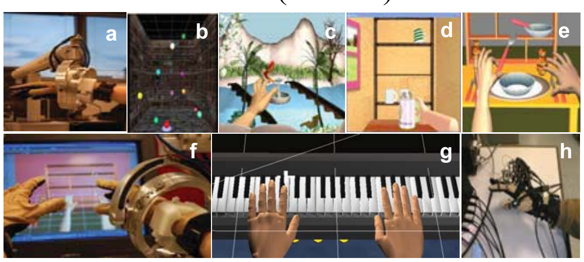

Figure 1. a. Hand & Arm Training System using CyberGlove instrumented gloves and Haptic Master robotic interfaces. **b-e.** Examples of interactive haptic environments to train reaching, grasping and object manipulation in 3D space. **f.** Close view of the haptic interface in a bimanual task. **g.** The piano trainer consists of a complete virtual piano that plays the appropriate notes as they are pressed by the virtual fingers, with haptic assistance if needed (**h**).

RESULTS AND DISCUSSION

Subjects on average showed an 18% and a 41% improvement in JTHF and WMFT clinical tests. The group that practiced arm and hand tasks separately (HAS) showed a 14% and a 9% improvement in the WMFT and in the JTHF whereas the group that practiced using the simulations that trained the arm and hand together (HAT) showed a 23% and a 29% improvement, respectively.

There were also notable changes in the secondary outcome measures; the kinematic and force data derived from the virtual reality simulations during training. Subjects in both groups showed similar changes in the time to complete each game, 39% decrease, and in the smoothness of their hand trajectories by 56% indicating better control (Rohrer, 2002). However, the subjects in the HAT group showed a more pronounced decrease in the path length (40% versus 19% in the HAS group). This suggests a reduction in extraneous and inaccurate arm movement with more efficient limb segment interactions. We speculate that this may be a result of the whole arm practice in our protocol. For training on the virtual piano simulations, subjects showed similar improvements in key press accuracy (percent change HAS=20%; HAT=17%). However, the subjects that trained using the arm and the hand together were able to complete the task much more quickly (percent change HAS=60%; HAT=151%).

SUMMARY/CONCLUSIONS

We trained patients using three different paradigms, hand alone, hand and arm separate and hand and arm together. Our initial findings point to the possibility that training the arm and hand as a unit may provide a greater advantage for improving functional activities over the other two methods.

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THE USE OF A CAVE-LIKE ENVIRONMENT FOR THE TREATMENT OF PERSONS WITH VESTIBULAR DISORDERS.

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INTRODUCTION

Persons with vestibular disorders fall frequently, regardless of age (Whitney, et al, 2000; Herdman et al., 2000), with potentially catastrophic consequences (Boult et al, 1991; Tinetti et al, 1988). Balance rehabilitation therapy has been shown to be helpful for patients with balance disorders. However, balance therapy has several limitations, particularly related to quantifying the therapy and deciding when to increase the difficulty of a patient's exercise regimen. Development of rehabilitation methodologies that can improve balance could have a great impact on public health.

Patients with vestibular disorders often complain of having difficulty with their balance and have increased symptoms in situations where there are complex visual scenes and changing visual stimuli (e.g., supermarkets, shopping malls). Walking while making head movements is one of the most difficult tasks for persons with vestibular dysfunction. Walking in and around large groups of people is also a very difficult and stressful situation for many persons with balance disorders. It is currently impossible to replicate these situations in the rehabilitation environment.

Based upon favorable results of others using VR therapy for adapting the vestibule-ocular reflex, VR appears to have the potential to address some of the limitations of balance rehabilitation therapy. Specifically, VR can be used to create increasingly challenging environments in a controlled and safe setting (Ring, 1998). Also, the therapist can dose the exposure to the provocative stimuli. Virtual reality allows the physical therapist a degree of control over the environment that is not normally possible (Ring, 1998).

The purpose of this study is to begin to test the effectiveness of virtual reality as a therapeutic tool for persons with vestibular disorders. We will describe the study design and preliminary results that have been obtained in subjects with vestibular disorders.

METHODS

Subjects were randomized into one of two possible treatment groups: A) conventional vestibular rehabilitation therapy, and B) virtual reality therapy. All subjects were referred by a neurotologist (JMF), who confirmed the presence of vestibular dysfunction based upon vestibular laboratory testing, including ocular motor screening, positional testing, rotational chair testing, caloric irrigation, an audiogram, vision testing, and computerized dynamic posturography. The type of vestibular dysfunction could be peripheral, central or mixed origin.

Prior to either intervention, subjects had a baseline assessment of self-reported symptom and disability measures, and a balance and dizziness assessment. The following self-report questionnaires were completed: the Chambless Mobility Inventory, the Simulator Sickness Questionnaire, the Activities-specific Balance Confidence Scale, the Dizziness

Handicap Inventory, the Vestibular ADL, and the Situational Characteristics Questionnaire. Balance/gait measures include gait velocity, the Dynamic Gait Index, the Functional Gait Assessment and Timed Up and Go. In addition, subjects reported their dizziness severity in normal room light and when experiencing full-field visual stimulation.

Each group completed 6 treatment sessions. For the vestibular rehabilitation group, a licensed physical therapist with over 10 years of experience in vestibular rehabilitation provided the treatments according to her clinical expertise.

For the virtual reality therapy, patients underwent 6 virtual reality exposures of 4 minutes duration, with a rest break between each exposure period. The virtual environment consisted of a grocery store modeled in 3D Studio Max and imported into Unreal Tournament (UT2004), adapted for multi-screen environments with the CaveUT modification. The store was displayed on 3 screens that surrounded the patients in a full-field CAVE-like environment. The store contained 16 aisles with that were 20 m long, and had 8 levels of visual complexity that depended on the spatial frequency and contrast of the product textures. To move through the store, patients walked on a self-paced treadmill that was controlled by the amount of anterior force exerted on an instrumented shopping cart fixed to the treadmill. The patients were asked to find products on the shelves as they ambulated down the aisle.

Subjective units of discomfort scores were taken throughout the session, and heart rate and blood pressure were monitored during the rest periods using an electronic blood pressure cuff. Initially, subjects ambulated down the aisles with least visual complexity. As the therapy sessions progressed within and between sessions, more visually complex scenes (e.g. virtual grocery store) were projected depending on the patient's symptom tolerance.

In both groups, patients were given home exercises and are asked to keep a daily exercise diary. A post-treatment evaluation was conducted one week after the final treatment session.

RESULTS AND DISCUSSION

Currently there are 6 patients enrolled in the study and findings of their experience will be discussed. To date, all subjects have tolerated the virtual reality treatments well and no adverse events have occurred.

SUMMARY/CONCLUSIONS

We are currently running clinical trials to examine if virtual reality therapy is a useful tool for treating people with vestibular disorders.

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ACKNOWLEDGEMENTS

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Track 01

Electrode Arrays (EA)

RELIABILITY OF MUAP PROPERTIES IN MULTI-CHANNEL ARRAY EMG RECORDINGS OF UPPER TRAPEZIUS AND STERNOCLEIDOMASTOID

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INTRODUCTION

Muscle activity can be assessed non-invasively by means of electrodes placed at the skin overlying a muscle. When an array of many closely spaced electrodes is applied, it is possible to extract motor unit action potentials (MUAPs) from the EMG signals. These MUAPs can be characterized by parameters describing their amplitude (RMS_{MUAP}) and their frequency content ($FMED_{MUAP}$, median frequency of the power spectrum of the MUAP shape). RMS_{MUAP} is related to the size of the motor unit (MU) and the frequency content is related to the conduction velocity of the MU (Lindstrom and Magnusson 1977). In addition, the number of MUAPs per second (MUAP Rate) provides a measure for the input of the central nervous system to the muscle (Kallenberg and Hermens, 2006). Surface array EMG is getting more common in research focusing on motor control and in clinical applications. However, not much is known about the reproducibility of the parameters obtained in this way. The objective of the present study was to determine the test-retest reliability of the mentioned parameters during different tasks of the shoulder and neck muscles.

METHODS

Eight-channel linear arrays (Merletti et al., 2003) were placed on the upper trapezius (UT) and sternocleidomastoid (SCM) muscles of 12 healthy subjects. Subjects performed 3 tasks: shoulder abduction (90 degrees), ironing (subjects had to repetitively touch two ends of a horizontal bar positioned in front of them), and 90 degrees head turning towards the non-dominant side. The protocol was performed twice while electrodes remained in place (trial 1 & 2) and repeated a week later (trial 3).
Global root-mean-square value (RMS_G) and median frequency of the power spectrum ($FMED_G$) of the EMG signal were calculated. MUAPs were extracted with the segmentation part of the decomposition software described in Gazzoni et al. (2004) and MUAP Rate was calculated. RMS and median frequency of the power spectrum of the extracted MUAP shapes were averaged across all MUAPs (RMS_{MUAP} and $FMED_{MUAP}$). Intra-class correlation coefficients (ICC), F statistics, and within-subject normalised standard errors of the mean (nSEM) were calculated to assess repeatability (trial 1 x 2) and test-retest reproducibility (trial 1 x 3). ICCs provide an indication of the ability of a parameter to distinguish between subjects but are only valid when the between-subject variability is larger than the within-subject variability. This can be confirmed with the F statistic (being the ratio of the between-subject and within-subject variability). Within-subject nSEMs reflect the experimental noise due to trial-to-trial variability (trial 1 x 2) and day-to-day variability (trial 1 x 3) and therefore can be used to quantify repeatability.

RESULTS AND DISCUSSION

In 3 subjects, UT was not active during the ironing task and in 3 other subjects, SCM data quality was insufficient during the head turn task. Data from these subjects were not included in the analysis. Tables 1 to 3 show the ICCs, F statistics and the nSEMs.
All parameters except MUAP Rate showed good reliability and acceptable nSEM values. nSEM was somewhat higher for trial 1 x 3 than for trial 1 x 2, meaning that replacement of the array introduces additional noise, as would be expected. ICC estimation for MUAP Rate was not valid for the shoulder abduction tasks due to equal within- and between-subject variances, resulting in negative ICC values. It is likely that the high contraction level required for this task resulted in superimposed MUAPs, leading to detection errors. This may explain the low reliability for MUAP Rate.

CONCLUSION

Even though MUAP Rate could not be reliably estimated in all cases, MUAP parameters showed a good reliability, similar to that of the global variables.

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Table 1 ICCs, nSEMs and F values for the shoulder abduction task (upper trapezius)

	ICC		Within-subject nSEM		F statistic	
	Trial 1 x 2	Trial 1 x 3	trial 1 x 2	Trial 1 x 3	Trial 1 x 2	Trial 1 x 3
RMS_G	0.97	0.86	3.04	6.72	54.5	13.4
$Fmed_G$	0.98	0.96	1.00	2.13	92.9	59.3
MUAP Rate	0.92		3.67	11.2	24.2	0.811
RMS_{MUAP}	0.95	0.78	3.92	8.27	38.2	7.78
$Fmed_{MUAP}$	0.97	0.87	0.87	2.07	105	15.0

Table 2 ICCs, nSEMs and F values for the iron task (upper trapezius)

	ICC		Within-subject nSEM		F statistic	
	Trial 1 x 2	Trial 2 x 3	trial 1 x 2	Trial 1 x 3	Trial 1 x 2	Trial 1 x 3
RMS_G	0.91	0.86	5.97	12.5	49.2	12.2
$Fmed_G$	0.95	0.93	3.40	4.05	35.3	25.4
MUAP Rate	0.64	0.84	14.50	16.5	4.30	10.7
RMS_{MUAP}	0.78	0.91	10.05	11.1	11.7	19.7
$Fmed_{MUAP}$	0.93	0.79	1.18	1.89	28.9	8.62

Table 3 ICCs, nSEMs and F values for the head turn task (sternocleidomastoid)

	ICC		Within-subject nSEM		F statistic	
	Trial 1 x 2	Trial 1 x 3	trial 1 x 2	Trial 1 x 3	Trial 1 x 2	Trial 1 x 3
RMS_G	0.96	0.84	8.46	14.9	41.3	12.8
$Fmed_G$	0.91	0.76	2.00	2.65	21.5	6.65
MUAP Rate	0.93	0.60	3.15	13.7	24.1	3.88
RMS_{MUAP}	0.95	0.83	11.21	15.1	36.8	9.87
$Fmed_{MUAP}$	0.95	0.76	1.33	4.08	65.9	7.42

INNERVATION ZONES OF THE BICEPS BRACHII MAY SHIFT WITH INCREASING TORQUE DURING ISOMETRIC FOREARM FLEXION

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INTRODUCTION

Innervation zones (IZ; points at which the peripheral nerve terminals connect with the muscle fibers) are an important factor that can affect the surface electromyogram (EMG) (Rainoldi et al. 2004). EMG signals acquired from bipolar electrodes that span the IZ often have lower amplitudes and higher center frequencies (Farina et al. 2001). Even normalized EMG values cannot be compared among individuals if the IZ confounds the signals (Beck et al. 2006). Thus, determining the location of the IZ to avoid it during EMG acquisitions has become critically important.

A recent study indicated that IZs shift proximally as the biceps brachii muscle is shortened (Martin and MacIsaac, 2006). However, they did not observe force-related changes in the IZ location. Since 10 mm interelectrode distances were used, the authors suggested that an array with smaller interelectrode distances (< 10 mm) may provide more precise information about potential shifts in the IZ.

Therefore, the purpose of this study was to track the locations of biceps brachii IZs during isometric muscle actions at 20, 40, 60, 80, and 100% of the maximal voluntary contraction (MVC) using an electrode array with 2.5 mm interelectrode distances.

METHODS

Nine subjects were included in this report (8 men, 1 woman, age range = 18-35 years). All participants signed informed consent forms. This study was approved by the University of Oklahoma Institutional Review Board.

Isometric torque of the right forearm flexors was measured on a Biodex System 3 dynamometer. Each muscle action was 6-s in duration and performed at a 120º joint angle (between the arm and forearm). Each subject performed isometric MVCs and submaximal isometric forearm flexions at 20, 40, 60, and 80% MVC in random order.

Surface EMG signals were recorded from the biceps brachii muscle with a 16-channel linear electrode array acquisition system (EMG16, LISiN-Prima Biomedical & Sport, Treviso, Italy). A 16-electrode probe (5 mm x 1 mm, 10 mm interelectrode distance) identified the initial IZ location. A smaller 16-electrode probe (1 x 5 mm prongs, 2.5 mm interelectrode distance) was then taped over the initial IZ location to record all subsequent EMG signals (bipolar recording mode, 1000x gain, 2,048 Hz sampling frequency).

Precise IZ locations were determined off-line by displaying the EMG signals for each contraction intensity with custom-written software (LabVIEW 8.5, National Instruments, Austin, TX). A trained investigator (JMD) evaluated all consecutive 0.25-s epochs throughout the entire 6-s contractions. The channel that consistently demonstrated (a) the location of phase reversal and (b) low amplitude was denoted as the IZ.

A repeated measures ANOVA was used to examine the IZ location (channel position) versus torque relationship (Figure 1).

RESULTS AND DISCUSSION

Table 1 shows the shifts in IZ locations relative to the IZs at 20% MVC.

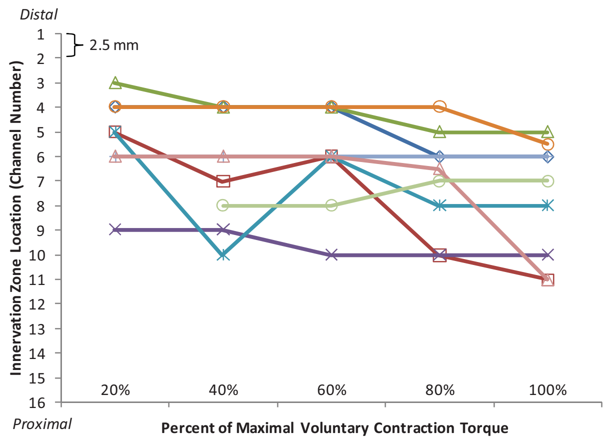

Figure 1: Absolute innervation zone locations for each subject (y-axis) as a function of isometric torque (x-axis).

The ANOVA indicated a significant (P=0.002) shift in IZ location from distal to proximal as isometric torque increased from 20% to 100% MVC. Only three of the 9 subjects demonstrated a shift greater than 10 mm (Table 1), and the mean (± SD) proximal shift was 2.2 ± 4.2 mm at 40%, 1.1 ± 1.3 mm at 60%, 3.5 ± 4.6 mm at 80%, and 5.4 ± 5.6 mm at 100% MVC.

SUMMARY/CONCLUSIONS

These findings indicated that there were small, but significant ($P \leq 0.05$), proximal shifts (range: 1.25 – 15 mm) in the average IZ location with increases in isometric torque. With the exception of 3 subjects, most IZ shifts would not have been detectible with an interelectrode distance of ≥10 mm. It is possible that slight decreases in elbow joint angle during the isometric testing may have caused the IZ shifts.

Surface EMG electrodes that assess muscle activation of the biceps brachii should be located at least 15 mm away from the nearest IZ during standard forearm flexion isometric assessments.

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Table 1: Changes in IZ location. Values are Δ scores in millimeters (mm) from the IZ location at 20% MVC. Positive values indicate proximal shifts, while negative values represent distal shifts.

Subject	20% MVC	40% MVC	60% MVC	80% MVC	100% MVC
1	0 mm	0 mm	0 mm	5 mm	5 mm
2	0 mm	5 mm	2.5 mm	12.5 mm	15 mm
3	0 mm	2.5 mm	2.5 mm	5 mm	5 mm
4	0 mm	0 mm	2.5 mm	2.5 mm	2.5 mm
5	0 mm	12.5 mm	2.5 mm	7.5 mm	7.5 mm
6	0 mm	0 mm	0 mm	0 mm	3.75 mm
7	0 mm	0 mm	0 mm	0 mm	0 mm
8	0 mm	0 mm	0 mm	1.25 mm	12.5 mm
9	#	0 mm	0 mm	-2.5 mm	-2.5 mm

 # No single innervation zone could be determined from this trial.

High Density MES Mapping of the Shoulder Complex Musculature

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INTRODUCTION

The shoulder is one of the most sophisticated and complex joints in the human body. Unlike the knee and elbow, which can be defined as a hinged joint, the shoulder is best described as a ball and socket joint capable of a series of motions with a wide range of mobility (Taylor '54). These movements are caused by a large array of muscles that act upon and displace the humerus, clavicle and scapula.

Recent work at the Institute of Biomedical Engineering at the University of New Brunswick has been investigating the potential pattern classification ability of the remaining musculature of high level amputees to be used for the control of a prosthetic limb (Buerkle 2006, Losier 2007). The electrode placements during these experiments have been largely based on literary references, which often outline which muscles are active when performing various motions (Hartigan 2004). Unfortunately, these references do not highlight the MES intensity of the musculature while performing these various movements. Having such information would allow investigators to strategically place the surface electrodes to be used for pattern classification.

In this study, MES amplitude information is presented for the shoulder complex of an able-bodied subject using a multi-channel high density surface MES electrode array. The work presented demonstrates the ability of this method to provide investigators with a general guideline by which electrodes can be positioned for pattern classification.

METHODS

A total of 101 monopolar electrodes were placed over the back, deltoid and pectoralis muscle regions of the subject's dominant side (Figure 1). The electrodes were spaced as to provide adequate coverage of the musculature regions of interest. MES were collected at a sampling rate of 2 kHz using a REFA multi channel amplifier system (TMS International, Oldenzaal, The Netherlands).

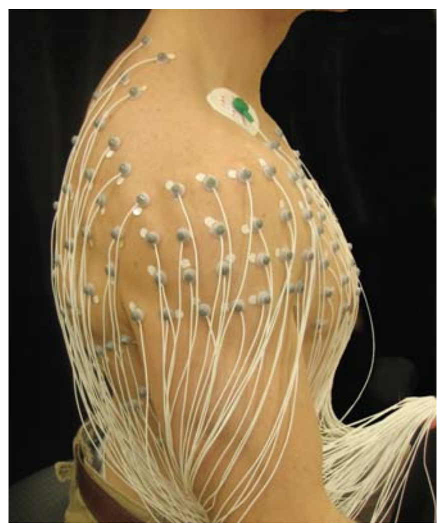

Figure 1: High density monopolar MES experimental setup.

The subject was prompted to perform twelve different movements which involved the use of muscles from the shoulder complex. Each motion was held for a four second duration and repeated twice over the course of the experiment. The subject was allocated an additional four seconds between motions to rest. The mapping of the MES data was performed offline.

SIGNAL PROCESSING

The MES data were processed using a third order high-pass Butterworth filter at 20 Hz to eliminate motion artifact and notch filtered between 55 and 65 Hz to reduce power line interference. The MES intensity was calculated using the root mean square (RMS) for each electrode.

RESULTS

The data were used to create several activity color maps corresponding to the movements performed by the subject (Figure 2). Each MES channel's intensity color was based on its associated normalized RMS value while the color gradient between points was produced by an interpolation function. It can be seen that distinct muscle activation is elicited for each movement performed by the subject. It should also be noted that the areas of high activation appear to appropriately correspond with the musculature highlighted by referenced literature.

SUMMARY

Intensity color maps are presented for multiple shoulder movements based on the calculated RMS values of high density MES data. This information will provide useful information for strategic placement of MES electrodes for future shoulder motion based pattern classification studies.

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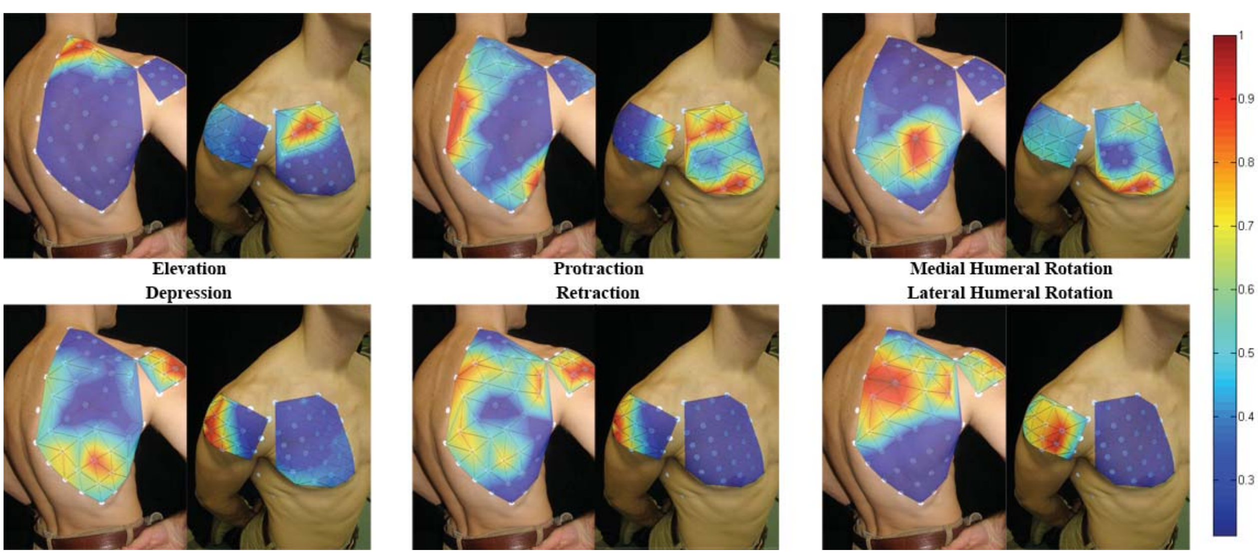

Figure 2: Examples of color maps based on MES RMS values for 6 different motions.

AUTOMATIC LOCALIZATION OF INNERVATION ZONES OF SPHINCTER MUSCLES: RELIABILITY AND REPEATABILITY OF A NEW METHOD

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INTRODUCTION

Lesions of the innervation zone (IZ) of the external anal sphincter (EAS), e.g. during delivery, can promote the development of fecal incontinence. Recently developed probes allow high-resolution detection of electromyographic (EMG) signals from the EAS. The analysis of pelvic floor muscles by surface EMG (in particular, the estimation of the location of the IZ) has potential applications in investigation of the mechanisms of incontinence and in prevention of lesions resulting from episiotomy. The level of asymmetry of IZ distribution is related to an increased risk of faecal incontinence in case of traumas (Wietek et al. 2007).
A method for the automatic estimation of the IZs of EAS from surface EMG is proposed. The performance of the method is tested on simulated signals. Furthermore, the repeatability of the indications provided is assessed on experimental signals.

METHODS

The algorithm for the estimation of the IZ distribution uses a continuous wavelet transform to extract motor unit action potentials (MUAP) from multi-channel signals. The position of the IZ of each extracted MUAP is determined considering the channel from which signal propagation starts (see Figure A).
Simulations: Simulated signals were obtained by an analytical model of generation of surface EMG signals from EAS including two layers, mucosa and muscle. Simulations were performed varying the length of the fibers, the thickness of the mucosa, the position of the motor units (MUs), and the force level. Different distributions of IZ were simulated in two sets of simulations. Simulation 1: two IZs in different positions. Simulation 2: multiple IZs, with different levels of asymmetry.
Experimental protocol: Thirteen healthy female subjects participated in the measurements. Surface EMG signals were acquired using an anal probe with three arrays (of 16 channels each) at different locations within the anal canal (15 mm distance between the centers of adjacent arrays), during four independent experimental sessions. Three maximal voluntary contractions (MVC) of 10 s were performed for each session. Repeatability of the estimation of the distribution of IZ was tested by evaluating the coefficient of multiple correlation (CMC) between the IZ distributions estimated from the signals recorded from each subject (see Figure B). Asymmetry of the IZ distribution was defined as the 2D vector barycenter of the IZ distribution in the plane orthogonal to the axis of the probe. An asymmetry index (AI), defined as the distance of the barycenter from the center of the probe divided by the radius of the probe, was also defined. Repeatability of asymmetry estimation was

assessed by the interclass correlation coefficient (ICC).

RESULTS AND DISCUSSION

The performance of the method, tested on simulations, was affected by surface MUAP amplitude (as was the identification of IZ distribution by visual inspection). The performance was affected by mucosa thickness, decreased when fiber length was higher and was affected by the distribution of MU size within the muscle. Nevertheless, the method was able to identify the locations of the two simulated IZ and to measure asymmetry of the IZ distribution.

A high repeatability (CMC>0.8) was found comparing IZ distributions estimated from signals recorded by each array during different contractions within the same session. A slightly lower value was obtained considering signals recorded during different sessions (CMC>0.7), but a higher value (CMC>0.8) was obtained after aligning the estimated IZ distributions (realignment needed to overcome operator's error in repositioning the probe exactly in the same place).

The estimation of asymmetry was repeatable (ICC about 70%) when estimated by the 2D vector barycenter, after alignment of the IZ distributions. Repeatability of AI was lower (ICC about 55%).

SUMMARY/CONCLUSIONS

A reliable and repeatable estimation of IZ distribution of EAS can be obtained by the proposed automatic algorithm. This result strengthens the potential applications of high density surface EMG in the prevention and investigation of incontinence.

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ACKNOWLEDGEMENTS

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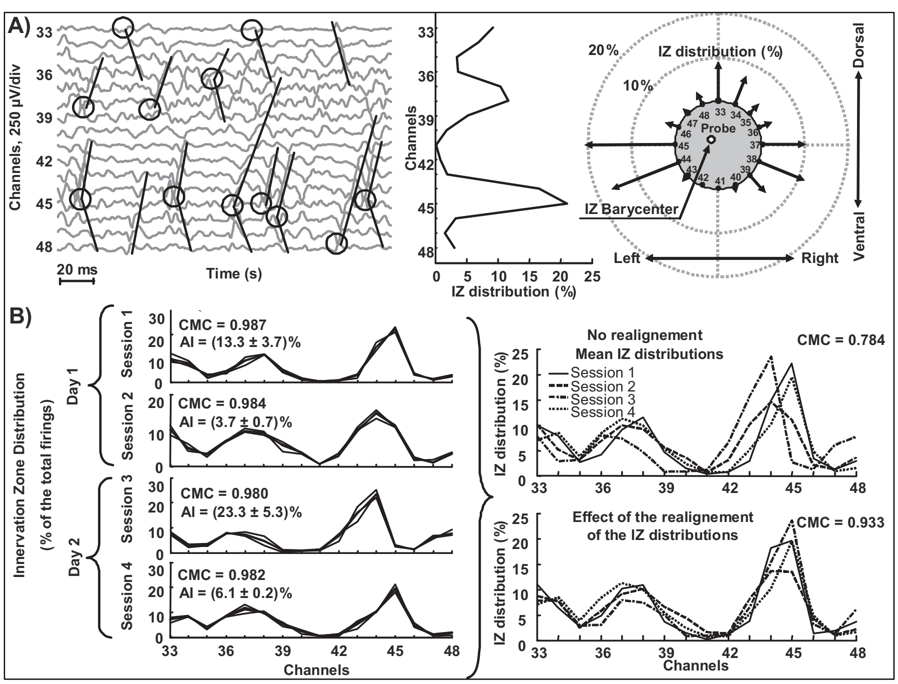

A) Example of an epoch of EMG on the external array during MVC with indication of the identified MUAPs and position of IZs (left). Estimated IZ distribution for the whole signal and indication of asymmetry (right). B) IZ distributions for each trial and session for a subject, with indication of coefficient of multiple correlation (CMC) with and without realignment between the IZ distributions and asymmetry index (AI).

INTER-RATER RELIABILITY IN LOCATING THE INNERVATION ZONE USING SEMG SIGNALS IN TRAPEZIUS MUSCLE

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INTRODUCTION

Localization of the motor endplate zone by sampling the surface EMG signal detected on the skin with a linear array or a matrix of electrodes has been proposed in the literature (Masuda et al., 1985; Merletti et al., 2003). The innervation zone (IZ) location is established by placing electrodes along the muscle fiber direction to detect the generation of the action potentials and their spatial propagation.

The aim of this study was to assess the inter-rater reliability between two operators in locating the IZ in the trapezius muscle using single differential electromyographic signals detected with a matrix of electrodes (64 equally spaced electrodes disposed on 5 linear arrays with 8mm IDE).

Reliability of this procedure is a necessary pre-condition for the use of surface EMG to describe the relationship between painful spots and motor endplate region in the trapezius muscle

METHODS

Two operators (MB,RG), involved in the data collection, underwent a training session on electrode positioning and surface EMG signal analysis. A standard reference system was defined by marking the middle point of the line between C7 and the acromial angle. The central electrode was positioned at this point. Ten healthy subjects participated in this study. Subjects were asked to sit on a special chair with the trunk against the back, and with the arms resting vertically the body. Each hands grasped a handles connected to a load cell. Maximal voluntary contraction level was determined by asking the subjects to perform three maximal voluntary contractions of the upper trapezium muscle during shoulder elevation and then taking the maximum of the three values. Both operators applied the matrix of electrodes and collected the surface EMG signal during 6 isometric contractions of 10 seconds each, 3 at 20% MVC and 3 at 40% MVC, during two separate experimental sessions. Operators and contraction type were selected randomly. Fourty signals out of 100 were then randomly selected using the following criteria:

- 20 signals collected by operator MB, 10 at 20% MVC and 10 at 40% MVC
- 20 signals collected by operator RG, 10 at 20% MVC and 10 at 40% MVC

The two operators established in a blind way the IZ location by analysing 5 epochs of 0.250 s each for each selected single differential EMG signal. The 5 linear arrays of electrodes were numbered 1 to 11, 14 to 24, 26 to 37, 40 to 50, and 53 to 63 and the IZ location indicated using the corresponding electrode number. If the IZ was detected between two consecutive

electrodes e, e+1, it was indicated as e.5 (see Figure 1).
The Bland-Altman method was used to assess the degree of agreement between the two operators (Bland and Altman, 1986).

RESULTS

Perfect agreement was reached on 169 out of 200 estimates (84.5%) . Agreement was not reached on 31 estimates based on 15 contractions at 20% MVC and 16 contractions at 40%. Bland Altman analysis showed that the mean of differences between operators did not differ significantly from zero (-0.01) and the concordance interval ranged from -0.51 to 0.49 (See Figure 2). All the estimates differences were within the concordance interval except two.

CONCLUSIONS

Our results indicate that the protocol adopted guarantees a good inter-rater agreement between two operators in the localization of the IZ of the trapezius muscle. Data also showed that the contraction level (% MVC) did not affect the quality of IZ detection and the degree of agreement between operators.

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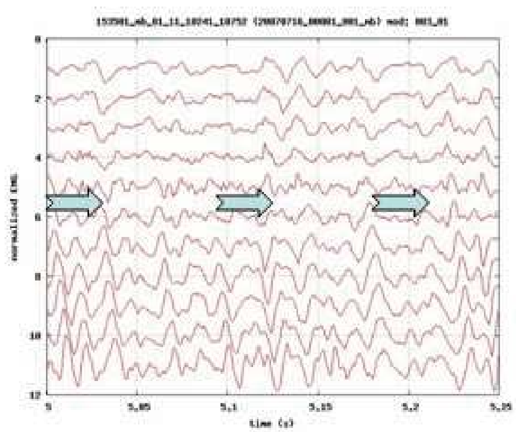
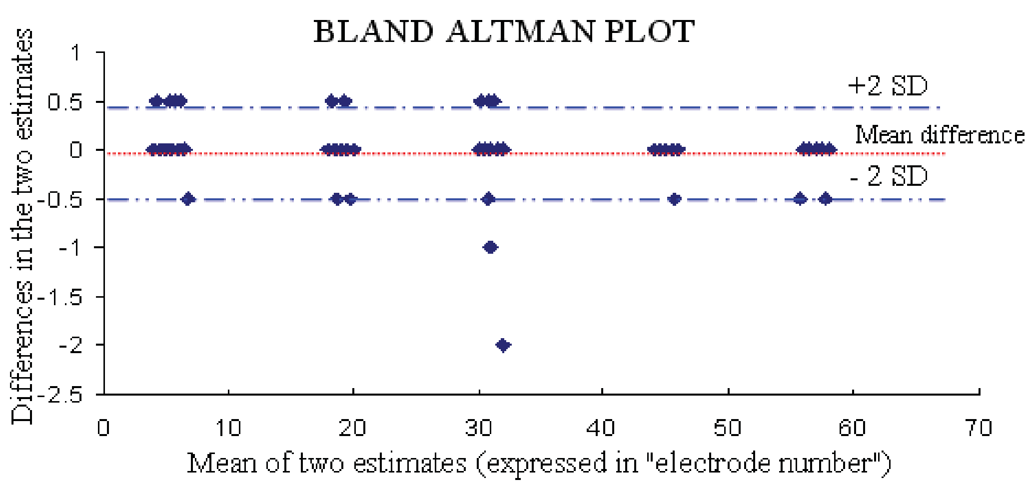


Figure 1: surface EMG epoch showing an innervation zone (IZ) between electrode 5 and 6.

Figure 2: Dotted lines represent: the limits of agreement (+/-2 SD) and the mean of differences between operators. Units on the both axes are expressed in "electrode number" (see text).



QUANTITATIVE EVALUATION OF CROSS-TALK USING HIGH-DENSITY SURFACE EMG

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INTRODUCTION

Cross-talk in surface EMG (sEMG) can be defined as the unwanted registration of myoelectric activity of adjacent muscles. Especially when close topographical relations exist between synergistic and antagonistic muscles, this phenomenon may significantly bias the quantitative estimation of the contributions of different muscles to the examined function. As a result, conclusions drawn from corresponding EMG data may be limited as well.

The most plausible strategies to suppress cross-talk and increase measurement selectivity respectively are 1) to use bipolar or higher order electrode montages with small inter-electrode distances and 2) to optimize EMG sensor placement. However, these strategies may be insufficient in recordings from challenging areas, as e.g. the lower face, where several muscles interdigitate and overlap in a relatively small area.

In a previous study (Lapatki et al., 2003), cross-talk in the facial area was quantitatively estimated on the basis of selective muscle contractions. The disadvantage of this approach is that even in trained subjects, co-activation can never be completely avoided resulting in a bias of the estimated amount of cross-talk.

To overcome these limitations we developed a new method for quantitative evaluation of cross-talk using high-density sEMG. This technique provides complete topographical information on a whole muscle area and allows for selectively decomposing motor unit (MU) action potentials (MUAPs). The purpose of the present study was to apply this method in the lower facial musculature to evaluate the cross-talk occurring at optimal bipolar electrode positions in this area.

METHODS

Signals were recorded from thirteen specially trained, healthy subjects using flexible high-density sEMG electrode grids. Selective contractions of four different lower facial muscles (m. depressor labii inferioris, m. depressor anguli oris, m. mentalis and m. orbicularis oris inferior) were performed.

Data analysis consisted of 1) extraction of MUAPs separately for each recorded raw data file (decomposition), 2) determination of MU endplate positions and muscle fiber orientations for representative individual MUs by evaluating the corresponding MUAPs as interpolated (monopolar) amplitude map sequences (Lapatki et al., 2006), 3) construction of bipolar electrode montages from these interpolated amplitude maps at different locations along the MU's main muscle fiber direction, 4) localization of the optimal bipolar recording position for each MU at the location where a bipolar signal with maximal RMS value was achieved within the electrode grid, 5) spatial normalization and inter-individual averaging of optimal bipolar electrode positions, and 6) quantitative evaluation of cross-talk by calculating for each MUAP the bipolar potentials for the averaged optimal recording positions on adjacent muscles. As a quantitative measure, cross-talk ratios (CRs) for the different MUAPs (i) and optimal sensor positions on the adjacent three muscles (j) were defined as

$$CR_{MUAP\ (i) \rightarrow Sensor\ position\ (j)} =$$

$$\frac{RMS_{Sensor\ position\ (j)}}{RMS_{Sensor\ position\ (i)}}$$

Finally, CRs were intra- and inter-individually averaged separately for the 12 possible combinations of muscles and adjacent muscles' optimal recording position.

RESULTS AND DISCUSSION

Our data show that measurement selectivity can be significantly improved by optimal bipolar sEMG sensor positioning.

The determined CRs were lower than those determined on the basis of conventional bipolar recordings and attempted "selective" muscle contractions, respectively (Lapatki et al., 2003). This can be explained by the limitations of such muscle-selective contractions, even in specially trained subjects, and by the suboptimal sensor locations in this previous study (due to the lack of objective electrode placement guidelines).

SUMMARY/CONCLUSIONS

Decomposed MUAPs are characterized by "full muscle-selectivity" and therefore allow accurate quantitative estimations of cross-talk. In this study, such quantitative information was established for a challenging area for sEMG recordings, i.e. the lower face.

The determined CRs describe the limits of measurement selectivity. In theory individual CRs allow to separate the individual contributions of single muscle subcomponents to the EMG interference pattern recorded during an experimental task or function, respectively.

Our results contribute substantially to the interpretation of sEMG data from future studies in the lower facial area. The proposed method can also be applied to other recording areas with significant occurrence of cross-talk.

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ELECTRODE ARRAY BASED EMG MEASUREMENTS COMBINED WITH WAVELET ANALYSIS AS A TOOL TO ASSESS SMALLER LOWER LEG MUSCLES.

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INTRODUCTION

Studying smaller muscles is important for detecting the degree of activation of these muscles and their interplay with respect to human stability or with respect to training and rehabilitation. Modern electronics make EMG measurements by electrode arrays possible. Arrays can detect muscle areas that were activated with different priorities. In turn, the wavelet analysis is able to detect muscular events within fast physiologically relevant times. Our purpose was to develop an electrode array that can be used to study the interplay of lower leg muscles and intra muscular domains. The development of such an array also requires the adaptation of appropriate analysis methods. These methods have to consider that muscular events are of short durations (about 40 ms per event). In this work we explored the correlation between the muscular events that were detected by the different electrode pairs.

METHODS

An array consisting of a band with16 parallel bipolar Ag/AgCl electrodes (electrode diameter 2 mm) was built with an inter electrode distance of 11 mm and a distance between electrode pairs of 11 mm. This band was wrapped around the leg at a position which was determined from MRI images showing the cross sectional muscle distribution. The selected area was below the gastrocnemius muscle and contained the muscles flexor digitorum longus (FDL), peroneus brevis (PB), soleus (SO) and tibialis anterior (TA). The EMG signals were submitted to a wavelet analysis and the total intensity was extracted thereafter (von Tscharner, 2000). The intensities for each electrode pair were aligned as rows in a matrix and displayed as contour plot for selected time periods and movements. These contour plots represent activity maps of the lower leg muscles. The elements of the matrix were squared and the cross correlation between the rows was used to indicate how the muscular events interacted.

RESULTS

The arrangement of the electrode positions around the lower leg are shown in figure 1.

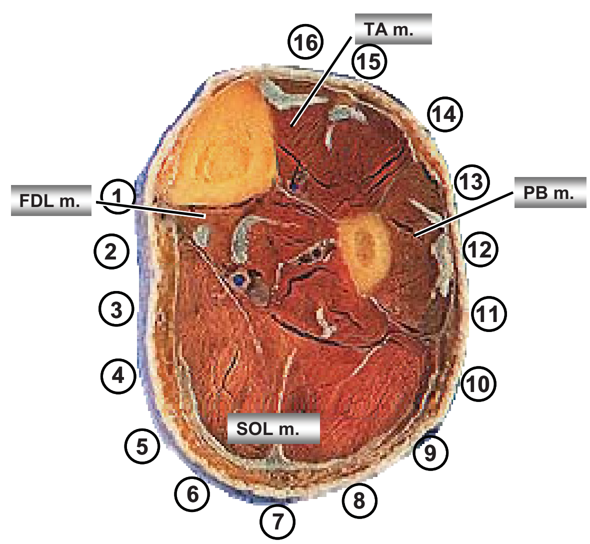

Figure 1. Cross sectional image around the distal third of the right leg and electrodes positioning.

Different movements activated different muscles and using the contour plots and the correlation matrix it was possible to assign which electrode was located on what muscle. The cross correlation obtained when rotating the foot clearly showed very high cross correlations when the electrodes were on the same muscle ($r > 0.8$) and much less ($r < 0.5$) when they were on different muscles. However, the correlation varied significantly among the electrodes located on the soleus muscle and 3 subparts of this muscle can be detected with a rotational movement of the foot. A much higher correlation between muscles was observed while walking.

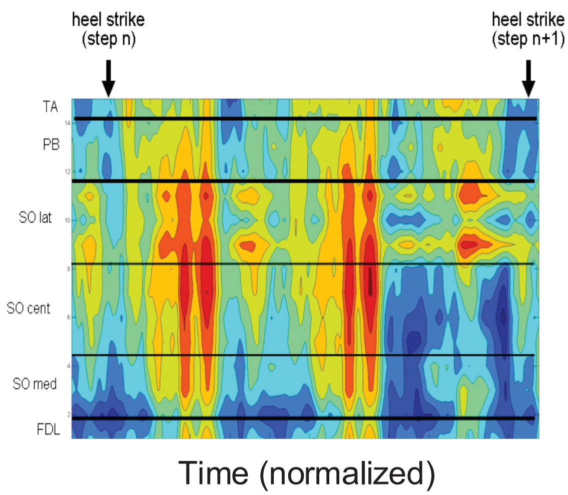

Figure 2. Activation map of one step while walking. The log of the intensity is represented as color code where blue represents low and red high intensity. The contour map interpolates between neighboring intensities. The thick horizontal lines represent the separation of the muscles the thin lines separate areas within the soleus muscle.

Figure 2 shows the muscle activation map reflecting these correlations for certain muscular activities. However, not all muscular events were correlated across all muscles or across all parts of the soleus muscle.

DISCUSSION

The EMG measured with the array of electrodes combined with the wavelet analysis clearly resolved the muscles around the lower leg. The correlations indicate different aspects of these muscles. a) a strong correlation indicates that the electrodes are on the same muscle and muscular events are correlated; b) a small correlation among electrodes that are not on the same muscle is most likely caused by crosstalk between neighboring muscles. c) two electrode pairs that showed only minor correlations during the basic calibration movements and then become highly correlated e.g. while walking may indicate that the muscular events get synchronized by the central nervous system. Thus working with such an array opens the door to studying how the muscular events get coordinated in the lower leg muscles.

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ESTIMATION OF MOTOR UNIT ACTION POTENTIAL CONDUCTION VELOCITYS USING THREE DIMENSIONAL TEMPLATE FROM GRID SURFACE ELECTROMYOGRAM

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INTRODUCTION

A muscle contracts when potential is generated in the muscle fiber that belongs to the motor unit. When the position of a muscle fiber that belongs to an individual Motor Unit (MU) is different, the action potential that appears in the skin surface has spatially different distribution. For example, when two motor units with different depths are compared, deep portions are distributed in the motor unit action potential (MUAP) that appears in the skin surface more widely than superficial portions. In other words, when the motor unit action potential is observed with a multi-channel surface electrode, the Motor Unit Action Potential that appears in each channel will contain information on three dimensions at the position and temporal. However, a method of using information on three dimensions for the technique of identifying the Motor Unit by using template matching from multi-channel surface Electromyography has not been found.

The purpose of this study is development of a Motor Unit identification technique that uses grid multipoint induced surface Electromyography. Therefore, we developed a technique for identifying the Motor Unit that applied three-dimensional (3D) templates to a 7 x 8 channel surface Electromyogram (sEMG) with a grid surface electrode, searched for the temporal and the electrode positions in which the action potential waveform that belonged in each single motor appeared, and classified it.

METHODS

Figure 1 a) shows picture of the grid surface electrode. Figure 1 b) shows location of muscle fiber and the grid surface electrode of 56 channels. The sEMG signals of 56-channels were detected with a grid of 8 x 8 (64) electrodes which seven differentials sEMGs were arranged linearly (column: x direction) as one unit and these eight units were set parallel (row: y direction) with each other. The arrangement of grid surface electrode was as follows: the diameter of electrode was 1mm, the distance of electrode of bipolar sEMG channels was 5mm for along the line of muscle fiber and distance between each column was 2.5mm.

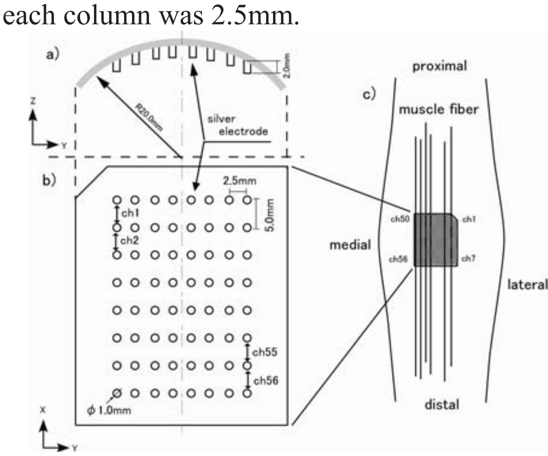

Figure 1: The appearance of the grid surface electrodes manufactured which shown the arrangement and configurations. a) side view of the electrode-box. b) Under view of the electrode-box c) The location of muscle fiber and the grid surface electrode of 56 channels.

The 56-channel sEMG signals were recorded on the biceps brachii muscle for 5 seconds by the grid surface electrode. All the signals were acquired simultaneously and continuously with sampling frequency of 10 kHz. The subject was instructed to perform isometric contraction at 20% Maximal Voluntary Contraction (MVC) in their right biceps brachii muscle.

We provide 3D template matching method that is configured to temporal, x-direction and y-direction. A 3D template is selected 3 x 3 channels from surface potential distribution of figure2 and at 15 sample time points in the vicinity. Figure2 shows surface potential distribution waves by grid sEMG signals. Template area is defined based on the surface potential distribution shown in Figure2.

It searched for the common feature by using the correlation coefficient to specify the temporal-spatial position of MU0. The temporal-spatial poison can shows to conducted trajectory of MUAP. We can understand that MUAP was conducted from which position at which time from spacio-temporal positions.

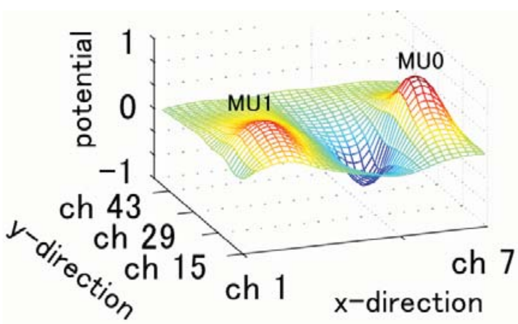

Figure2: Surface potential distribution waves by grid sEMG signals.

RESULTS

Figure3 shows results of spatio-temporal points and conduction velocity times of each firing.

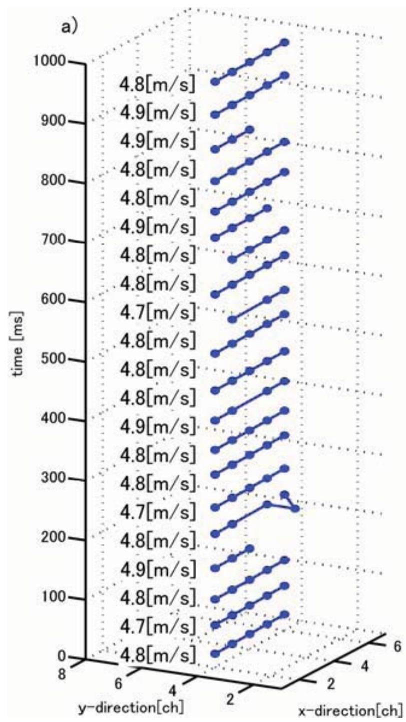

Figure3: Results of spatio-temporal positions from MU0 and conduction velocity of each firing

SUMMARY/CONCLUSIONS

We proposed new identification method of a MU using 3D template. We showed that we can get to MUAP firing rate and each conduction velocity of single MU from grid surface electrodes using 3D template.

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DEVELOPMENT OF BIOELECTRIC SIGNAL SENSOR WITH CHANGEABLE ELECTRODES MOUNTED ON CLOTHES

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INTRODUCTION

Recently many measurement methods in an unconscious and unconstraint state are developed for health care and behavior monitoring in a long term of bioelectricity signal. For example, Paradiso et al. (2005) developed new type of electrodes which are made of electric conductive fiber in one part of clothes. In addition, Fujiwara et al. (1992) demonstrated usefulness of the amplifier near electrodes. When we measure weak bioelectricity signal such as myoelectric signal, we have to put an amplifier near electrodes for improving signal-to-noise ratio. Because it is important to amplify an electronic signal before getting mixed up noise. Hoffmann, K-P. (2007) developed flexible surface-electrodes which provided a very good wearing comfort.

The problems of these developments were the necessity that a subject must wear the specific clothes that electrode and an amplifier were set up. The clothes are not suitable for bioelectricity signal measurements for long terms such as several months. In a long term measurements, clothes suck in perspiration and are stained and there is possibility to change electric characteristics of electric conductive fiber in electrodes by absorption of perspiration. Clothes with electrodes needs to be changed when it is stained or the electrode characteristic changes. However disposable clothes are too expensive for a long term measurement.

On the other hand, particle such as carbon of electric conductive fiber in electrode is washed out when clothes for measurement are washed. As the results electrodes deteriorate. In addition, we have to think about electrodes fixing methods to the human body. At first we fix electrodes to clothes and establish an amplifier near electrodes. It is necessary not to fix and to be able to remove an amplifier easily. By this method, only short duration measurements can be performed with suppressed noise, using electrodes fixed by clothes to human body. For long terms such as several months, an electrical characteristic of electric conductive fiber will change; as a result, bioelectricity signals can not be measured.

The other important point is to alter size of electrodes according to purposes. For example, when we want to know the whole muscle activities, we must use large electrodes. However when we measure motor unit activities by surface electrodes we must use very small electrodes.

From the consideration mentioned above, we concluded that it is necessary for electrodes and amplifiers to be established and removed easily.

The purpose of this study is to develop a new type bioelectricity signal sensor which is possible to change electrodes size and to mount on any clothes.

CONFIGURATION OF A SENSOR FOR BIOELECTRICITY SIGNAL

Figure 1 shows the configuration of a sensor for bioelectricity signal which we developed in this study. In this sensor, the electrodes which are made of the electric conductive fiber mounted on the pins, which can go through clothes, are possible to link to the amplifier via connection units. Figure 2 is photo of a pin type electrode.

The below is the concrete description of each part.

The electrode is made of the thick and elastic sponge with an electric conductive characteristic. As the electric conductive sponge can easily change into form along skin shape, it does not make discomfort to touched skin. Furthermore, it can transform its shape following the skin deformation by exercise of subjects. As a result, we can expect a motion artifactitious reduction effect.

Pin is apical sharp electrically-conductive rigid body and goes through clothes and can be linked to connection units. The connection unit is made of an electric conductive material and is fixed a pin physically and input an electronic signal from electrodes into an amplifier. When a pin is inserted into connection unit, it is fixed with friction. In addition, we can stab a pin again and again. We use a conductive rubber for connection unit.

With the above configuration, subjects can wear any kind of clothes, such as underwear, a T-shirt, socks, gloves, an athletic supporter, a hat and so on.

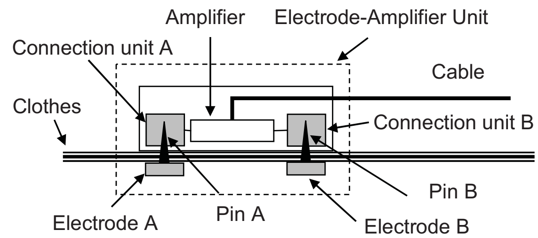

Figure 1: Configuration of bioelectric signal sensor with changeable electrodes on clothes.

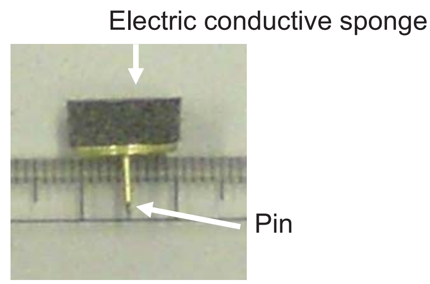

Figure 2: Pin type electrode

CONCLUSIONS

We developed a new type bioelectricity signal sensor which is possible to alter electrode size and is fixed to any kind of clothes. The first characteristic of this sensor is to be able to measure bioelectricity signal for a long term. The second characteristic is to reduce the noise that is caused by the direct contact of the skin and an external cable.

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MULTIFUNCTION PROSTHESIS CONTROL USING IMPLANTED MYOELECTRIC SENSORS (IMES)

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INTRODUCTION

We have developed a novel multichannel/multi-function prosthetic hand/arm control system capable of receiving and processing signals from up to thirty-two implantable bipolar differential electromyographic (EMG) electrodes via transcutaneous inductive and radio-frequency links. An implantable wireless EMG system with the ability to provide much finer resolution for determining specific muscle activity than surface EMG while at the same time addressing the traditional disadvantages of percutaneous fine wire EMG recording, will greatly increase the number of control sites available to amputees for control of their prostheses.
The current generation of EMG controlled hand prostheses are generally single degree-of-freedom (DOF) (opening and closing) devices. Control for these devices is achieved by measuring the surface EMG in the residual limb from a small number of control sites. This produces a simple control set sufficient to reliably determine whether the user wishes to open, close, or keep the hand in a static position, but not suited to simultaneous control of multiple DOFs. Persons with recent hand amputations expect modern hand prostheses to be like hands; when single DOF prostheses fail to meet user expectations they tend to be under-used or rejected (personal communication).

We believe intra-muscular EMG signals from multiple residual muscles can be used to provide simultaneous control of multiple active DOFs. More importantly, we believe that a larger set of independent control sites will allow the user to control the prosthesis in a more meaningful and coordinated fashion. We also feel that the ability to provide an accurate, precise set of controls for a large number of DOF will also encourage the development of more anthropomorphically correct and functional prosthetic devices. It has been presented that by using up to four surface EMGs (the maximum which could be isolated without cross-talk becoming unacceptable) up to four DOF can be controlled **Aijiboye *et al.* (2002)**. It has also been predicted that a sensor with the characteristics of the IMES implant will have an EMG pickup range ranging from 4.8-7.5 mm, **Lowery et al. (2005)** which correlates closely to the size of several of the smaller muscles in the forearm. By virtue of being able to isolate the EMG activity of individual muscles, we propose that an IMES controlled system will be able to increase the number of control sites to all muscles that are at least partially intact, greatly increasing the number of simultaneous controllable DOF.

SYSTEM DESCRIPTION

An IMES system consists of multiple implants (up to 32 devices per system), each of which contains a differential electrode, an amplifier with programmable gain, filter corner frequency, sampling rate, integration or raw EMG transmission selection circuitry, analog to digital converter, and RF transmitter; all of which is completely encased in a RF BION® ceramic capsule *(Alfred Mann, Valencia, CA)*. The IMES implants are inductively coupled to a litz wire coil which will be encased within the laminate of the prosthetic socket. This external coil transmits the power at an efficient drive frequency (121 KHz nominal) and forward telemetry via modulation of the drive frequency to the implant; it is also the receiving antenna for the RF (6.8 MHz nominal) reverse telemetry EMG data. The final component of the design is the Telemetry Controller which encodes command information for the implant onto the drive signal, and decodes the EMG data for passing to a prosthetic controller or external unit.

METHODS

Benchtop testing was conducted to show substantial equivalence to wireless EMG systems currently in clinical use (Telemyo 2400). Acute animal testing in which we examined the crossed extension reflex during plantarflexion (Lateral Gastrocnemius LG, Medial Gastrocnemius MG, and Soleus) as recorded by the IMES and Telemyo systems. We have implanted IMES implants into the LG plantarflexion and tibialis anterior (TA) dorsiflexion muscles of cats for chronic study of IMES performance during normal walking.

RESULTS AND DISCUSSION

We have shown substantial equivalence in frequency response between the IMES system and the Telemyo 2400. We have found a degree of cross-correlation and magnitude squared coherence (MSC) in the acute experiments between the IMES and Telemyo, while at the same time showing a lower degree of MSC and cross-correlation between IMES implants which shows signal independence. The cross-correlation and magnitude squared coherence (MSC) values of <0.1 between raw EMG signals acquired in the chronic animals during walking.

CONCLUSIONS

The IMES system provides a comparable quality of EMG signal to commercially available EMG systems. The IMES pick-up field is suitable to acquire independent signals from the residual muscles of the forearm.

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ELECTROMIOGRAPHIC SIGNAL RELIABILITY ANALYSIS DURING ISOMETRIC AND DYNAMIC ACTIONS PERFORMED IN DIFFERENT ENVIRONMENTS

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INTRODUCTION

In attempt to reproduce the electromyographic signal (EMG) on land, investigations were performed during isometric and dynamic contractions. In aquatic environment, the research of Pöyhönen et al. (1999) was the unique found in literature that had analyzed the EMG signal reliability in maximal and submaximal isometric contractions. Independent of environment, the action type during the exercise may difficult the EMG signal reliability (Sbriccoli et al., 2003).

The objective of the present study was to analyze the EMG signal reliability of the vastus lateralis and biceps femoris muscles during isometric and dynamic actions performed in aquatic environment and on land.

METHODS

The study sample consisted of six young women (22 ± 1 yrs) water aerobics experts, that performed four experimental sessions (2 in aquatic environment and 2 on land).

In each experimental session, a maximal voluntary isometric contraction (MVIC) was performed on land for knee extensors and flexors. Subsequently, the dynamic exercise (Stationary running) was performed in aquatic environment (first week) and on land (second week) in three different cadences (100, 80, and 60 bpm). The interval between each session in both environments was 24 to 48 hours.

Electrodes were placed on the belly of vastus lateralis (VL) and biceps femoris (BF) muscles with a 3-cm center-to-center spacing. Transparent dressing was used to insulate electrodes for the water condition trials.

The EMG signals were registered with a 4-channel EMG system (Miotool400 USB, Brazil), with a common mode rejection ratio >110 dB and a sampling rate of 2000 Hz by channel. The filtering of the raw EMG was performed with a filter Butterworth type, with a bandwidth of 25–500 Hz. The EMG data of each muscle were normalized by MVIC. The root mean square (rms) values were used to analyses.

Intertester reliability measurements were determined using Intraclass Correlation Coefficients (ICC) for rmsEMG values during MVIC and for normalized rmsEMG values during dynamic exercise. The significance level adopted was $p < 0.05$ (SPSS vs 13.0).

RESULTS AND DISCUSSION

For isometric exercise, high ICC values were found for both muscles (VL: ICC=0.86; BF: ICC=0.98). However, for dynamic exercise high ICC values were found only in aquatic environment in cadence of 100 bpm (see Table 1).

The results suggested an excellent reliability for VL and BF muscles activity during MVIC with short intervals of 24 and 48hs. The high ICC found in stationary running, only at 100 bpm in water, might be explained due to the wide use of this cadence in water aerobics routines.

However, some characteristics of dynamic exercise such as electrodes placement, control of the range of motion, and of the angular velocity can difficult their reliability in the other cadences in water and in all cadences on land. These difficults appears to be more related to the characteristic of the dynamic exercise than with the environment itself.

SUMMARY

Thus, in maximal voluntary isometric contraction on land and in stationary running at 100 bpm in water was found a high reliability in contrast to other cadences in water and all cadences on land.

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Table 1 – Intraclass Correlation Coefficients (ICC) for normalized rmsEMG values during dynamic exercise.

Muscle	Cadence	rmsEMG (%) – in water	rmsEMG(%) – on land
		ICC	ICC
VL	60 bpm	0.68	0.13
	80 bpm	0.29	0.04
	100 bpm	0.98*	0.02
BF	60 bpm	0.64	0.44
	80 bpm	0.64	0.34
	100 bpm	0.99*	0.13

Note: * $p < 0.05$.

ELECTROMIOGRAPHIC ACTIVITY ALTERATIONS DURING ISOMETRIC ACTIONS PERFORMED IN WATER AND ON DRY LAND

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INTRODUCTION

Investigations in literature demonstrate lower neuromuscular responses in aquatic environment for different situations and muscle group in dynamic (Masumoto et al., 2004; Miyoshi et al., 2004) or isometric (Fujisawa et al., 1998; Pöyhönen et al., 1999) exercises.

However, recently the approaches of Rainoldi et al. (2004) and Veneziano et al. (2006) have shown that these responses may be similar between the environments if factors were controlled, such as protocol's type, skin temperature and adoption of the waterproof adhesive protection.

The objective of the present study was to verify the recording electromiographic (EMG) signal alterations of the vastus lateralis (VL) muscle during isometric actions performed in water and on land.

METHODS

The study sample consisted of eight young women (23.13 ± 1.13 years), experienced in water exercises.

Electrodes were placed on the belly of VL muscle, with a 3-cm center-to-center spacing. The EMG signals were registered with a 4-channel EMG system (Miotool400 USB, Brazil), with a common mode rejection ratio >110 dB and a sampling rate of 2000 Hz by channel. The filtering of the raw EMG was performed with a filter Butterworth type, with a bandwidth of 25–500 Hz.

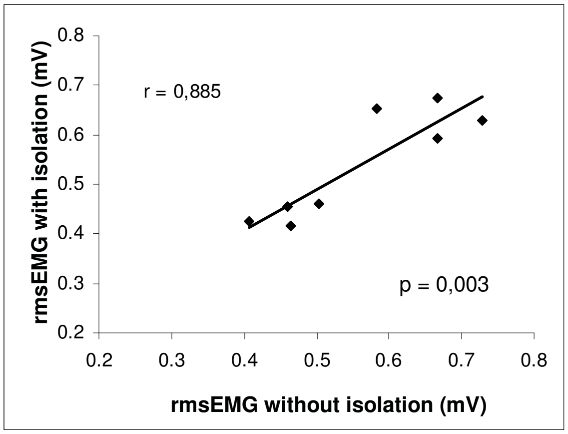
The subjects performed maximal voluntary isometric actions (MVIC) in an isokinetic dynamometer (Cybex Norm, USA) to record the peak torque for knee extensors (90° flexion knee) with (WITH) and without (WITHOUT) superficial electrodes isolations to verify the influence of the water-resistant adhesive taping (TEGADERM, 3M). Subsequently, another MVIC was performed, with the subjects seated, against manual resistance for the same EMG recordings with isolation on land (D) and in water (W) immersion up to navel to verify the influence of the environment. The root mean square (rms) values were used to analyses.

Paired T-Test and Pearson correlation were used to statistical analyses. The significance level adopted was $p < 0.05$ (SPSS vs 13.0).

RESULTS AND DISCUSSION

No statistical difference was found between VL rmsEMG values for WITH (0.560 ± 0.118 mV) and WITHOUT (0.538 ± 0.110 mV) isolation situations ($p = 0.306$), for similar peak torque values ($p = 0.191$). In addition, rmsEMG and peak torque values showed high and significantly correlation between with and without isolation situations (Figure 1).

The same results were found comparing the rmsEMG values for VL muscle between D (0.428 ± 0.054 mV) and W (0.388 ± 0.105 mV) situations (p = 0.446).



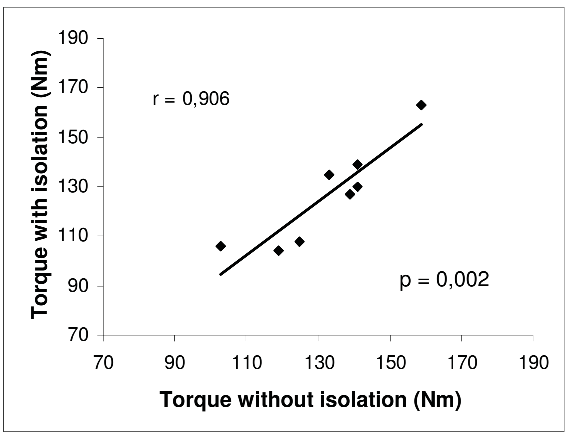


Figure 1 – Correlation for rmsEMG and peak torque values between with and without isolation situations.

The water-resistant adhesive taping did not alterate the signal EMG registered. Based in this finding, we found no differences on rmsEMG values during isometric action performed both in water and on land, suggesting that the environment did not influence the rmsEMG amplitude. Our data corroborate with similar studies (Rainoldi et al., 2004; Veneziano et al., 2006) that stated the use of the isolation to register EMG signal in water.

SUMMARY

Thus, this methodology suggests that future EMG signal comparisons might be employed with the subjects in different environments.

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Track 02

Modeling and Signal Processing (MS)

DEVELOPMENT OF AN SVM CLASSIFIER FOR DETECTING MERGED MOTOR UNIT POTENTIAL TRAINS

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INTRODUCTION

Electromyographic (EMG) decomposition is the process of resolving a composite EMG signal into its constituent motor unit potential trains (MUPT). For motor unit potential (MUP) clustering during EMG signal decomposition the only data available are MUP shapes and their times of occurrence. During EMG signal acquisition, the shape of MUPs can change due to electrode movement, or due to interfering contributions from the MUPs of other active MUs (i.e. superposition); therefore, some times a MUPT is clustered into two trains erroneously. In addition, the shapes of the MUPs created by two or more MUs can be very similar to each other such that the decomposition algorithm erroneously considers these MUPs as one train. Finding these merged trains can improve the robustness of an EMG decomposition algorithm. Here, a Support Vector Machine (SVM) classifier was developed to assess whether a MUPT is representative of the occurrence times of a single MU (i.e. a single MUPT) or if it is a merged train.

METHODOLOGY

MU inter-discharge intervals (IDIs) follow a Gaussian distribution with mean around 100ms and deviation (STD) of 15%-25% of the mean. To develop the classifiers, trains of IDIs were created using a wide variety of parameters. Trains of 75 IDIs were initially, independently generated using a Gaussian distribution with a mean IDI of 80, 90, 100, 110, or 120 ms and coefficient of variation (CoV) ranging from 10% to 30%. In addition, 5% false classification and from 0% to 70% misclassification noise was added to the true trains. In the second step, each possible pair of the generated trains (different mean, STD, false and misclassification rate) was merged. On the whole, 90,000 single trains and 90,000 merged trains were generated. The firing pattern features listed in Table 1 were used to discriminate between merged and single MUPTs. Three classification treatments were examined; Fisher linear discriminate analysis (FDA), SVM and Pattern Discovery (PD). We used a SVM because it minimizes an upper bound on the generalization error, which has been shown to be superior to conventional learning algorithms (e.g. neural

Table 1: Firing Pattern Features

Feature	Description
Skewness	A measure of symmetry of the IDI histogram.
1stSCorr	First coefficient of serial correlation.
CoV_L	Lower coefficient of variation.
CoV_L/CoV_U	The ratio of lower and upper coefficient of variation.
FR_MCD	Firing Rate mean consecutive difference.
IDI_MCD	IDI mean consecutive difference.
IDI_L	Lower IDI ratio.
IDrate	Identification rate.
PI	Percent of inconsistency.

networks) that minimize training data error (Vapnik, 2000). FDA is a simple and powerful linear classifier, it is easy and computationally cheap to implement. PD (Yang and Wang, 2003) is an associative rule-based classifier. Patterns discovered in training data and present in a feature vector to be classified are combined using information theory metrics for classification. To estimate the accuracy of each classifier, 10-fold cross-validation was performed.

RESULTS AND DISCUSSION

Table 2 shows summary results for this experiment. Each row describes the confusion matrix of each classifier. The first two columns show the accuracy of the three classifiers in classifying a MUPT correctly and the last two columns show the percentage of MUPTs that were misclassified by these three classifiers. As this table shows, SVM has a very high accuracy in classifying an IDI train correctly. It has an average accuracy of 99.3%, FDA 98.95% and PD 99%. The accuracy of PD is compatible with SVM, but the probability of error for this classifier in classifying a single train is higher which can cause duplication of MUPTs during EMG decomposition. In comparison to the other two methods, SVM has the lowest misclassification error. On average the misclassification error of SVM, FDA and PD is 0.71%, 1.05% and 1%, respectively.

An interesting aspect of the performance of the SVM is that it works very well even when the misclassification rate of the IDI trains was 70% and only 25 true IDI were available. As shown in Figure 1, in this worse case the accuracy of the SVM in classifying a merged train correctly was 95%. If the misclassification rate of one train decreases to 50%, the accuracy of the SVM increases to 99%, which shows how

Table 2: Confusion Matrix of three Classifiers (SasS: Single as Single, MasM: Merged as Merged)

	SasS	MasM	SasM	MasS
SVM	99.45	99.13	0.55	0.87
PD	99.00	99.00	1.00	1.00
FDA	98.80	99.10	1.20	0.90

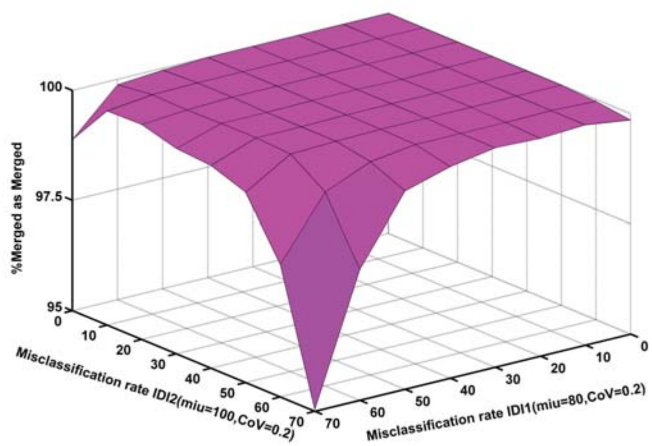

Figure 1: Accuracy of SVM in classifying a merged train as a function of the rate of misclassification noise in each train.

well the classifier is able to recognize merged trains during signal decomposition.

SUMMARY/CONCLUSIONS

We compared the accuracy of three classifiers for recognizing a merged train during EMG decomposition. Performance of the SVM is encouraging and it outperforms the two other classifiers studied.

ACKNOWLEDGEMENTS

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CRITICAL EVALUATION OF MOTOR UNIT ARCHITECTURE MODELS

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INTRODUCTION

Motor unit modeling is central in the understanding of neuromuscular function, as it affects both the mechanical and the electrical activity originated in muscle contraction. In a recent work (Keenan and Valero-Cuevas, 2007), the need for critical evaluation and refinement of force and EMG models was pointed out. In the trend of fragmenting the validation effort and concentrating in specific parts of the model, our work focuses on the evaluation of motor unit architecture models. We present an intensive simulation study of eight motor unit architecture schemes derived from the combination of four different motor unit placement approaches and two innervation pattern models. Our study allows to test the degree of compliance of the different models with well-established physiological assumptions (Farina et al, 2004).

METHODS

For the motor unit territory placement we study four different approaches. The **uniform-restricted** takes a uniform distribution of the territories but with the restriction for the territories to lie completely inside the muscle cross-section; the **uniform-cut off** allows the territories to partially exceed the muscle boundary and cuts off the exceeding part; the **uniform-augmented** augments the territory radius of the exiting territories until the inside region equals the original territory area; the **DSVM-augmented** (density spatial variance minimization-augmented) applies an optimization algorithm to place the territories in such locations so that a constant muscle fiber density within the muscle cross section is satisfied, including radius augmentation. For the innervation process we compare two different approaches. In **random innervation** for each of the muscle fibers, an innervating motor unit is selected from the set of units whose territories cover the fiber position. In **scatter innervation** an exact number of motor unit fibers is assigned to each motor unit uniformly distributed over the motor unit territory.

We studied the behavior of three different physiological parameters: we expect to find a constant muscle fiber density (MFD) within the muscle cross-section; a constant motor unit fiber density (MUFD) for all the motor units within the motor unit territories; and an exponential distribution of the motor unit fiber number (MUFN or innervation ratio) of the motor units of a muscle (Fuglevand et al., 1993). A further reasonable property is that motor unit parameters are supposed to be independent of the location of the motor unit.

RESULTS AND DISCUSSION

A constant MFD within the muscle cross-section is guaranteed by the random innervation models, as they use a predefined muscle fiber grid that faithfully represents the muscle fibers. In contrast, MFD in scattering innervation models is proportional to the overlapping. This makes them suffer from an edge effect (Fig. 1): MFD on the periphery of the muscle cross-section is much lower than in deepest locations, something not observed in real muscles. In this case only the DSVM-augmented model preserves a constant MFD.

For the MUFN, scatter innervation models ensure that each motor unit innervates exactly the intended number of fibers as long as the motor unit territory area (MUTA) also remains as designed. This will discard the uniform-cut off model to be used with scatter innervation as MUTA, and consequently MUFN, is dramatically reduced for the motor units close to the muscle cross-section boundary. For the rest of the motor unit placement models, muscles simulated with scatter innervation will present a distribution of motor unit twitch forces that follow exactly an exponential law. For the random innervation models, when overlapping suffers from the previously mentioned edge effect, a severe dependence of the MUFN on the radial position of the motor unit territory center is observed. In this case, only the DSVM-augmented model presents MUFN highly independent from the position of the motor unit.

The MUFD is always preserved at its defined value in scatter models. On the other hand, MUFD in random innervation models suffer from the overlapping edge effect unless the DSVM-augmented model is used (Fig. 2). When the edge effect is present, motor units placed next to the muscle cross-section boundary tend to have a higher MUFD than those placed in inner parts of the muscle.

SUMMARY

The statistical evaluation of the simulation outcomes of motor unit architecture models provides an objective way to test whether widely accepted physiological properties concerning muscle architecture are well reproduced by these models. After the evaluation, DSVM-augmented model, used both with scatter and random innervation models, provides a constant MFD and allows to obtain the closest results to assumed motor unit architecture data.

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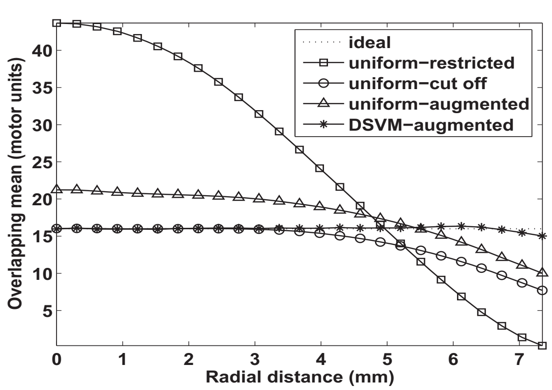

Figure 1: Overlapping motor units as a function of the radial position within the muscle cross-section.

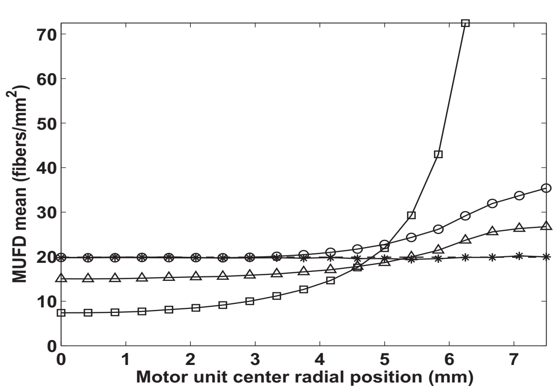

Figure 2: MUFD as a function of the radial position of the motor unit territory center with random innervation (R_{MU}=1mm).

MODEL BASED ESTIMATON OF MOTOR UNIT PROPERTIES

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INTRODUCTION

High-density surface EMG (HD-sEMG) measurements allow dynamical and topographical evaluation of motor units (MUs). The spatial interrelation between the electrode grid and the underlying muscle fibers is of importance for several reasons. First, this interrelation determines how anatomical parameters such as endplate zone location and muscle fiber direction can be estimated from the HD-sEMG signal (Lapatki et al., 2006). In addition, spatial filters and conduction velocity estimation are sensitive to the orientation of the electrode grid relative to the muscle's main fiber direction (Grönlund et al., 2005). Hence, methods are desired that are independent of grid (mis)alignment to improve accuracy and to reduce bias of estimates from HD-sEMG signals.

Here we propose a method to estimate mean fiber orientation, conduction velocity (CV), and center of innervation zone of motor units from sEMG recordings.

METHODS

We used a generative model for the motor unit action potential (MUAP) based on propagating tripoles that comprise mean CV, three-dimensional muscle fiber orientation, and location of MU endplate zone as parameters. These parameters were subsequently estimated by fitting the model MUAPs to measured and decomposed (monopolar) MUAPs, by minimizing the associated weighted least squares objective function. Estimation errors were obtained by a bootstrap procedure, which accounts for spatial correlated noise.

Measured and simulated signals were used to validate the proposed method, i.e. to quantify the deviation between grid orientation and muscle fiber direction. Signals were recorded from an electrode grid of 10x13 electrodes placed above the medial head of the biceps brachii muscle of a healthy subject. Simultaneously, two pairs of wire EMG electrodes were inserted into this muscle. Based on visual estimation we aligned the electrode grid as good as possible parallel to the muscle fibers. Next, the grid was rotated around its center in steps of 7.5 degrees up to 22.5 degrees. At each position, invasive and HD-sEMG signals were recorded simultaneously for 1% and 5% maximum voluntary contraction force estimated from root mean square values of selected signals. Single MUAPs were identified by decomposing selected wire EMG signals using EMGLAB Version 0.9 (Florestal J.R. et al., 2007). Four MUs were decomposed for each recording position. Surface MUAPs were determined by spike triggered averaging using the firing events of decomposed invasive MUAPs.

In addition, simulated MUAPs were obtained using the MUAPs from the non-rotated grid position and calculating "rotated" MUAPs in steps of 7.5 degrees by spline interpolation. MU parameters were estimated from both simulated and measured MUAPs.

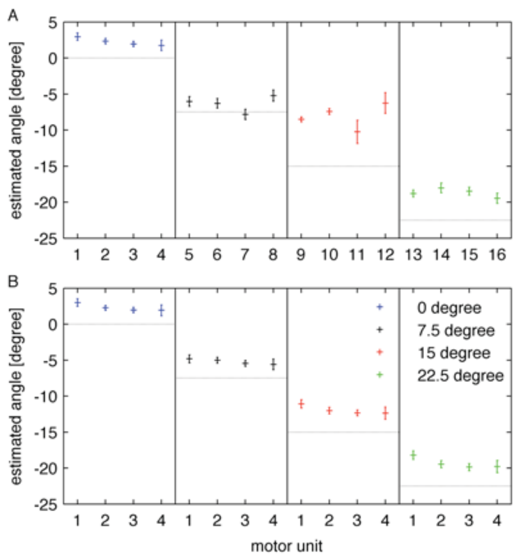

Figure 1 Estimated angles of muscle fiber orientation with one sigma standard deviation for different grid rotations (0, 7.5, 15, 22.5 degree) and motor units from measured (A) and simulated signals (B).

RESULTS AND DISCUSSION

For the given subject and muscle the proposed method performed well. Angular displacements were detected within the experimentally specified range between measurements of different rotational degrees. Evaluation of measured signals (Figure 1, A) resulted in higher variability between different MUs than in the evaluation of simulated signals (Figure 1, B). This enhanced variation may result from alterations in action potential shape of MUs due to 1) slight wire displacement during grid reattachment between the different measurements and 2) muscle tissue movements during the experiments. Such alterations of wire signals hamper unambiguous identification of single MUs making it extremely difficult or sometimes even impossible to trace MUs in subsequent measurements. As a result, the evaluated MUAP samples may vary between different measurements.

In addition, there are experimental inaccuracies when placing the electrode grid at different angles above the biceps. These problems are connected e.g. with the flexibility of tissue and changes in curvature of the skin surface over the examined muscle and may lead to a bias of the results, too.

SUMMARY/CONCLUSIONS

We have proposed a method to estimate mean fiber orientation, conduction velocity, and the center of the innervation zone of motor units from monopolar HD-sEMG signals. This method uses the information of all channels and does not require parallel alignment between the electrode grid and the muscle fibers. The proposed method may facilitate future measurement and may improve the accuracy of estimations of MU parameters. Moreover, anatomical parameters such as muscle fiber direction can be determined more easily for small muscles having distinct fiber orientations (e.g. in the face). Further data from different subjects and muscles are required to confirm the general applicability of this method.

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CONTROLLING CLASSIFIER ENSEMBLE MEMBERS FOR MOTOR UNIT POTENTIAL CLASSIFICATION THROUGH DIVERSITY MEASURE

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INTRODUCTION

In (Rasheed et al., 2007) we presented a hybrid classifier fusion system for motor unit potential (MUP) classification during EMG signal decomposition. It uses a fixed set of classifiers and its both combiners act on the outputs of the same base classifiers. The major problem that needs to be solved in the hybrid classifier fusion system is the choice of an appropriate ensemble of base classifiers to be combined. This paper provides a solution to the base classifiers choice problem in the hybrid classifier fusion system through developing a system that considers the selection of classifiers based on a diversity measure. The system selects the classifier ensemble members by exploiting kappa statistic diversity measure for designing classifier teams through estimating the level of agreement between base classifier outputs.

The developed diversity-based hybrid classifier fusion system has been evaluated and applied for the MUP classification task. Synthetic simulated EMG signals of known properties and real EMG signals have been used for the performance evaluation of the developed system and then compared with the performance of the constituent base classifiers. Across the EMG signal data sets used, the diversity-based hybrid approach had better average classification performance overall, specially in terms of reducing the number of classification errors.

METHODS

The overall accuracy of classifier fusion system depends not only on the way the base classifiers are fused but also on the selection of the classifiers used in the fusion. Choosing base classifiers can be performed directly through exhaustive search with the performance of the fusion being the objective function. As the number of base classifiers increases, this approach becomes computationally too expensive. Instead of the exhaustive search, we exploited for this purpose the kappa statistic as a diversity measure through assessing the base classifiers agreement. Agreement can be assessed on a per MUPT basis for an ensemble of K base classifiers $e_k, k = 1, 2, ..., K$ used to classify a set of N MUP patterns into M MUPTs and the unassigned category $\omega_i \in \{\omega_1, \omega_2, ..., \omega_M, \omega_{M+1}\}$. We want to estimate the strength of the association among them through measuring the degree of agreement among dependent classifiers.

For j = 1, 2, ..., N; i = 1, 2, ..., M+1 denote by d_{ji} the number of classifiers which assign candidate MUP m_j to class ω_i. Based on the per MUP pattern diversity matrix of K classifiers, the degree of agreement among dependent classifiers $e_k(m_j), k = 1, 2, ..., K$ in classifying MUP pattern m_j is measured using the following kappa statistic formula for multiple outcomes and multiple classifiers (Fleiss et al., 2003):

$$\hat{\kappa} = 1 - \frac{NK^2 - \sum_{j=1}^{N} \sum_{i=1}^{M+1} d_{ji}^2}{KN(K-1)\sum_{i=1}^{M+1} \overline{p}_i \overline{q}_i}$$

where $\overline{p}_i = \frac{\sum_{j=1}^{N} d_{ji}^2}{NK}$ represents the overall proportions of outputs of classifiers in MUPT ω_i and $\overline{q}_i = 1 - \overline{p}_i$.

RESULTS AND DISCUSSION

The base classifiers used for experimentation belong to the types described in (Rasheed et al., 2007). We used a base classifier pool containing eight base classifiers e_1, e_2, e_3, e_4, e_5, e_6, e_7, e_8 from which we selected six classifiers to work as a team in the ensemble for every EMG signal in the used data sets described in (Rasheed et al., 2007) and at each stage combiners. The number of classifier ensembles that can be created is 28 ensembles. Classifiers e_1, e_2, e_3, e_4 are adaptive certainty classifiers and e_5, e_6, e_7, e_8 are adaptive fuzzy k-NN classifiers.

SUMMARY/CONCLUSIONS

A diversity-based hybrid classifier fusion approach was proposed to overcome the limitation of selecting the base classifiers comprising an ensemble. It exploits a diversity measure for designing classifier teams. We chose the kappa statistics measure for this purpose to estimate the level of agreement between the base classifier outputs. The developed approach has been evaluated using simulated and real EMG signals and compared with the performance of the base classifiers. Across the used EMG signal data sets, the diversity-based hybrid schemes outperform the best base classifier and as shown in Table 1.

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Table 1: Mean and mean absolute deviation (MAD) of performance indices for the different classification approaches across the three EMG signal data sets.

Classifier	Independent simulated singnals		Independent simulated singnals		Real signals	
	E_r%	CC_r%	E_r%	CC_r%	E_r%	CC_r%
Base Classifier e_1	**1.8 (0.9)**	**83.7 (5.8)**	**4.9 (2.2)**	**79.9 (4.7)**	**2.9 (2.7)**	**81.4 (3.6)**
Base Classifier e_2	**1.6 (0.9)**	**85.5 (5.8)**	**5.1 (2.3)**	**81.6 (4.1)**	**4.7 (5.8)**	**76.6 (6.1)**
Base Classifier e_3	**2.5 (1.3)**	**84.8 (5.4)**	**5.7 (2.6)**	**81.3 (4.2)**	**5.9 (5.2)**	**77.8 (5.8)**
Base Classifier e_4	**2.1 (1.3)**	**86.8 (5.3)**	**6.3 (2.7)**	**82.7 (4.2)**	**7.5 (7.1)**	**74.6 (8.5)**
Base Classifier e_5	**4.1 (2.4)**	**89.3 (4.9)**	**8.4 (2.5)**	**82.1 (3.5)**	**6.4 (2.6)**	**87.3 (3.5)**
Best Base Classifier e_6	**3.2 (1.8)**	**93.6 (3.5)**	**5.6 (1.9)**	**86.3 (4.2)**	**7.1 (5.4)**	**84.6 (8.0)**
Base Classifier e_7	**4.7 (2.4)**	**87.8 (4.8)**	**8.7 (2.3)**	**82.0 (2.8)**	**9.4 (5.3)**	**82.9 (5.7)**
Base Classifier e_8	**3.6 (2.0)**	**92.5 (3.7)**	**6.2 (1.8)**	**85.2 (3.6)**	**8.8 (6.8)**	**82.2 (9.2)**
Average of Base Classifiers	**3.0 (1.6)**	**88.0 (4.9)**	**6.4 (2.3)**	**82.6 (3.6)**	**6.6 (5.1)**	**80.9 (6.3)**
ADMVAFR-6/8	**2.2 (1.2)**	**93.8 (3.0)**	**4.7 (1.9)**	**89.1 (2.6)**	**3.9 (2.2)**	**89.4 (3.1)**
ADMVSFI-6/8	**2.3 (1.2)**	**93.7 (3.1)**	**4.9 (1.8)**	**88.9 (2.6)**	**4.0 (2.0)**	**89.2 (2.9)**

ADMVAFR - stands for Adaptive Diversity-Based Majority Voting with Average Fixed Rule hybrid classifier fusion scheme.
ADMVSFI - stands for Adaptive Diversity-Based Majority Voting with Sugeno Fuzzy Integral hybrid classifier fusion scheme.
6/8 - stands for selecting 6 base classifiers from the classifier pool containing 8 classifiers.

FINITE ELEMENT SIMULATION OF SURFACE POTENTIAL DISTRIBUTIONS AT VARIABLE LIMB JOINT ANGLE

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INTRODUCTION

Modeling the generation of surface electromyographic (sEMG) signals is essential for understanding the type of information that can be extracted from these signals on the underlying physiological mechanisms. In sEMG modeling, the volume conductor is either described analytically (Farina et al. 2004) or numerically. The latter approach allows the representation of more complex geometries than it is possible by analytical derivations (Lowery et al. 2004). However, no studies have yet considered modifications to the volume conductor due to changes in limb joint angle, which is part of all dynamic tasks.
The aim of this study was to investigate the effect of changes in limb joint angle on surface electromyographic potential distributions, using a novel anatomically-based finite element (FE) model.

METHODS

Magnetic Resonance Image (MRI) scans (82 slices, in-plane resolution: 512×512, 0.39 mm^2/pxl, slice thickness: 3 mm) of the left human upper arm at two elbow joint angles (180° and 90°) were acquired from a healthy man, lying on his back (age: 25 yr, height: 1.77 m, weight: 72 kg) using a 1.5 T Siemens scanner.
From the transversal MR slices (Fig. 1A) a 3-dimensional volume conductor (Fig. 1B) was reconstructed for both joint angles. The models consisted of three tissue layers:

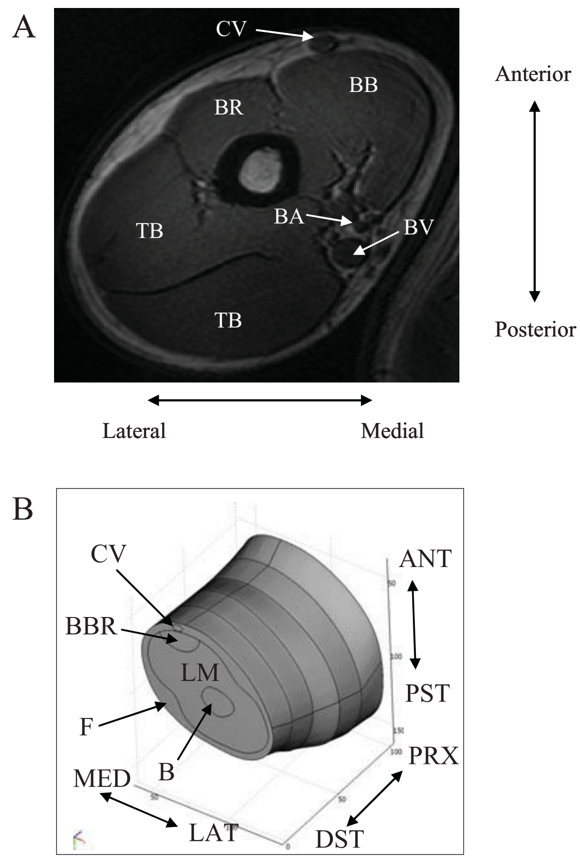

Figure 1: (A) Transversal MR slice of the left upper arm (180°). CV: Cephalic vein, BB: Biceps Brachii, BR: Brachialis, TB: Triceps Brachii, BA: Brachial Artery, BV: Brachial Vein. (B) Volume conductor geometry. F: subcutaneous tissue, LM: Lumped muscle tissue, B: Bone, BBR: Local Biceps Brachii region.

muscle, bone, and subcutaneous tissue. The geometries of the layers were obtained by segmentation of the MR images. A commercial FE software (COMSOL MultiPhysics v3.4, COMSOL A/S, Denmark), was used to generate FE meshes and solve the mathematical model, i.e. Poisson's equation, of the volume conductor. Approximately 400.000 tetrahedral mesh elements were used for the models of both joint angles. The boundary conditions were similar to those discussed in Lowery et al. (2004). The muscle tissue was described as anisotropic, being approximately five times more conductive in the longitudinal (proximal-distal z-direction) than in the radial direction. Conductivity values of the tissues were fixed according to literature data. A single muscle fiber was simulated within the local biceps brachii muscle region (BBR), i.e. a region that could be tracked and segmented across joint angles, of each FE model. In order to reduce computational time, a tripole approximation of the action potential (AP) shape was used. The models were solved for all spatial positions of the tripole traveling from the proximal to the distal part of the model at a velocity of 4 m/s.

RESULTS AND DISCUSSION

Representative results are shown in Fig. 2. The surface potential distributions generated by the tripole source change in a different way for the two models during propagation. The different geometry as a consequence of change in joint angle has thus an effect on the surface recorded signal. Similar results were obtained for other positions of the fibers within the muscle tissue.

SUMMARY/CONCLUSIONS

In this simulation study surface potential distributions were generated by muscle fibers in the same muscle tissue but different volume conductor geometry due to a change in joint angle. This change in volume conductor geometry would occur in dynamic tasks. The results have shown that the change in geometry has an important effect on the surface signal. This effect determines variability in the signal properties exclusively due to geometric changes.

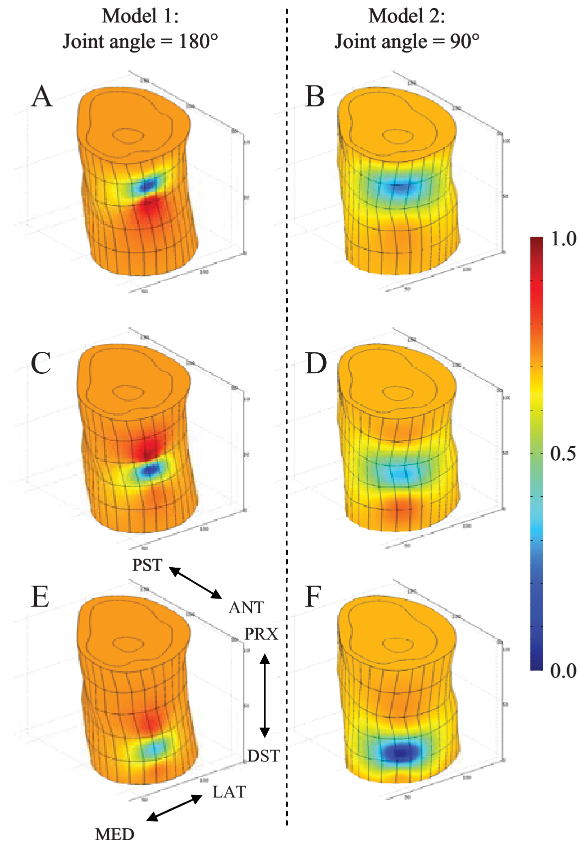

Figure 2: Surface potential distributions as generated by a traveling AP from PRX to DST in the two models (A, C, E and B, D, F). (A,B) AP at t = 6 ms, (C,D) t = 12 ms, and (E,F) t = 18 ms. The data have been normalized for each model, i.e. each model (column) has its own color scale corresponding to amplitudes in the interval [0;1] a.u..

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USE OF HILL-TYPE MUSCLE MODELS IN THE FAST ORTHOGONAL SEARCH ALGORITHM FOR WRIST FORCE ESTIMATION

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INTRODUCTION

The aim of this research was to use Hill-type muscle models in the Fast Orthogonal Search (FOS) algorithm to predict force at the wrist from flexion and extension torque at the elbow. A previous study used the FOS algorithm with surface EMG (sEMG) and joint kinematic data as inputs to predict the force at the wrist (Mobasser et al., 2007). In this study, FOS candidate functions were tailored to reflect neuro-muscular behaviour and thus obtain more accurate wrist force predictions. Results using the new candidate functions are compared to previous results.

METHODS

sEMG data from the biceps brachii, triceps brachii and brachioradialis muscles, and elbow joint angle and angular velocity data were collected with the arm positioned in the horizontal plane. Linear force at the wrist (F_w) was also obtained. The sEMG data were normalized and processed to obtain the linear envelope $e_i(t)$, $i=Bi, Tri, Brd$, as in Mobasser et al., (2007). Muscle activation $u_i(t)$ was calculated using:

$$u(t) = \alpha e(t-d) - \beta_1 u(t-1) - \beta_2 u(t-2) \quad (1)$$

where the electromechanical delay d=40ms, and dynamic parameters $\alpha=1$, $\beta_1=0.25$ and $\beta_2=2.25$ (Buchanan et al., 2004).

The measured F_w was used in a FOS training algorithm (Fig.1), which forms the sum of M non-linear basis functions $p_m(n)$ with coefficients a_m (Mobasser et al., 2007):

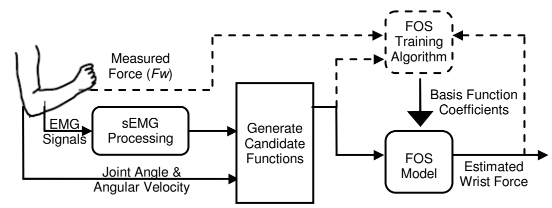

Figure 1: Proposed force observer. Dashed lines represent signals used for training.

$$F_w = \sum_{m=1}^{M} a_m p_m(n) \quad (2)$$

Basis functions were iteratively selected from a pool of candidate functions (>>M) to minimize the mean square error between the system output and F_w. Previously identified candidate functions consisted of groupings of linear, cross and square root terms and sinusoidal functions of the inputs (Mobasser et al., 2007). A new set of functions was created based on the classic Hill model comprised of a contractile element (CE) and a parallel elastic element (PE) arranged in parallel, with the total muscle force equal to the sum of the forces generated by the CE (F^{CE}) and PE (F^{PE}) in order to create a more physiologically relevant estimation. The new functions include equations for wrist force due to contributions of F^{CE} and F^{PE} to elbow moment for the three muscles. A general form of F^{CE} is expressed as the product of peak isometric muscle force F_0^M, force-length (f_l,) and force-velocity (f_v) relationships and muscle activation $u(t)$ (Eq.3) (Buchanan et al., 2004).

$$F^{CE} = \left(F_0^M \cdot f_l \cdot f_v \cdot u_i(t)\right) \quad (3)$$

A modified force-length relationship (Eq.4)

(Cavallaro et al., 2006) was calculated as a function of joint angle θ^M and optimal joint angle, θ_0^M for a range of θ_0^M spanning 40°-100° of flexion. φ_{mT} and φ_{vT} are Gaussian fit functions with values of 0.1-0.3 and 1. The force-velocity relationship (Eq.5) was approximated as a sigmoid function using shortening velocity v_{ce}, σ = 1.33 and a scaling factor γ dependent on v_{ce}.

$$f_l = e^{\left(-0.5\left(\frac{\frac{\theta^M}{\theta_0^M}-\varphi_{mT}}{\varphi_{vT}}\right)^2\right)} \tag{4}$$

$$f_v = \frac{\sigma}{1+(\sigma-1)e^{-v_{ce}/\gamma}} \tag{5}$$

F^{PE} was approximated as a 2nd-order poly-fit of a model (Cavallaro et al., 2006) for a range of θ_0^M spanning 40°-100° of flexion.

RESULTS AND DISCUSSION

The experimental system was implemented using a 1-dof exoskeleton testbed that holds the shoulder and wrist of each subject in a fixed position, and constrains flexion and extension of the arm to the horizontal plane. Up to 19 datasets were collected from each of three subjects during a series of random flexion and extension tasks with and without the application of an external torque at the elbow (no-load and isotonic conditions), and a set of isometric contractions at specific joint angles. Seven datasets were used to train the FOS models with both the muscle-model-based (MMB) candidate functions and those used in Mobasser et al., (2007). Models were formed with M=7 to minimize evaluation %RMSE. Optimal joint angle for each muscle was defined for each subject as the angle producing the lowest evaluation %RMSE. Figure 2 illustrates the measured and estimated force using both models for an isometric phase from subject M1. In each case a model was generated using one dataset and evaluated with those remaining. Table 1 summarizes the mean evaluation results using full dataset and the isometric subset for training. Results show that the MMB models produced lower %RMSE and SD for subjects F1 and M1. More accurate estimation for subject F2 may be obtained using function parameters that more closely mimic the physiology of subject F2.

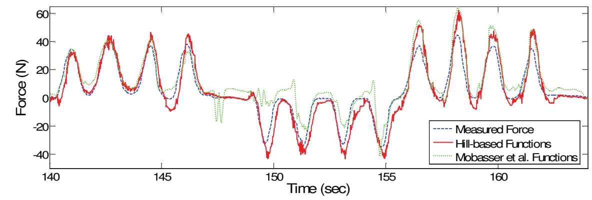

Figure 2: Isometric phase F_w for subject M1 and predicted force using both models.

Table 1: Evaluation %RMSE (±SD) using full dataset and isometric phase for training.

Subject	FOS Candidate F^{ns}	Trained and Evaluated with: All Data	Isometric Phase
F1	Mobasser et al.	28.0 ± 13.5	33.6 ± 26.1
	MMB	25.7 ± 9.2	31.2 ± 19.5
F2	Mobasser et al.	15.6 ± 6.9	16.1 ± 9.8
	MMB	17.9± 6.1	18.3 ± 9.2
M1	Mobasser et al.	12.5 ± 5.6	16.9 ± 6.6
	MMB	12.4 ± 3.8	15.3 ± 8.8

SUMMARY/CONCLUSIONS

This study suggests that Hill-based FOS models can provide better mappings between sEMG and joint kinematics, and measured wrist force. Future work will statistically evaluate the FOS method using MMB functions over a wider subject pool.

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DIFFERENTIATION BETWEEN CROSSTALK AND CO-ACTIVATION: A FUZZY-INFERENCE BASED APPROACH

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INTRODUCTION

Muscular coordination represents the basis of every implemented movement of the human body. Therefore, the evaluation of muscular coordination pattern is important for clinical decision making. Conventional surface electromyography (SEMG) is a frequently used methodology, which allows the noninvasive assessment of the muscular activation pattern. Although standards for electrode localization and signal processing have been developed (SENIAM), the problem of the occurrence of crosstalk in SEMG-signals have not been solved so far. Different approaches to minimize crosstalk or to completely eliminate it have not lead to satisfying results. At the Helmholtz-Institute a methodology has been developed, which allows the determination of a confidence value for the assumption that an EMG signal is not affected by crosstalk.

METHODS

A basic assumption in SEMG is that the recorded electric potentials originate from the muscle directly under the electrodes. However, crosstalk occurs as a result of the volume conduction properties of biological tissue. The contribution of the individual signal sources to the derived signal thereby depends on a multitude of different properties of the volume conductor. These properties are highly individual for each subject and in most cases they are not separately quantifiable. Therefore, it can be assumed that crosstalk also is an individual measure, which varies from one person to another.

Crosstalk Risk Factor-CRF:
In order to quantify the probability for the occurrence of crosstalk individually for each subject, we introduced a "Crosstalk Risk Factor" (CRF). It combines the high number of subject specific properties and integrates them into one objective variable.

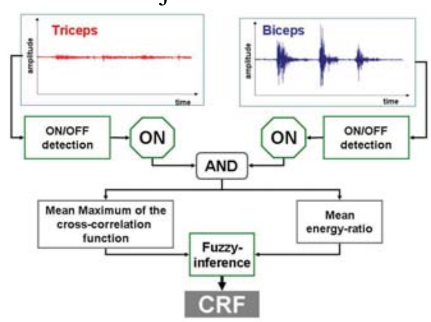

Figure 1: Calculation of the Crosstalk Risk Factor CRF as an individual variable.

For that, fuzzy inference has been applied on measured SEMG data which were recorded on a muscle pair of each specific subject. Movements of such types in which physiologically no muscular co-activation takes place were used (Figure 1). Both SEMG signals have been rectified and smoothed according to the SENIAM standard. If the envelope exceeds a signal-to-noise ratio of 18dB the muscle was considered as active (ON). If both muscles have been considered ON, the ratio of the signal energies derived from both muscles has been calculated. The signal energy ratio has been used as one input parameter for the fuzzy inference (Figure 1). Additionally, for all simultaneous detected ON-phases in both muscles the maximum of the cross-correlation function between both signals has been calculated. The mean value of all maxima has been used as the second in-put parameter for the fuzzy inference (Figure 1). The fuzzy inference links the information about the cross-correlation and the energy ratio on the basis of the fuzzy sets theory, which is tolerant of imprecise data and, therefore, well suited to predict potential crosstalk. The out-put variable of the fuzzy-inference is CRF, which predicts the probability for the occurrence of crosstalk individually for each subject.

Confidence of Co-activation – CCA:
To predict muscular co-activation, which is often essential in clinical applications the "Confidence of Co-activation - CCA" has been introduced which is, in contrast to the CRF, calculated for movements, in which muscular co-activation takes place potentially.

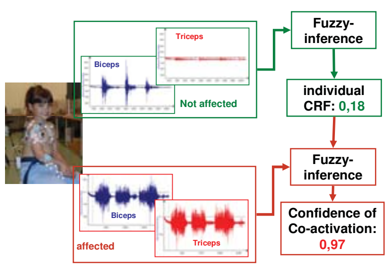

Figure 2: Calculation of the confidence of co-activation on the affected side of a patient with plexus lesion.

Similar to the CRF the CCA is build by fuzzy inference. As input variables the CRF as well as the mean maximal cross-correlation coefficient of the measured SEMG signals is used (Figure 2). In this way the fuzzy inference reflects the knowledge about the individual affinity for crosstalk as well as the fact that SEMG signals originating from co-activated muscles are only little correlated. The CCA reflects likelihood that co-activation of the two muscles and not crosstalk occurs.

RESULTS AND DISCUSSION

The CRF as well as the CCA were calculated in a case study consisting of 20 children, suffering from a plexus brachialis nerve lesion and a resulting co-activation of the biceps and the triceps muscle. The CRF has been calculated for each subject on the non-affected side. The measurements have been repeated two times between three and six month after the initial investigation. Figure 3 shows the calculated CRF of 6 subjects of all repetitions. The results show clearly the individuality of the CRF as well as a high intra-individual reproducibility (except of one) of the measure with time.

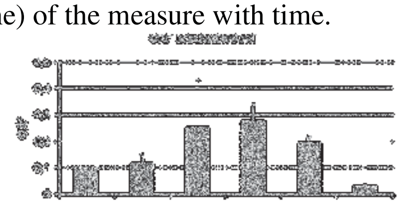

Figure 3: Intra-individual CRF of 6 exemplary subjects.

Within the group of healthy elbow-flexion-movements values lower than 0.4 on a scale from 0 to 1 have been calculated for the CCA, which means that there is only slight likelihood of co-activation between biceps and triceps. In contrast to that, the CCA, was calculated very high (>0.8 on a scale from 0 to 1) on the affected side of the subjects suffering from a plexus brachialis nerve lesion. A double-sided t-test showed with a confidence < 0.01, that there was a significant difference between the healthy and affected subjects with respect to the calculated CCA.

SUMMARY/CONCLUSIONS

The results show that by utilizing the procedure presented, discrimination between crosstalk and co-activation becomes possible and a prediction of muscular co-activation can be done. This is an important step towards clinical decision making based on SEMG.

NOVEL POWER SPECTRUM-BASED FRACTAL INDICATORS FOR MYOELECTRIC PARAMETERS

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INTRODUCTION

This paper presents a novel methodology for fractal analysis of surface myoelectric signals (MESs) named the general power spectrum method (GPSM). This method exploits the persistence among the distribution of MES's physical quantities and provides important information about the underlying single muscle fiber action potentials reflected in the geometry of the interference pattern. This type of analysis is well suited for MESs as they are composed from a strong non-linear summation of a number of self-similar distributions and the functional dependence on independent physiological mechanisms is unknown or highly complex (Merletti, 2004). GPSM provides a multi-scale self-affine power-law that is capable of accurately representing the MES power spectrum. We have previously shown that this method provides fractal indicators (FIs) that are capable of sensing force and joint angle separately (Talebinejad, 2007a). This method also provides a measure for power distribution based on the dominant fractal dimension of MES and asymptotic forms of power spectrum, which is termed the characteristic frequency (CF). We have shown that the CF is independent from the changes of force and joint angle. We have also shown, using simulated MESs, that the CF is sensitive to changes of muscle fiber conduction velocity (CV) (Talebinejad, 2007b). In this paper we evaluate CF when the motor unit (MU) recruitment (i.e. number of active MUs and firing rate), MU depth, and muscle fiber CV are changed, using simulated MESs. Results show that the CF is insensitive to MU recruitment strategy and shape. This indicates good potential of this analysis methodology for fatigue assessment during dynamic contractions.

METHODS

GPSM is based on irrational *Wiener* spectra in this form,

$$\hat{P}(f) = c\left(\frac{f}{f_0}\right)^{2g} \Bigg/ \left(\left(\frac{f}{f_0}\right)^2 + 1\right)^{q+g} \qquad (1)$$

In Eq.(1), c is a scaling factor related to the total power, f_0 is the CF, q is the high frequency indicator, and g corresponds the low frequency power spectrum. Note, the low frequency indicator g is decoupled from the high frequency indicator q; that is, for frequencies higher than f_0, the power spectrum is asymptotically independent from the low frequency indicator g, similar to the *Bode* plot of frequency response for Eq.(1), with a knee frequency equal to the CF. In a similar manner, for frequencies lower than f_0, the power spectrum is asymptotically independent from the high frequency indicator q. This model represents a multi-scale self-affine behavior in this form,

$$\Pr[S(t, f_0)] = \frac{\lambda^g}{\lambda^q} \Pr[S(\lambda t, f_0/\lambda)] \qquad (2)$$

In Eq.(2), S is a temporal multi-scale self-affine signal, λ specifies the constant magnification when zooming into the signal by scaling the time t, which also results into a shifted CF, and Pr[.] is the probability distribution of the samples. This also suggests as we zoom into the signal, the distribution of samples is not changed and is only subjected to a scaling. The four parameters c, f_0, q, and g can be used as indicators related to the self-affine characteristics and the geometrical distribution of signal S. Multiple FIs provide a more comprehensive representation for stochastic signals with long term correlation and fractional characteristics, which are not necessarily strictly self-affine in form of fractional *Brownian* motion and thus, a single-scale fractal dimension is insufficient for characterizing the actual fractal geometry. The CF, f_0 is a critical exponent reflecting a frequency in which the asymptotic form of the power-law is changed. It is expected that this parameter is not changed when the underlying contributions to the MES power spectrum are not changed regardless of recruitment and depth of MUs.

For our experiment, MESs were simulated with characteristics close to a moderate isotonic, isometric *biceps* contraction. The power spectrum estimation, preprocessing and simulation methodology are similar to (Talebinejad, 2007b). The parameter ranges are summarized in Table 1.

Table 1: Simulation parameters

Muscle fiber CV	4-8 m/s ($\sigma = \pm 1$ %)
Number of MUs	20-50
Firing rate	5-20 Hz ($\sigma = \pm 1$ %)
MU Depth	20-21 mm ($\sigma = \pm 1$ %)

The statistics are based on a single repeated measure ANOVA with a null hypothesis for p-values higher than $\alpha_T = 0.05$ which indicates insensitivity. The objective of the experiment is to examine how the CF is affected by the MU recruitment, depth and muscle fiber CV.

RESULTS AND DISCUSSIONS

Results show that the CF is sensitive to changes of muscle fiber CV ($p < 0.001$) but insensitive to number of active MUs ($p = 0.42$), firing rate ($p = 0.53$) and MU depth ($p = 0.14$). This is consistent with our previous studies and expected as the spectral distribution of single muscle fiber action potentials does not vary with the MU recruitment and depth; however, a lower CV results into a shift of single muscle fiber action potentials' spectral content towards lower frequencies (De Luca, 1984) which is reflected in the CF.

CONCLUSIONS

A novel methodology for fractal analysis of MESs was presented. This method shows potential as a powerful tool for fatigue assessment during dynamic contractions when conventional measures such as median and mean frequency are confounded by the effects of muscle shape, force and fatigue (MacIsaac, 2006).

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EFFECT OF RECTIFICATION ON SURFACE EMG COHERENCE

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INTRODUCTION

Coherence analysis is widely used to examine coupling between EMG signals simultaneously recorded from synergistic muscle pairs, or between EMG signals and EEG/MEG signals recorded from the contralateral motor cortex. Rectification of the surface EMG signal prior to coherence analysis is commonly performed in order to enhance the information regarding common inputs to the EMG or EEG/ MEG signals, particularly where data within this frequency range has been discarded as a result of filtering during the data acquisition process. The reasons why rectification should enhance the desired coherence information, however, are not clear. In general, since rectification is a non-linear process, it may be expected to distort the frequency content of the signal, altering the distribution of the power spectrum in an unpredictable manner. For this reason, the validity of rectification prior to coherence analysis has several times been brought into question.

In a previous study it was shown that rectification of simulated surface EMG signals altered the shape of the resulting power spectrum but could enhance peaks in the power spectrum corresponding to mean motor unit firing rates (Myers *et al*., 2003). Analysis of experimental data has also shown that rectification of EMG, despite being a non-linear operation, does not significantly affect the estimated coherence between experimentally recorded surface EMG and EEG or MEG signals (Yao *et al*., 2007). These studies suggest that there may be something inherent within the structure of the EMG signal that can cause rectification to enhance components of the signal arising from periodic inputs.

The aim of this study was to examine the validity of rectifying the surface EMG signal prior to coherence analysis, using an EMG and motoneuron pool model. Coherence between simulated EMG signals with shared neural inputs was examined and the effect of rectification on the resulting coherence spectra was explored.

METHODS

Surface EMG signals from two simultaneously active muscles were simulated as the level of neural input common to both muscles was varied.

The motor unit firing patterns of each muscle in response to descending excitatory inputs were simulated using a previously developed model of the motoneuron pool (Lowery and Erim, 2005). The motor unit firing times were then coupled to a surface EMG model which simulated the motor unit action potentials detected at the skin surface when each motor unit of the muscle was stimulated. The EMG signals were generated using a finite element model which simulated the volume conduction effects of the surrounding muscle, skin and fat tissues and the recording electrode configuration (Lowery et al. 2003).

The basal firing rates of the motoneuron pool were determined by the mean value of the effective synaptic current. A filtered random noise component was also added to the membrane voltage of each motoneuron to represent fluctuations in membrane voltage due to the temporal and spatial summation of randomly occurring post-synaptic potentials. In addition to the random noise component, motoneurons received a second, common component that occurred simultaneously in several neurons. This represented the branched pre-synaptic inputs, believed to be responsible for short-term synchronization of motor units, and was simulated using band-limited Gaussian noise. Coherence between simulated EMG signals was compared for the raw EMG, rectified EMG, EMG high-pass filtered at 50 Hz, and rectified EMG following high-pass filtering. The simulations were repeated for 100 different muscle pairs.

RESULTS AND DISCUSSION

Significant peaks were detected in the coherence spectra between the simulated EMG signals at frequencies corresponding to the frequency of the shared neural inputs. Rectification of the raw EMG altered the distribution of the EMG power spectrum, resulting in a shift in power towards lower frequencies. The presence of significant peaks in the coherence spectra at frequencies corresponding to the shared neural input was, however, maintained following rectification, although the magnitudes of the peaks in the coherence spectra of the rectified data were lower than were altered, Figure 1.

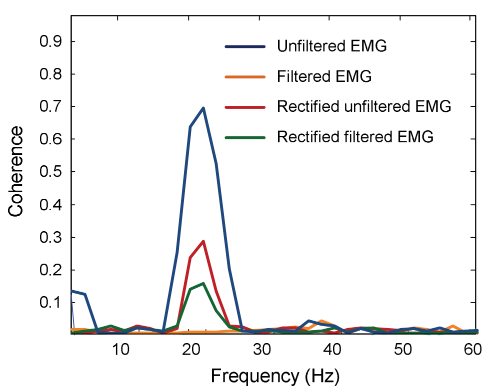

Figure 1: Example of coherence spectra between simulated EMG signals with a 17-23 Hz common motoneuron input.

High-pass filtering above the frequency of the shared neural input removed the presence of coherent activity, as expected. However, rectification of the filtered EMG was found to restore the presence of significant peaks in the coherence spectra at frequencies corresponding to the frequency of the common neural inputs. Rectification of neither the raw nor filtered EMG signals did not result in the presence of spurious peaks in the EMG coherence spectra at frequencies other than those at which shared neural inputs were present.

SUMMARY/CONCLUSIONS

The effect of rectification on coherence between surface EMG signals from muscles receiving shared neural inputs was examined using a model of the motoneuron pool and surface EMG. The presence of significant peaks in the EMG coherence spectra at frequencies corresponding to the frequency of the common input was maintained following rectification. Furthermore, rectification was found to restore peaks in the coherence spectra that had been removed following high-pass filtering of the EMG signal. The results indicate that despite the non-linear nature of the process, rectification of surface EMG is an appropriate means of enhancing the detection of peaks in the EMG coherence spectrum due to common neural inputs. The magnitude of the coherence that is observed is, however, altered by the rectification process.

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FEATURE SPACE TRAJECTORIES FOR USE WITH MYOELECTRIC CONTROL OF A PROSTHETIC LIMB

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INTRODUCTION

The investigation into the use of MES pattern recognition based prosthetic control schemes has been ongoing by several research groups for the past few decades (Parker 2006). Studies have shown the potential for high classification accuracy of various continuous sequential multi degree of freedom control strategies during offline data processing (Englehart 2003). The evaluation of these proposed schemes, however, have only recently begun within a clinical setting (Kuiken 2004).

Preliminary observations have shown that, although highly accurate classification does occur during a steady state elicited contraction, some misclassification does occur during the transition between classes. This results in the undesired joint motion associated with another class from the pattern recognition system. Although generally of short duration, it would be preferable to remove these non-elicited movements when a user transitions between two different motion classes. The work presented in this paper attempts to provide a method by which to observe the real-time behavior of the extracted MES features in an attempt to explain the occurrence of the undesired classification outputs.

METHODS

MES data were measured from 1 mid-length transradial amputee and 1 normally limbed subject while they were prompted to elicit nine different contractions. The experiment was approved by the University of New Brunswick's Research Ethics Board. Six electrodes were placed on the proximal portion of the forearm / residual limb as illustrated in Figure 1. The electrode locations were chosen based on previous research which yielded high classification accuracies.

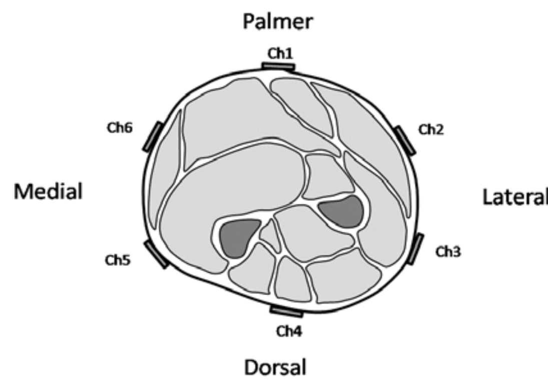

Figure 1: Cross-section of the upper forearm showing the 6 channel electrode placement.

The subjects were instructed to elicit constant force contractions from the rest state for motions corresponding to forearm flexion, forearm extension, forearm pronation, forearm supination, hand open, chuck grip, key grip and fine pinch grip. Data were collected for two cases: 1) data were sampled after the subject was performing the steady state contraction, and 2) data were sampled during the transition from rest to the active motion. All data were collected using a custom built pre-amplification system, a 16-bit DAQ and custom data acquisition software, sampled at 1 kHz per channel. Time domain statistics (Englehart 2003) were used to create a feature vector to represent the MES data. Furthermore, the steady state data were used to compute an uncorrelated linear discriminant analysis (ULDA) feature reduction matrix (Chan 2007). The first 2 components were retained and plotted to graphically represent the signal.

RESULTS AND DISCUSSION

Figures 2 and 3 display the plots of the first 2 ULDA components for the normally limbed and amputee subjects respectively.

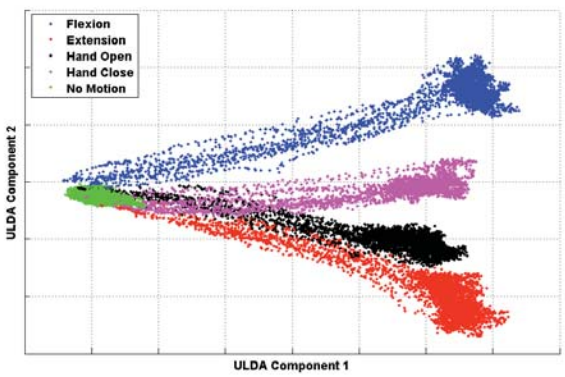

Figure 2: Feature trajectories resulting from the normally limbed subject.

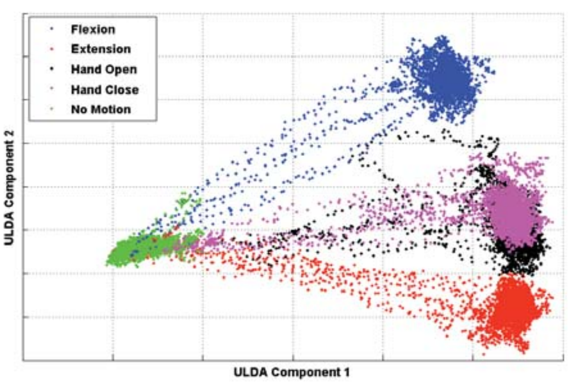

Figure 3: Feature trajectories resulting from the transradial amputee subject.

Both subjects displayed similar feature trajectories; the start in the rest location and move along repeatable trajectories as they enter the motion. Once in the desired class, the subjects were able to maintain the location in feature space. It should be noted that 10 repetitions for each motion were collected for the normally limbed subject while only 4 were collected for the amputee. The trajectories and steady state clusters were less variable and more distinct for the normally limbed subject compared to the amputee. This is expected as they normally limbed subject retained all musculature from which to measure the MES. Furthermore, the normally limbed subject had visual and proprioceptive feedback to ensure they were making repeatable contractions. It should be noted that only the first 2 ULDA dimensions are shown in Figures 1 and 2. There is additional discriminatory information in the additional ULDA dimensions.

SUMMARY

This work highlighted the repeatability of trajectories through feature space as subjects elicited motions. Although only the trajectories from the rest motions are shown, similar trends exist when transitioning between all motion classes. Future work will investigate how to best use trajectory information to improve myoelectric control for prosthetic limbs.

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ACKNOWLEDGEMENTS

This work was supported by NSERC Discovery Grants 171368-03 and 217354-01, and the Atlantic Innovation Fund.

SPECTRAL MULTIDIP CONDUCTION VELOCITY ESTIMATION WITH HIGH-ORDER SPATIAL FILTERS

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INTRODUCTION

Muscle fiber conduction velocity can be estimated from surface electromyographic (SEMG) signals and provides important information on the fiber properties (Farina & Merletti, 2004). Recently, a method based on spectral multidips has been proposed for the estimation of conduction velocity (Farina & Negro 2007). This method is based on the linear relation between the spatial and the temporal frequencies for propagating signals. Imposing zeros (dips) in the spatial power spectrum, it is possible to estimate the corresponding dips in the temporal power spectrum and thus measure conduction velocity. The approach leads to a closed analytical expression for conduction velocity calculation as a function of the auto and cross spectra of the recorded surface EMG signals (Farina & Negro 2007).
The method proposed by Farina & Negro (2007) was limited to the introduction of dips by spatial filters made of four electrodes. In this study we present a generalization of the multidip approach to any number of electrodes.

METHODS

N recorded signals will be considered. Each spatial filter applied to these signals introduces *N*/2 dips by its transfer function. All the spatial filters considered in this study have transfer function with a zero in DC.

In the temporal domain, the linear combination of the *N* detected signals is written as:

$$s_T(t) = \sum_{i=1}^{N} a_i x_i \qquad (1)$$

where $a_1 \ldots a_N$ are the coefficients of the spatial filter and $x_1 \ldots x_N$ the *N* detected signals.
In case of spatial differentiator filters with linear phase, the coefficients are antisymmetric:

$$a_i = -a_{N+1-i} \qquad (2)$$

Moreover, we impose the middle coefficient $a_{\frac{N}{2}} = 1$. The minimum of the power spectrum of the linear combination in Eq. (1) can be computed by nulling the derivatives. Calculations (omitted) lead to the first partial derivatives of the temporal power spectrum of (1) in the following matrix form:

$$\frac{1}{2}\frac{\partial P_S(f_t)}{\partial a_i} = \begin{bmatrix} A_{1,1} & \cdots & A_{ND,1} \\ \vdots & \ddots & \vdots \\ A_{1,ND} & \cdots & A_{ND,ND} \end{bmatrix} \bullet \begin{bmatrix} a_1 \\ \vdots \\ a_{ND} \end{bmatrix} - \begin{bmatrix} C_1 \\ \vdots \\ C_{ND} \end{bmatrix} \qquad (3)$$

where

$$\begin{cases} A_{i,i}(f_t) = S^R_{i,i}(f_t) + S^R_{N+1-i,N+1-i}(f_t) - 2S^R_{i,N+1-i}(f_t) \\ A_{i,j}(f_t) = S^R_{i,j}(f_t) - S^R_{i,N+1-j}(f_t) - S^R_{j,N+1-i}(f_t) + S^R_{N+1-j,N+1-i}(f_t) \\ C_i(f_t) = S^R_{i,\frac{N}{2}}(f_t) - S^R_{i,\frac{N}{2}+1}(f_t) - S^R_{\frac{N}{2},N+1-i}(f_t) + S^R_{\frac{N}{2}+1,N+1-i}(f_t) \end{cases} \qquad (4)$$

with $S^R_{i,i}(f_t)$ and $S^R_{i,j}(f_t)$ the real part respectively of the auto and cross spectra and $ND = \frac{N}{2} - 1$ the number of imposed dips in the spatial transfer function.
A dip in the temporal power spectrum is a local minimum, therefore the vector of the

coefficients $\left[a_1 \quad \dots \quad a_{ND}\right]$ can be calculated imposing the derivative in Eq. (3) equal to zero.
The positive phases of the roots (zeros) of the polynomial with $\left[a_1 \quad \dots \quad a_N\right]$ coefficients provide the vector $\left[f_{Z_1}^D \quad \dots \quad f_{Z_{ND}}^D\right]$ of the spatial frequency dip estimates.
The corresponding conduction velocity estimations can be calculated by the relationship

$$v_1 = \frac{f_{T_i}^D}{f_{Z_i}^D} \qquad (5)$$

where $\left[f_{T_1}^D \quad \dots \quad f_{T_{ND}}^D\right]$ are the temporal frequency dips imposed in Eq.(3).

RESULTS AND DISCUSSION

Figure 1 shows the results of the proposed method applied to a simulated EMG signal (CV=3.24, number of fibers 15 with density 20 fibers/mm^2, interelectrode distance 5 mm, length of fibers 60 mm and mean depth of the motor unit approximately 5 mm) for one single step. The model adopted was described in (Farina 2004). Imposing three dips [29.0, 141.0, 198.0] Hz in the temporal power spectrum, the corresponding dips in the spatial transfer function are [9.2, 42.9, 61.2] 1/m. From Eq. (5) the corresponding conduction velocity estimate is 3.22 ± 0.07 m/s.

SUMMARY/CONCLUSIONS

The generalization of a previous method for the estimation of muscle fiber conduction velocity using a multidip approach has been proposed. The new method allows the use of any number of electrodes and thus reduces the variance of estimation. The estimations are a function of the auto and cross spectra of the recorded signals; therefore the method can be easily extended to time-frequency representation (instantaneous conduction velocity estimation).

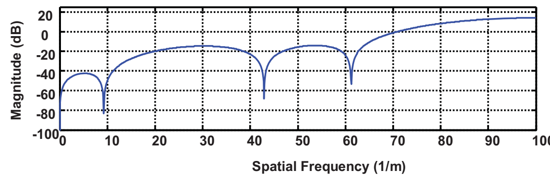

Figure 1:.Spatial transfer function for a simulated signal with three dips imposed in the temporal power spectrum.

Figure 2 shows the mean and the standard deviations of conduction velocity estimates using the proposed algorithm with 4 and 6 electrodes (1 or 2 dips in the spatial transfer function). The signal used is the same as in Figure 1 with SNR = 5 dB. A set of 100 frequencies was used in both cases. The standard deviation of the estimations decreased from 0.22 to 0.10 m/s with 4 and 6 electrodes, respectively.

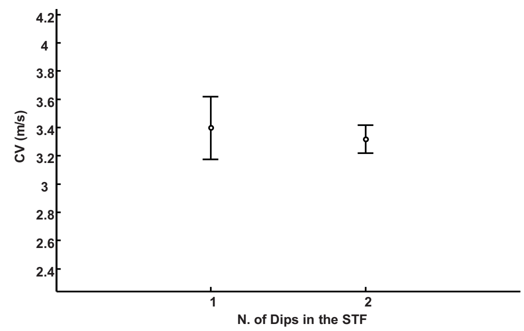

Figure 2: Mean and standard deviation of conduction velocity estimates for 1 and 2 dip cases (4 and 6 electrodes). 100 noise realizations.

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IMPROVING THE REPRESENTATION OF A MUSCLE IN SURFACE EMG USING "EXTREME" HIGH PASS FILTERING

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INTRODUCTION

A muscle would be well represented in the surface EMG signal, if all active motor units (MUs) contribute to that signal, independent of their depth in the muscle. It is well known that in practice such is not the case. Recently, Potvin and Brown (2004) showed a substantial improvement of muscle force prediction from EMG after "extreme" high pass (HP) filtering (HP ≈ 400Hz), removing 99% of the signal power. As an explanation of this finding, based on the insight that non-propagating components of MU potentials (MUAPs) contain the highest frequencies and are less suppressed by the tissue, we hypothesized that HP filtering favors deeper MUs more than the superficial ones (Staudenmann et al., 2007). The current study tests this hypothesis using both a simulation model and an experimental approach.

METHODS

Model study
The ANVOLCON model (Blok et al., 2002) was used to simulate 300 active MUAPs (Figure 1). Bipolarly recorded MUAPS (IED = 2cm) were simulated.

Experimental study
Scanning EMG (Gootzen et al. 1992), was used to obtain intramuscular MU properties, including their depth. A combination with High-Density sEMG also gave properties of the accompanying surface MUAP signals. The effect of additional HP filtering on the signal power of the detected surface MUAPs was determined.

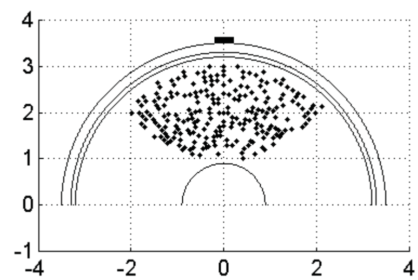

Figure 1: Graphical representation of the transversal cross section of a simulated muscle (muscle, subcutaneous tissue and skin). 300 MU (centered at black dots) where randomly positioned at least 5 mm below the surface and maximally 25 mm from the recording electrode (black bar). Recording electrodes above the muscle.

RESULTS AND DISCUSSION

Figure 2 shows the relative decrease of the contribution of MUs to the EMG signal as a function of their depth below the skin. This is shown for regularly filtered EMG (dark blue) and for different HP settings (4^{th} order Butterworth, 100 – 400Hz). The model prediction is rather complex. For HP settings 100-200Hz, the operation is counterproductive. The deeper MUs are losing instead of gaining relative contribution. Only the "extreme" HP value of 400Hz gives the expected result (purple line, highest of the five lines for depth >1.2 cm).

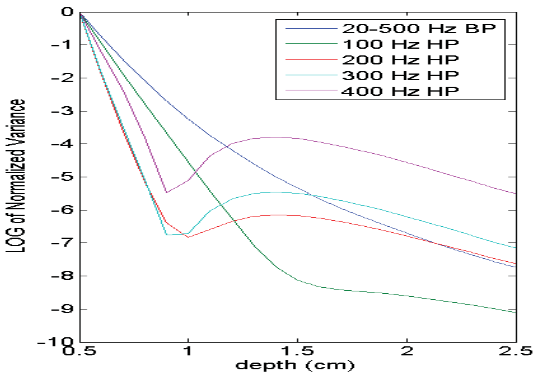

Figure 2: The results of the simulation. Log representation of the relative contribution to EMG power of MUs as a function of their depth below the skin (x-axis) for different HP cutoff frequencies.

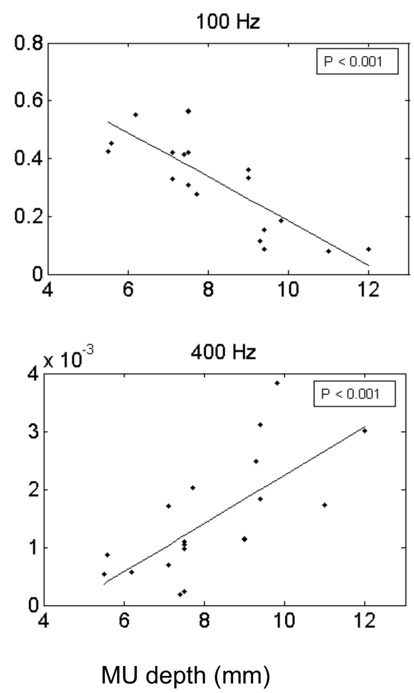

Figure 3: EMG power contribution of the filtered versus the not filtered bipolarly recorded MUAPs as a function of depth for 100 Hz and 400Hz HP filtering. No-influence would mean a horizontal fit line. Note the changes in the vertical axis. It shows that overall the power of the filtered MUAPs strongly decreases with increasing cut-off frequency.

In line with the simulation results, the experimental approach also shows a counterproductive result for the intermediate HP setting (100Hz). The deeper units loose power. Also here, however, the HP setting of 400Hz gave a gain in power for deeper MUs. It should be noted that the range of depths is different between the two approaches.
Varying the parameters in the model (not shown) revealed that high pass filtering strongly depends on the frequency content of the most superficial MUAPs. When the large (propagating) components of these MUAPs are "preserved" by the HP filtering, the operation may become counterproductive. We also studied these phenomena in monopolarly recorded MUAPs. The effects are even more pronounced then, which leads to the conclusion that indeed non-propagating components of MUAPs are largely responsible for the resulting effects.

SUMMARY/CONCLUSIONS

Using both a simulation model and an experimental approach, we could give biophysical evidence for the finding that "extreme" HP filtering favors deeper MUs in comparison to more superficial ones. The effect and its size strongly depend on several intrinsic (e.g thickness subcutaneous fat layer) and extrinsic (e.g. electrode montage) factors.

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VARIANCE-BASED SIGNAL CONDITIONING IMPROVES SPIKE DETECTION IN MULTI-UNIT INTRA-FASCICULAR RECORDINGS

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INTRODUCTION

Multi unit intrafascicular recording techniques allow the detection and tracking of single nerve fiber action potentials (or spikes) , which may be of paramount importance for advanced degree of freedom prosthetic limb control (Dhillon G.S. et al 2005). However, the signal-to-noise ratio (SNR) of intra-fascicular recordings is known to be < 6 dB. Novel signal conditioning techniques are therefore needed in order to increase the SNR to help and improve performance of later spike detection stages and classification algorithms.

Aim: The aim of the study was to develop a variance based signal (VBT) conditioning technique to be compared with a previously described wavelet-based technique (WBT) (Diedrich A et al. 2003).

METHODS

Variance technique: It is assumed that each recorded signal comprises many consecutive APs generated by several, different nerve fibers at different time instants. We further assume that the recorded APs are embedded in different, independent, white, and zero mean additive Gaussian noises (WGN). We propose a single channel based conditioning technique, which uses the simplified mathematical model given in Eq. (1).

$$y(n) = w(n) \times x(n), \qquad n = 1, \ldots N \qquad (1)$$

where *x(n)* is the recorded input signal, *w(n)* a weighting vector and *y(n)* the weighting vector and *y(n)* the conditioned signal, N is the number of samples.

Now let *X* be the space containing sample indexes (1 to N) of the recorded signal *x(n)*, and *P* a subspace of *X*, containing the sample indexes belonging to APs, then there exists a weighting vector that satisfies the following condition*:* *w(i)* = 1 if *i* Є *P*, and *w(i)* = 0 elsewhere. This condition is sufficient in order to attenuate the background noise occurring between APs and thereby enhance the appearance of the APs.

We have chosen to determine the elements of *w* using the second order statistics (variance) of the signal in hand, because it is not possible to know a priori neither the time of occurrence nor the duration of the spikes. We did also determined that the variance of an individual AP is higher than the variance of white Gaussian noise samples of the same power, when computed on the rectified signal (results not shown). Thus, the weighting vector *w* can be estimated by computing the variances of all consecutive portions (or windows overlapped by one sample) of the rectified signal *rx(n)*. This technique utilizes a window (of size *Z*) that moves down the recorded samples one sample at a time, as described in Eq. (2).

$$C(n) = \frac{1}{Z-1} \sum_{i=n-\frac{Z}{2}}^{n+\frac{Z}{2}-1} \left(rx(i) - \mu\right)^2 \qquad (2)$$

where μ is the estimated mean of the window, *C(n)* the set of variance values, *rx(n)* the rectified signal, *Z* the window size and N the number of samples. Note *rx(i)* = 0 for *i* < 0.
The next step is to find a threshold measure as a factor α times the standard deviation (SD) of *C(n)* as given in Eq. (3A).The weighting vector *w* is obtained as given in Eq. (3B). The weighting vector is now multiplied to the raw signal *x(n)* (see Eq. (1)) to obtain the conditioned output *y(n)* with background noise attenuated.

$$T_1 = \alpha * SD(C) \quad (3A)$$

$$w(n) = \begin{cases} 1 & if\ C(n) \geq Thr \\ C(n) & elsewhere \end{cases} \quad (3B)$$

Simulation: 650 isolated action potentials of 4 ms duration were selected from 3 acute rabbit experiments from another study (Djilas M. et al 2007). APs were sorted into 4 classes and a template was obtained for each class, which was used to simulate trains of data including 100 APs randomly separated between 2 and 10 ms. WGN with zero mean and equal variances, of different signal to noise ratios (SNR: 0 to 6 dB) was added and 30 noise realizations were obtained for each SNR.

Detection Performance: The percentage of correct detection (PCD) is defined as the number of correctly detected APs with respect to the total number in the simulated signal. The percentage of error (PE) is defined as the combination of false (no APs existed) and missed (undetected APs) detections. The given PE is the ratio between the PE of the applied technique and the PE of the raw signal (APs detected without conditioning). The detection algorithm is a simple thresholding where the threshold is set to 1 x SD of the input signal. Three mother wavelets were investigated.

RESULTS

Figure 1 summarizes the results where PE of WBT >> 2 x PE of VBT for SNR > 3 dB.

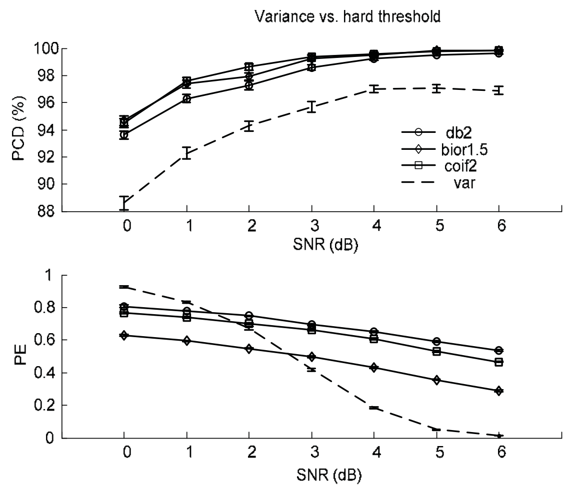

Figure 1: Performance of variance technique with window size fixed to 0.5ms and α = 1; and wavelet technique with decomposition. level = 2 and hard-threshold = 0.8.

CONCLUSION

We have proposed a variance based conditioning technique, where the threshold (α) has to be optimized to meet the SNR of the actual signal. The technique was compared to wavelet-based technique and it was shown that wavelet technique was superior for SNR below 3 dB. However the variance technique is superior in terms of detection error for SNR above 3 dB.

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USE OF SPATIAL INFORMATION IN 2D ARRAY SEMG DECOMPOSITION

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INTRODUCTION

With high density surface EMG (HD-sEMG) it is possible to investigate both single Motor Units (MUs) as well overall sEMG parameters. This can be useful in various fields like rehabilitation, sports and the diagnosis of myogenic and neurogenic diseases.

To extract MU properties from a HD-sEMG signal, the signals have to be decomposed. An often used approach for this is application of the following steps: Segmentation – (Feature extraction) – Classification. Segmentation is the extraction of the MUAPs from the HD-sEMG signal. Features (characteristic properties of a MUAP) are not always extracted; classification (assigning MUAPs to their corresponding MU) can be applied either to features or directly to the MUAPs.

In literature, 2D spatial information is only marginally used for feature extraction and classification. In this study, spatial energy distribution and MUAP shape information is used as base for feature extraction to discriminate between MU classes. The aim of this study was to obtain reliable MUAP templates of HD-sEMG signals obtained during moderate static contractions. The decomposition application was applied to experimental HD-sEMG data recorded with a 2D electrode grid consisting of 32 electrodes (Figure 1).

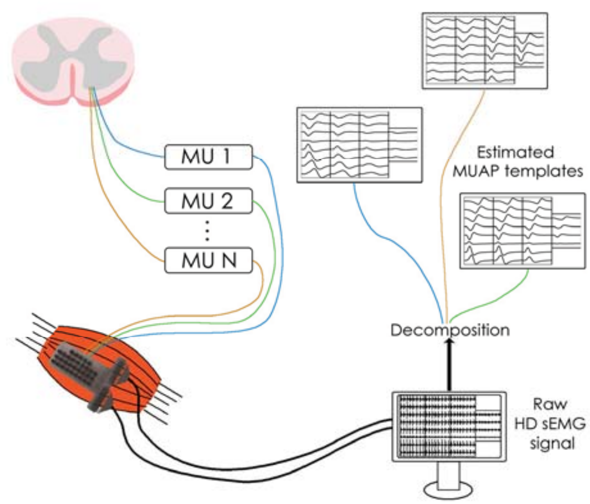

Figure 1: HD-sEMG decomposition: obtaining MUAP templates from the HD-sEMG signal. MUs fire asynchronously, the resulting raw HD-sEMG signals are processed by the decomposition algorithm resulting in MUAP templates of the individual MUs.

METHODS

MUAPs differ from each other by shape, amplitude, distribution over the electrode grid and innervation zone. These properties can be summarized in features.

Continuous Wavelet Template matching is a proven method to locate MUAPs in an EMG signal and describes both shape and amplitude (Gazzoni et al., 2004). Using two different mother wavelets (first and second Gaussian derivative) as feature extractors, different MUAP shapes can be described.

Energy distribution across the electrode grid in both directions can be described by the energy feature of each HD-sEMG channel.

Principal component analysis (PCA) was used to reduce dimensionality by removing redundant information of the estimated features. The resulting combined feature vector was suitable for clustering.

Classification of the MUAPs was performed by clustering the feature space with a classical iterative Expectation – Maximization method (Bishop, 1995). Given a fixed (over estimated) number of classes EM tries to find (Gaussian) clusters of data in the feature space. Using a template matching method clusters are merged afterwards.

RESULTS AND DISCUSSION

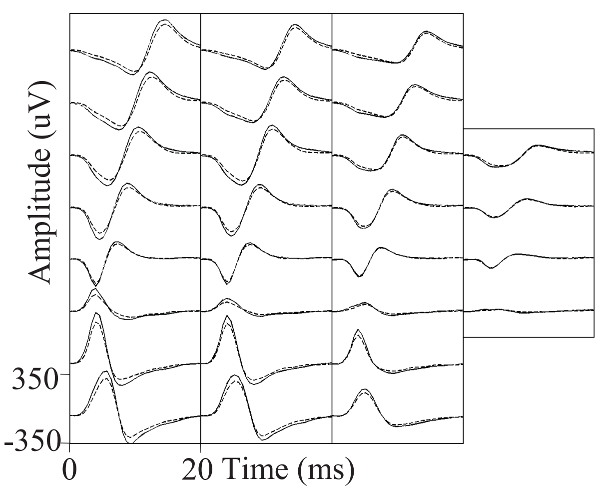

Figure 2: An estimated template (dashed) compared with the original template. Relative error between the two templates is 0.02.

An experimentally obtained sEMG signal from the m. Vastus Lateralis (10s, 30%MVC) was segmented using an adapted version of the method described in (Gazzoni et al., 2004). 349 MUAPs were detected and an EMG expert classified these into 11 clusters (MUs). With the presented method it was possible to estimate 4 MUAP templates correctly (relative error < 0.05) and 4 additional templates reasonably (error < 0.10). The classes that were not detected contained in total only 14 MUAPs. An example of a detected MUAP template matched with the accompanying original template is shown in Figure 2.

This study did not focus on complete decomposition (i.e. find all firings of each MU and resolve superimpositions). Even then, using the presented method it was possible to obtain sufficiently reliable MUAP templates.

SUMMARY/CONCLUSIONS

The goal of this research was to obtain reliable MUAP templates from HD-sEMG signals. To accomplish this, a new feature extraction / classification method for HD-sEMG classification was presented, which highly incorporates spatial information present in the HD-sEMG signals. CWT template features and energy features were used to discriminate between MU classes. A classical EM-method is used to find the resulting clusters. Good results were obtained with a single dataset; currently the method is further validated using larger datasets in different muscles and at different contraction levels.

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EXTRACTION OF REPRESENTATIVE POTENTIALS IN MUAP SETS

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INTRODUCTION

MUAP decomposition techniques can isolate the discharges present in continuous EMG signals and separate them in sets corresponding to different MUAP trains. For quantitative EMG analysis in a neurophysiological context, representative MUAP waveforms must be obtained from MUAP sets. The potentials of a MUAP train are usually corrupted by noise, baseline wander and potentials from other MUAP (MUAP interferences), making the extraction of representative waveforms a problematic task. Traditional statistical approaches such as the mean, median, mode and weighted average often lose significant details of the MUAP waveform and suffer severe distortion with baseline wander and interference potentials. In a previous work we presented a novel technique for potential selection based on an iterative elimination of potentials (IEP) from the MUAP set (Navallas, 2004). Here we present a new algorithm that builds up a MUAP representative potential following a sample-by-sample selection procedure.

METHODS

We present three versions of our MUAP representative extraction (MRE) algorithm:
MRE 1: at each sample time, the median value of all the samples in the MUAP set is obtained. The subset of these samples closer to the median than a fixed value ('delta') is retained. The mean of this subset constitutes the representative amplitude at that time.
MRE 2: Similar to MRE 1, but 'delta' varies according to the standard deviation (std) of the set of samples at each sample time.
MRE 3: The median of all the MUAP set segments viewed through a moving window is obtained at each sample time. The segments closer to this median segment are retained and averaged to compose the representative potential. Segment distances are based on the Mahalanobis distance.

To test these techniques we used both real and simulated data. As real data we used 5 s-long continuous EMG signals recorded from tibialis anterior normal muscles, using concentric needle electrodes and different contraction levels. 175 MUAP trains were obtained from these EMG signals using a decomposition algorithm developed at Montreal University (Florestal, 2006). MUAP potentials were segmented by an EMG expert who manually identified their initial and end points. The expert also selected a reduced subset of the potentials with very close waveshapes. The average of these potentials was considered as the "gold standard" representative potential for the MUAP set.
As simulated signals we used 25 of the previous 'gold standard' potentials that were taken as 'clean' potentials. For each of these, 20 copies were taken to compose the MUAP set. They were independently corrupted by (a) white Gaussian noise, (b) baseline wander obtained from an AR process, (c) interference potentials taken from other 'gold standard' potentials and (d) a combinations of the three corruption sources. We put to work our algorithms and

compared them to the IEP for the four types of corruption. The representative potential obtained by each method was compared to the 'gold standard' for the real data, and to the 'clean' signal for the simulated data. Mean square error (MSE) was measured for all algorithms and cases. Signal distortion ratio (SDR) was also measured (Table 1).

RESULTS AND DISCUSSION

Results from the simulated data show a better behavior of the IEP compared to the MRE method for the only-noise (column 3) and only-baseline (column 4-5) cases, whereas the three MRE versions clearly outperform the IEP in the interference cases (columns 6-7). When the three corrupting sources are present (columns 8-9), results are dependent of their relative weight. IEP outperforms MRE when noise or baseline wander are factors more relevant than interference potentials and vice versa. The three MRE versions have similar performance throughout the set of tests. With respect to real data (column 2), IEP yields a significant lower error than MRE. This result has to be considered cautiously as measurements are obtained with respect to a manually determined 'gold standard' prone to subjective bias. Besides, the selection technique used by the expert was made on a potential-by-potential selection basis, the same as for the IEP method. Therefore, simulated and real data give complementary insight for this comparative study.

SUMMARY/CONCLUSIONS

A new algorithm for obtaining a representative waveform from a MUAP set has been proposed and tested with real and simulated data which included noise, baseline wander and interferences from secondary potentials. A previously presented algorithm (IEP) has also been tested for comparative purposes. Our new algorithm is seen as a valid technique and is particularly useful when secondary potential interferences are present in the MUAP set.

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Table 1: Mean and std (brackets) of the MSE (raws 2-5) and SDR (raw 6). (MSE x 1.0E-5), (σ_{noise} x 1.0E-4), (σ_{BL} x 1.0E-3) (Att= attenuation factor for the interference potentials).

	Real	σ_{noise}**=1.0**	σ_{BL}**=1.0**	σ_{BL}**=2.5**	**Att=0.25**	**Att=0.5**	σ_{noise}**=0.25** σ_{BL}**=2.5** **Att=0.5**	σ_{noise}**=1.0** σ_{BL}**=2.5** **Att=0.25**
IEP	1.21E-1 (2.2E-1)	2.22 (5.9E-3)	2.01 (1.4)	4.86 (4.3)	4.27 (1.8)	8.00 (3.4)	9.57 (3.9)	6.90 (2.4)
MRE 1	2.86E-1 (3.8E-1)	3.01 (9.8E-2)	2.37 (1.0)	6.65 (3.9)	0.0837 (0.14)	0.196 (0.33)	8.57 (3.8)	8.81 (3.0)
MRE 2	2.62E-1 (3.7E-1)	3.30 (9.1E-2)	3.12 (1.0)	7.65 (3.5)	0.0866 (0.14)	0.176 (0.29)	10.0 (4.1)	9.90 (3.2)
MRE 3	2.45E-1 (3.6E-1)	2.77 (2.3E-1)	2.71 (1.0)	6.90 (3.5)	0.101 (0.14)	0.220 (0.22)	8.95 (4.0)	8.42 (3.8)
SDR		20.17 (8.1)	22.18 (8.9)	14.20 (8.6)	19.68 (10.3)	20.83 (9.6)	11.36 (9.1)	11.86 (8.8)

TEAGER-KAISER ENERGY OPERATOR IMPROVES SIGNAL-TO-NOISE RATIO OF EMG SIGNAL IN YOUNG AND OLD ADULTS

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INTRODUCTION

Surface electromyography (sEMG) is widely used to determine muscle activation non-invasively. Temporal analysis of the sEMG signal provides insights into the recruitment patterns of muscle activation, including the onset, offset, and activation duration of muscles during tasks performed in response to an external perturbation (Henry et al. 1998) or executed voluntarily such as locomotion (Hortobágyi and DeVita 2000). Such analyses, however require an accurate and reliable computational method to determine the temporal properties of the sEMG bursts.

Although several algorithms have been proposed to improve automated muscle activity detection (Staude et al. 2001), due to their computational demands, the “standard deviation” method (Hodges and Bui. 1996) is still used most often. This technique however requires a high signal-to-noise ratio (SNR) to accurately distinguish between sEMG and background noise.

The recording of high quality sEMG signal during dynamic tasks can be challenging even in healthy lean individuals and collecting sEMG data from old subjects can be even more complicated. Increased impedance of the skin and subcutaneous fat (Kohrt et al. 1992) and decreased volume of the muscles can significantly reduce SNR and eventually affect the analysis onset detection.

Recently, a new method was proposed to improve SNR in the sEMG, the Teager-Kaiser Energy Operator (TKEO) calculated from the instantaneous amplitude and frequency of the sEMG signal (Kaiser 1990). This method produces remarkable improvements in SNR and increases the accuracy onset and offset detection from sEMG recorded during dynamic movements as demonstrated in a previous study using predominantly EMG simulation (Li et al. 2007).

The purpose of this study was to examine if TKEO analysis improves signal-to-noise ratio of the recorded sEMG signals and whether the amount of improvement is greater in sEMG recorded from old vs young adults during level walking.

METHODS

Eleven young (age range 19 to 25 y) and eleven old (71 to 86 y) adults participated in the study.Bipolar, disposable, pre-gelled Ag/AgCl surface electrodes with 2.0 cm spacing were placed on the belly of the vastus lateralis muscle to record sEMG during gait at 1.5 m/s. Signals were collected with the TeleMyo 900 telemetric hardware system (Noraxon USA, Inc., Scottsdale, AZ), sampled at 1 kHz, amplified 2000x and band-passed at 30-500Hz. TKEO of the EMG signals was defined as:

$$\Psi_d[x(n)] \approx A^2 sin^2[\omega_0 n]$$

where A and ω_0 represent instantaneous amplitude and instantaneous frequency, respectively. SNR calculation was used to quantify difference between rectified EMG and signals after TKEO analysis.

RESULTS AND DISCUSSION

Table 1 shows the results for group (young, old) by method of computing SNR (raw EMG, TKEO) analysis of variance.

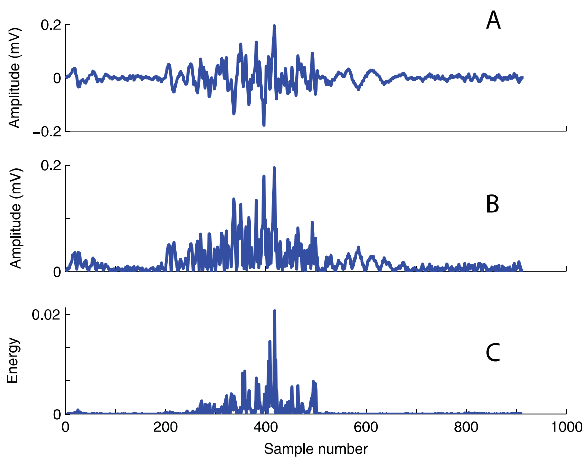

Figure 1. Example of the data analysis. A – raw sEMG, B – rectified sEMG, C – TKEO of the sEMG.

Table1. SNR data in young and old subjects

Group	SNR_{RAW}	SNR_{TKE}
Young	11.6 ± 3.1	219.8 ± 44.8
Old	4.7 ± 0.4	106.3 ± 22.8

Values are mean±SE (unitless). All values are different from one another ($p < 0.05$)

Young vs. old adults had 52% ($F = 5.7$, $p = 0.026$) higher SNR. The SNR was 190% greater using the TKEO method compared with the SNR of the raw signal ($F=37.7$, $p=0.001$). The interaction effect ($F=4.5$, $p=0.047$) indicated that SNR in old adults was significantly worse (lower) in the raw signal and TKEO improved SNR in both groups but the improvement was greater in young vs. old.

SUMMARY/CONCLUSIONS

TKEO is a highly sensitive and computationally less demanding form of data conditioning. TKEO can increase the SNR by an order of magnitude. Such improvement in the quality of data may be especially important and necessary when the data are collected from patients' muscles, who are old or suffer from other clinical conditions. Our experience shows that conditioning raw sEMG signals with TKEO is highly effective and reliable in determining burst onset and offset and computing the magnitude of agonist and antagonist muscle co-activation.

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REAL-TIME REDUCTION OF POWER LINE INTERFERENCE IN MULTI-CHANNEL SURFACE EMG

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INTRODUCTION

Electrophysiological signals are generally noisy because of their low amplitudes and the body's susceptibility to power line interference. Hardware techniques (active electrodes, DRL circuits, virtual ground circuit, electrode-skin impedance equalization, and guarding systems) have been proposed to reduce the acquisition of power line interference. Residual power line interference superimposed to the acquired data can be removed by digital signal processing techniques. Power line interference may vary slowly in both frequency and amplitude. A new digital processing method to remove power line interference is proposed, which is a multi-channel modified version of the method presented in Widrow et al. 1975.

METHODS

A block diagram of the proposed method is shown in Figure 1. Widrow's method requires a reference sinusoid with the frequency of the interference to be removed. The reference signal was obtained using a Phase Locked Loop (PLL) (see Figure 1a). Reference signals can be generated for each harmonic (see Figure 1b), by multiplication of the output of the PLL by the harmonic number and duplicating the Widrow method. In multi-channel systems the common mode can be enhanced by averaging across all channels, which improves the estimation of the interference (see Figure 1c). This estimate can then be subtracted from each channel, with a gain correction (see Figure 1d).

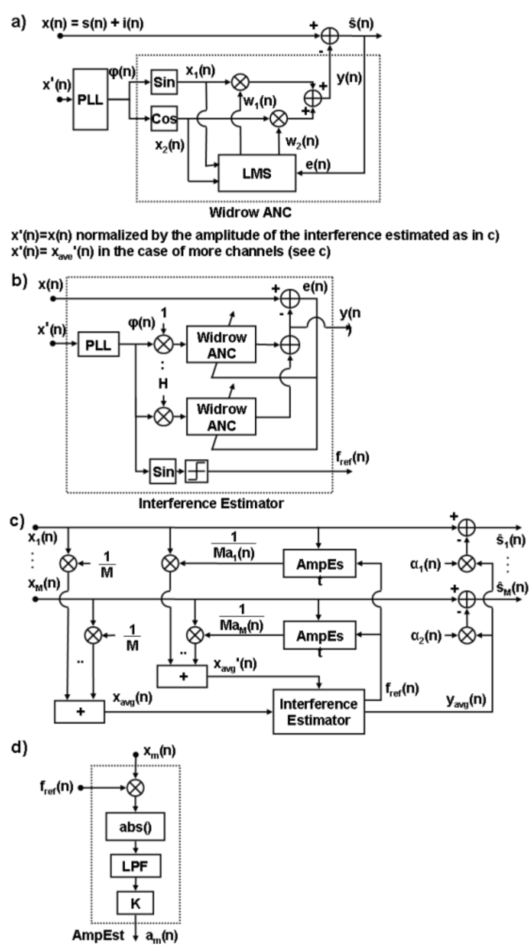

Figure 1: Block diagram of the method

In this way, an arbitrary number of harmonics can be removed from an arbitrary number of channels.
The method is tested by adding simulated EMG signals (using the plane layer model proposed in Farina and Merletti, 2001) on both simulated and experimental interference (acquired from a subject during rest – no EMG activity).

RESULTS AND DISCUSSION

The performance of the method was described in terms of the Signal to Interference Ratio (SIR), defined as the ratio between the power of the signal and that of the interference. Figure 2 shows SIR before and after the application of the method to remove simulated (Figure 2a) or experimental (Figure 2b) interference added to simulated signals.
In the case of simulated interference, performance improves with increasing number of channels. The 8-channel configuration shows an output SIR above 18 dB for input SIR in the range between -20 and 20 dB. In the case of experimental interference, performance resembles the case of simulated interference for high input SIR. The lower performance for low input SIR can be ascribed to the fact that the recorded interference contains also other noise sources which are not removed.

SUMMARY/CONCLUSIONS

A new method to remove power line interference is applied to surface EMG signals. A multi-channel adaptive filter technique is used, with reference signal obtained by a PLL directly applied to the raw data. Results show that the method improves output SIR with respect to input SIR both in the case of simulated and experimental interference.
The performance improves increasing the number of channels.

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ACKNOWLEDGEMENTS

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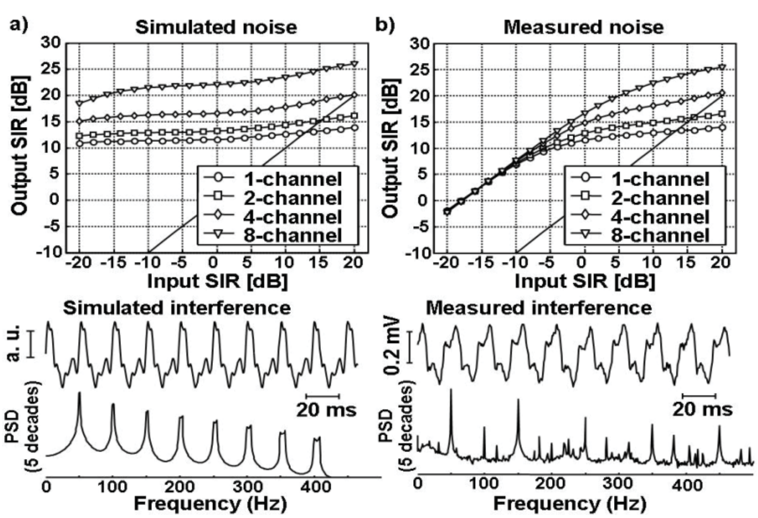

Figure 2: Effect of varying input SIR. a) Effect of varying input SIR when using simulated EMG and interference. b) Same simulated EMG as in a), but with interference signal acquired from a subject. The lower part of the figure shows 200 ms of the time series of the interference signal and the PSD estimate (by the Welch averaging method) in a semi logarithmic plot.

A MAXIMUM LIKELIHOOD APPROACH FOR THE DETECTION OF MUSCULAR ACTIVATION TIMING

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INTRODUCTION

The interest for surface electromyography (sEMG) has widely increased over the last years. In biomechanics sEMG allows for a deeper understanding of muscles' functionality during motor tasks. Among the information (De Luca, 1997) sEMG can provide, the muscular activation timing in dynamic conditions has a real clinical impact (Benedetti, 2001), especially in fields such as orthopedic surgery (Perry, 1998) and rehabilitation (Benedetti et al, 1999).

Therefore, a reliable measure of the muscular activation timing to be provided automatically and in real-time could be of great usefulness in clinical applications. At the moment, in routine work, activation timing is detected by using single threshold methods based on a comparison between a selected threshold and either the raw sEMG signal or its envelope (Bogey et al, 1992; Hodges and Buy, 1996). The technical settings of these methods (low-pass filter band, threshold's value) have not been yet, and cannot be, standardized thus giving rise to a great variability of the results. An algorithmic solution of the problem has been attempted by different and sophisticated approaches ranging from statistical to time-frequency methods (Bonato et al, 1998, Merlo et al, 2003). None of the proposed approaches succeeded in obtaining an user-independent, real-time and consistent detection of muscular timing in dynamic conditions. Aiming at this, in this work we propose the use of a method based on the so-called generalized likelihood ratio test (GLR), which has been designed by improving and extending the proposal by Micera and co-workers (Micera et al, 1997),.

METHODS

The sEMG signal during voluntary dynamic contractions may be considered as a non stationary zero-mean Gaussian process $s_k \in N(0,\sigma_s)$ modulated by the muscle activity and corrupted by an independent zero-mean Gaussian additive noise $n_k \in N(0, \sigma_n)$. In the algorithm, sEMG has to be whitened to obtain statistical independent samples. The GLR works in a Bayesian framework where H_0 and H_1 hypotheses deal respectively with the silent and the activated state of the muscle. The probability that the k-th signal sample corresponds to a transition toward a muscular activation is given by $L(0,1,k,s_0^n) = p_0(s_0^{k-1})p_1(s_k^n)$ where $p_o(s_k)$ and $p_1(s_k)$ are the probability density functions of the hypothesis H_0 and H_1. Since the s_k samples are statistical independent the decision function can be written as

$$DF(k) = \frac{L(0,1,k,s_0^n)}{L(0,s_0^n)} = \log\left(\prod_{i=k}^{n} \frac{p_1(s_i)}{p_0(s_i)}\right),$$

where $L(0,s_0^n) = p_0(s_0^n)$ is the probability that the signal does not contain any muscular activation. The latter is estimated by using the first 50 ms of the signal where only noise is supposed to be, while $p_1(s_k)$ is evaluated over a window increasing its length with the index k. The samples where the DF(k) function assumes minimal and maximal relative values represent the transitions between silent and activity state (muscular onset) and vice versa.

sEMG has been simulated by modulating a s_k time series by gaussian waveform (σ=50, 100, 150) ms, truncated at ±ασ (α=1.5 and α=2.4) and adding the n_k series. SNR values have been designed in the range (10, 15, 20) dB. Ten realizations for every triplet (α, σ, SNR) have been processed.

RESULTS AND DISCUSSION

Performance of the algorithm has been assessed in terms of bias (Table 1) and standard deviation (Table 2) of the estimation of the muscular onset.
The results underlined by the bold format are the ones that can be accepted in real protocols. Bias and standard deviation (respectively less than 10 ms and 15 ms) are satisfactory for α=1.5 and for all the values of SNR. For α=2.4 the detection performance fail.

SUMMARY/CONCLUSIONS

The proposed algorithm allows the detection of the muscular activation timing with bias lower than 10 ms, and standard deviation lower than 15 ms for SNR values typical in most applications. Future studies will deal with the application of this algorithm to real experimental condition.

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Table 1: Bias (ms) for the estimation of the transitions onset as function of (α,σ,SNR)

Bias (ms) of the estimation of the transitions onset as function of (α, σ, SNR)						
SNR	σ=50 ms		σ=100 ms		σ=150 ms	
	α=1.5	α=2.4	α=1.5	α=2.4	α=1.5	α=2.4
20	**-2.3**	11.9	**-3.9**	21.5	**-3.4**	35.6
15	**-0.9**	17.0	**-2.2**	38.3	**-2.0**	51.3
10	**1.5**	20.1	**1.6**	47.2	**0.6**	56.7

Table 2: Standard deviation (ms) for the estimation of the transitions onset as function of (α,σ,SNR)

Standard deviation (ms) of the estimation of the transitions onset as function of (α, σ, SNR)						
SNR	σ=50 ms		σ=100 ms		σ=150 ms	
	α=1.5	α=2.4	α=1.5	α=2.4	α=1.5	α=2.4
20	**2.83**	**10.82**	**2.51**	17.12	**3.5**	17.19
15	**3.7**	**8.35**	**3.82**	15.92	**5.0**	19.29
10	**4.79**	**7.69**	**5.42**	18.37	**8.7**	22.15

THE EFFECT OF BACKGROUND MUSCLE ACTIVITY ON COMPUTERIZED DETECTION OF EMG ONSET AND OFFSET

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INTRODUCTION

Muscle reflex response is often used to study pathology such as low back pain (LBP). A number of studies have reported that reflex responses are delayed in people with LBP (e.g. Radebold et al., 2000).

There is evidence that higher level of background activity causes detection artifact in terms of longer latencies for muscle onsets (Staude et al., 2001). Given that the differences between LBP patients and healthy controls were primarily with muscle offset, the effects of background activity on latencies and number of muscles responding requires investigation to ensure that findings reported in the literature did not arise solely from detection artifact. We hypothesized that the muscle onset and offset latencies will be longer and the number of muscles responding will be smaller as background activity increases, but different computerized detection methods would be affected differently.

METHODS

The performance of two computerized algorithms for the detection of muscle onset and offset was compared. Standard deviation (SD) method (Hodges and Bui, 1996), a commonly used algorithm, and approximated generalized likelihood ratio (AGLR) method, a more recently developed algorithm (Staude et al., 2001), were evaluated at different levels of background surface EMG (sEMG) activity. For this purpose, the amplitude ratio between the period of muscle inactivity vs. activity (onset) and activity vs. inactivity (offset) was varied from 0.125 to 1 in 100 artificially assembled sEMG traces. In addition, 1230 real sEMG signals, obtained from various trunk muscles, were raised to a power of 3 to change the relative amplitude ratio. Forty-one quick release trials from each of 3 directions (flexion, extension, and left lateral bending) were randomly selected from data collected previously (Cholewicki et al., 2005).

RESULTS AND DISCUSSION

As the relative level of background activity increased, both the SD and AGLR methods produced longer latencies and detected fewer muscle responses, suggesting that a detection artifact can be introduced if the subject populations being compared have different levels of background muscle activity (Figure 1). Of the two methods, AGLR appears to be the least affected by background activity. However, above the ratio 0.8, results from AGLR are also unreliable particularly in detecting offsets. Average latency artifacts near this ratio were 8 ms for AGLR and 46 ms for SD.

For the real signals, there was a general trend for SD to be affected more by level of background activity than AGLR. Average muscle onset and offset latencies, detected with SD were significantly longer for the original sEMG signal as compared to the cubed signal (Table 1). However, with AGLR, only onset latencies detected for original signals were significantly longer

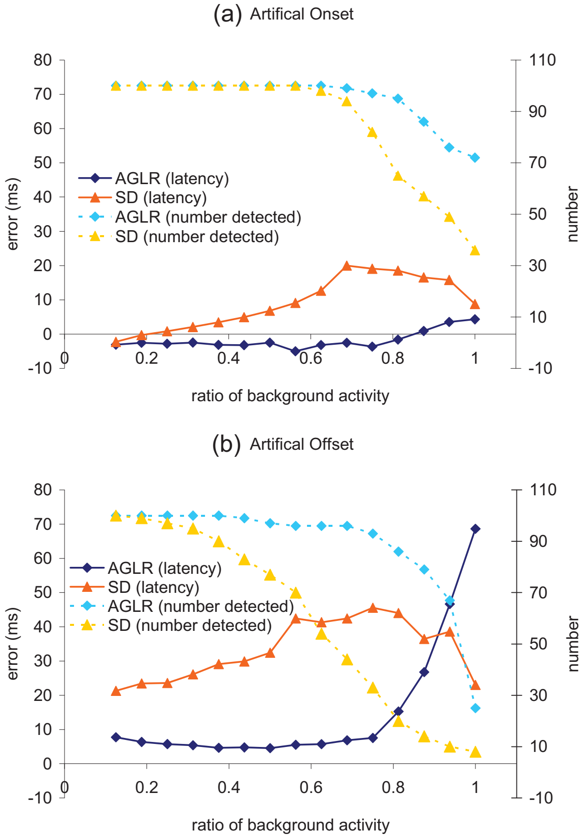

Figure 1: Detection error for artificial onset (a), and offset (b). Positive error indicates longer latency.

than for cubed sEMG signals. The numbers of detected offset responses were significantly increased regardless of the direction of pull with SD, whereas with AGLR, only the extension direction was affected (Table 1).

SUMMARY/CONCLUSIONS

With more background EMG activity, the detected muscle onset and offset latencies became longer. The AGLR method appears to be more robust than the SD method.

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Table 1: Mean (std) number of detected trunk muscle responses and their latencies separated by exertion direction, and onset and offset of activity. Maximum possible number of onset muscles for extension, flexion and lateral bending were 6, 4, and 5 respectively and for offset muscles, they were 4, 6, and 5. Bold typeface indicates significant differences between original and cubed EMG signals ($p<0.05$).

		AGLR			SD		
		Extension	Flexion	Lateral B.	Extension	Flexion	Lateral B.
# of onsets	Original	6.0(0)	4.0(0.2)	**4.3(1.0)**	5.7(0.7)	3.9(0.5)	4.7(0.8)
	Cubed	6.0(0)	4.0(0)	**5.0(0.2)**	6.0(0.2)	4.0(0.2)	4.9(0.4)
# of offsets	Original	**2.2(1.2)**	2.2(1.7)	1.6(1.3)	**1.4(1.2)**	**1.7(1.5)**	**0.9(1.0)**
	Cubed	**1.5(1.1)**	2.3(1.4)	1.1(1.0)	**3.5(0.7)**	**4.3(1.4)**	**2.8(1.4)**
Onset (ms)	Original	**54(9)**	**58(12)**	**51(18)**	**54(14)**	**51(24)**	**56(19)**
	Cubed	**51(8)**	**52(10)**	**35(11)**	**40(12)**	**35(17)**	**45(16)**
Offset (ms)	Original	43(28)	82(37)	93(36)	**60(23)**	**67(22)**	**51(18)**
	Cubed	35(16)	74(39)	105(38)	**33(13)**	**41(16)**	**41(18)**

APPLICATION OF THE HILBERT SPECTRUM FOR FEATURE EXTRACTION FROM ELECTROMYOGRAPHIC SIGNALS

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INTRODUCTION

Most of biological signals are inherently nonlinear and nonstationary. Therefore, when analyzing these signals it is desirable to employ digital signal processing tools that take into account their characteristics. In this context joint time-frequency distributions, which allow for the visualization of the energy of signals as a function of time and frequency, are important tools that have been successfully applied.

This paper shows how a recent developed time-frequency distribution, the Hilbert Spectrum (HS) (Huang et al., 1998), may be applied for feature extraction from electromyographic (EMG) signals. In particular, we show that the median of the Instantaneous Mean Frequency (IMNF), as defined in (Bonato et al., 2002) and estimated from the HS, is a relevant feature in clinical practice.

METHODS

Surface EMG signals were collected from the First Dorsal Interosseous (FDI), which is a very superficial muscle of the hand located between the thumb and index finger, of 11 subjects. A pair of standard passive surface electrodes (MedTech Systems, Ag/AgCl, diameter of 1 mm, inter-electrode distance of 1 mm, type pellet) was used for signal detection.

All subjects were asked to keep isometric contraction during abduction of the index finger for 30 s while applying a force equivalent to a load of a 100 g on a customized force platform. The same procedure was also repeated with subjects applying a force equivalent to a load of 500 g. Tailored-made biofeedback software provided visual information about the exact level of force to be exerted.

The IMNF, estimated from the HS, was obtained for the EMG signals of all subjects in both experimental conditions, i.e., with subjects applying forces equivalent to 100 g and 500 g on the force platform.

RESULTS AND DISCUSSION

Each EMG signal was a 30-second time-series that was divided into 30 windows of one second each. For each window the median of the IMNF was estimated, and the mean and standard deviation were obtained. The results considering subjects applying two distinct force levels on the force platform are given in Table 1. From them it is possible to conclude that the median of the IMNF reduces when the force level of the muscle contraction increases.

The study of changes in spectral EMG parameters as a function of force level of the muscle contraction is a problem that has been addressed in many studies.
For instance, in (Bartuzi et al., 2007) the authors illustrate applications with this aim. In addition, Farina et al. (2004) report in a review article that there is a diversity of experimental findings on changes of spectral parameters with the changes of the level of muscle contraction. In (Rainold et al., 1999)

the authors found that there is a decrease of spectral parameters with increasing force. In (Farina et al., 2004), it is reported conclusions from studies that show both no increase and increase of spectral parameters when increasing the force level of muscle contraction. The authors also mention that this relation should be identified on a subject-by-subject and muscle-by-muscle basis because of the number of variables that may be considered in this relation.

Our results are in accordance to those reported in (Westbury et al., 1987; Rainold et al., 1999) that observed a decrease in spectral parameters with increasing force. However, as pointed out above they should not be generalized, especially because the volume conductor theory indicates that any relationship between recruitment and spectral features can be masked by anatomical factors (Merletti and Parker, 2004). Furthermore, more studies considering other experimental protocols, e.g. subjects executing ramp contractions, have to be carried out in the future to fully investigate the behavior of the IMNF obtained from the HS.

Table 1: Mean and standard deviation (std) of the median of the IMNF (in Hz) estimated for each subject supporting loads of 100 g and 500 g during a 30 s isometric contraction of the First Dorsal Interosseous.

Subject	Load = 100 g	Load = 500 g
S1	181.09 ± 5.22	152.58 ± 5.66
S2	184.00 ± 4.24	167.17 ± 5.61
S3	151.43 ± 7.22	147.81 ± 6.85
S4	178.40 ± 6.06	173.48 ± 7.33
S5	147.51 ± 11.35	134.30 ± 10.72
S6	195.15 ± 4.06	182.51 ± 5.58
S7	167.11 ± 6.19	144.87 ± 8.11
S8	161.19 ± 5.99	153.74 ± 12.06
S9	178.15 ± 4.67	175.16 ± 6.12
S10	170.11 ± 6.34	167.10 ± 8.96
S11	161.37 ± 4.20	153.55 ± 7.34

SUMMARY/CONCLUSIONS

This work showed the application of the HS as a tool for feature extraction from EMG signals. Our results showed that there is a reduction in the median of the IMNF, estimated from the HS, when the force applied during the muscle contraction increases.

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A NOTE ON THE SIGNAL-TO-NOISE RATIO OF MYO-ELECTRIC CHANNEL

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INTRODUCTION

The Signal-to-noise Ratio (SNR) of a myoelectric channel has been studied in the literature (Parker & Scott, 1986, Zhang *et al*, 1990). These studies have shown that the SNR of a myoelectric channel is bonded by the product of the bandwidth of the signal and the time constant of the low-pass filtering. One of the key assumptions of these studies is that a myoelectric channel contains a considerable number of active motor units. In the present study, a more general form of the SNR is derived, without the above mentioned assumption.

THEORETICAL DERIVATION

Consider a myoelectric signal (MES), $X(t)$, contains *N* active motor units. Without loss of generation, assuming all units have identical MUAP, $p(t)$, and $X(t)$ can be expressed as:

$$X(t) = \sum_{i=0}^{N} x_i(t) = \sum_{i=0}^{N} \sum_{j=-\infty}^{-\infty} p(t)\delta(t - t_{ij}), \quad (1)$$

where $x_i(t)$ is the MUAP train (MUAPt) of the *ith* unit. The only difference between the trains are their firing times, t_{ij}. Usually, the square of $X(t)$ is low-pass filtered, through, for example, a *T*-seconds long square window:

$$Z(t) = \int_{t}^{t+T} X^2(t)dt \quad (2)$$

The SNR of a myoelectric channel is defined as:

$$SNR = \left(\frac{E[Z(t)]^2}{\sigma_z^2}\right). \quad (3)$$

Since $p(t)$ is zeros mean, and any two MUAP trains, $x_i(t)$ and $x_j(t)$, are uncorrelated, it is straight forward to show that:

$$E[Z(t)] = N\lambda_0 T\alpha, \quad (4)$$

where λ_0 is the mean firing rate of all units, and α is the area of $p^2(t)$:

$$\alpha = \int p^2(t)dt. \quad (5)$$

When the duration of $p(t)$ is much shorter than T, such that the end effects of the low-pass filtering can be ignored, there is always an integral number of pulses that fall into the integration interval $[t, t+T]$. Denoting this number by m, and it can be shown that σ_z^2 is:

$$\sigma_z^2 \simeq N\alpha^2 Var[m] + 2E\left[\sum_{i=1}^{N}\sum_{\substack{k=1 \\ k\neq j}}^{N} \left(\int_T x_i(t)x_j(t)dt\right)^2\right]. \quad (6)$$

It was shown in (Cox, 1962) that:

$$Var[m] = \lambda_0 T\rho^2 + \frac{1}{6} + \frac{\rho^1}{2}, \quad (7)$$

where ρ is the coefficient of variation of the inter-pulse intervals of the MUAP train. The term inside of the double sum in the right hand side of (6) is approximately:

$$\int_T x_i(t)x_j(t)dt \simeq \beta\lambda^2 T, \quad (8)$$

where β is:

$$\beta = \int \left(\int p(t)p(t+\tau)d\tau\right)^2 dt. \quad (9)$$

From (7), (8) and (9), (6) becomes:

$$\begin{aligned}\overline{\sigma_z^2} &= N\alpha^2(\lambda_0 T\rho^2 + \frac{1}{6} + \frac{\rho^1}{2}) + 2\beta N(N-1)\lambda^2 T \\ &= N\alpha^2 A + 2\beta N(N-1)\lambda^2 T\end{aligned} \quad (10)$$

The SNR defined in (3) becomes:

$$SNR = \frac{N\lambda_0 T}{\frac{A}{\lambda T} + \frac{(N-1)\lambda_0}{B}}, \quad (11)$$

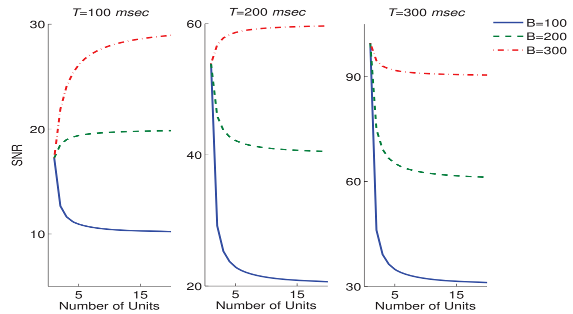

Figure 1. The Theoretical SNR from (11). The firing statistics are $\bar{\lambda} = 20\ Hz$ and $\rho = 18\%$. From left to right, $T = 100\ ms$, $200\ ms$ and $300\ ms$, respectively. In each plot, the three lines are for $B = 100\ Hz$, $200\ Hz$, and $300\ Hz$, respectively.

where B is statistical equivalent bandwidth of $p(t)$:

$$B = \frac{\alpha^2}{2\beta}. \qquad (12)$$

From (11), when N is large:

$$SNR \simeq BT, \qquad (13)$$

which is the results from the previous studies. In Fig. 1, some SNR curves are plotted according to (11) with different B and T. From the figure, the SNR is not necessarily a monotonically increasing function of N, which is the case before low-pass filtering (Parker & Scott, 1986). In certain cases, the SNR becomes a monotonically decreasing function of N. This is an important result for applications such as intra-muscular recordings, and neural recordings, where only a few units exist in one channel.

SIMULATION STUDIES

Several computer simulations were carried out to verify these theoretical results. The simulated EMG contained from 1 to 100 motor units. The mean firing rate of these units was $20\ Hz$, and the coefficient of variation of the firing rate was 18%. Two types of MUAP were simulated, with $B_1 = 130\ Hz$ and $B_2 = 380\ Hz$. The low pass filter was a $200\ ms$ long rectangular window. The resulting SNRs are plotted against the theoretical SNRs in Fig. 2. As shown in the figure, the theoretical results match the simulated results reasonably well.

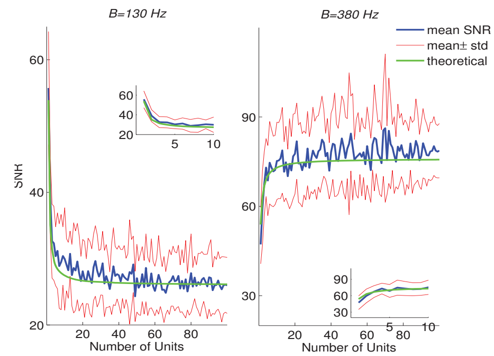

Figure 2. The SNR of simulated EMG. The Left plot is for $B_1 = 130\ Hz$, and the right plot is for $B_2 = 380\ Hz$. The inserts in each plot are zoomed-in plots with $N \leq 10$.

SUMMARY

In this study, the SNR of a myoelectric channel is derived without the assumption of large number of active motor units. It is shown both in theory and simulation that the SNR could be a monotonically decreasing function of N.

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ASSESSMENT OF SURFACE EMG RMS AMPLITUDE DURING MAXIMAL VOLUNTARY CONTRACTION

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INTRODUCTION

Descriptors of surface EMG amplitude estimated from a maximal voluntary contraction (MVC) effort, are extensively applied for normalization purposes (Bamman et al., 1997).

Since sEMG signal features depend on anatomical, geometrical and physiological factors (Merletti and Parker, 2004), between and within subject comparisons demand a normalization factor, which typically corresponds to the sEMG amplitude estimated from a MVC test. On the other hand, the choice of sEMG signal epoch to estimate the highest amplitude may not be established on empirical basis (i.e. an arbitrarily chosen epoch matching a period of constant force), but account for the non-linearities of EMG-force relationship (Vieira et al., 2007). Thus, which epoch and how long should it be to provide a reliable maximum sEMG amplitude descriptor for a MVC effort?

Force and sEMG RMS experimental data are fitted with an exponential function to assess the influence of test duration on the reliability of maximal sEMG amplitude during a prolonged MVC test.

METHODS

Fifteen healthy non-trained male subjects (23±3 years, 86±8kg, 181±0,05cm) were enrolled in the experiment after providing informed consent. Each subject was instructed to perform three sustained elbow flexion MVCs (10s) in seated position, with the elbow at right angle, the fist attached to a load cell fixed on the ground through a rigid cable and the left arm relaxed over the tight. Force feedback, verbal incentive, rest period (5min) and an acoustic stimulus to start contraction were provided.

Single differential sEMG signals were recorded using Ag-AgCl electrodes with 20 mm interelectrode distance, positioned 1/3 distant from cubital crease and oriented along muscle fibers. EMG amplifier gain was 1k, with 120 dB CMRR and band limited 10–400Hz. Both force and sEMG signals were synchronously digitized with 2 kHz sampling rate and 12 bits resolution using a custom script written in LabView 6.0 (National Instruments, Dallas, U.S.A).

Mean force and sEMG RMS time series were estimated using 250ms epochs and fitted with the forced first order model:

$$\mathrm{F}_{\mathrm{RMS,Force}}(t) = A\left(1 - e^{\frac{-1}{\tau_c}(t - t_o)}\right)u(t - t_o)$$

where A, τ_c and t_o correspond respectively to RMS and Force plateau, time constant (how fast subjects achieve their maximum) and the beginning of sEMG and Force activity.

RESULTS AND DISCUSSION

Normalized mean square error (NMSE) between RMS and force fitted curves and experimental data was less than 5% of maximum RMS and force values. The time response t_o decreased significantly from the first to both second and third trials for F_{RMS} and F_{Force} functions, suggesting a learning effect across trials (Table 1).

RMS plateau and time constant did not differ across trials. However, the time limit (t_{lim}) that provided the lowest NMSE between observed and fitted data was lower for RMS than force (Table 1). Even though subjects sustained a constant force level up to 6s of MVC the sEMG amplitude linearly increased after the first second of MVC (Figure 1), a phenomenon known as myolectric manifestation of muscle fatigue (Merletti and Parker, 2004).

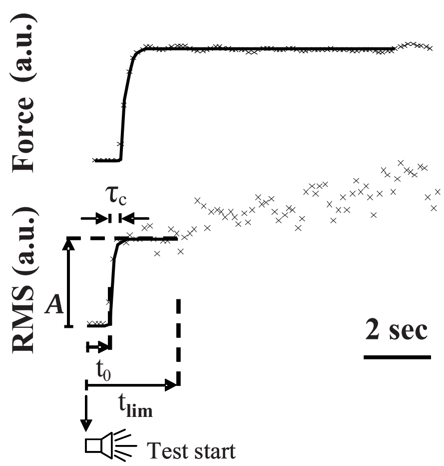

Figure 1: Force (top) and sEMG RMS (bottom) fitted and experimental data. Parameterization procedure is depicted on bottom. Note the sEMG RMS increase while force is constant.

Any time epoch chosen after t_{lim} would provide overestimated sEMG amplitude, since only during a short period of circa 1s (Figure 1) sEMG RMS and force maxima were stationary. Present results indicate that a time epoch of ≈1s, after achieving maximum force, provide a reliable sEMG amplitude estimate.

CONCLUSIONS

The time epoch chosen for sEMG amplitude estimation is critical. A single 1s epoch after reaching the maximum force seems to be adequate to provide a reliable sEMG amplitude estimate.

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Table 1: Mean (SD) of RMS and force fitted parameters for each trial

	$F_{RMS}(t)$			$F_{Force}(t)$		
	Trial 1	**Trial 2**	**Trial 3**	**Trial 1**	**Trial 2**	**Trial 3**
τ_c (s^{-1})	0.20(0.09)	0.25(0.16)	0.30(0.21)	0.434(0.36)	0.265(0.08)	0.27(0.14)
t_o† (s)	0.99(0.23)*	0.79(0.21)	0.86(0.17)	1.13(0.19)*	0.98(0.18)	1.01(0.15)
A (mV|N)	0.17(0.06)	0.19(0.07)	0.15(0.04)	294.9(62.9)*	285.7(52.3)	263.8(54.1)
t_{lim}† (s)	3,03(1,28)	2,76(0,45)	2,66(0,20)	4,71(2,49)	3,71(1,97)	4,43(2,57)

* $p<0.05$ for between trials effect. † $p<0.05$ additive effect between EMG and Force.

RELIABILITY OF THE MECHANOMYOGRAM DURING INCREMENTAL ISOMETRIC MUSCLE ACTIONS

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INTRODUCTION

The mechanomyogram (MMG) is a signal recorded from the surface of the skin that quantifies the oscillations or vibrations of contracting skeletal muscle (Orizio et al. 2003). The time domain (amplitude) is thought to reflect motor unit recruitment (Beck et al. 2006), while the frequency domain may provide a global estimation of motor unit firing rate (Beck et al. 2006). However, the MMG signal can be influenced by muscle temperature, mass, stiffness, and intramuscular pressure (Beck et al. 2006).

Since it may be useful to use MMG to monitor changes in recruitment (Evetovich et al. 1998), rate coding (Evetovich et al. 1998), muscle temperature (Muro et al. 2007), stiffness (Marek et al. 2005), and pressure (Blangsted et al. 2005), studies are needed to determine the reliability and external validity of the MMG signal. However, very few studies have investigated the test-retest reliability of MMG (Smith et al. 1997).

Therefore, the purpose of this study was to examine the test-retest reliability of MMG amplitude and mean power frequency recorded from the vastus lateralis during submaximal and maximal isometric muscle actions of the leg extensors.

METHODS

Nineteen healthy subjects (mean±*SD* age=24±4 yrs; stature= 173±9 cm; mass =75±13 kg) volunteered for this investigation. All participants signed informed consent forms. This study was approved by the University of Oklahoma Institutional Review Board.

Each subject visited the laboratory on 4 occasions: 1 familiarization trial and 3 experimental trials. Each trial was performed at the same time of day (±2 hours) separated by 2 – 5 days.

During each experimental trial, isometric torque of the right leg extensors was measured on a Biodex System 3 dynamometer. Each muscle action was 4-s in duration and performed at a 120° joint angle (between the leg and thigh). Each subject performed isometric maximal voluntary contractions (MVCs) and submaximal isometric leg extensions at 15, 25, 35, 45, 55, 65, 75, 85, and 95% MVC in random order.

MMG signals were recorded with an active miniature accelerometer (EGAS-FS-10-/V05, Measurement Specialties, Inc., Hampton, VA). The accelerometer was taped to the skin over the lateral/anterior portion of the vastus lateralis muscle. The most stable 1-s epoch of the 4-s torque plateau was selected to calculate the submaximal torque level, MMG amplitude (MMG_{RMS}), and MMG mean power frequency (MMG_{MPF}) values.

Repeated measures ANOVAs assessed the level of systematic variability among the trials. The intraclass correlation coefficient

(ICC) model 2,1 (Shrout and Fleiss, 1979) represented the relative consistency (test-retest reliability) and the standard error of measurement (SEM) represented absolute consistency (Weir, 2005) of the dependent variables (isometric torque, MMG_{RMS}, and MMG_{MPF}).

RESULTS AND DISCUSSION

Table 1 shows the type I error rate (*P*-value), ICC, and SEM for isometric torque, MMG_{RMS}, and MMG_{MPF} at all the submaximal and maximal torque levels.

The one-way repeated measures ANOVA indicated that there was systematic variability among the trials during the step contraction at 45% of MVC (*P*=0.034). There was no other systematic variability present.

SUMMARY/CONCLUSIONS

For MMG_{RMS} and MMG_{MPF}, the magnitude of the ICCs were moderate to high and were generally strongest from 35% to 100% MVC, while the SEMs ranged from 14.9 to 28.8% and 6.9 to 13.8% of the mean values for MMG_{RMS} and MMG_{MPF}, respectively.

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Table 1: Measures of reliability and measurement variability for isometric torque, mechanomyographic amplitude (MMG_{RMS}) and frequency (MMG_{MPF}) for step muscle actions at each percent of maximal voluntary contraction (MVC).

Percentage of MVC		15%	25%	35%	45%	55%	65%	75%	85%	95%	100%
Isometric Torque	*P-value*	0.052	0.071	0.085	*0.034	0.128	0.137	0.052	0.074	0.098	0.622
	ICC	0.90	0.91	0.91	0.92	0.92	0.92	0.92	0.92	0.92	0.91
	SEM	7.0	6.7	6.6	6.3	6.8	6.7	6.5	6.6	6.4	7.5
MMG_{RMS}	*P-value*	0.406	0.739	0.67	0.335	0.599	0.744	0.613	0.529	0.315	0.353
	ICC	0.54	0.65	0.81	0.74	0.83	0.78	0.87	0.89	0.88	0.86
	SEM	18.2	20.3	21.2	28.8	20.6	22.7	16.6	15.7	14.9	15.3
MMG_{MPF}	*P-value*	0.479	0.861	0.346	0.476	0.64	0.38	0.814	0.246	0.522	0.328
	ICC	0.84	0.60	0.72	0.73	0.72	0.80	0.69	0.65	0.76	0.71
	SEM	6.9	12.4	11.3	10.7	11.4	10.1	11.7	13.5	10.4	13.8

P-value = type I error rate for the one-way repeated measures ANOVA across trials 1, 2, and 3.
$ICC_{2,1}$= intraclass correlation coefficient, model 2,1 (Shrout and Fliess, 1979).
SEM=standard error of measurement, expressed as a percentage of the mean.
*reflects as significant (P≤0.05) difference across trials.

IN VIVO PASSIVE FORCES OF THE INDEX FINGER MUSCLES: VALIDATING CURRENT MODELS

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INTRODUCTION

The passive properties of the muscle have been found to be an important element in the biomechanical modeling of hand movements (Dennerlein et al., 1998; Keir et al., 1996; Sancho-Bru et al., 2001). However, due to the unique physiological characteristics of the intrinsic hand muscles and the complicated structure of the extensor hood mechanism, it is difficult to model the passive forces of the hand. Although few studies have measured the extrinsic muscle passive length-force property from cadaver fingers (Keir et al., 1996; Lee et al., 1990), the biomechanical parameters that predict the passive forces from these cadaver studies have limited in vivo validation. Therefore, we utilized a standard biomechanical model that incorporates typical exponential force-length relationship and *in vivo* fingertip forces. Through comparing the two, we implemented a particle swarm optimization (PSO) to determine the parameters of the exponential relationship. PSO is a population based stochastic computer algorithm developed by Eberhart and Kennedy in 1995 and it has been successfully applied in many research and application areas. It has been demonstrated that PSO gets better results in a faster, cheaper way compared with other methods (Schutte et al., 2005).

METHODS

We conducted a laboratory experiment measuring the passive fingertip forces of index finger in 18 postures in ten subjects. EMG confirmed no activity in the forearm muscles. Fingertip forces were averaged across the ten subjects for each of the 18 postures. For each posture we computed change of muscle length of six muscles -- flexor digitorum profundus (FDP), flexor digitorum superficialis (FDS), radial interosseous (RI), ulnar interosseous (UI), lumbrical (LU) and extensor digitorum communis (EDC) -- based on the excursions of the tendons of the finger adopted from the method used by Lee et al. (1990). Based on these length changes, we calculated muscle passive forces via the exponential relationship with its length excursion (e.g. Lee et al., 1990).

$$F_p = \beta_1 \cdot e^{(\beta_2 \cdot \Delta l)} \quad (1)$$

where, F_p is the passive muscle force, β_1 and β_2 are biomechanical coefficients, and Δl is the muscle length excursion from reference position. From these passive forces we calculated the fingertip force via

$$F_{tip} = (J^T)^{-1} R F_p \quad (2)$$

where, F_{tip} is the fingertip force, J is the Jacobian matrix for the one of the 18 postures, and R is the moment arm matrix for the posture.

Particle swarm optimization searched the parameter space for the biomechanical coefficients β_1 and β_2 which yield the passive fingertip forces that are closest to the measured experimental data.

RESULTS AND DISCUSSION

The average values from ten sets of the global best solutions of β_1 and β_2 after 10000 iterations were different than values presented by Lee and Rim (Table 1), which are based on fitting the length-force curves from three cadaver fingers.

The simulated betas of FDP, FDS and LU from the model were higher than their reported values, and betas of RI, UI and EDC were lower. This may due to the reason that the 18 postures measured in the experiment did not cover the whole range of muscle excursions. The excursions of FDP, FDS and LU calculated from the postures are on the lower side of the exponential curve and the excursions of RI, UI and EDC are on the higher side. We also only explored flexion forces and no extension due to the experimental protocol. This may explain the railing of the EDC passive force parameters.

Another reason that may cause the difference is subject variability. The subjects in this study were young and healthy (average age = 29.9±6.4 years), while the average age was 72 years for the cadavers.

The average best global fitness value of the ten runs was 0.0966 N / posture, and the average passive fingertip force across all postures was 0.4957 N. The optimization time for 10000 iterations is about 30 minutes on an IBM computer (Duo processor 6600 at 2.40 GHz).

In addition, we have assumed a specific configurations for the tendons of the finger; however, the extensor mechanism of the finger is quite complex and hence the model may not be completely representative.

CONCLUSIONS

A biomechanical model of passive muscle forces of the index finger using PSO algorithm was proposed. This model produced promising results to estimate and predict the passive forces and moments for different postures. Future work needs to be done on collecting data on a wider range of excursion distribution to improve the robustness of the model.

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Table 1: Average values of β_1 and β_2 of ten runs with 10000 iterations for index finger.

	Original Coefficients*		Coefficients after PSO	
	β_1	β_2	β_1	β_2
FDP	1.072	0.122	2.418	0.568
FDS	1.575	0.112	2.360	0.382
LU	0.198	0.255	2.407	0.431
RI	2.474	0.360	0.906	0.095
UI	1.966	0.408	0.549	0.098
EDC	4.164	0.118	1.939	0.001**

* from Lee. and Rim (1990)
** Minimum allowable value during PSO

A PHYSIOLOGICALLY ACCURATE MOTOR UNIT ARCHITECTURE MODEL

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INTRODUCTION

Motor unit architecture models are concerned with the statistical distribution and relationships of several physiological parameters, specifically, the motor unit fiber number (MUFN or innervation ratio), the motor unit territory area (MUTA), or the motor unit fiber density (MUFD). Available models assume a constant MUFD for all the motor units of the pool. However, there are physiological evidences that suggest that MUFD distribution is not the same for all MUs, but follows a different distribution for each MU type, characterized a by normal probability density function (PDF) with a different mean and variance (Kanda and Hashizume, 1992). We propose a new model in which the MUFN and MUFD distributions of different motor unit types can be established separately, allowing to match physiological data. The model is evaluated with the data reported by Kanda and Hashizume for the cat medialis gastrocnemius muscle.

METHODS

Muscle cross-section is modeled by a circle of radius *R*, filled with a set of *M* muscle fibers disposed in a hexagonal grid. A set of *N* motor units is then created. MUFN is distributed exponentially for the motor unit pool from n_{min} to n_{max}, satisfying Fuglevand exponential law, with the additional restriction that summation of the MUFN of all the motor units should match the number of muscle fibers in the muscle, *M*. Motor units of each type are ordered according to increasing MUFN: the smallest MUFN being S-type, intermediate MUFN being FR-type, and highest MUFN being FF-type motor units. MUFD is distributed normally for each motor unit type. Motor unit territories are modeled as circles, with an area that satisfies MUFN / MUFD. The territories are placed according to the DSVM-augmented model (Schnetzer et al., 2001), which attempts to minimize the spatial variance of the muscle fiber density, given the individual densities of the overlapping motor units at each point. For each muscle fiber, the innervating motor unit is selected at random from the ones covering its position (Stashuk, 1993).

In order to adequate the output MUFD to the design values, complete statistical analysis of the model was performed. In random innervation models, the innervation process generates an unavoidable variability in the outcomes of the MUFN for each motor unit. On the other hand, MUTA outcomes exactly match the design values due to a radius augmentation process, but are random variables themselves due to the random nature of the input MUFD. Incorporating the statistical description of the model itself to adjust the input parameters renders the advantage of adjusting the output MUFDs of each motor unit type to match the physiological distributions.

RESULTS AND DISCUSSION

Results, summarized in Table 1, show a high degree of accuracy reproducing the physiological data. Up to our knowledge, no previous attempt to fit motor unit architectural data including MUFD variability has been performed.

Two important implications stem from the simulation results. Firstly, if the motor unit pool is divided in types according to increasing MUFN values (Fig. 1(a)), and MUFD values are drawn from independent normal distributions for each motor unit type (Fig. 1(b)), the resulting MUTA values may not strictly follow the so called "size principle" (Fig. 1(c)). We can observe the high degree of overlapping of the MUTA values between the type S and type FR motor units, indicating that motor unit territory size may not be determinant for the classification of motor units. Secondly, although the overall fit is very accurate, the slight deviations of the MUFN and MUTA ranges suggest that a MUFN distribution slightly different from the exponential could still improve the fit.

The presented model can be applied to simulation studies relating muscle force and EMG activity. Some of these studies indicate an overestimation of the EMG amplitude at low-to-moderate levels of contraction. In the light of the present results, modeling different motor unit types with different MUFD distributions may lead to more accurate EMG amplitude estimations at these activation levels, as the number of generators would decrease accordingly to the lower MUFD of S and FR motor unit types.

SUMMARY

The presented model allows creating muscle architecture instances with a precise control of several motor unit architectural properties, i.e.: MUFN, MUFD, and MUTA.

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Table 1: Comparison of physiological and simulated data.

	Physiological data			**Simulated data**		
MU type	**S**	**FR**	**FF**	**S**	**FR**	**FF**
MUFN range (fibers):	41-80	116-198	220-356	42-71	77-177	212-426
MUFD m±sd (f./mm^2):	14.7±2.2	24.6±3.8	28.2±7.0	14.7±2.2	24.6±3.8	28.2±7.0
MUTA m±sd (mm^2):	4.2±1.1	5.6±1.0	9.3±3.3	3.9±0.6	5.1±1.6	11.8±4.8

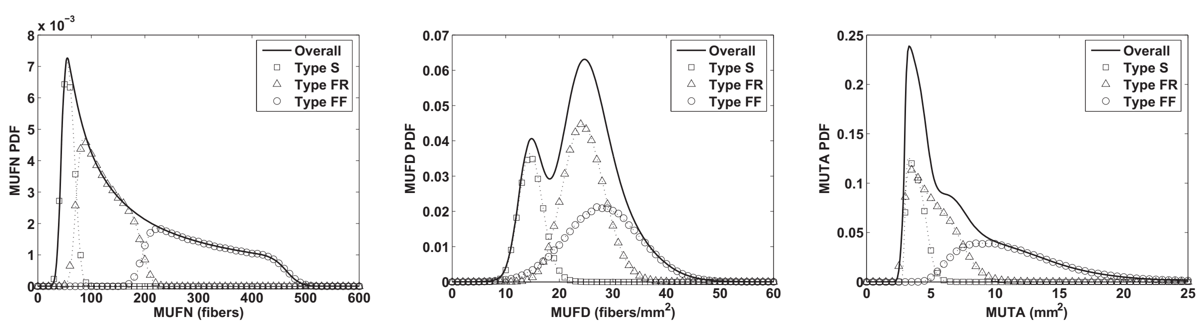

Figure 1: Probability density functions of: (a) MUFN, (b) MUFD, and (c) MUTA. The overall distributions are the composition of the individual distribution of each motor unit type.

ESTIMATION OF MONOPOLAR SIGNALS FROM SPHINCTER MUSCLES

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INTRODUCTION

Surface electromyogram (EMG) is usually recorded by means of spatial filters with vanishing sum of weights. More information could be extracted from monopolar signals measured with respect to a reference electrode away from the muscle.
Under some assumptions, surface EMG detected along a line parallel to the fiber path has zero mean value in space at any time. This property is a constraint which can be used to estimate monopolar signals from single differential (SD) EMG signals.

METHODS

Single differential (SD) signals $s_i(t)$ from an array of N electrodes can be expressed in terms of monopolar signals $m_i(t)$ as follows

$$\vec{s}(t) = \underline{\underline{A}}\,\vec{m}(t) \qquad (1)$$

where

$$\underline{\underline{A}} = \begin{pmatrix} 1 & -1 & 0 & \cdots & 0 \\ 0 & 1 & -1 & \vdots & \vdots \\ \vdots & \cdots & \ddots & \ddots & 0 \\ 0 & \cdots & 0 & 1 & -1 \end{pmatrix} \qquad (2)$$

where $\underline{\underline{A}}$ is a matrix with dimensions $(N-1)\times N$.

Matrix $\underline{\underline{A}}$ cannot be inverted as it has a vanishing eigenvalue, associated to an eigenvector with constant entries.
Nevertheless, pseudoinverse $\underline{\underline{A}}^{\#}$ of matrix $\underline{\underline{A}}$ can be evaluated and monopolar signals can be estimated as

$$\vec{m}_{est}(t) = \underline{\underline{A}}^{\#}\vec{s}(t) \qquad (3)$$

The resulting estimated monopolar signals $\vec{m}_{est}(t)$ have zero spatial mean. Thus, the estimation can be considered acceptable only if the monopolar signals $\vec{m}(t)$ to be estimated have zero spatial mean.
Under the assumption that the volume conductor is space invariant, the monopolar surface EMG detected along a curve parallel to the fiber path has zero mean value in space at any time. This is approximately true also when such a potential is detected by an array of electrodes covering the entire spatial support of the signal.
The method is particularly suitable for EMG signals from sphincter muscles, recorded with an array of electrodes located on a cylindrical probe along the circular path of the fibers, so that the spatial support of the signal is covered.

RESULTS AND DISCUSSION

The method was tested on signals simulated by a model of sphincter muscle proposed by Farina et al., 2004. A representative example of application to a simulated single fiber action potential (SFAP) with additive noise is shown in Figure 1. Performances of the method were assessed on SFAPs corresponding to fibers with different geometry and placed in different positions within the muscle, varying the number of detection channels and considering different perturbations of the simulated SD signals: A) white noise at different SNR, B) amplitude variations across SD channels, C) common mode sinusoidal interference with random amplitude variations across different channels.

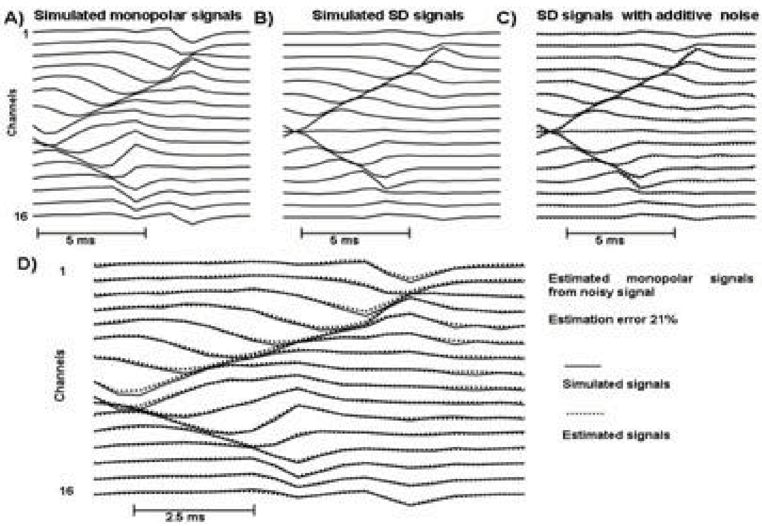

Figure 1
A) Example of simulation of monopolar SFAP, using a model of sphincter.
B) SD signals obtained from the monopolar signals. C) SD signals with 15 dB white noise.
D) Estimate of monopolar signals from perturbed SD signals.

The estimation error of monopolar signals decreased by increasing the number of channels and was negligible when at least 12 electrodes were used. Both propagating and non propagating components were estimated.
Equivalent results are obtained considering a motor unit action potential (MUAP) instead of a SFAP (Figure 2A) or interference EMG signals (Figure 2B).

SUMMARY/CONCLUSIONS

The application of pseudoinversion to reconstruct monopolar surface EMG signals from SD signals detected by a circular array along the direction of muscle fibers is discussed. Under the hypothesis of space invariance of the volume conductor, it is proved that monopolar signals detected along the direction of the muscle fibers with an array covering the entire spatial support of the potential distribution have vanishing spatial mean at any time. This provides a constraint for estimating monopolar from SD signals from sphincter muscles.

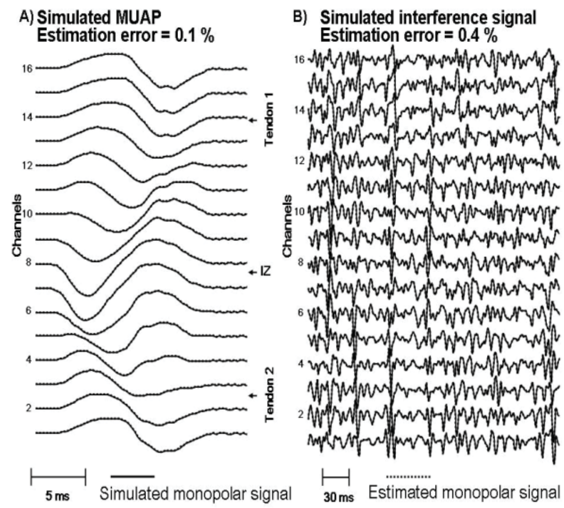

Figure 2: A) Application of the method to a MUAP (MU constituted by 154 fibers) and B) to an interference signal (60 MUs, force level 40% MVC). SD signals were obtained from the simulated monopolar signals, and then monopolar signals were reconstructed.

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EMG-TO-TORQUE MODELING: INTERPRETING THE ROLE OF MUSCLES CROSSING THE KNEE DURING AN IMPACT LANDING

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INTRODUCTION:

Investigators have proposed theories regarding the role of muscles crossing the knee for different loading scenarios based on the EMG magnitude relative to a reference contraction (Steele & Brown, 1999; Chmielewski, et al., 2002). Joint angles change considerably during these activities, thereby continually modulating force independent of the muscle activation level. Incorporating muscle mechanics with EMG is problematic and requires sophisticated muscle calibration measures and muscle modeling parameters (White & Winter, 1993). This study shows how a simple correction for joint angle affects the EMG-to-torque relation for medial hamstring, and affects the interpretation of that muscle's role in a sudden deceleration impact landing.

METHODS:

Ten male subjects performed a series of muscle EMG-to-torque calibration trials on an isokinetic dynamometer. After the calibration trials each subject performed a series of deceleration landing trials that involved stepping down and out from a 10 cm high bench to land in single leg support on a piezoelectric force plate. Subjects were instructed to land within three different ranges of knee flexion ($0 – 25^0$, $25 – 50^0$ or $50 – 75^0$). Three trials were collected and averaged for each landing angle. Surface EMG was collected using bipolar Ag/AgCl electrodes placed over the vastus medialis and lateralis, medial and lateral hamstrings, and medial gastrocnemius centrally on the muscle after standard skin preparation. The EMG signal was analog to digitally sampled (16 bit) at 2400 Hz after being amplified and bandpassed from 10 to 500 Hz. EMG data collection and processing was identical for the calibration trials and the landing trials. The EMG-to-torque calibration required the subject to perform varying effort, slow velocity (15 ^{0}/s) concentric contractions for knee flexion, knee extension and ankle plantarflexion. Calibrations were performed with the subject lying supine or prone with a neutral hip angle (~15^0 degrees of flexion). A 0.5 s root mean square (RMS) value was determined from the raw EMG signal at 15^0, 35^0 and 65^0 of knee flexion, and at 3^0 of ankle plantar flexion. Linear, best fit regression equations for EMG-to-torque were calculated based on at least 6 data points (Fig. 1). The regression equations were used to calculate muscle torque values during the landing trials based on 0.5 s RMS EMG values starting 0.25 s prior to the peak vertical ground reaction force at landing.

RESULTS AND DISCUSSION:

EMG-to-torque equations vary with joint angle. This angle effect was consistent across all subjects for each of the quadriceps and hamstrings monitored but only the results for medial hamstrings are reported here (Fig. 1). Muscle length and its moment arm vary with a changing joint angle, affecting muscle torque for a given muscle activation (EMG) value. With the hip at a relatively constant neutral angle, the hamstrings shorten with increased knee flexion while their moment arm increases (Herzog & Read, 1993). For submaximal efforts, torque for a given level of muscle activation decreases if the net mechanical advantage decreases. The medial hamstrings EMG-to-torque equations reflect this net decline (Fig. 1). These results are consistent with torque-angle data for knee flexors reported by others (Mohamed, et al., 2002).

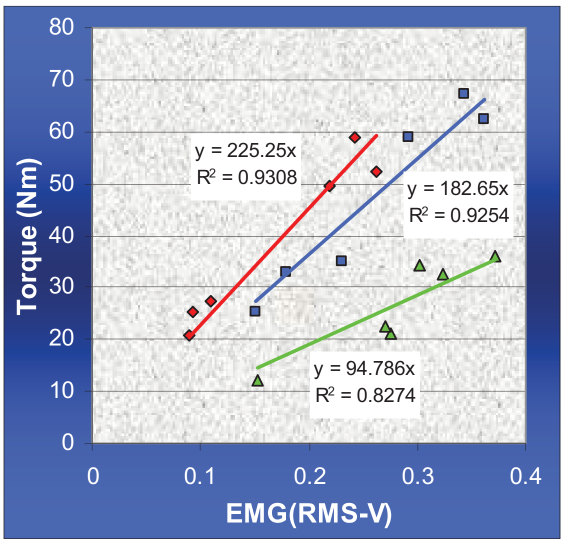
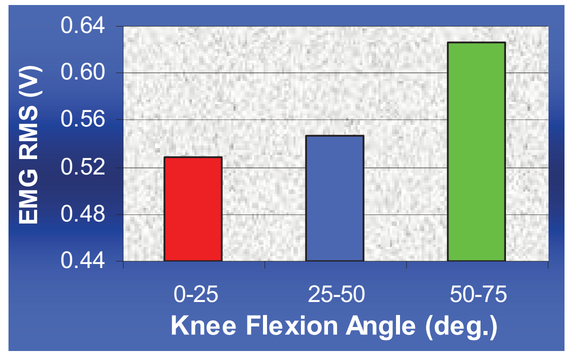


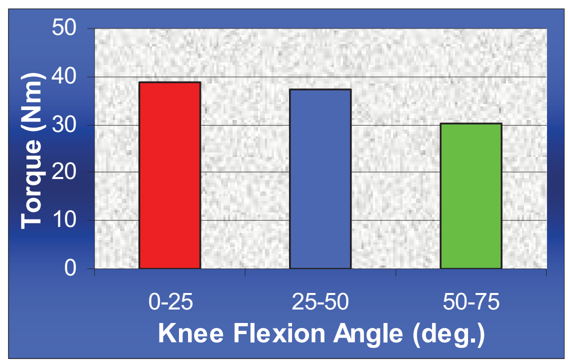
Fig. 1: Medial hamstring torque-to-EMG regression equations at 15^0 (red diamond), 35^0 (blue squares) and 65^0 (green triangles).

The average (N = 10) RMS EMG for the medial hamstrings increased with increased knee flexion at landing (Fig. 2a), suggesting that the hamstrings have a greater stabilizing or controlling role. When the hamstring RMS EMG data were converted using the EMG-to-torque equations based on an equivalent knee joint angle, there was actually a decrease in medial hamstring torque as knee flexion angle at landing increased (Fig. 2b).

Joint angles vary considerably during dynamic activities. Muscle moment arm and length are altered confounding the EMG-to-torque relationship under submaximal activation conditions. The effect of joint angle, in addition to other factors, need to be considered in conjunction with the EMG, particularly when calculating quantitative values such as co-contraction indices.



2a



2b

Fig. 2: Medial hamstring RMS EMG (2a) and torque (2b) when landing at three different knee flexion angles for a sudden deceleration task.

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Track 03

Ergonomics (ER)

NONLINEAR ANALYSIS OF REPETITIVE ARM MOVEMENT: EFFECTS OF PAIN AND EMPLOYMENT DURATION ON VARIABILITY

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INTRODUCTION

Variability is an important topic. Indeed, motor variability is though to have a key prognostic value with respect to the development of work-related musculoskeletal disorders (WMSD). Studies investigating motor control variability while performing a repetitive motor task are rare. Moreover, a lack of standardization often makes differentiation between task and motor control variability impossible. In controlled laboratory conditions, it has been shown that both experience and pain influence motor strategies during simulated repetitive work (Madeleine et al. 2003). Pain has been recently reported to influence the size of variability (Madeleine et al. 2007) but no studies have investigated the structure of variability.

Our aim was to assess the structure of motor variability during standardized repetitive work to delineate the effects of pain and work experience. For this purpose, we reanalyzed data from a prospective study (Madeleine et al. 2003).

METHODS

Twelve female workers (filetering butchers) took part in two recording sessions within one month of employment (0 month) and after six months work (6 month). At time 0 month all workers were pain free while 6 out of 12 had pain at time 6 months.

A time-paced repetitive work task was designed to simulate real work situations in a laboratory setting for 3 min 3 times at time 0 and 6 months.
A 3D motion analysis system (McReflex, Qualysis A/S, Partille, Sweden) was used for upper right arm and trunk movement recordings (Fs 30 Hz). Relative motion of the right arm and the trunk were expressed in 3D (Madeleine et al. 2003).

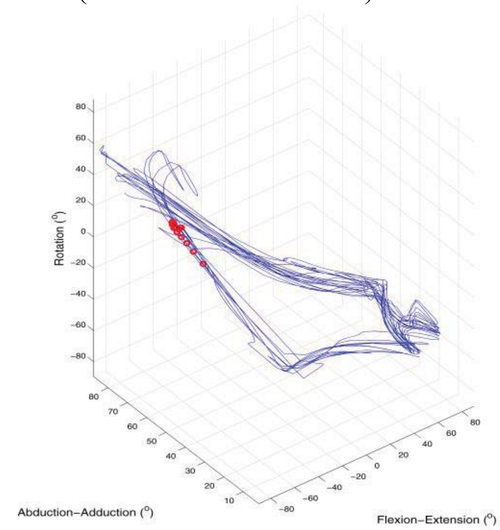

Figure 1: Example of 1-min arm movement in 3D with cycle start indicated by "**o**".

Motor variability was assessed by computing the unbiased autocorrelation, the sample entropy and the fractal dimension of the movement. Autocorrelation at lag 1 was computed to estimate the regularity between cycles. Sample entropy was computed to analyze the structure of variability. Finally, fractal dimension was computed by dispersion analysis to estimate the chaotic structure of the work.

Statistical analyses: Three-way analysis of variance (ANOVA) was applied.

RESULTS AND DISCUSSION

Table 1 summarizes the results of the statistical analysis. The autocorrelation at lag 1 of arm movement increased after six months work compared with 0 month (0.73 vs. 0.6 SEM 0.006) and was higher in workers with pain compared with no pain (0.72 vs. 0.62 SEM 0.006). Moreover, there was a significant interaction between pain and experience. The sample entropy of arm movement decreased after six months work compared with 0 month (2.55 vs. 2.66 SEM 0.008) and was higher in workers with pain compared with no pain (2.66 vs. 2.56 SEM 0.008). The fractal dimension of arm movement decreased after six months work compared with 0 month (1.21 vs. 1.26 SEM 0.003).
For the trunk, the autocorrelation at lag 1 was higher in workers with pain compared with no pain (0.8 vs. 0.69 SEM 0.007). Moreover, there was a significant interaction between pain and experience. The sample entropy of trunk movement decreased after six months work compared with 0 month (2.56 vs. 2.67 SEM 0.007) and was smaller in workers with pain compared with no pain (2.56 vs. 2.67 SEM 0.007). The fractal dimension of arm movement decreased after six months work compared with 0 month (1.26 vs. 1.29 SEM 0.004).

For the arm, the similarity among cycles increased after 6 months. This contrasted with the more heterogeneous structure of cycles (sample entropy) observed for the arm and trunk after 6 months work. While the decrease in fractal dimension with experience underlining more constant signal was in line with higher similarity among cycles.
In workers with pain (developed after 6 months work), the similarity among cycles was higher for both arm and trunk movement in line with a study investigating the degree of variability in pain conditions (Madeleine et al. 2007). Nonlinear analysis corroborated this finding only for the arm as a more homogenous structure of the movement was observed.

SUMMARY/CONCLUSIONS

The present results underlined that the structure of motor variability is also influenced by experience and pain. Nonlinear analysis might reveal important characteristics of motor control helping to prevent WMSD.

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Table 1: Statistical results on the autocorrelation at lag 1 (Ad1), sample entropy (SEnt) and fractal dimension (FD) for the arm and trunk (F: variance ratio, P: level of significance)

	Arm						Trunk					
	Pain		Exp		Pain*Exp		Pain		Exp		Pain*Exp	
	F	P	F	P	F	P	F	P	F	P	F	P
Ad1	10.1	**.002**	16.0	**<.001**	8.5	**.004**	11.5	**<.001**	0.1	.96	4.2	**.04**
SEnt	7.0	**.009**	7.6	**.007**	1.5	.23	8.3	**.005**	8.7	**.004**	0.6	.44
FD	0.1	.98	11.0	**.001**	0.43	.51	0.8	.36	4.3	**.04**	0.1	.77

ASSESSING GAIT CHANGES DUE TO FATIGUE AND PROTECTIVE CLOTHING

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INTRODUCTION

Sufficient mobility and balance abilities are important for safety and efficient work performance of workers such as firefighters (FFs) who must perform physically demanding tasks, often in hot and hostile environments. FFs wear personal protective equipment (PPE) consisting of specialized clothing and a self-contained breathing apparatus (SCBA) to protect themselves from the environment. However, the use of PPE presents potential problems due to heat stress, increased fatigue rate and impaired mobility. The goal of this study was to assess the effect of fatigue, via a simulated firefighting activity (FFA), and restrictive protective clothing (firefighting PPE) on mobility, balance, and gait. Specifically, we investigated changes in gait parameters due to PPE use, PPE configuration and 18 min of strenuous FFA.

METHODS

Thirty-four male FFs (ages 18-50 years) participated in this study. Subjects were divided into control (n=15) and intervention (n=19) groups. The control group (standard PPE) wore PPE typically used by US FFs, i.e., helmet, heavily insulated hood, bunker coat and pants, and rubber boots. The intervention group (enhanced PPE), wore PPE (designed with industrial partners) that incorporated the same components but included: 1) a lighter helmet, 2) more breathable (nomex) hood, 3) bunker coat and pants with reduced insulation, maximum breathability, and a passive cooling system to assist with heat transfer; and 4) lightweight leather/Kevlar boots. Both PPE ensembles meet current guidelines for thermal protection and breathability. Both groups wore identical SCBA and masks. To assess the effect of wearing PPE and FFA, subjects were evaluated at three testing periods: initially in normal clothing (baseline), immediately before FFA in PPE (pre-activity), and immediately after FFA in PPE (post-activity). FFA consisted of alternating work-rest cycles that involved four simulated firefighting activities in a burn tower that contained live fires for 18 min. Post-activity testing occurred within 12 min after completing the simulated FFA. For the gait assessment, subjects were instructed to walk at either of two speeds (“normal, comfortable pace” or “as fast as you can without running”) on an 8 m instrumented gait mat (GAITRite Platinum, CIR Systems Inc). Outcome gait parameters were based on average values over two trials per condition.

The following measures were examined using repeated-measure multivariate analysis of variance (MANOVA) tests to assess the effect of PPE group, testing period, and instructed speed on average 1) overall gait speed, 2) step length (normalized by leg length), 3) step width, and 4) single leg stance time (expressed as percent of the gait cycle); labeled as GS, SL, SW, and SLST, respectively.

RESULTS AND DISCUSSION

The MANOVA found significant main

effects associated with testing period (baseline, pre-activity, and post-activity, *p<0.001*), instructed speed (normal vs. fast, *p<0.001*), and PPE (standard vs. enhanced, *p=0.025*). Subsequent univariate ANOVAs identified specific relationships between testing factors and outcome gait parameters.

Only SLST decreased with each testing period ($p<0.001$; Table 1). Differences in other gait parameters (GS, SL, and SW; $p<0.001$) were only due to use of PPE (baseline to pre-activity) and not FFA (pre- to post-activity). Previous studies have also found that PPE use decreases FF performance during balance tasks (Kincl, 2002; Punakallio, 2003). We are not aware of other studies that have investigated the effect of FFA on gait parameters. As expected, walking speed (from normal to fast) significantly increased GS, SL, and SLST ($p<0.001$), but not SW. Significant interactions (instructed speed × testing period; $p \leq 0.002$) were found for GS, SL, and SLST such that faster speed accentuated declines in these parameters due to wearing PPE (i.e., at fast speed, values decreased substantially between baseline and pre-activity, but not between pre- and post-activity).

Subjects wearing the enhanced PPE were found to have smaller SW ($p=0.002$) and longer SLST ($p=0.03$) than control subjects in standard PPE (Table 1). PPE style did not affect GS nor SL significantly ($p>0.05$), though subjects with enhanced PPE tended to use faster gait speeds and longer step lengths. These results suggest that the enhanced PPE may allow greater gait performance.

SUMMARY/CONCLUSIONS

The use of both PPE ensembles was found to significantly impact gait performance. The use of new enhanced PPE, however, appears to minimize these effects. Further study is necessary to determine which parts of the ensemble (clothing, helmet, boots) provide the greatest impact. Participation in strenuous FFA did not appear to affect gait performance except for reducing single leg support time. These results may suggest that the heat and work conditions encountered during these FFA do not significantly alter gait behavior of firefighters (even though body temperature increased significantly). Significant differences in some parameters between the two PPE ensembles warrant further investigation, particularly considering the wide range of activities that firefighters may engage in.

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Table 1. Mean ± Standard Error for gait parameters. Superscript denotes significant difference from indicated group (*p<0.05*).

		Gait Speed (cm/s)	Step length (%LL)	Single support (%GC)	Step width (cm)
PPE	Standard (A)	178.9±3.3	93.3±1.6	38.2 ± 0.2^{B}	13.1 ± 0.5^{B}
	Enhanced (B)	186.0±2.9	96.3±1.4	39.9 ± 0.2^{A}	10.8 ± 0.5^{A}
Testing Period	Baseline (A)	187.4 ± 2.8^{BC}	96.6 ± 1.3^{BC}	39.6 ± 0.2^{BC}	11.0 ± 0.4^{BC}
	Pre-activity (B)	179.1 ± 2.1^{A}	93.9 ± 1.1^{A}	38.2 ± 0.2^{AC}	12.3 ± 0.4^{A}
	Post-activity(C)	180.8 ± 2.5^{A}	93.9 ± 1.1^{A}	37.9 ± 0.2^{AB}	12.6 ± 0.4^{A}
Instructed Speed	Normal (A)	155.9 ± 2.4^{B}	88.9 ± 1.2^{B}	37.6 ± 0.2^{B}	11.8±0.4
	Fast (B)	209.0 ± 3.3^{A}	100.7 ± 1.1^{A}	39.5 ± 0.2^{A}	12.1±0.4

EVALUATION OF EASE OF TABLET INTAKE BY SENSORY TEST AND EMG

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INTRODUCTION

Japanese society is aging rapidly, and is expected to become an extremely aged society in which approximately one quarter of the population are of old age. Additionally, society has a growing interest in self-medication, and many health supplements are distributed in tablet or capsule form. Therefore, there are increasingly more and more opportunities for medicinal drug use. Consumers expect drugs which are not only efficacious but drugs which are easy to take.

There have been previous studies on the ease of tablet intake that examined the swallowing time for tablets using X-ray photography, but these methods involve a good deal of difficulty and/or discomfort for the subject. Therefore we sought to develop a technique to evaluate ease of tablet intake by physiologic measurement and sensory tests.

The purpose of this paper is to investigate how the size and number of tablets influences the ease of swallowing. We examined the correlations between subjective tests and the electromyography or the vibration of the larynx.

METHODS

Samples were tablets which were made of lactose (67.2%), cornstarch (28.8%), binder (3.5%) and lubricant (0.5%). We selected three kinds of tablets (TA, TB, and TC) which were different in diameter, as shown in Table 1. We tested each of these tablets (TA, TB, TC) using 1 to 5 tablets per sample, labeled TA1, TB1, TC1 for 1 tablet samples, to TA5, TB5, TC5 for 5 tablet samples.

Table 1: Details sample.

	TA	**TB**	**TC**
Diameter (mm)	6	8	10
Curvature radius (mm)	4.5	12	17
Thickness (mm)	3.78	3.66	4.16
Weight (mg)	101±1	199±1	355±1
Volume (mm^3)	76.1	149.8	224.0

Subjects were six healthy adult males who took a pre-test to fully explain the contents and gave informed consent. The subjects each sat on a chair and each sample was placed on a table with 20ml of water. The samples were presented at random. Before and after taking the sample, the subjects evaluated ease of intake using five grades, from easy to swallow to hard to swallow. Before intake, the samples were evaluated only by only visual evaluation without touching the sample.

We used surface electromyography (EMG) and vibration to measure the muscle activities of the larynx. We chose four muscles: two sides of the geniohyoid muscle and two sides of the sternothyroid muscle. The three axis acceleration sensor for measuring vibration was patched at the bottom of the thyroid cartilage. The horizontal axis was defined as X and the vertical axis as Y. When taking the sample, the EMG and the vibration were measured by the analog data recording system (DataLINK, Biometrics Ltd.) with the EMG amplifier (SX230) and the three axis acceleration sensors (ACL 300). The EMG was measured at a sample rate of 1kHz and the vibration measured at 100Hz.

RESULTS AND DISCUSSION

From sensory tests after taking the sample, TA1 and TA2 were evaluated as very easy, and TC5 was evaluated as very hard. All of the TA samples were evaluated as easy, and all of the TC samples were evaluated as hard. From the results of TA3 and TC1, which were of near identical weight, it can been seen that TA3 which was smaller was easier to swallow than TC1. The same also applies to TA2 and TB1, TA4 and TB2. For TB4 and TC2, both of samples were evaluated as a little hard, but TB4 was easier than TC2. This implies that if we are taking tablets that are of near identical weight, we should take many smaller ones rather than a small number of larger ones. From the correlation between the sensory value of before taking and that of after taking, it was found that there were significant correlations at 1% level.

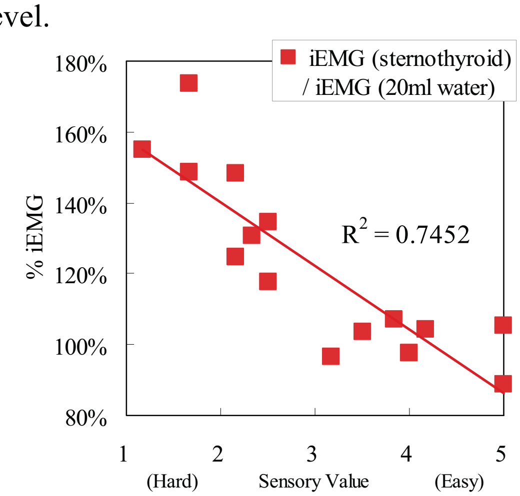

Figure 1: The relationship between sensory value and %iEMG of sternothyroid.

The raw EMG signals were sampled at 1kHz, full-wave rectified and integrated EMG (iEMG) from the start point of measurement to the end point. Each iEMG was normalized by the iEMG by having a subject drink 20mL water without any tablets, and was defined as %iEMG. The %iEMG of the sternothyroid of TA1 to TA5, TB1 and TB2 were less than 110%, which is very small. The samples which were small %iEMG correlated with the samples which were easy to swallow for sensory test. The relationship between sensory values after intake and %iEMG of sternothyroid is shown in Figure 1. From Fig.1, it was found that there were significant correlations at 1% level. And we could evaluate the ease of intake of tablets by using the %iEMG of sternothyroid.

We calculated the vibration dose value (VDV) for each axis from the vibration signals of the larynx. Each VDV was normalized in the same way as the iEMG, and we defined it as %VDV. The correlation coefficient between the sensory value and %VDV was -0.58 (significant at 5% level).

SUMMARY/CONCLUSIONS

From the correlation between the sensory value of before intake and that of after intake, we have seen it is possible to predict the ease of tablet intake by only visual evaluation.
From the correlation between the sensory value and the %iEMG, it was found that we could evaluate the ease of intake of tablets by the %iEMG of sternothyroid.

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TEMPORAL ACTIVATION PATTERNS DURING A CONTROLLED HORIZONTAL TRANSFER TASK: COMPARISON OF MEN AND WOMEN

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INTRODUCTION

Coordination of the forces produced by the trunk musculature in response to changing torques on the spine is important to maintain spinal stability and produce motion. Studies have evaluated the temporal features of the entire electromyographic (EMG) waveforms from a number of trunk muscles during dynamic tasks (Hubley-Kozey and Vezina, 2002; Lamoth et al., 2006), but none have been performed on functional load handling tasks. Therefore the primary purpose of this study was to determine the magnitude and temporal activation patterns of the major trunk muscles while performing a controlled horizontal transfer task. A secondary purpose was to determine if differences in patterns exist between men and woman.

METHODOLOGY

Twenty healthy women (age 31±8yrs,BMI =22.6±4) and 18 men (age 30±7 yrs, BMI =24.5 6±3) were recruited through local advertisements. Participants provided institutional approved informed consent. Meditrace™ electrodes (0.79cm^2) were placed over 12 trunk muscle sites bilaterally: upper and lower rectus abdominus; anterior, lateral and posterior external oblique; internal oblique; longisimus and iliocostalis at two spinal levels (L1,L3); multifidus and quadradus lumborum. Participants performed a horizontal transfer task in random order for two conditions i) from right to left and ii) left to right. The task was performed in the maximum reach using a 2.9 kg load. Subject's spine and trunk motion were monitored throughout the transfer task using a FOB™ system. Then, maximal voluntary isometric contractions (MVIC) were performed for normalization purposes. EMG signals were amplified (AMTI-8, Bortec, Canada) and digitized at 1000 Hz using Labview™ and the angular motions were recorded at 50 Hz. Data processing algorithms were written in Matlab™. An event marker identified when the load was i) lifted off the table, ii) as it passed the body midline and iii) when it was set back down. The EMG signals were corrected for subject bias and gain, full wave rectified, low pass filtered (Butterworth, zero lag, 6Hz), time normalized to represent 100% of the task (lift off- back down) and amplitude normalized to MVIC.

Principal components (PC) were extracted from the measured EMG waveforms using pattern recognition techniques and *PC-scores* were computed for each waveform (Hubley-Kozey and Vezina, 2002). Separate analysis of variance models tested for sex, muscle and condition main effects and interactions ($\alpha = 0.05$) for the abdominal and back extensor *PC-scores*. Bonferonni post hoc tests determined pair wise differences.

RESULTS AND DISCUSSION

Ninety-seven percent of the variance was explained by three PCs (See Figure 1). PC1 captured the overall magnitude, PC2 captured an asymmetric response and PC3 a gradual increase as the load was moved toward the midline and then gradual decrease as it moved away.

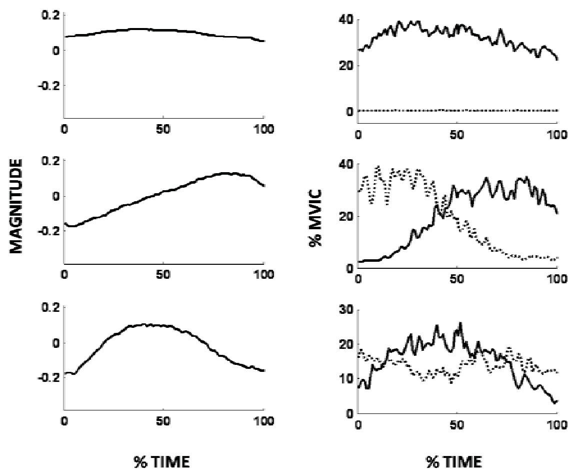

Figure 1: PCs are in the left column. On the right, EMG waveforms associated with a high PC *score* are depicted by the solid line and low *score* the dotted line.

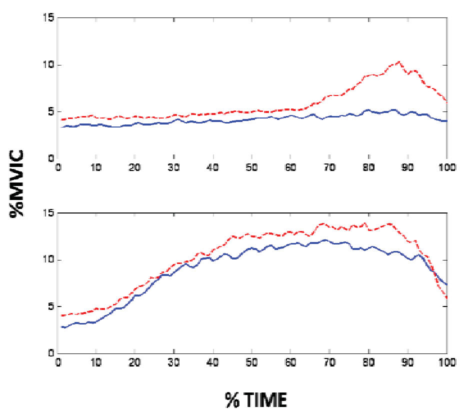

Figure 2: Right posterior external oblique is in upper panel and longissimus at L3 in lower panel. Red dotted is for women and blue for men.

There were statistically significant ($p<0.05$) muscle by sex and condition by muscle interactions for PC1 and PC3 *scores* for the abdominals and a muscle by sex by condition interaction for PC2. Women recruited their abdominal muscles with a higher overall magnitude and a difference in lateral and posterior external oblique responses to the changes in loading compared to men. See upper panel in Figure 2 for a right to left transfer task.

For back extensors, only a muscle by sex by condition interaction for PC2 *scores* was found ($p<0.05$) and no other sex effects. Similarities in waveform amplitude and shape between men and women are in the lower panel of Figure 2. PC1 had a significant condition by muscle affect and PC3 a significant muscle affect only. These results captured a pattern of higher activity for those muscles closer to the midline of the body, similar to a high PC3 *score* (Figure 1, lower right) whereas those farther away had higher PC2 *scores* i.e. more responsive to load location (Figure 1 middle right).

SUMMARY/CONCLUSIONS

In this healthy sample, amplitude and temporal activation patterns of 24 trunk muscle sites were altered based on muscle location and load location during the horizontal transfer task. Coordination of activation patterns was specific to muscle groupings, indicative of distinctive roles during this relatively simple task. As well, there were more differences between men and women for abdominal muscle amplitude and temporal patterns than for back extensor patterns. These data provide a template for developing methods to assess and train neuromuscular control of functional tasks.

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A STUDY OF EMG AMPLITUDES DURING COMPUTER MOUSE USE IN SYMPTOMATIC COMPUTER USERS

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INTRODUCTION

With widespread use of computers, the "mouse" is becoming an essential input device in daily office work. Intensive mouse operation is reported to be associated with disorders such as forearm tenosynovitis and carpal tunnel syndrome. Previous research has demonstrated that the neck-shoulder muscles such as the upper trapezius displayed consistent differences in muscle activities between symptomatic and asymptomatic computer users (Szeto et al, 2005). These muscle activity differences were thought to be an important mechanism in musculoskeletal disorders. The present study aimed to examine whether such muscle activity differences also existed in other muscles such as the forearm extensors and flexors – which are heavily involved in operating the computer mouse.

METHOD

Office workers with computer-related hand/wrist disorders were recruited as Case Group subjects (n=10) and those without symptoms participated as Control Group (n=10). Each subject performed 4 multi-directional mouse clicking tasks with: (1) high precision (HP), (2) low precision (LP), (3) fastest possible speed (ASAP) and (4) constant speed (CS).

Surface electromyography (sEMG) of right flexor carpi ulnaris (FCU), flexor carpi radialis (FCR), extensor carpi radialis (ECR) and extensor carpi ulnaris (ECU) were measured using the Noraxon Telemyo system (Noraxon, U.S.A. Inc., U.S.A.) with a sampling frequency of 1500Hz and band-width of 10-500Hz. Bipolar Ag-AgCl (3M™ Infant Red Dot™) surface electrodes of 15mm diameter (3M Hong Kong Ltd, Hong Kong) were placed on the muscles with an inter- electrode distance fixed at 20mm. Standard normalization procedures were performing with 3 trials of maximum voluntary contractions (MVC) of 5 sec for each muscle. After this, the subjects performed the 4 mousing tasks for 5 min each in a random order. The same computer equipment and workstation was used by all subjects.

Signals were demeaned, rectified, down-sampled to 10Hz root mean square (RMS) values and normalised to compute the %Maximum EMG (MEMG). The dependent variables of 10^{th}%, 50^{th}%, 90^{th}% and the APDF range were compared between groups and between tasks. The amplitude range is the difference between the 10^{th}% and 90^{th}%APDF and it can be considered as an indicator of the extent of variation of the EMG amplitudes.

RESULTS AND DISCUSSION

Case Group generally exhibited higher muscle activities in all 4 muscles among the 4 tasks compared to Control Group, and there is a statistically significant difference between groups (p=0.01). In particular, the ECR and ECU muscles displayed greater Case-Control differences in all 3 levels of APDF. For the 4 mousing tasks with different precision and speed demands, the forearm extensor muscles (ECU and ECR) generally worked in the range of 5-15% MEMG, while the flexor muscles (FCU and FCR) activities were lower, in the range of 1-10%MEMG.

Comparing the different task demands, the results showed that the high speed demand elicited the greatest increase in muscle activities, followed by the high and low precision and lastly, constant speed. Fig 1 shows the median activities (50^{th}%APDF) of the 4 muscles during the ASAP task. Comparisons of the Case and Control amplitude range (90^{th}%-10^{th}%) of the ECR muscles are illustrated in Fig. 2. The amplitude range is a new variable being reported and it has shown good evidence of greater variability in muscle activation in symptomatic individuals. The group differences also displayed similar patterns for the other muscles.

CONCLUSION

The present results showed consistent differences in muscle activities between symptomatic and asymptomatic computer users. These differences were consistent in both the median activities as well as the amplitude range. The results would provide further evidence to support the important role of motor control in contributing to the development of work-related musculo-skeletal disorders in the forearm and wrist region.

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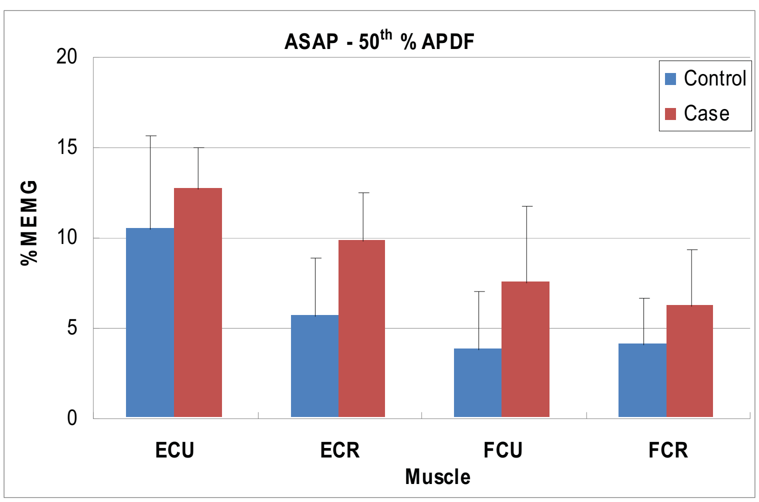

Figure 1: Mean 50^{th}%APDF in 4 muscles of Case and Control Groups during the fastest speed demand (ASAP)

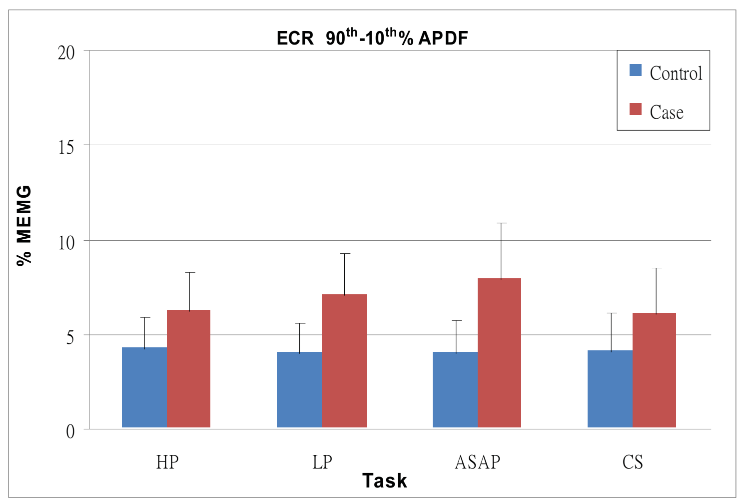

Figure 2: Comparison of amplitude range in ECR muscle between Case and Control Groups

HIGH DENSITY SEMG RECORDINGS FOR ONLINE BIOFEEDBACK APPLICATIONS

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INTRODUCTION

Muscle fatigue can be regarded as a precursor of muscle pain and disorder. Therefore biofeedback used as a mean to prevent muscle pain in the neck and shoulder region has primarily been based on parameters related to fatigue development. One of the most commonly used source for biofeedback is the crude measure of surface electromyographical (SEMG) amplitude. It is well known that increase in amplitude is associated with fatigue development during well-controlled laboratory sustained contractions. However, in occupational settings the variations in amplitude are much larger and fatigue related changes not so consistent.

Recent studies have shown not only amplitude increases, but a reorganization of the spatio-temporal activity patterns of the upper trapezius muscle in relation to both fatigue and experimental pain. (Farina et al 2006, Madeleine et al 2006) Therefore a promising source for biofeedback could be high-density SEMG recordings providing data on the changes in spatial distribution of activity above the upper trapezius. However, to be applied as online biofeedback, the complex data processing from the high density matrix requires reductions in the quantity of the recorded signals. The aim of the study was to investigate the ability to extract spatio-temporal features from the high density SEMG for application as online biofeedback.

METHODS

Eleven healthy, right-handed male subjects (body mass 71.9 ± 6.6 kg, and stature 1.79 ± 0.06 m, body mass index 22.6 ± 2.2) participated in the study.

The study was conducted in accordance with the declaration of Helsinki and approved by the Local Ethics Committee, and written informed consent was obtained from all participants prior to inclusion.

SEMG signals were detected with a semi-disposable adhesive grid of 64 electrodes. The grid consists of 13 rows and 5 columns of electrodes (2-mm diameter, 8-mm inter-electrode distance in both directions) with a missing electrode at the upper right corner. The missing electrode was considered the origin of the coordinate system to define electrode location. The SEMG signals were amplified in bipolar configurations for 5000 times and sampled at 4096 Hz. The 64-electrode grid was then placed on the upper trapezius muscle with the 4th row aligned with the C7-acromion line, parallel to the muscle fiber direction. The lateral edge of the grid was 10 mm medial to the identified innervation zone. A reference electrode was placed at the right wrist of the subject. Bipolar signals were computed in the fiber direction, thus 51 bipolar derivations arranged in 13×4 were obtained. Before data analysis, the SEMG signals were digitally band-pass filtered in the frequency bandwidth 10–400 Hz (2nd order Butterworth filter).

Subjects sat comfortably on a chair. Subjects were then asked to abduct both arms at 90^o with elbows fully extended and forearms 90^o pronated, with palm facing toward the ground and without hand load. Two flexible bars placed at shoulder level were used to provide tactile position feedback to the subject. The static contraction was sustained until it was not possible for the subject to maintain the arms in contact with the flexible bars.

Topographical maps along time axis are calculated by applying a number of rectangular windows on each data channel over the whole data set equally distributed in time. The number of windows was set to 5. Each window consists of 10 sec of SEMG signal and divides into 10 non-overlapping 1 second epochs of signal. For each channel of data, averaged rectified values (ARV) are calculated over each epoch. Then for each window, the average value is considered as ARV value for the window at time 0-25-50-75-100% of contraction time.

To represent each window with a single value, the center of gravity (COG) corresponding to the 2D was computed.

A reduced electrode grid was used; meaning that only a limited number of SEMG channels were applied for extracting the features, i.e. cranial- caudal shift. For this purpose, the 1st and last rows of the electrode grid were selected. As the COG shift mainly occurs in the cranial-caudal direction, all ARV values along medial-lateral were summed for each row. A linear regression analysis was performed to investigate the relation between the shift of the center of gravity and linear combination of selected rows. Significance was set to $p < 0.05$.

RESULTS AND DISCUSSION

10 out of 11 subjects showed significant fit between the COG from all channels and the linear combination mean between the first and last row of the grid. Figure 1 illustrates how the mean of the linear combination from all subjects of the first and the last row can be fitted on the mean shifts in the COG curve.

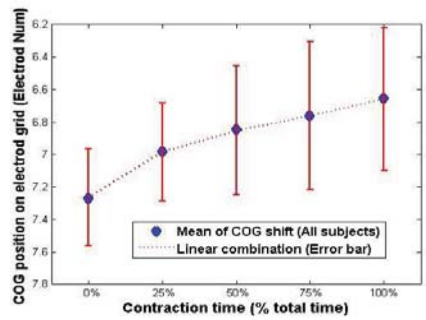

Figure 1: Comparing shift of COG and linear combination mean of the first and the last row. Horizontal axis represents time windows (0 to 100%, 25% increment) of contraction time.

SUMMARY/CONCLUSIONS

The study confirmed the possibility of extracting representative information from a few SEMG channels. Spatial changes in muscle activity are thus available as source for online applications such as biofeedback.

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TRUNK MUSCLE RECRUITMENT STRATEGIES AND HANDEDNESS

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INTRODUCTION

In many ergonomic studies, it is assumed when lifting with the right hand that the trunk muscle responses are mirror images of the responses when lifting with the left hand. However, evidence suggests that preferential use of the dominant hand (handedness) may change muscle properties that affect muscle fatigue variables [Merletti et al., 1994; Sung et al., 2004] and spinal loading [Marras and Davis, 1998]. Presently, the affect of handedness on trunk muscle amplitude recruitment strategies during work related tasks has not been fully explored. Therefore, the purpose of this study was to quantify how activation amplitude patterns consisting of 24-trunk muscle sites during a one handed asymmetrical lift and replace task were affected by different work conditions (lifting with the right or left hand) and whether the patterns changed during different task demands (horizontal reach).

METHODS

Healthy, volunteers aged 20-50 years without a history of low back pain were recruited. Ag/AgCL (Meditrace .79 cm^2) surface electrodes were placed in a bipolar fashion parallel to the fiber direction of 24 trunk muscle sites (Right (R) and Left (L)); rectus abdominis lower (LRA) and upper (URA), external oblique representing the anterior (EO1), lateral (EO2) and posterior fibers (EO3), internal oblique (IO), back extensors at different lumbar levels L1, L3, L4 and L5. For L1 and L3 electrodes were placed at 3 and 6-cm from the midline. For L4 and L5 electrodes were placed 8 and 1-cm from the midline, respectively. All sites were based on standard placements but were adjusted for individual anthropometrics. The EMG signals were amplified (3 AMTI-8 Bortec, Canada) and sampled at 1000Hz using Labview™ and processed using Matlab™ software.

From a standing position the subjects were asked to lift a 2.9kg load 5-cm off the table in a normal and maximum reach position [Butler et al, 2007]. The load was positioned either 45-degrees to the right (RASYM) or left (LASYM) of the body midline. Three trials were recorded for each side. Subjects were asked to maintain a comfortable upright trunk position while minimizing spine and pelvic motion. Lumbar and pelvic position was monitored by a magnetic sensing device. The root mean square (RMS) amplitude over the lift phase was calculated for each site for all trials. For normalization purposes the subjects performed 9 different maximum voluntary isometric contractions (MVIC) [Butler et al, 2006]. The overall maximum RMS amplitude from a 500-msec moving average window was used to normalize (NRMS) the lifting trials as a %MVIC.

The 24 NRMS amplitudes made up the activation amplitude pattern, which is unique to the order of the muscles, and was used in the pattern recognition algorithms [Hubley-Kozey and Smits, 1998]. The principal patterns (PPi) and scores were derived with the scores used in an ANOVA repeated measures model to determine if there were significant differences in shape and amplitude of the patterns for reach and hand. All significant effects ($p<0.05$) were tested using a Bonferonni post hoc test.

RESULTS AND DISCUSSION

Twenty-nine healthy individuals with a mean age of 30.9±9.1 and mean BMI 23.5±3.6 participated in this study. Figure 1 shows the mean NRMS amplitude pattern of the 24 trunk muscle sites for RASYM in normal and LASYM in maximum reach.

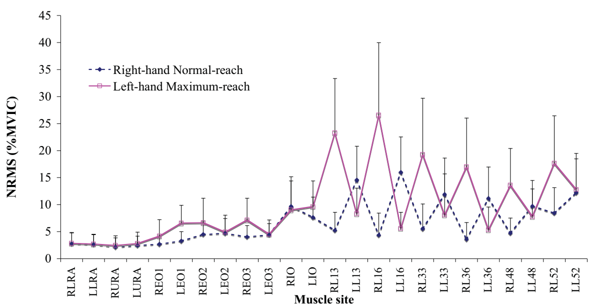

Figure 1: Mean NRMS amplitude patterns.

Using the pattern recognition technique, 3-principal patterns were found to explain 94% of the variance in the measured amplitude patterns (Figure 2). The results from the statistical analysis revealed that there were significant differences ($p<0.05$) for reach in PP1 and significant reach-by-hand interactions for PP2 and PP3.

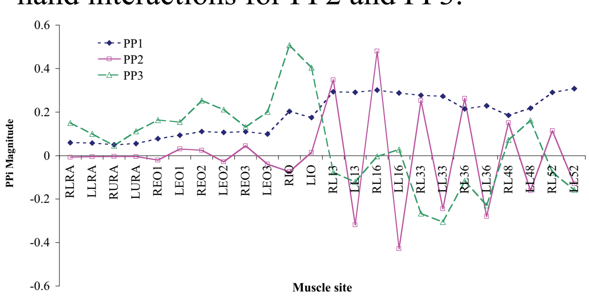

Figure 2: Principal patterns.

Overall, PP1 captured the majority of the variation in the data with higher scores associated with maximum reach. However, PP2 and PP3 captured subtle changes in amplitude recruitment strategies, which depended on both reach and hand.

SUMMARY/CONCLUSIONS

Pattern recognition techniques quantified the amplitude pattern of 24-trunk muscle sites with 3 numbers, which described the specific neuromuscular control strategies used for a one-handed asymmetrical task. In particular, PP1 characterized the general shape and amplitude differences for work demands (reach). However, the asymmetrical shape of PP2 clearly characterized the amplitude difference between the back sites on different sides of the body. PP2 also captured differential recruitment for the back (L1 and L3) and abdominal (among EO) sites, which was featured during conditions with greater biomechanical demands (maximum reach), while PP3 captured co-activation between the IO and back muscle sites (bracing) during the lighter demands (normal reach).

While the results showed a statistical effect for handedness for PP3, clinical relevance was not demonstrated since similar activation amplitudes (<1% MVIC) for the agonist and antagonist muscles were observed for the right and left handed lifts.

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LOAD ON THE LUMBAR SPINE FOR FLIGHT ATTENDANTS WHEN HANDLING TROLLEYS ABOARD AIRCRAFT

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INTRODUCTION

Flight attendants report on high physical load and complaints, particularly focussing on the lower back. These findings are mainly ascribed to the handling of trolleys – containing meal and beverage items – during the flight phases combined with an inclined cabin floor. Increasingly in the recent years, passengers' services are extended into the ascent and descent phases on short and medium-distance flights. For these service operations, more or less laden trolleys are moved along the aisles of the aircraft. Previous studies in this context were performed to quantify external-load indicators like the forces applied to transport carts, to the handle height of the handled trolley, to the moving speed or to the floor properties (e.g. Winkel 1983, Lee et al. 1991, Al-Eisawi et al. 1999, Laursen and Schibye 2002). In other investigations, knowledge on lumbar load is restricted on the handling activities of carts or two-wheeled containers on horizontal surfaces (e.g. Hoozemans et al. 2004).

According to the mainly involved body area, the provided part of an interdisciplinary experimental study – followed by detailed biomechanical model calculations – aims to quantify the load on the lumbar spine, to estimate lumbar overload risk, to identify disadvantageous task conditions and to derive biomechanically justified hints for work design in order to prevent low-back overload for flight attendants.

METHODS

Task conditions were recorded aboard aircraft, subjective perception of musculoskeletal load and complaints were examined via questionnaires (592 flight attendants), the working capability and the biometric data of larger samples of German flight attendants were recorded, and typical push-or-pull manoeuvres considering various floor-gradient angles, trolley types and weights were investigated in a specific laboratory set-up with an adjustable walkway.

The musculoskeletal loads from moving trolleys on planes were studied by observation of trolley handling aboard aircraft and subsequent video analyses in the laboratory. Approximately 150 to 250 trolley manoeuvres in the aisle can be supposed for a typical working shift, based on the analyses of the on-flight observations on typical task conditions, such as frequency and performance properties during trolley handling performed by a total of 15 female flight attendants on 10 flights in different types of aircraft.

Regarding postures and exerted forces, comprehensive three-dimensional measurements were performed in a specifically established laboratory set-up. The subjects wore a measuring system ('CUELA': see Ellegast and Kupfer 2000) containing several goniometers and inclinometers enabling continuous ambulatory postural data recording of 26 indicators. The forces at the trolley were measured via bars equipped each with three-axial force sensors at both ends. The recorded data served as input measures at subsequent three-dimensional biomechanical model calculations ('THE DORTMUNDER': see Jäger et al. 2001) for the prediction of several lumbar-load indicators for 458 manoeuvres performed by totally 25 selected flight attendants (22 female, 3 male) recruited from 5 German airlines. Airlines' compilations were examined regarding the biometric data of 2,347 flight attendants, and the physical strength of 510 persons was measured via the maximal isometric force-production capability to guarantee a representative sub-sample for the laboratory experiments.

RESULTS AND DISCUSSION

The laboratory measurements regarding posture and action forces during trolley handling and the subsequent biomechanical computations reveal that the load on the lumbar spine varies according to the handling mode (pushing, pulling), to the grade of floor inclination (0°, 2°, 5°, 8°), to the trolley type (half-, full-size), to trolley loading (empty, medium, full) and to the individual execution technique. For each of the 48 task configurations, lumbar load was evaluated with respect to potential biomechanical overload via work-design recommendations for disc compression and moment. Irrespective of floor gradient, trolley mass and individual performance, pushing of small trolleys is combined with 'acceptable' lumbar load, whereas pulling lead to 'critical' load. The latter is ascribed to the considerable amplitudes of the vertical hand-force component superimposing the horizontal force component necessary for motion; in some cases the upward-directed force, applied to avoid tilting of the half-size trolley, was about the force in movement direction. Moving the large trolleys occasionally lead to a 'critical' lumbar load, in particular, when heavy containers are handled on a considerably inclined floor.

CONCLUSIONS

Paired comparisons of moving actions resulting in extreme lumbar-load values, i.e. to a minimum or maximum despite of identical task configurations, demonstrated that top-edge grasp positions should be avoided for pulling of the 'short' half-size trolleys in order to diminish the biomechanical overload risk for the flight attendants relevantly. By contrast, for all the other cases – i.e. pushing small containers and pulling or pushing the 'large' full-size trolleys –, grasping at the upper edge of the trolley is recommended.

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POSTURE ANALYSIS FOR SURGEONS DURING MINIMAL INVASIVE OPERATIONS IN UROLOGY

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INTRODUCTION

Surgical operations are performed to an increasing extent using minimal invasive operation methods. In such operations, the visual inspection of the operation area is performed via endoscopes introduced into the body via small orifices. The application of minimal invasive techniques has the advantage of relatively small injuries to be sustained by the patients; the surgeons, however, have often to adopt specific and - in part - disadvantageous postures when performing the operations.

For urological operations in the bladder region a so-called resectoscope is used consisting of a rod-shaped endoscope and a wire loop mounted with the endoscope in the same shaft. For the dissection of tissue from the bladder or prostate the wire loop is charged with a high-frequency current and moved through the tissue. The instrument is normally held in the left hand and the right hand is used to manipulate the wire loop. Visual control of the operating area is possible via “direct endoscopy” or “monitor endoscopy”. In the direct method the eye of the surgeon is permanently in contact with the aperture of the endoscope. In the monitor method a video camera is mounted on the tip of the endoscope and the observation of the operation field is achieved using a monitor. The application of both methods requires different postures of the surgeons: During direct endoscope a close coupling between the endoscope and the eye is needed and the head, the hands and the endoscope of the surgeon form a relatively “rigid unit”. In consequence, especially during removal of tissue from the ventral part of the bladder or prostate, the surgeon has to incline the trunk and head steeply and to maintain this posture for a long period of time.

The aim of this study was to analyse the posture of the surgeons quantitatively during the execution of urological operations with direct and monitor endoscopy.

METHODS

The quantitative analysis described in the study on hand is based on video recordings performed in the operation theatre and subsequent posture classifications in the laboratory. Analyses were carried out for 10 operations with direct endoscopy and for 9 operations with monitor endoscopy; the analysed operation periods lasted between 15 and 79 min (mean 40 min). Posture analysis was performed by visual inspection of the videos and the application of an encoding procedure posture. In the code the angular positions of the various body segments were described using classification methods derived from literature (OWAS: Stoffert 1985, ISO 11226) as well as a newly developed special classification procedure. The development of a new classification schema was necessary, since the considered body segments and the differentiation between various angular positions in the methods from literature were proven as insufficient.

The new classification system is based on a method developed by Jäger et al. (2000) for the posture evaluation of manual handling tasks. The procedure was adapted to the situations in this area and a 26-digit code was applied with one digit each for the segmental position of all relevant body parts. For example, the first 3 digits of the code were used for the description of the trunk position with respect to the sagittal and lateral inclination as well as the torsion; for these movements the classification was performed in steps of 20°. Similarly the positions of the head, shoulders, upper and lower arm, hands were encoded.

The encoding was executed by visual inspection of the videos recordings by the same person. The advantage of the off-line encoding procedure was the possibility of multiple inspections of the recordings and a stepwise rating of the posture.

RESULTS AND DISCUSSION

Comparison of the results for the postures adopted by the surgeons during the application of both operation methods demonstrate that the portion of time with distinct inclination of the trunk and the head in forward and lateral directions and with raised shoulders and upper arms are higher during direct endoscopy than during monitor endoscopy. In figure 1 an example is presented for the forward inclination of the head. Maximum time portions are found for the inclination range between 0° and 10° backward in monitor endoscopy and for up to 20° in the forward direction for direct endoscopy.

Furthermore during monitor endoscopy a chair equipped with a back support and armrests was applied. The support for the back and arms was used for about a quarter of the operation time. In direct endoscopy the use of arm rests hampers, since a free movement of the trunk and arm is needed to guarantee the permanent contact between endoscope, eye and hands.

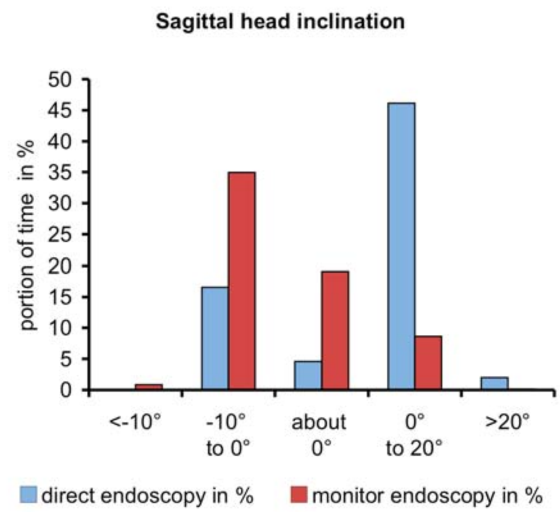

Figure 1: Forward inclination of the head

CONCLUSION

It is concluded that postural load is diminished during monitor endoscopy in comparison to direct endoscopy. This result confirms previous electromyographical findings regarding muscular strain and fatigue during endoscopic surgical work (Luttmann et al. 1996 a,b).

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ASSOCIATION BETWEEN MUSCULAR ACTIVITY AND UPPER LIMB POSTURE DURING LIFTING WITH HEIGHT AND WEIGHT VARIATION

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INTRODUCTION

Work-related upper limb disorders are common among several worker groups. Although workers performing Manual Material Handling (MMH) are also included in those groups, very little is known about upper limb biomechanics during MMH.

In order to contribute with the understanding on upper limb and MMH, the objective of this study was to evaluate the association between the activity of deltoid muscle and the movement of shoulder abduction, as well as biceps activity and elbow/shoulder flexion during lifting, attempting to weight and height variation.

METHODS

Thirty-two healthy subjects gave their informed consent to participate. Their mean age; height; and weight was 22.6 (±4.3) years; 1.69 (±0.04) m; and 68.3 (±8.7) kg, respectively. EMG and video were recorded during four lifting activities: a box weighting 7 and 15kg was handled from a low shelf (62.5cm) to an intermediate one (102.5cm) and from the intermediate to a high shelf (142.5cm). Surface EMG was recorded from right biceps (BI) and right deltoid (DE) muscles. Active single differential surface electrodes (Model #DE-2.1, DelSys®) were attached to the skin using a double-sided interface (DelSys®). An adhesive reference pad with diameter of 5 cm was placed on the right wrist.

Electrodes positioning and attachment were defined according to SENIAM. During the experiment the subject's movement in the sagittal (right view) and the frontal plane (posterior view) was recorded at 50Hz using two digital cameras (GR-DV 1800, JVC). Passive markers were used to reconstruct right shoulder abduction (SA), and shoulder SF) and elbow (EF) flexion (Fig. 1).

Kinematics data were digitalized, low-pass filtered at 5Hz, and reconstructed using Ariel Performance Analysis System (APAS®, Ariel Dynamics) software. The reconstruction of real coordinates was done using direct linear transformation (DLT). EMG data were processed using MatLab® (7.0.1, MathWorks). All signals were band-pass filtered using a 4th order and zero-lag Butterworth filter at 20-400 Hz, and then full wave rectified and low-pass filtered at 5 Hz to obtain the linear envelop. At the end of the processing both integrated kinematics and EMG were calculated.

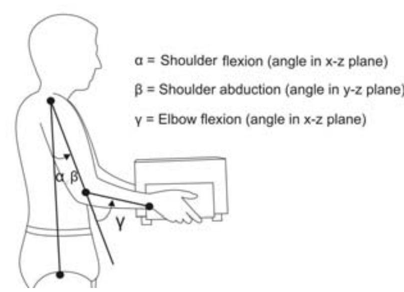

Figure 1: Markers position and angular reconstruction.

The duration of the liftings was compared by 1-way ANOVA. Two-way ANOVA was applied to compare lifting conditions. The association between integrated EMG and posture was calculated using Pearson Product-Moment Correlation. An alpha level of 0.05 was set for all statistical tests, which were performed using STATISTICA software (5.5, StatSoft).

RESULTS AND DISCUSSION

There were no differences on the duration of lifting, thus the identified differences are not due task duration. BI and DE activity changed ($p<0.05$) according to weight and height variation. The box weight (heavy) and the shelf height (high) increased EMG for both muscles. Habes et al (1985) and Nielsen et al (1998) also found similar data regarding height variation. SA and SF had significant change only due height. Lifting to high surface was associated with higher ROM. On the other hand, EF significantly changed only due weight variation.

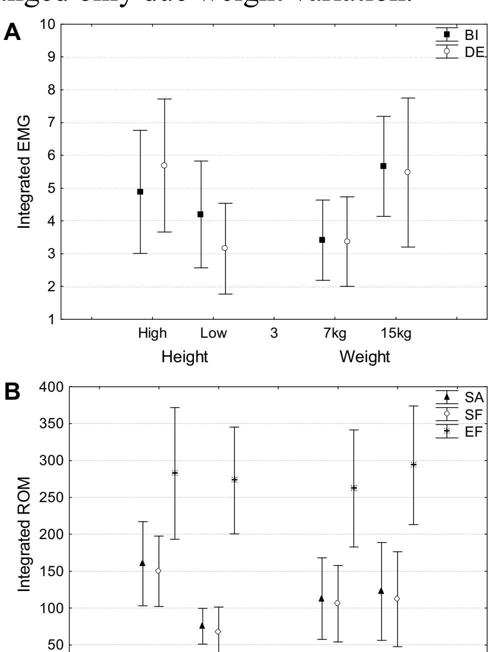

Figure 2: Mean (± 1 SD) for EMG (A) and posture (B) for each weight and height lifting condition. BI=biceps; DE=deltoid; SA=shoulder abduction; SF=shoulder flexion; EF=elbow flexion.

Table 1 presents *p* and r values for the correlation test applied for each condition. There was weak/moderate correlation for 9 of the 12 comparisons of EMG and upper limb postures analyzed. The correlation was not constant for all conditions, except for the association of BI and EF. For this test, moderate correlation was identified for all conditions. This result can be explained by the fact that BI is the main muscle responsible for EF. Furthermore the variation of EF across conditions was very small (Fig. 2). This result was not observed for SA and SF. Lifting height had a significant effect on these movements and it was important in determining correlation.

Table1: r and *p* values for each weight (W) and height (H) lifting condition.

W	H	DE x AS		BI x SF		BI x EF	
		r	*p*	r	*p*	r	*p*
7kg	HS	0,40	0,02*	0,39	0,03*	0,59	0,00*
	LS	0,57	0,00*	0,18	0,31	0,67	0,00*
15kg	HS	0,28	0,13	0,44	0,01*	0,64	0,00*
	LS	0,44	0,01*	0,27	0,14	0,63	0,00*

SUMMARY/CONCLUSIONS

Only weak/moderate association was identified between upper limb EMG and posture. A pattern across conditions was not observed, except for biceps and elbow posture. Upper limb seems to be influenced by task parameters, particularly height of surface. This fact should be considered when planning or assessing MMH activities.

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EFFECT OF LOAD CARRIAGE ON GROUND REACTION FORCES DURING OBSTACLE CROSSING: AN INVESTIGATION OF FIREFIGHTER AIR BOTTLES

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INTRODUCTION

Personnel such as firefighters and hazardous material workers often use a self-contained breathing apparatus (SCBA) that includes a face mask and shoulder pack with air bottle. Choice of pressurized air bottle is currently a controversial topic due to investment costs for lightweight, more compact expensive bottles vs. heavier, larger but less expensive designs. Prior studies have examined the effect on balance of wearing protective clothing with SCBA packs for firefighter (Punakallio, 2003) and haz-mat workers (Kincl, 2002). Others have looked at gait changes due varying carried load by military soldiers (Tilbury-Davis, 1999; Birrell, in press). However, no systematic assessment has investigated how modest changes in load carriage due to different bottle configurations (bottle weight and size) might affect gait behavior, especially when crossing obstacles. We hypothesized that reduced weight and size would improve gait performance. In this report, we focus on kinetic parameters based on ground reaction force (GRF) measurements while walking over level ground or a stationary obstacle.

METHODS

Four 30-minute air bottles were tested. An aluminum bottle (AL) representing current low-budget, heavy and large designs: 9.1kg, 53.3cm (L), 17.2cm (dia). A carbon fiber bottle (CF) representing current expensive, light and small designs: 5.4 kg, 47.0 cm (L), 14.0 cm (dia). To assess the effect of weight, a fiberglass bottle (FG) with similar size as CF but weight of AL was constructed: 9.1kg, 49.5cm (L), 14.0cm (dia). To assess center of mass location, a novel redesigned bottle (RD) was constructed to provide a light, short design that sat lower on the back: 5.4 kg, 31.8cm (L), 19.0cm (dia). Nineteen male firefighters (age 28±5 yrs) walked on a 9.8 m walkway with an embedded force plate (BP600900, AMTI) at either of two speeds ("normal, comfortable pace" or "as fast as can without running"). Three obstacle conditions were tested (no obstacle, 10 cm, or 30cm obstacle). Either obstacle, 10cm(W)×113cm(L), was placed such that the trailing foot contacted the force plate. Outcome parameters were averaged over two trials per condition. Each subject wore his own bunker coat, pants, and boots. Helmet and SCBA pack were provided. Five kinetic parameters were obtained from the vertical GRF for the trailing foot when crossing the obstacle: time to 1st peak, peak force in early and late stance, impulse in early and late stance. Early and late stance were defined as the breaking (heelstrike) and propulsion (toe-off) portions of the vertical GRF curve, respectively. Impulse was the integral of contact force with respect to time. GRF data were sampled at 1000 Hz. Multivariate analysis of variance tests examined whether bottle configuration, obstacle height, and walking speed affected the kinetic parameters.

RESULTS AND DISCUSSION

There were significant main effects on both early and late stance peak GRFs associated with bottle configuration ($p \leq 0.004$), obstacle height ($p<0.001$), and walking speed ($p< 0.001$), Table 1. GRF impulses were not influenced by the bottle. However, obstacle height and walking speed significantly affected impulse at early and late stance ($p \leq 0.007, p \leq 0.001$). No interaction effects were found except speed × obstacle for time to peak GRF ($p=0.039$) and peak GRF at late stance ($p=0.002$).

Since differences in peak GRF were found between the lightweight CF bottle and heavier AL and FG bottles, but not light RD bottle, these results suggest that only weight affects peak GRF. A recent study also found that peak vertical GRF increased with applied load during gait (Birrell, in press). Load length (or location) does not appear to affect peak GRF or impulse, although a trend appeared suggesting slower time to 1st peak GRF with a lower load (RD bottle).

All gait parameters except time to 1st peak increased with obstacle height. High peak and impulse values suggest that greater effort is necessary by the trailing leg as obstacle height increases. Firefighters took longer to reach the peak on no obstacle walking than obstacle crossing trials, although this finding failed to reach statistical significance.

SUMMARY/CONCLUSIONS

Load configuration, obstacle height, and gait speed affected peak GRFs. More specifically increasing load weight increased peak GRF, and increasing obstacle height and gait speed also increased peak GRFs and GRF impulse. Lowering load location with the RD bottle did not affect peak GRF and GRF impulse significantly, but might affect time to peak force. Future studies include kinematic assessments of the experimental data.

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Table 1. Gait parameters mean ± SD. Superscript denotes significant difference from indicated ($p<0.05$).

		Peak GRF time (ms)	Early stance peak GRF (%BW)	Late stance peak GRF (%BW)	Early stance impulse (%BW·s)	Late stance impulse (%BW·s)
Bottle	CF (A)	156±5	1.59 ±0.04^{CD}	1.53±0.03^{CD}	0.38±0.01	0.36±0.01
	RD (B)	173±12	1.60±0.04^{C}	1.52±0.03^{CD}	0.41±0.03	0.41±0.03
	AL (C)	160±4	1.66 ±0.05^{AB}	1.59±0.03^{AB}	0.39±0.01	0.38±0.01
	FG (D)	163±4	1.64±0.05^{A}	1.60±0.03^{AB}	0.39±0.01	0.39±0.017
Obstacle	0 cm (A)	175±9	1.52±0.03^{BC}	1.43±0.03^{BC}	0.36±0.02^{C}	0.32±0.02^{C}
	10 cm (B)	157±4	1.64±0.04^{AC}	1.56±0.03^{AC}	0.39±0.01^{C}	0.37±0.01^{C}
	30 cm (C)	158±5	1.72±0.05^{AB}	1.68±0.04^{AB}	0.42±0.01^{AB}	0.46±0.01^{AB}
Spd	Normal (A)	196±7^{B}	1.48±0.03^{B}	1.50±0.03^{B}	0.42±0.02^{B}	0.46±0.02^{B}
	Fast (B)	130±4^{A}	1.77±0.06^{A}	1.62±0.04^{A}	0.36±0.01^{A}	0.31±0.01^{A}

FUNCTIONAL STRENGTH TESTING OF RADIAL AND ULNAR DEVIATION BEFORE AND AFTER CMC ARTHROPLASTY

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INTRODUCTION

Records of hand strength have proven valuable in diagnosis, in the assessment of surgical outcome and in rehabilitation. (Mathiowetz, 1985) Grip and pinch tests are often used in clinical and ergonomic tests (Young, 1989; Su, 1994; Voorbij, 2002; Massy-Westropp, 2004) but their use does not focus on many activities of daily living that require twisting strength while gripping, such as opening and closing a jar or pill bottle. A patented dynamometer was developed at Allegheny General Hospital (AGH) to quantify this twisting strength and was applied to study pathological and normal hand/wrist function. (Miller, 2005)

The AGH dynamometer includes five disks of different diameters that can be mounted on the torque sensor to quantify radial/ulnar deviation. The torque output recorded on the conditioner measures the functional strength of the subject. Application of the device showed that handedness, gender and disk size had significant effects, while direction did not. A test of repeatability proved that the test was reliable with normal subjects. (Miller, 2005)

Carpal-metacarpal (CMC) interposition arthroplasty is one means of restoring hand functionality after thumb CMC joint degeneration. The purpose of the present study was to assess the functional outcome and compare these outcomes to pinch and grip results.

METHODS

Seventy-two consecutive CMC patients in the practice of the clinical investigator (MEB) were tested with the AGH dynamometer, a pinch tester, and a grip strength tester. Testing was planned for four visits: one week pre-operatively and at 3-, 6- and 12-months post-operatively. As of 28 December 2007, 54 patients had reached the 12-month time period, but only ten had completed all four tests with all five disks. Of these 10, only six completed all four pinch and grip tests. The drop-out rate was high and attributable to a shift in residency requirements that caused patients to be missed at the 6 month time period when a clinical visit was not medically required.

All subjects were seated and held the elbow at 90º of flexion with the upper arm adducted. For the twisting tests, the wrist was pronated; for the pinch and grip, the wrist was at neutral in both pronation/ supination and radial/ulnar deviation.

Repeated measures ANOVA's were performed to test differences, working from the null hypothesis that there were no differences due to time of testing, disk size or direction of twisting. The missed tests could not be recovered, of course, so that the number of cases differed between the pinch/grip tests and the radial/ulnar deviation tests. Gender was not considered in these tests. For the disk tests, there were 2 males and for the pinch/grip tests, 1 male.

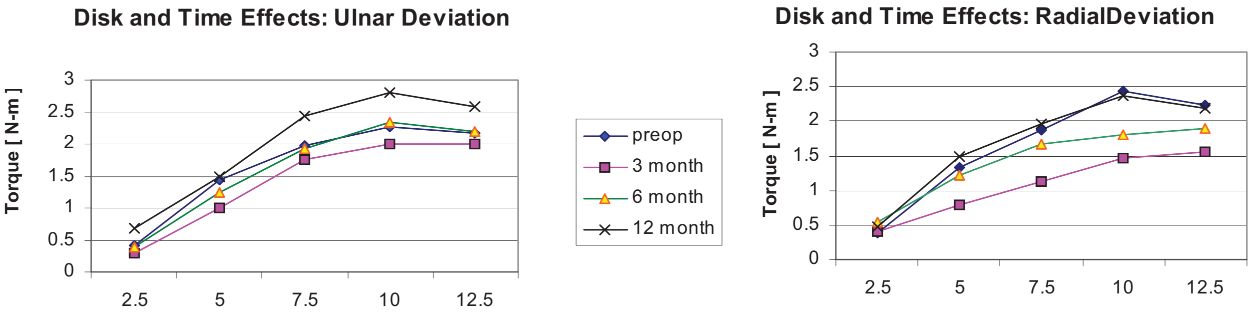

Figure 1: Torque Testing (n = 10)

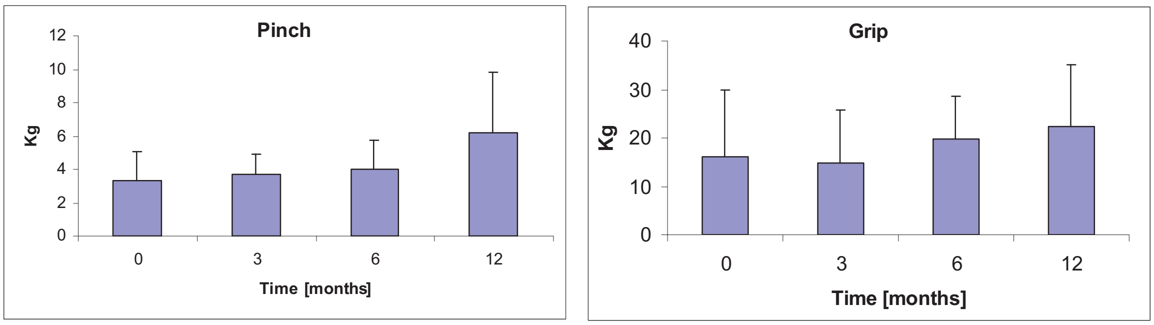

Figure 2: Pinch and Grip Results (n = 6)

RESULTS AND DISCUSSION

Figure 1 shows the results of the disk torque testing. Time of the testing ($p = 0.025$), disk size ($p = 0.006$) and direction of rotation ($p < 0.001$) were statistically significant. Tukey's Honest Significant Difference method found that only the three month time period differed from the others ($p = 0.017$) and that the 2.5 cm and 5 cm disk results differed from all others. Figure 2 shows pinch and grip testing results. No significant differences resulted from the pinch testing. In grip testing, the three month time period differed from all others ($p = 0.043$).

Post-operatively, twisting, pinch and grip returned to or exceeded pre-operative levels by 12 months. The significance of disk size and time in torque testing were expected but the increase in ulnar deviation 12 months post-operatively was not.

The number of subjects is small, but no means of recovering missed tests is possible because of the time effects. A new series of testing is underway to increase the sample size. The small sample size also inhibited testing of correlation between grip and twisting strength.

SUMMARY/CONCLUSIONS

Functional testing of twisting strength may show aspects of rehabilitation and recovery not found by pinch or grip testing. However, further testing may show that grip testing alone also correlates with recovery of radial/ ulnar deviation strength.

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INFLUENCE ON EMG ACTIVITY OF TREADMILL WALKING WEARING POSTURE CORRECTION WEAR AND LUMBAR SUPPORTER

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INTRODUCTION

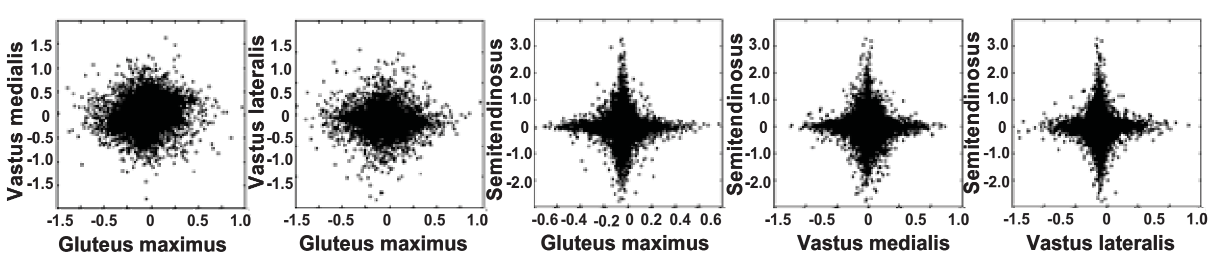
According to comprehensive survey by Japanese government, the patient of low back pain is increasing in Japan. Most of the patient has experience of orthotic treatment as a lumbar supporter which is effective on correction of excessive curve of spinal. However for the psychological relief by the wearing of lumbar supporter, many patients continue to use it after improvement of the symptom. There is fear that the habit to use the lumbar supporter can reduce the muscle which is necessary for ADL.
In this study, we pre-produced the posture correction wear which was expected to prevent the low back pain, and measured EMG activity during treadmill walking comparing that of commercial lumbar supporter.

METHODS

Healthy five Japanese male subjects were enrolled in Textile courses at Shinshu University. The averaged height and weight was 175cm, 66.8kg, and the averaged Body Mass Index (BMI) calculated as body weight divided by height squared was 21.8kg/m^2.
After put the surface electrode, subject wore either a commercial lumbar supporter (Sacro Wide-FX, ALCARE) or a posture correction wear which was sewed the high elastic fabric diagonally on the back of the body and walked for 10 minutes on treadmill (velocity was 5km/h, slope angle was 5 degree up). Then the EMG amplitude during Maximum Voluntary Contraction (MVC) was measured. Finally, sensory evaluation was carried out in semantic differential method.
The subjective muscles were Gluteus maximus (GM), Vastus medialis (VM), Vastus lateralis (VL), Semitendinosus (ST) at the left half of the body. EMG and ECG was measured by MP100 (Biopack). Sampling frequency was 1200Hz.

RESULTS AND DISCUSSION

From the sensory evaluation, both samples were assessed as 'feel stability slightly at trunk'. On the evaluation of the 'ease of walk', the posture correction wear was assessed as 'slightly easy'. But the lumbar supporter was assessed as 'neither'.



(a) GM ×VM (b) GM ×VL (c) GM ×ST (d) VM ×ST (e) VL ×ST

Figure 1: Lissajous diagram among the EMG amplitude of each muscle.

From the ECG analysis, physiological cost index (PCI) which indicates energy consumption in the walking exercise was calculated by ((heartbeat in walking) – (heartbeat in rest)) / velocity of walking). The averaged PCI in lumbar supporter and that in posture correction wear were 0.64 and 0.62 bpm/m /min. (i.e. the energy consumption in the walking exercise in posture correction wear was smaller).
Figure 1 shows the lissajous diagram among the EMG amplitude of each muscle.
EMG amplitude was not correlated in (a) GM and VL, (b) GM and VM (i.e. VL and VM were mainly related to the behavior of knee joint). On the other hand, the EMG amplitude were scattered around vertical and horizontal axis in (c) GM and ST, (d) VM and ST, (e) VL and ST. Therefore ST was an antagonist of GM, VM and VL (i.e., ST was related to both behavior of coxa and knee joint).
The oscillation which occurred at the heel contact was measured as trigger signal by oscillometer which was placed on the treadmill. By using the trigger signal, the EMG amplitude was divided by a walking cycle (i.e. one cycle was heel contact ➡ stance phase ➡ toe off ➡ swing phase ➡ heel contact). And %MVC was calculated as the Route Mean Square (RMS) of a walking cycle was divided by that of MVC as shown in Figure 2 (GM).
The EMG amplitude of GM increased over 20% of MVC when the beginning of stance phase and ending of swing phase. The EMG amplitude of VM, VL and ST also increased in this phase. Therefore Mean value and SD of %MVC was calculated on each muscle in this phase (See Table 1).
%MVC of GM was larger in the lumbar supporter than that in the posture correction were for all subjects. From the result, it was found that the lumbar supporter gives large constraint on behavior of coxa.
%MVC of ST, VM and VL were smaller in the lumbar supporter than that in the posture correction wear.

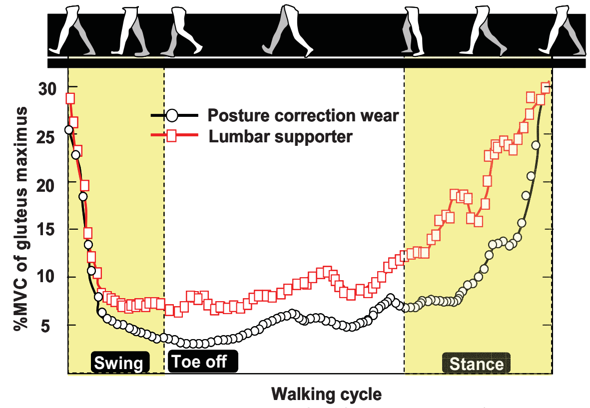


Figure 2: %MVC of Gluteus maximus during walking cycle.

Table 1: %MVC of each subject & muscle. PCW : Posture Correction Wear, LS: Lumbar supporter (unit : %)

Sub.	Gluteus maximus		Semitendinosus		Vastus medialis		Vastus lateralis	
	PCW	LS	PCW	LS	PCW	LS	PCW	LS
A	9	14	27	18	8	6	12	7
B	7	28	20	21	18	11	21	17
C	7	16	28	17	6	6	10	6
D	8	11	14	20	9	8	8	8
E	11	16	9	17	5	8	8	11
Mean	8	17	20	19	9	8	12	10
S.D.	2	6	8	2	5	2	5	4

From these result, it was supposed that the constraint of the lumbar supporter at coxa cause the decrease of ROM at knee joint. However the variances among individuals were large. Therefore we need more investigation at this point.

SUMMARY

The results are summarized as follows:
(1) Subjects feel no difference on the stability of trunk' among the lumbar supporter and the posture correction wear, but they feel easy to walk in the posture correction wear.
(2) PCI in lumbar supporter is slightly larger than that in posture correction wear. Namely the energy consumption in walking is smaller in posture correction wear.
(3) %MVC of GM is larger in the lumbar supporter. Namely the lumbar supporter gives large constraint on behavior of coxa.

MEASUREMENT OF FATIGUE FEELING DURING CAR DRIVING BY USING ELECTROMYOGRAM (EMG)

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INTRODUCTION

The purpose of this study is to construct an objective evaluation method of automobile drivability. The automobile drivability should be evaluated by both of the vehicle dynamics and physiological and psychological response of human for the dynamics. Driving the automobile with different vehicle handling and stability, physical stress can be estimated by measuring electromyogram (EMG) and the vehicle dynamics : speed, handle angle, handling torque, yaw rate. The actual drivability for human can be evaluated by comparing the vehicle dynamics with the muscular activity. Moreover, psychological stress can be estimated by EMG analysis.

In this paper, we studied the method used to assess the relation between driver's Bio-signal and mental load. The mental load state at the time of automobile operation is the result of continuous accumulating the stress to which cognition, judgment, and operation of a driver are caused repeatedly, and it is known that the influence will attain to even the muscles, circulatory and respiratory system.

METHODS

We implemented driving experiment using the driving simulator (DS) for one hour. The subjects executed the task of driving the track that meandered right and left for one hour with a driving simulator at a constant speed. We investigated whether the tiredness felt by driving with DS is able to be evaluated by using EMG. The subjects were healthy eight university students. The sensory inspection was executed every ten minutes on six factors of the mental work-load: difficulty, concentration, boredom, sleepiness, fatigue feeling, gratification. The EMG and the electrocardiogram (ECG) were measured as a physiological response. The EMG at sternocleidomastoid muscle (SCM), the upper part of the trapezius muscle(TM), and the front part of the deltoid muscle (DM) was measured by the bio-amplifier in the sampling frequency of 2000Hz. The trapezius muscle is directly related to the handling motion, and two another muscles doesn't relate directly to the driving action. It is well-known that the driver feels the fatigue feeling in the neck and the shoulder during car driving. The EMG of each muscle in maximal voluntary contraction (MVC) was measured before and after the driving task execution. And also, the EMG and ECG during the driving task were measured for one hour continuously. The ECG was recorded into the computer via the bio-amplifier in the sampling rate of 100Hz as a physiological activity indication of the mental load.

RESULTS AND DISCUSSION

Figure 1 shows the result of the sensory evaluation concerning the fatigue feeling factor. The fatigue feeling factor increased according to change of time. Other elements were the results similar to the fatigue factor. There was a significant difference in the tiredness feeling by driving in DS after beginning drive and one hour.

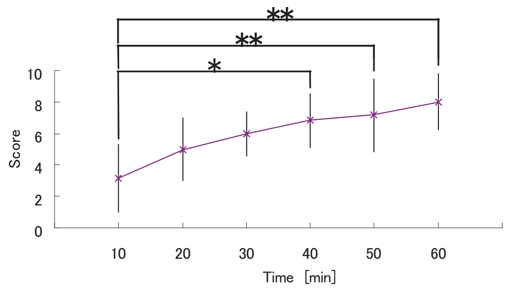
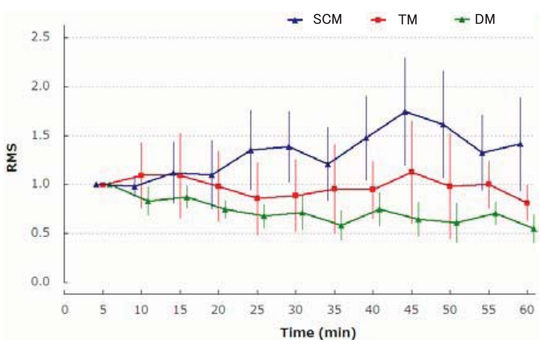
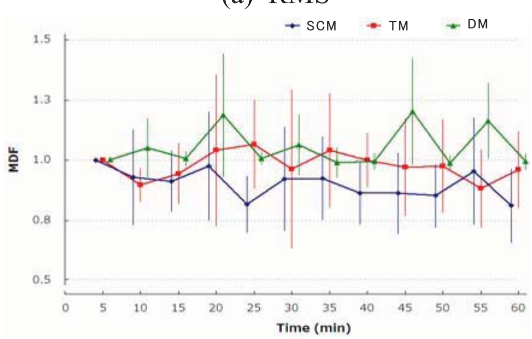


Figure 1 : Results of Sensory evaluation on fatigue feeling



(a) RMS



(b) MDF

Figure 2 : Results of RMS and MDF

We calculated R wave interval from the electrocardiogram, and obtained the variation of RRI as an index of a mental load. Since there was no significant difference among the variation of RRI according to change of time, we guessed that the task of DS was not strong stress for the subjects.

Figure 2 shows root mean square and median power frequency (MDF) of surface EMG during operation of DS and indicates a relative value based on the first mean value of five minutes in the section. It is well known that median power frequency (MDF) of surface EMG shifts to the low-frequency side and the muscle fiber conduction velocity (MFCV) decreases and the amplitude (RMS) of EMG increases with fatiguing contraction. In this experiment, SCM had a tendency of muscle fatigue.

Figure 3 shows the change rate of MCV measured before the task is executed and after it. SCM and TM had a tendency of muscle fatigue. We obtained a possibility that the muscle fatigue of SCM and TM indicate fatigue feeling.

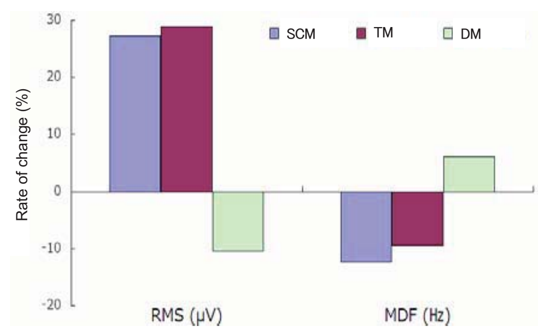


Figure 3 : Difference of MVC

CONCLUSIONS

In this study, we investigated the possibility to evaluate the fatigue felt by driving by the measurement of surface EMG of muscle that didn't relate directly to driving. The results are summarized as follows.

(1) The change of fatigue was able to be confirmed in the muscle activity of SCM, and "fatigue feeling" factor of the sensory evaluation and a high correlation were able to be obtained.

(2) The possibility that the evaluation index corresponding to the sensory evaluation was able to be made was obtained by measuring an unrelated muscle activity rather than the muscle activity related to the driving operation.

EVALUATING A CASE OF EPICONDYLITIS IN A CLEANER

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INTRODUCTION

In Spain work related musculoskeletal disorder (MSD) can be considered either an accident or a work related disease (WRD). Epicondylitis can be categorized as WRD only for certain working groups, that do not include explicitly cleaner, if they perform certain movements or postures that produce impact on the elbow or are associated to repeated prono-supination of the arm against resistance or if there are forced flexo-extension of the wrist. The case we present is a complicated one. She was diagnosed of radial epicondylitis and was categorized as WRD by the third party payer. As her progress was not good she underwent surgery. We saw her for the first time in our office because she was suffering from pain and she was going to send back to work. Clinically she had and epicondylitis and was under analgesic and anti-inflammatory medication. The objective of the study was to identify the tasks where she could be over demanding the extensor muscles of the wrist as to modify her work. Also, we thought that tasks that could place a risk for the epicondyle should be avoided for the rest of the workers. Then, proving that a risk existed needed objective data.

METHODS

Permission was obtained from the worker. We interviewed the supervisor and asked her about her impression of the problem and the task assigned to each worker. A questionnaire was put to the 5 workers in her working site and asked other workers to volunteer for the study. The identified tasks that could pose a risk in the epicondyle were: vacuum cleaning of the carpet, combing the carpet with the vacuum, washing the floor (carpet and marble) with a small mop. The patient thought that the worst task was the combing which they do because after vacuuming the carpet hair is oriented in several direction: ; It's when they reach with the vacuum, supporting the weight, when they feel pain in the epicondyle. We used DelSys SEMG hardware for the signal acquisition, and Delays software for the analysis of the signal and SPSS 12.0 for the statistical evaluation. We placed the surface electrodes in the extensor carpi ulnaris, extensor digitorum and third one in the extensor carpi radialis brevis as shown in the picture. After a rest recording, we asked them to perform a maximal voluntary isometric contraction, extending the wrist against resistance with the wrist in neutral position. Finally we asked them to perform the tasks while they were videotaped. We studied the patient and another worker, the only male in the group. We used the RMS of the MVC to normalize the infield studies. The following calculation were done: percentage of time that the muscle is activated, percentage of time that the muscle is activated a proportion of the MVC: 20%; 40%; 60%; 80%; 100%. MSD can occur as a result of the combination of force, repetition and posture. In our case, we hypothesize that sustained contraction can produce fatigue of the supporting tissues.

SUMMARY/CONCLUSIONS

All muscles are active over 99% of the time while performing all 4 tasks. We present here the results on the patient. The 90th percentile of rest voltage RMS was very similar in the three electrodes sites, between 6.8 E-06 and 7.2 E-06. MVC RMS peak was 0.00015 for ECU; 0.00017 for ECRB and 0.00070 for ED. It seems that ED gets more signal or has a more potent contraction performing this isometric contraction. Posterior cubitalis ECU is highly activated in all four tasks, though while carpet mopping the activation is less intense, table. Radialis ECRB is the next muscle with more sustained activation, highest in the carpet mopping. Finally ED has very little activation.

The proportion of time that a muscle is activated a percentage of the MVC indicates how much traction is the supporting tissue standing. Of course, a more potent muscle, with the same MVC proportion activation, will have a larger demand. Also, the demand of all muscles adds up, but we cannot create a summary variable. If ED is the most potent muscle, we could hypothesize that the whole demand is not large, however, the large sustained activation of the other two muscles sites makes us consider all these tasks as potentially risky.

We are aware of the technical limitations of this study. Ideally, we should compare the isometric MVC done in a posture with the infield contraction observed in this posture. To do so we are recording postures and activation with a 3 axis goniometer and doing MVC in the postures where more time and activation is found. This study is underway.

In summary, we think the cleaner work might pose a risk of epicondylitis even that the Spanish legislation does not explicitly recognizes so.

Proportion of time the selected muscles are activated as a percentage of the MVC

Activity	**Carpet vacuum**			**Carpet combing**			**Carpet mopping**			**Marble mopping**		
Muscle	**% time RMS is above the MVC stated %**			**% time RMS is above the MVC stated %**			**%% time RMS is above the MVC stated %**			**% time RMS is above the MVC stated %**		
	60%	40%	20%	60%	40%	20%	60%	40%	20%	60%	40%	20%
ECU	0.5	17.1	51	1.2	28.5	67.1	0.5	4.3	26.9	2.6	20.9	53.3
ED	0	0	0	0	0	0.6	0	0	0.9	0	0	0.7
ECRB	0.2	1.5	17.1	4.6	8.7	24.7	1.3	4.7	26.2	0	0.4	15.5

ECU extensor carpi ulnaris; ED extensor digitorum; ECRB extensor carpi radialis brevis

CONCURRENT TASK EFFECTS ON SHOULDER MUSCLE ACTIVITY

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INTRODUCTION

The shoulder is an often injured joint and is often difficult to rehabilitate. Working above chest or shoulder level or working with elbows raised is a risk factor for injury to the shoulder. While general guidelines are sound, subtle variations in superficial muscle activity indicate that certain additional tasks likely increase the loads on the rotator cuff musculature. These subtle changes may help us better understand injury mechanisms, improve rehabilitation and assist in return to work.

A series of shoulder and gripping studies have been completed. These studies have examined isometric and dynamic shoulder exertions in numerous postures under a variety of loading conditions including simultaneous gripping and mental tasks (Au & Keir, 2007; Antony, 2006; DiDomizio, 2006; MacDonell & Keir, 2005). The purpose of this communication is to highlight differences noted between protocols.

METHODS

Study 1: Arm elevation with grip
Participants performed static and dynamic shoulder flexion exertions with their right arm, elbow extended, and a neutral wrist posture while standing. Shoulder exertions were also completed with each of three different hand loading conditions: (i) no load in the hand, (ii) holding a grip dynamometer (0.5 kg, MIE Medical Research Ltd., Leeds UK) and (iii) exerting a grip force of 30 ± 3% of maximum (Antony, 2006).

Study 2: Push-pull with grip
The experimental protocol consisted of 25 10 s trials which included: (i) isolated hand grip at 15, 30 and 50% MVC, (ii) isolated 30 N push, (iii) isolated 30 N pull, (iv) 30 N push with 15% hand grip and (v) 30 N pull with 15% hand grip. All tasks were performed in three forearm postures (supination, neutral and pronation). Pushing and pulling tasks were repeated with a neutral forearm with the grip dynamometer rotated 180º to examine its effect on “assisting” grip force (DiDomizio, 2006).

In both studies, bipolar surface electromyography was collected from eight upper extremity muscles (AMT-8, Bortec Biomedical, AB, Canada). After preparing the site, disposable Ag-AgCl electrodes were placed over the muscle belly along the fibre direction with 3 cm spacing for muscles of the forearm, upper arm, and shoulder. Bias was determined from a quiet trial and removed from all EMG data prior to normalization to muscle specific MVEs. Muscle activity was analog linear enveloped at 3 Hz and, along with forces, moments and wrist angles were sampled at 100 Hz.

RESULTS AND DISCUSSION

When a 30 % grip force was added to a static (or dynamic) shoulder raise, a significant reduction in deltoid activity was seen, while infraspinatus activity increased. (Figure 1). These changes were dependent on the plane of activity (flexion or abduction). The findings were consistent during static and two speeds of dynamic arm raises.

When the grip effort was "coupled" to the shoulder effort with the push and pull tasks, the same effect was not seen (Figure 2). In the push task, the anterior deltoid activity remained at the same level regardless of grip condition. A similar effect was noted with the posterior deltoid with simultaneous grip during the pull task. Interestingly, the posterior deltoid had noticeable AEMG during the push tasks.

The findings from study 1 support our previous work on maximal and sub-maximal static shoulder exertions with gripping which also found a reduction in deltoid activity and increase in infraspinatus activity (Au & Keir, 2007; MacDonell & Keir, 2005). Those studies also found changes of similar magnitude and direction when a mental task (modified Stroop test) was combined with shoulder exertions.

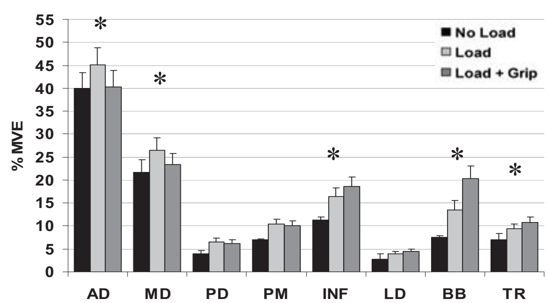

Figure 1. Mean AEMG (% MVE ± S.D.) for shoulder exertions at 90° flexion (n = 16). * indicates difference ($P < 0.002$) between load and load + grip conditions.

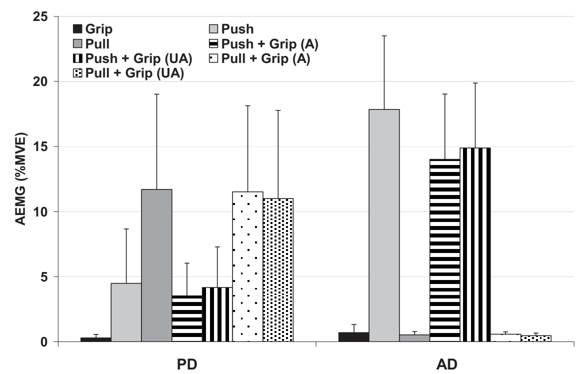

Figure 2. Mean AEMG (% MVE ± S.D.) for during push and pull tasks with and without grip (n = 12).

The intent of the push-pull tasks combined with gripping was to attempt to functionally couple the grip effort with the shoulder effort. However, the experimental design may have simply constrained the task such that deltoid activity was necessary for the task.

SUMMARY/CONCLUSIONS

While further work is required to elucidate the full effect, there appears muscle loading redistribution from the deltoids to the rotator cuff with concurrent gripping under certain conditions.

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ACKNOWLEDGEMENTS

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RISK OF MUSCULOSKELETAL DISORDERS ASSOCIATED TO THE DESIGN OF THE WORK STATION

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INTRODUCTION

Work related musculoskeletal injuries is the first cause of injuries at work in Spain. In 2003 it accounted for 32.4% of the total. From 1987 the INSHT performs “The National Survey on Work Condition” . In 2003 the fifth survey was performed: VENCT.
The objective of this study is to examine in the VENCT the association between musculoskeletal (MS) disorders and accident affecting the MS system with postures at work.
METHODS
There were 12.606.478 workers in Spain in 2003. The VENCT consisted in a stratified sample of 5.236. From the workers questionnaire we selected for this study the questions pertaining the experiencing frequently one of the following situations at work: lack of working space, chair very uncomfortable, having to reach for tools, work in area of difficult access for the hands, working surface unstable or irregular
Description of the distribution of the variables was performed before proceeding to produce contingency tables to examine the associations by means of the Odds Ratio and 95% confidence interval. Exploratory multivariate models were constructed.

RESULTS AND DISCUSSION
Only 18.7% of the workers interviewed perceived musculoskeletal risk at work. In the other side, almost 89% of the workers manifest musculoskeletal complaints attributed to postures or efforts performed at work.

Almost 64% of the workers consider their working site is correct. Of those that identify problems, lack of space is the most important one

Having a very uncomfortable chair, that occurs in 5% of the cases, is associated with a high risk of MSD complains, OR 7.10, Other circumstances of the job design as outreaching and working on areas of difficult access for the hands, lack of illumination or instable surface of work are also associated to higher risk of MSD complains

Having suffered a musculoskeletal accident the previous two years, among those that kept the same post these two years, is not associated to an uncomfortable chair, neither with lack of illumination, but it´s associated to lack of space, outreaching and having to work with the hands in areas of difficult access, and is borderline associated to instable surface,

The chair has become one of the most important working tools in the XXI century. In this large survey, it is shown that an uncomfortable chair is associated to MSD complaints but not to accidents as it is expected.. Accidents are associated to situation where the worker has to adopt forced or awkward postures as in outreaching. In this cases there is an association between force and postures that produces the MSD

ASSOCIATION BETWEEN MSD COMPLAINTS AND SITUATIONS AT WORK

WORK SITUATION	SITUATION AND MSD	SITUATION AND NO MSD	NO SITUATION AND MSD	NO SITUATION AND NO MSD RIESGO	ODDS RATIO	95% CONFIDENCE INTERVAL
LACK OF SPACE TO WORK COMFORTABLY	861	75	3291	1008	3.51	2.74-4.99
OUTREACHING	431	3722	40	1043	3.09	2.16-4.20
AREAS OF DIFFICULT ACCESS FOR THE HANDS	307	3486	42	1041	1.99	1.43-2.75
VERY INCONFORTABLE CHAIR	282	3870	11	1072	7.10	3.87-13.09
ILLUMINATION	406	3747	27	1056	4,21	2.81-6.32
IRREGULAR SURFACE	409	3744	45	1038	2.50	1.84-3.45

ASSOCIATION BETWEEN HAVING SUFFERED A MSD OCCUPATION ACCIDENT IN THE PREVIOUS TWO YEARS AND SITUATIONS AT WORK AMONG WORKERS THAT REMAINED IN THE SAME POST THE PREVIOUS TWO YEARS

WORK SITUATION	ACCIDENT AND SITUATION	ACCIDENT AND NO SITUATION	NO ACCIDENT AND SITUATION	NO ACCIDENT NO AND SITUATION	ODDS RATIO	95% CONFIDENCE INTERVAL
LACK OF SPACE	60	101	129	520	2.39	1.69-3.47
OUTREACHING	37	124	81	569	2.09	1.35-3.27
AREAS OF DIFFICULT ACCESS FOR THE HANDS	27	133	50	599	2.43	1.46-4.02
VERY INCONFORTABLE CHAIR	8	153	43	606	0.73	0.33-1.6
IRREGULAR SURFACE	31	130	91	559	1.46	0.93-2.30
ILLUMINATION	17	43	68	581	1.01	0.57-1.78

RISK OF MUSCULOSKELETAL DISORDERS ASSOCIATED TO WORKING POSTURE

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INTRODUCTION

Work related musculoskeletal injuries is the first cause of injuries at work in Spain. In 2003 it accounted for 32.4% of the total. From 1987 the INSHT performs “The National Survey on Work Condition” . In 2003 the fifth survey was performed: VENCT.
The objective of this study is to examine in the VENCT the association between musculoskeletal (MS) disorders and accident affecting the MS system with postures at work.

METHODS

There were 12.606.478 workers in Spain in 2003. The VENCT consisted in a stratified sample of 5.236. From the workers questionnaire we selected for this study the questions pertaining the main posture at work: it includes, standing, sitting, knees somehow bent, keeling, squatting and laying. Also questions about if the work implies to maintain a painful or tiring posture or a sustained posture. Description of the distribution of the variables was performed before proceeding to produce contingency tables to examine the associations by means of the Odds Ratio and 95% confidence interval. Exploratory multivariate models were constructed.

RESULTS AND DISCUSSION

Only 18.7% of the workers interviewed perceived musculoskeletal risk at work. In the other side, almost 89% of the workers manifest musculoskeletal complaints attributed to postures or efforts performed at work.

An awkward or painful posture has to be maintained more than half of the working day by 10% of the workers, while 33% has to maintain the same posture more than half of the working day. The most frequent working posture is the combination of standing and walking.

Having to work with the knees bent is associated to both MSD complaint and injury; also having to maintain and awkward or forced posture is heavily associated to both outcomes in a dose response relationship, however the maintenance of a posture is not associated to work accident. In the other side, a working postures that allows sitting and standing is associated to a lower risk of accident and MSD complaint

In this very large survey, it´s confirmed that musculoskeletal complaints are associated with postures. Also musculoskeletal occupational accidents are associated to these factors in a general similar pattern. There are important opportunities for prevention. For instance, 21% of the working population has to maintain awkward, fatiguing postures more than one quarter of the working day. This is associated to a four fold increase of the risk of a MS accident. An ergonomic design of the working station is a very important strategy to control MSD.

RELATIONSHIP BETWEEN COMPLAINTS OF MSD ATTRIBUTED TO POSTURES OR EFFORTS AT WORK AND WORKING POSTURES

FACTOR	MSD AND FACTOR	MSD AND NO FACTOR	NO MSD AND FACTOR	NO MSD AND NO FACTOR	ODDS RATIO	95% CONFIDENCE INTERVAL
Standing with knees bent a little bit	78	4065	7	1076	2.95	1.36-6.41
Sitting and standing often	1271	2872	413	670	0.71	0.62-0.82
Awkward tiring posture	504	2139	11	851	18.23	9.98-33.30
Maintain posture more than half of the working day vs nothing	212	2516	12	836	5.87	3.26-10.55

RELATIONSHIP BETWEEN THE OCCURANCE OF A OCCUPATIONAL ACCIDENT IN THE PREVIOUS TWO YEARS AND WORKING POSTURES, AT LEAST THE LAST TWO YEARS

FACTOR	MSD AND FACTOR	MSD AND NO FACTOR	NO MSD AND FACTOR	NO MSD AND NO FACTOR	ODDS RATIO	95% CONFIDENCE INTERVAL
Standing with knees bent a little bit	10	151	14	635	3.0	1.30-6.89
Sitting and standing often	18	142	169	480	0.36	0.21-0.60
Awkward tiring posture more than half of the working day vs nothing	46	46	470	2944	6.26	4,11-9.53
Maintain posture more than half of the working day vs nothing	48	49	1680	1637	0.995	0.64-1.43

Track 04

Mechanomyography and Muscle-Joint Mechanics (MM)

ARCHITECTURAL ANALYSIS AND INTRAOPERATIVE MEASUREMENTS DEMONSTRATE THE MULTIFIDUS' UNIQUE DESIGN FOR LUMBAR SPINE STABILITY

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INTRODUCTION

The posterior paraspinal muscles provide motion and dynamic stability to the multi-segmented, multi-articular spinal column. Spinal surgery that disrupts these muscles may lead to severe functional deficits or various pain syndromes. Paradoxically, surgery designed to treat these various spinal disorders actually disrupt these muscles, and in turn, may lead to significant functional deficits or various pain syndromes. The purpose of this study was to define the physiological operating range and architectural design of the multifidus muscle.

METHODS

Muscle architecture was determined according to the methods of Sacks and Roy (1982) as previously described by Lieber and colleagues (1990) for muscles of the upper extremity. Eight lumbar spines were harvested *en bloc*, stripped of superficial soft tissue and immersion fixed in 10% formalin for 72 hours. Cadaveric specimens were positioned in supine at the time of fixation to maintain a neutral lumbar spine position. After fixation, superficial lumbar fascia was excised and longissimus and iliocostalis lumborum muscles were reflected to reveal the multifidus muscle. After mapping locations for muscle fascicle harvesting, half of the muscle was dissected free of its bony attachments stored in saline to remove residual fixative. The remaining half was stored in PBS intact on the bony vertebral column from T11 to the sacrum. Multifidus muscle specimens were obtained from patients undergoing spinal surgery (n = 16). After skin incision, the dorsolumbar fascia was incised and the multifidus muscle identified by its position adjacent to the spinous process and the cranial/medial-to-caudal/lateral projection of its fibers. A small segment of the multifidus on the posterior, lateral region of the muscle belly was isolated by blunt dissection along natural fascicular planes and a specialized clamp was slipped over the bundle and the biopsy immediately placed in Formalin for fixation. Laser diffraction was then used to measure sarcomere length (Lieber *et al.* 1984). Intraoperative lumbar spine position was quantified and interpreted by measuring the intraoperative L1-S1 angle (Fig. 1B) and comparing it to the preoperative flexion, neutral, and extension lateral plain film radiographs (Fig. 1).

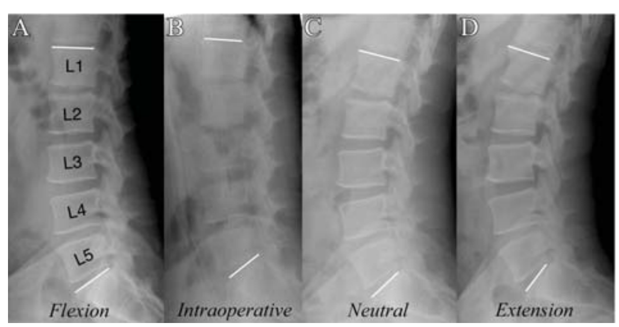

Figure 1: Radiographs of spine in (A) flexion , (C) neutral and (D) extension as

well as during intraoperative sarcomere length measurement (B).

RESULTS AND DISCUSSION

In contrast to the mass data that provided no unique insight into multifidus design relative to other lumbar muscles, physiological cross-sectional area (PCSA) was impressively large in the multifidus muscle (23.9 ± 3.0 cm^2). This value was over twice as large as any other muscle in the lumbar region ($p < 0.05$) in spite of the fact that it did not have as great a mass. In fact, the next closest muscle, the longissumus thoracis has a PCSA that is only about half of the multifidus (11.5 ± 1.1 cm^2) in spite of the fact that its mass is greater. This extremely large PCSA results from the very short multifidus fibers that are arranged along its length enabling very efficient packing of a large number of force generators into a relatively small volume

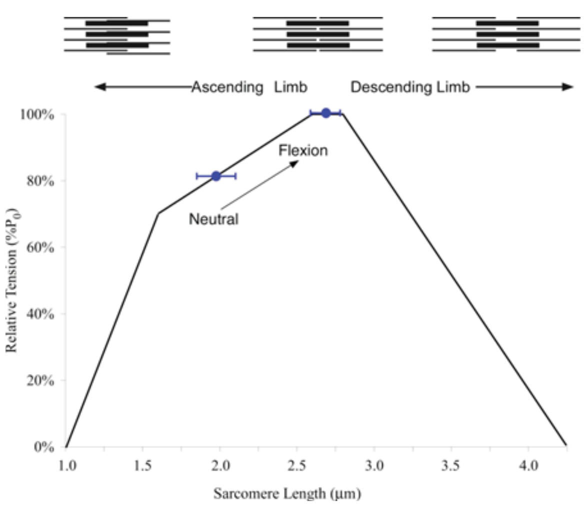

Figure 2: Sarcomere length operating range of the multifidus in terms of the human skeletal muscle sarcomere length tension curve (black line). Blue circles represent average sarcomere length obtained via biopsy in prone (neutral, n=8) or lumbar flexion (n= 5). These data indicate that the multifidus muscle operates on the ascending limb of the length tension curve and becomes intrinsically stronger as the spine in flexed (arrow).

The clamped muscle biopsies provided a unique opportunity to determine the *in vivo* sarcomere length operating range. In the neutral spine position sarcomere lengths were the shortest we have ever measured in human tissue (1.98 ± 0.15 µm), reinforcing the fact that the muscle operates on its ascending limb (Fig. 2). When the spine was flexed (41.4 ± 3.5°), significantly longer sarcomere lengths were observed (2.70 ± 0.11 µm; $p < 0.05$), as one would expect as the muscle was "lengthened." However, in the biomechanical context of the length-tension curve, the extent of the sarcomere length increase was small. Therefore, throughout the range of motion that could be achieved intraoperatively, the muscle operated exclusively on its ascending limb.

SUMMARY/CONCLUSIONS

This anatomical and intraoperative study clearly reveals the unique design of the multifidus to stabilize the spine based on its very high PCSA and operating range where sarcomere length increases on the ascending limb of the length-tension curve with flexion.

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ACKNOWLEDGEMENTS

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CHANGES IN LEG MUSCLE ACTIVATION DURING LOWER LIMB PERTURBATIONS

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INTRODUCTION

Voluntary and involuntary muscle force contributions are vital to the joint rotational stiffness (JRS). Even though the importance of reflexive muscle forces has been acknowledged, most studies concentrate on the voluntary forces (Moorhouse and Granata, 2007). Brown et al. (2003), Chiang and Potvin (2001) and Stokes et al. (2000) used sEMG as an analog to JRS and concluded that greater muscle activation, and presumably greater JRS. The purpose was to examine the responses of muscles crossing the before, and immediately after, the initiation of a sudden perturbation of the lower leg.

METHODS

Nineteen subjects participated in the study, (10 males: 26.4±2.8 years, 74.3±9.8 kg, 1.8±0.06 m; 9 females: 25.2±2.8 years, 62.9±7.8 kg, 1.7±0.06 m). sEMG from the biceps femoris (BF), semimembranosus (SM), medial gastrocnemius (MG), rectus femoris (RF), lateral gastrocnemius (LG), vastus lateralis (VL), vastus medialis (VM) of the right leg were recorded. Subjects were fitted with an ankle brace to resist ankle motion and positioned prone on a table where they maintained a right knee angle of 90° before the perturbation. A pneumatic perturbation (PERT) device forced each subject's right leg suddenly into extension via a rapid push force (78±2.2N) to the posterior of the ankle brace such that the forces acted directly through the ankle joint (minimize moment). Two PERT timing knowledge (TK) conditions were tested: 1) Known timing (KT) self-elected via control button, 2) Unknown timing (UT) delivered after a random duration between 1-15 sec). Three masses fastened to the top of the ankle brace to manipulate the Potential Energy (PE) of the lower leg/foot (0, 7.5 & 15% of leg mass). sEMG data (1024 Hz) were bandpassed filtered (2nd order Butterworth, 100-498 Hz), full wave rectified, normalized to MVE contraction amplitudes (AMP) and low pass filtered (2.5 Hz). A uni-axial goniometer measured the extensor rotation of the knee. The sEMG data was windowed into 4 time periods and the average AMP was calculated for each: 1) Baseline (BL) from 150-100 ms prior to the PERT, 2) Anticipatory (AN) from 15-0 ms prior to the PERT, 3) Reflex (RF) from 25-150 ms post PERT and 4) Peak (PK) as the highest level >150 ms post PERT. Peak sEMG AMP changes between successive time periods were calculated (AN-BL, RF-AN, PK-RF).

RESULTS AND DISCUSSION

Repeated measures ANOVA ($p<0.05$) revealed significance at each time period change. An interaction between KT and muscle was found during the AN-BL period. BF and SM showed increases in sEMG AMP for most muscles during KT but not UT.

A main effect of muscle was found in the RF-AN period. The BF showed greater activation differences, thus higher activation than all muscles except the SM, which showed a greater change in sEMG AMP than the LG, RF, VL & VM. An interaction effect was found between the PE condition and muscles during the PK-RF period. In each PE condition, BF had a greater sEMG AMP change than all 6 muscles and the same occurred for the SM (except BF). LG had greater sEMG AMP changes from MG, RF, VM and VL for PE/7.5% and PE/15%. MG had greater sEMG AMP changes than RF, VL & VM in each of the PE conditions.

Results suggest variation in muscle recruitment due to changes in PERT conditions for each time period. Prior to the PERT (as shown in AN-BL) the muscle activation depended on subjects' knowledge of the PERT timing. During KT conditions there was a greater change within the major knee flexors, leading to a speculated increase in JRS, such that subjects were able to prepare the knee joint for a disturbance.

Following the PERT, prioritization of muscle activation was found as BF and SM showed the greatest activation increase of all muscles. During this time period (which includes RF and AN) the change in activation may be attributed to the involuntary reflex twitches. It is shown through their involuntary increase in sEMG activation that BF and SM are preferentially activated, presumably due to their size and flexor moment arm, by the neuromuscular system to reduce the initial motion of the leg caused by the PERT. An interaction effect was found during the PK-RF time period change between the PE conditions and muscles. The increase in PE, decreased knee joint stability and thus activation in BF, SM, MG and LG increased. Overall as the PE increased, a greater disparity in activation between muscles was found, where the BF and SM had the greatest sEMG AMP difference in each condition.

SUMMARY/CONCLUSIONS

No indications were given to suggest KT increased the anticipatory activation levels of the muscles. Increases in PE did affect the activation levels of sEMG, seemingly by preparing the subjects for the PERT. BF and SM are presumed to have the greatest contribution to JRS, due to their high activity in each condition.

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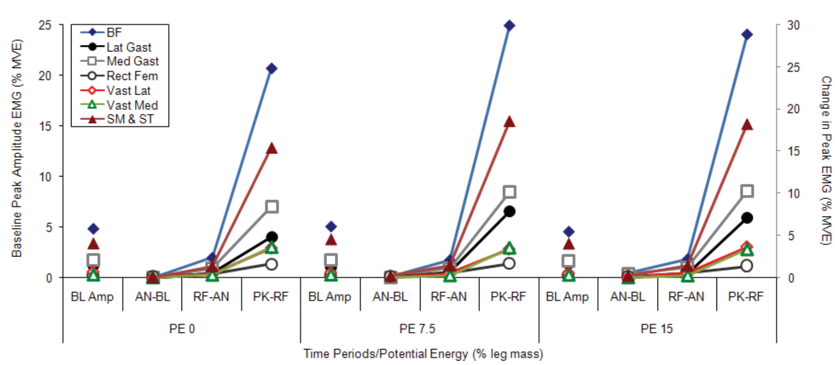

Figure 1. Peak sEMG AMP (%MVE) changes collapsed across timing knowledge for each muscle, over four time periods for each PE conditions along with BL AMP for reference (n=19).

FORCE AND MECHANOMYOGRAM POST-TETANIC POTENTIATION IN HUMAN TIBIALIS ANTERIOR

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INTRODUCTION

Post-tetanic potentiation (PTP) has been defined as the enhancement of the isometric twitch response that follows a tetanic contraction (Rassier and MacIntosh, 2000). It has been demonstrated that the potentiation phenomenon is typical of fast-twitch fibers.

For human Tibialis Anterior (TA), consisting of 73% fiber I type on the average (Johnson et al., 1973), it can be hypothesized only a slight enhancement in force twitch due to PTP.

Mechanomyogram (MMG) technique has been recently applied as an additional tool to assess muscle contractile function. MMG signal is due to the changes in the muscle geometry during contraction and monitors the acceleration (aMMG) and the displacement (dMMG) of the muscle surface depending on the transducer adopted for detection.

During muscle electrical stimulation, MMG has been widely used to study muscle mechanical properties and to follow their changes at fatigue. Scarce data are retrievable from literature regarding the application of MMG to investigate conditions of enhanced muscle function as PTP.

Aim of the study was to evaluate the changes in the mechanical responses (force, aMMG, dMMG) due to administration of a PTP protocol of stimulation in human TA muscle.

METHODS

According to the principles of the 1964 Helsinki Declaration on humans beings scientific research studies and after fully information about the aim and the experimental procedure, 8 healthy male subjects (age: 23-35 yo), with no history of neuromuscular or orthopedic problems, volunteered to participate in the study.

The leg of the subject was placed in a dynamometer with the ankle fixed at 120°. Force (FRC) produced by the dominant TA during transcutaneous monopolar supramaximal stimulation of the main motor point has been recorded by a load cell (Interface, SM-100N) strapped to the subject's foot. During the stimulated contractions MMG was detected over the muscle belly by means of an accelerometer (Entran, Egasy 25-D), while the surface muscle displacement was measured by a laser distance sensor (MEL, M5L/20) pointed closed to the accelerometer site. Before the potentiation started a series of 5 single twitches were delivered in basal conditions. The potentiation protocol (Fig.1) consisted of repeated cycles (5 s long) with an initial burst of 6 pulses at 100 Hz, 1 s of pause, 1 pulse (for single twitch peak estimation) and 4 s of pause. This stimulation pattern was suggested by Lee and Binder-Macleod (2000), except for the insertion of the single twitch (ST) in the cycle, which was needed for the online evaluation of the potentiation process cycle-by-cycle. When the ST amplitude increase levelled off, the stimulation stopped.

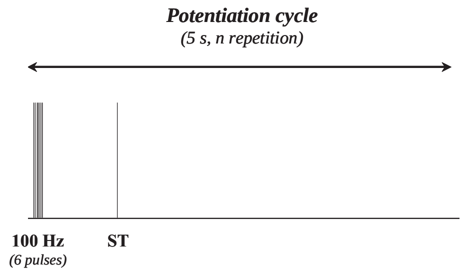

Figure 1. Potentiation protocol.

After conditioning the signals were stored (1024 Hz) on a PC.
The amplitude peak values of force (FRC-p) and muscle surface displacement (dMMG-p) together with the amplitude peak-to-peak of surface acceleration (aMMG-pp) were calculated from the ST of the last cycle of the potentiation procedure and compared to the average peak of the five basal STs.
For each signal amplitude parameters (pre- Vs. post-potentiation values), a paired t-test was conducted in the whole group.

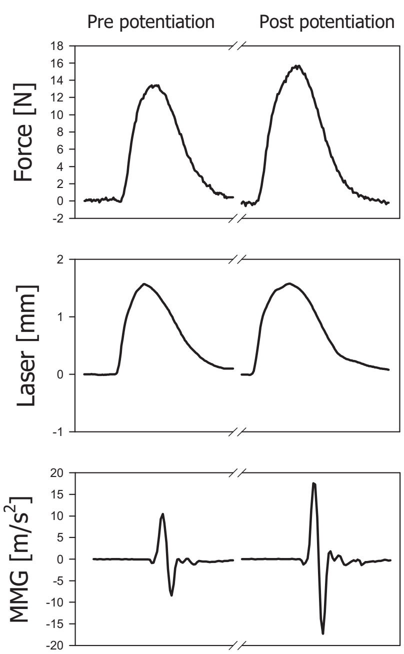

Figure 2. Mechanical signals for a representative subject: twitches before and after potentiation.

RESULTS

In figure 2 the FRC, aMMG and dMMG single twitch responses before and after potentiation are shown for a representative subject. It is evident that aMMG increases in amplitude much more than FRC and dMMG after potentiation.
The statistically significant average increase ($p<0.05$) were 7.9%, 9% and 42% for FRC-p, dMMG-p and aMMG-pp, respectively.

DISCUSSION

The potentiation protocol was aimed to gradually improve the twitch response until a plateau with constant maximum values of the elicited isometric twitch was reached.
TA is not a typical fast-twitch muscle, for this reason a not impressive potentiation in the force finds explanation, as hypothesized in Introduction. Accordingly, the amount of force level (and of dMMG of our data) is not proportional to the evident MMG-pp enhancement which could be explained by an improved rate of contractile activity development.
In general, the PTP phenomenon has been correlated to the increased phosphorylation of myosin regulatory light chains (RLC) which leads to potentiation of two mechanical properties: the maximal extent of isometric twitch tension and the rate of force development (Sweeney et al, 1993 and references within).
On these bases it could be suggested that, for the TA muscle, the adopted potentiation protocol influenced the phosphorylation of RLC predominantly enhancing the rate of tension development.

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EFFECTIVENESS OF AN EMG-ASSISTED MODEL TO QUANTIFY CO-CONTRACTION AT THE ANKLE DURING WALKING

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INTRODUCTION

The development of accurate, non-invasive methods to calculate individual force-time histories during movement across multiple joints is one of the greatest challenges in the study of human motion (White & Winter, 1993). In addition, the ability to precisely define muscle balance across a joint is of primary importance to clinicians who manage neurological or orthopedic conditions, such as cerebral palsy.

It was the purpose, therefore, of this study to introduce an EMG-assisted model to quantify co-contraction, i.e., muscle balance and imbalance among musculotendon units comprising the synergistic and antagonistic muscles involved in ankle plantar/dorsi flexion during normal gait, and assess the effectiveness by comparing it with a simpler model.

METHODS

Description of the mechanical response of the muscle model was based on previous work (Hill, 1938; Zajac, 1989), but incorporated individual muscle length, velocity, and excitation considerations. Processed EMG represented the neural input to the muscle. A musculoskeletal model defining joint kinematics, and line of action and architecture of the musculotendon units of the left lower limb, was developed by modifying a previously introduced model (Delp et al., 1990b). Muscle kinematics were calculated in conjunction with three dimensional cinematography. Individual muscle force as a function of length and level of excitation was also inquired as input to the model, and was established from a series of isokinetic calibration contractions and computer simulations. Co-contraction was measured as an index (CCI) using:

$$\frac{\sum F^{M}_{Total}}{\sum F^{M}_{Agonists}} - 1 = CCI \qquad (1)$$

where $\sum F^{M}_{Total}$ is the total muscle force at the joint, and $\sum F^{M}_{Agonists}$ is the total muscle force of the agonist muscle groups at the joint. The output of the model was validated using the model-predicted and inverse dynamics-measured joint moments. The value of the complexity of the model was assessed by comparing the co-contraction index output to the co-contraction index produced by using the normalized EMGs from the same muscles used in the EMG-assisted model.

Four subjects underwent instrumented gait analysis to compute joint kinematics and kinetics. Electromyography (EMG) was collected for the triceps surae, peroneous brevis, tibialis anterior and extensor digitorum. The EMG signals were processed and normalized using activity levels collected from maximum voluntary contractions performed during isokinetic calibration contractions.

RESULTS AND DISCUSSION

Results indicate that the moment curves, predicted and measured, matched closely in shape (see Figure 1 for an example). The correlation between moments derived from the two approaches ranged from 0.78 to 0.97 for the gait trials. The root mean square (RMS) difference between moment curves over one walking stride ranged between 5.3 N.m and 22.2 N.m. The results of this study were, in general, similar or better than those previously reported (White & Winter, 1993).

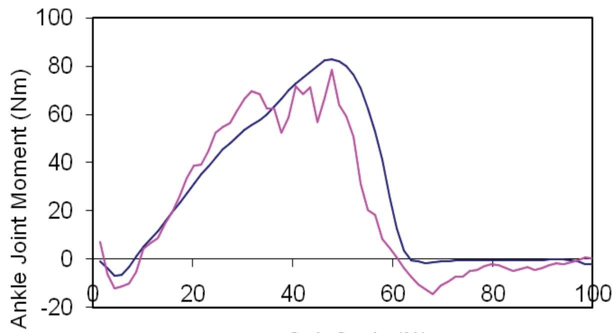

Figure 1. Comparison between the knee joint moment predicted by the EMG-driven model and the one measured from the inverse dynamics.

The timing of co-contraction is in agreement with previous studies (Falconer & Winter, 1985). However, to the best of our knowledge, no other studies in the past have presented the quantification of co-contraction at the ankle during walking gait by implementing the estimation of an index, which might allow comparison. Thus, from a functional perspective the results of the model suggest that co-contraction at the ankle during gait occurs at the time when stability is required. Furthermore, it appears that the EMG-assisted model maybe providing better information about the state of co-contraction state at the joint.

SUMMARY/CONCLUSIONS

While this study is limited by the number of

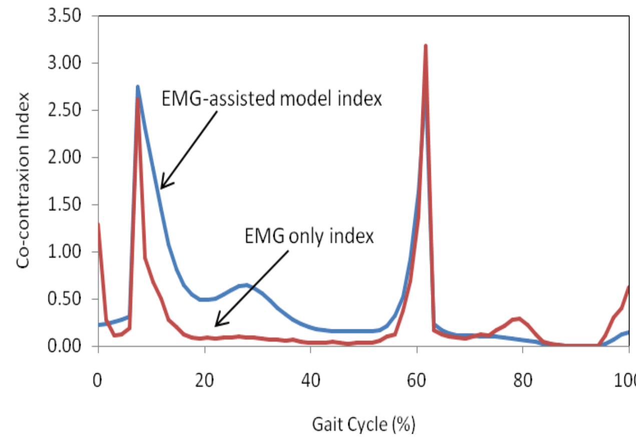

Figure 2. Co-contraction index at the knee during a complete gait cycle based on EMG-driven model the output.

subjects, the nature of the moment curve differences suggests that the present EMG-assisted model is essentially correct. However, there is room for improvement. For example, the temporal inconsistencies between the measured and predicted moments can be attributed to a variety of factors, such as the placement of the electrodes or the contribution of the passive structures at the joint.

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THE EFFECT OF PRECUES ON ATTENTION-DEMANDING PROCESSING IN WORKING MEMORY

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INTRODUCTION

The parameter precuing is a technique for studying motor programming. In this technique, partial or complete information about the response are presented in advance of a stimulus. The numerous experiments using this technique have shown that the precues reduce the reaction time (RT) (e.g., Anson et al. 2004; Eversheim & Bock, 2002), the mechanisms by which this reduction is achieved, however, remain disputed.

The effect of precue on different stages of sensorimotor processing was identified (e.g., Eversheim & Bock, 2002). However, according to cortical cell assembly theory of motor programming (Wickens et al.,1994) RT is a function of degree of overlap of muscle representations in the motor cortex, and in previous studies this factor was not controlled.

Therefore, the present study independently varied the number and nature of precued parameters, degree of overlap of cell assemblies, spatial extent of precues, and number of choices under 6 conditions, with the purpose of investigating the mechanisms of the effect of precue on RT in the stage of attention-demanding perceptual processing during foreperiod.

METHODS

Thirty-five (17 male, 18 female) non-athlete right-handed students of Islamic Azad University in age range 18-24 years selected randomly and participated in a within subjects design.

The task was isometric force production immediately after appearing target. They performed the task by parameter precuing apparatus under the following 5 precue and 1 control conditions:

1. Two -choice, large spatial extent, high overlap (known direction);
2. Two-choice, small spatial extent, high overlap (known direction);
3. Two -choice, small spatial extent, low overlap (unknown direction);
4. Three-choice, large spatial extent, high overlap (known direction);
5. Two -choice, small spatial extent, low overlap, mutually conditional combination of direction and force precues; and
6. Non-precue.

A Stroop task was added during foreperiod for interfering in the stage of attention-demanding perceptual processing. The participants repeated each condition 30 times with random order after performing 200 training trials.

RT of different conditions was measured and analyzed by 6 (condition) * 30 (repetition) ANOVA with repeated measures and Bonferroni post hoc test.

RESULTS AND DISCUSSION

The results indicated the significant main effect of condition ($F_{(3.9, 131.9)} = 57.3$, $p<.001$); but the main effect of repetition ($F_{(12.8, 435.5)}= 1.12$, $p= .34$) and interaction of 2 factors were not significant ($F_{(20.7, 705.7)}= .88$, $p= .62$).

According to results of post hoc tests, RT of the non-precue condition was significantly longer than other conditions (p<.01). RT of the 2-choice high spatial extent and overlap condition was significantly longer than other 2-choice conditions (p<.01). RT of the 2-choice small spatial extent high overlap condition was significantly shorter than the 2-choice low spatial extent and overlap with ambiguous precue and the 3-choice high spatial extent and overlap conditions (p<.01). But there were not significant differences between RT of the 2-choice high spatial extent and overlap, the 2-choice low spatial extent and overlap, and the 2-choice low spatial extent and overlap with ambiguous precue conditions (p>.05)(Fig.1).

Therefore, the effects of number of precued parameters, spatial extent of precues and degree of overlap of cell assemblies on RT were abolished by this interference, but the effect of number of choices remained significant.

SUMMARY/CONCLUSIONS

The results of this study indicated that the precues reduce RT of isometric force production task by involving of attention-demanding perceptual processes of sensorimotor processing. This finding was consistent with the results of Eversheim and Bock (2002).

Furthermore, the results related to the effect of degree of overlap of neuronal representations of muscles on RT were consistent with Anson et al. (2004) and supported the cortical cell assemblies theory (Wickens et al., 1994).

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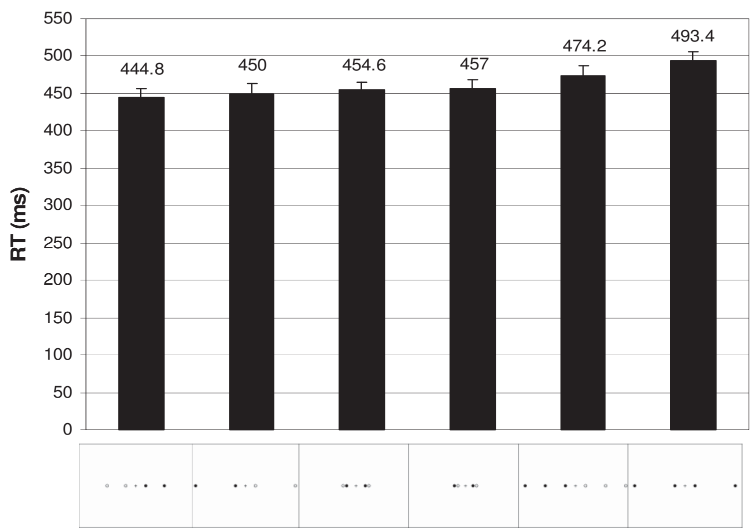

Figure 1: Mean and standard deviation of reaction time of different precue conditions.

MULTI-CHANNEL MECHANOMYOGRAPHY DURING NON FATIGUING AND FATIGUING CONTRACTIONS IN THREE HUMAN MUSCLES

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INTRODUCTION

Mechanomyography (MMG) originates from slow bulk movement of the muscle, oscillations created by the muscle at its own resonance frequency, and pressure waves due to muscle fiber dimensional changes (Orizio 1993). Surface MMG has been classically recorded with a single sensor located over the muscle belly in order to assess changes in muscle activity in various conditions, such as fatigue and pain. Multi-channel MMG has been recently used to assess muscle spatial reorganization following load changes and fatigue development (Cescon et al. 2004, Madeleine et al. 2006, 2007; Madeleine and Farina 2008).

The aim of the paper is to compare findings on spatial reorganization of MMG activity in the tibialis anterior, paraspinal and trapezius muscles during increasing force and fatiguing contractions.

METHODS

MMG signals were detected over the:
(i) dominant tibialis anterior muscle with 15 accelerometers on 11 healthy volunteers.
(ii) left and right paraspinal muscles with 12 accelerometers on 11 healthy volunteers.
(iii) dominant upper trapezius with 12 accelerometers on 12 healthy volunteers.

Isometric contractions with increasing load (randomized order): (i) 3-s long contractions at forces in the range 0-100 % MVC (10 % increment) for the tibialis anterior. (ii) 20-s long contractions with held loads in the range 0-15 kg (2.5 kg increments) for paraspinal muscles in a 20^{o} flexed position. (iii) 10-s long contraction at forces 10-20-40-60-80-100% MVC for the upper trapezius.

Fatiguing isometric contraction: (i) 6-min long contraction with 7.5 kg load for paraspinal muscles. (ii) 20 % MVC until task failure for the upper trapezius.

MMG signals were detected with accelerometers (ADXL202JE) connected to a multi-channel MMG amplifier (LISiN-Ottino Bioelettronica, Italy). Signals were sampled at 2048 Hz, converted in digital form by a 12 bit A/D converter, and stored on disk.
Maps of absolute and normalized average rectified value (ARV) and mean power frequency (MNF) were obtained from the two-dimensional MMG recordings. The centre of gravity and entropy (degree of homogeneity) were computed from the ARV and MNF maps.
Statistical analysis: ANOVA was applied to measure the effect of accelerometer location, force increase or contraction time on ARV and MNF values. $P < 0.05$ was considered significant.

RESULTS AND DISCUSSION

Effect of increasing force:

For the tibialis anterior, the absolute ARV did not depend on accelerometer location and the center of gravity of ARV did not change with force, while, normalized ARV did. Both absolute and normalized MNF depended on accelerometer location ($P < 0.05$) and the centre of gravity of MNF changed with contraction levels ($P < 0.001$). For both the paraspinal and upper trapezius muscles, the absolute ARV and MNF depended on accelerometer location ($P < 0.05$), while, normalized values did not.

Effect of fatigue development:
For both the paraspinal and upper trapezius muscles, the absolute ARV and MNF depended on accelerometer location ($P < 0.05$) during sustained contraction, while, normalized values did not. Moreover, lower ARV entropy values ($P < 0.05$) were found in the subjects with longer time to task failure (upper trapezius).

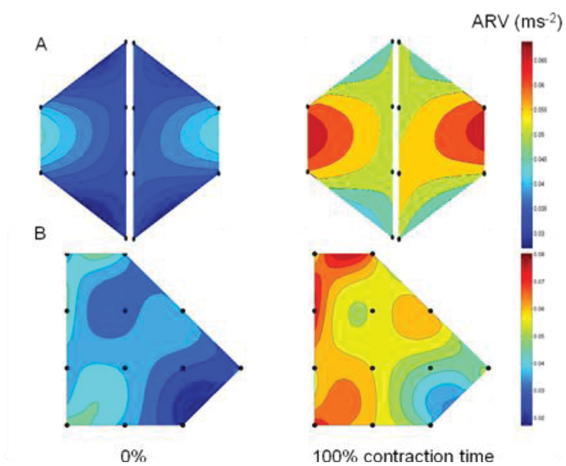

Figure 1: Changes in average rectified value (ARV) maps as a function of contraction time for paraspinal muscles (A) and upper trapezius (B).

The tibialis anterior was uniformly activated over the whole contraction range as no spatial changes were observed for ARVs. On the other hand, spatial dependency was found for the paraspinal and upper trapezius muscles. The muscle morphology and architecture most likely can explain the differences in MMG/force relationship. Multi-channel MMG assessment during sustained contraction demonstrated heterogeneous absolute activation of both paraspinal and upper trapezius muscles. Moreover, a functional role of spatial variations over time for the maintenance of force was found in the upper trapezius. Spatial changes were found in the frequency MMG maps highlighting the complexity of MMG spectral changes with increasing force and fatigue development. Furthermore, the present results confirmed that normalization procedure can mask changes during increasing force levels and sustained contractions.

SUMMARY/CONCLUSIONS

Two-dimensional arrays of accelerometers proved the importance of muscle morphology and architecture on the features of MMG signal. The technique may further contribute to a better understanding of the MMG signal.

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A METHOD FOR KINEMATIC AND KINETIC ANALYSIS OF UPPER EXTREMITY MOVEMENT

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INTRODUCTION

Measuring of 3-dimensional kinematic and kinetic data, parallel with surface EMG data is today a standard procedure in clinical gait analysis. Upper extremity movement is, in contrast to gait, not predefined, restricted, cyclic or repeatable. An additional problem is measuring of external forces, an important factor in detecting muscular coordination pattern, which are less defined and with lower magnitudes. Therefore, there is a need for a procedure that enables an objective analysis of unconstrained 3D upper extremity movement.

METHODS

Motion analysis system (Vicon 370) with 6 infrared cameras was used for acquiring kinetic data of upper extremity movement.

Figure 1: Measurement set-up

A rigid body model for the upper extremities was used to calculate the angles for the complete upper arm joint chain (hand, elbow, shoulder and sternoclavicular joint).In order to increase the reproducibility of the movement, a 5-DoF robot-arm presented a predefined 3D path that the subject should perform. The path can be adapted to the patient group and age. The 6-DoF force sensor was attached at the end of the robot-arm to acquire the external forces and torques in all three axes. To detect the muscular coordination, conventional surface EMG was used. The electrodes are placed on the muscles according to the SENIAM recommendations.
A visual force-feedback was used to help maintaining the desired force vector.

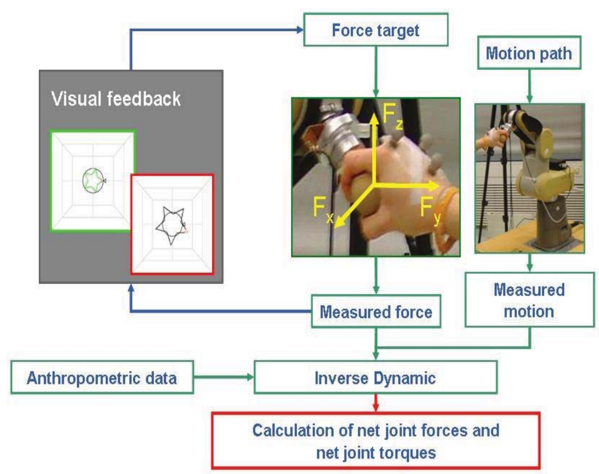

Figure 2: Concept

The position and magnitude of desired force can be predefined for each position and each axis. In this way, by choosing and controlling force direction and magnitude, a specific muscle group can be addressed and activated in different levels. For calculating kinetic data, anthropometric parameters and external forces were measured. Using those and kinematic data, forces and torques in each joint (hand, elbow, and shoulder) were calculated via inverse dynamic method.
As an example, a shoulder flexion was performed with 3 repetitions, with desired external force of 0N and 10N. Net joint forces for hand, elbow and shoulder joint were calculated. Muscular activation of brachioradialis, biceps brachii, deltoideus and trapezius was detected.

RESULTS AND DISCUSSION

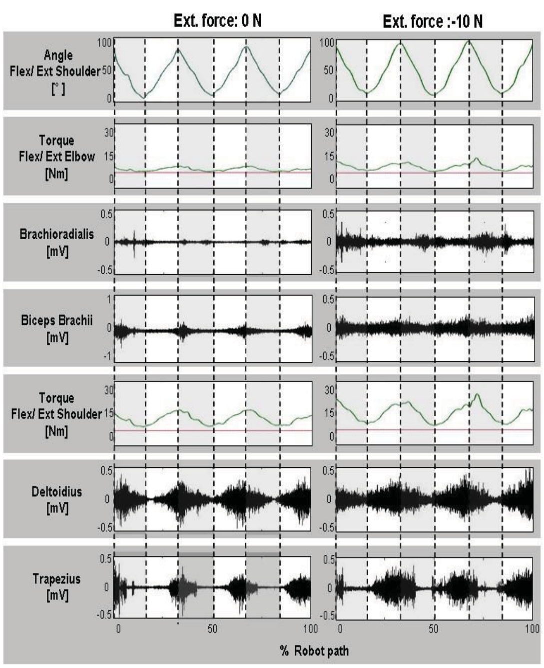

Figure 3: Angles, EMG and net joint torques for desired external force 0N (left) and 10N upwards (right).

The reproducibility of movement can be seen on Figure 2 which shows shoulder angle in flexion axis (first row) for two measurements. The net joint torques and muscle activation show the correlation between net joint torque and activated muscle group. The difference in amplitude of EMG signals between two measurements with different force levels confirms the assumption that choosing a force amplitude and direction influence the muscle activation.

SUMMARY/CONCLUSIONS

A procedure has for the analysis of kinematics, kinetics and muscular coordination of a free upper extremity movement has been developed. A force feedback has been used to improve the investigation of a muscular coordination pattern. This method can be used to evaluate operations, rehabilitation or treatment success.

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DEVELOPMENT OF DISPLACEMENT-MMG TRANSDUCER USING A PHOTO-REFLECTOR - DESIGN AND CHARACTERISTICS -

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INTRODUCTION

A piezoelectric accelerometer is generally used for the measurement of MMG (mechanomyogram). As body movement causes a large artifact, the body fields being measured must be fixed securely. When MMG measurement is applied to healthcare such as clinical medicine or sport science, it is necessary to measure the MMG during involuntary body movement or exercise (Silva et al. 2005). The measurement of displacement-MMG is useful to examine the characteristics of muscle contraction, because the continuous enlargement deformation of muscle as shown in tetanus can be observed by this method (Orizio et al. 2000). In this study, displacement-MMG transducer was newly developed, which was resistant to artifacts caused by involuntary movement and could be used to obtain measurements under exercise. Using this transducer, displacement-MMG of a single twitch following electro-stimulation was measured during exercise on an ergometer and while jogging on a treadmill.

METHODS

(1) Design of displacement-MMG transducer

The displacement-MMG transducer is composed of a small photo-reflector (LED and phototransistor (PTr), measuring 5.0x4.8x4.0mm and weighing 0.096g). Though the photo-reflector is originally a binary sensor, the authors utilized it as a displacement transducer, because the output of the transducer is almost proportional to the distance in the range of 7±2mm. The electronic circuits are composed of the driving circuit of LED, the amplifier and the filter for the PTr output, and they are mounted on the housing of the transducer. The effective range of irradiation from this photo-reflector was 7x4mm. The white circular seal, which measured 8mm in diameter and had a luster in order to avoid the influence of reflectivity of the skin surface, was adhered to the skin.

Fig.1 shows the design of displacement-MMG transducer. In this transducer, both legs of the housing are considered fixed points, and displacement of the central active point on the skin is calculated. The distance between the legs is 30mm, the transducer measures 37x17x23mm and weighs 4.8g including the sensor cables. The steady background noise is ±2μm and around 5μm for the external impact noise.

(2) Measurement of displacement-MMG at rest and during exercise

To confirm the effectiveness of the

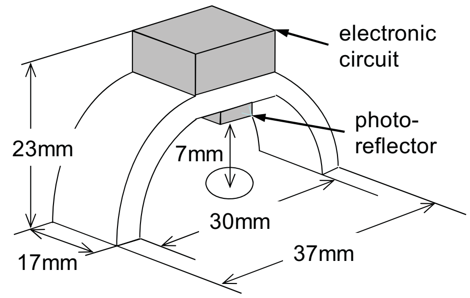

Figure 1: Design of displacement transducer using a photo-reflector.

transducer, the displacement-MMG in twitch contraction was measured when the femoral nerve was stimulated electrically for 1ms. The transducer was put on the musculus rectus femoris and MMG was measured at rest and during exercise (exercise on the ergometer and jogging on the treadmill). During measurement at rest, the hip joint and knee joint angles were fixed and displacement-MMG was also measured using the laser displacement sensor simultaneously. During exercise, the joint angle was measured using the goniometer and EMG was also measured to provide guidance for the electro-stimulation at the resting stage of muscle contraction.

RESULTS AND DISCUSSION

Fig.2 shows (a) electro-stimulation pulse, (b) EMG and (c) displacement-MMG of the musculus rectus femoris. The periodic MMG wave was found in the descending-phase of leg movement, based on periodic voluntary contraction of the muscle. However, twitch MMGs, indicated by open circles in the figure, were found during ascending-phase, corresponding to electro-stimulations at the resting stage of muscle contraction. Fig. 3(a) shows the twitch

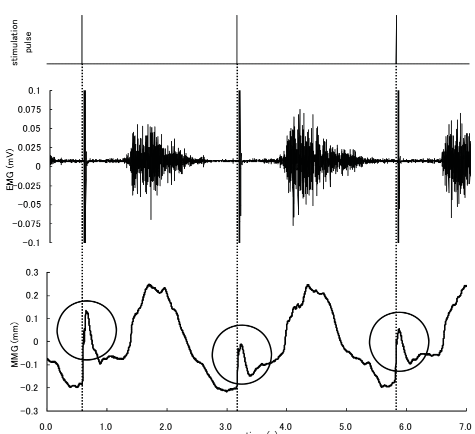

Figure 2: Electro-stimulation pulse, EMG and displacement-MMG in twitch contraction during exercise on an ergometer.

contraction MMG at rest, which was measured using the laser displacement sensor (dotted line) and this transducer (solid line). This displacement-MMG showed a wave that was almost similar to that but its magnitude was about one tenth of that, because it shows a difference in the central point and both legs. Fig.3(b) shows twitch contraction MMG during exercise on the ergometer (solid line) and while jogging on the treadmill (dotted line). Both MMGs were similar but there was a slight difference, because the knee joint angle was different in the two exercises.

SUMMARY/CONCLUSIONS

A displacement transducer using a small photo-reflector was developed. This transducer was resistant to artifacts caused by involuntary movement and it was possible to measure displacement-MMG under conditions such as exercise and jogging using this transducer.

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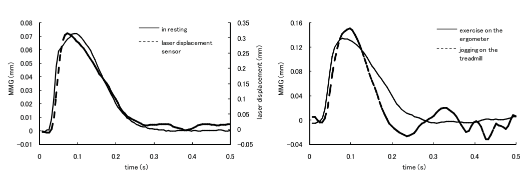

(a) at rest (b)exercise

Figure 3: Twitch contraction MMGs at rest and during exercise.

INCOMPLETE TETANUS PROGRESSION IN SKELETAL MUSCLE BASED ON DISPLACEMENT-MMG

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INTRODUCTION

A detached laser displacement sensor can measure displacement in the cross section direction of a muscle (displacement MMG) during tetanic contraction. Orizio et al. used a laser displacement sensor to measure displacement MMG while stimulating the gastrocnemius muscle of the cat electrically at 5～50Hz, and reported the mechanism of the skin surface deformation (Orizio et al 1999). However, these experiments were carried out only in the cat gastrocnemius muscle in which the type II fiber is predominant. Therefore, there are no reports that measured and discussed surface displacement due to tetanus in the skeletal muscle, which has a different component ratio of muscle fiber types.

In this study, we used laser displacement sensor to measure displacement MMG of the rat while continuously increasing from stimulation frequency 1～50Hz. Then, the tetanus process of the Vastus Intermedius muscle (VI) which is a mixed muscle composed of type I and type II fibers was examined after the gastrocnemius muscle (GC, mostly type II fiber) and Soleus muscle (SOL, mostly type I fiber).

METHODS

In the experiment, 230±40g rats were used for GC (n=8), VI (n=8) and SOL (n=6). Electric stimulation was performed using a negative pulse of 1ms duration. The pulse was given 2 times within 2 seconds for twitch, and a sweep of 1-50 Hz within 5 seconds to stimulate incomplete/complete tetanus, respectively (Figure 1). All motor units were ignited by the electric stimulation. The distal tendon was cut off, and was fixed at the muscle length at which the largest force was obtained. CCD type laser displacement sensor (LK-G155, KEYENCE) was used to measure displacement MMG. The laser beam vertically irradiated the central portion of the muscle. The fluctuation of MMG baseline that connected the minimum value between electric stimulation was measured, as shown by the dotted line in Fig.1. The maximum of displacement MMG magnitude is defined as MMGmax, the frequencies of 10% MMGmax and 90% MMGmax were obtained from the incomplete/complete tetanus experiments, respectively.

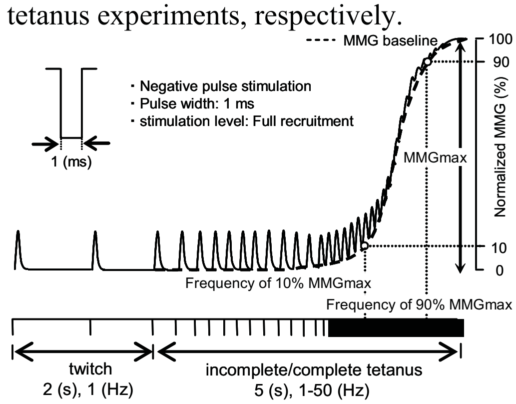

Figure 1: Incomplete tetanus progression at 1-50 Hz electro-stimulation.

RESULTS AND DISCUSSION

Figure 2 shows MMG baseline of GC, VI and SOL in tetanus progression by changing the stimulus frequency. Table 1 shows MMG max and frequency of 10%, 90% MMG max.

(1) GC (typeII fiber), SOL (typeI fiber)

MMGmax of GC is about 6 times as large as SOL. This was considered to have been caused by the component ratio of muscle fiber type and the muscle size differs in GC and SOL. Frequencies of 10%, 90% MMGmax of SOL were smaller than those of GC. The frequencies of 10%, 90% MMGmax are respectively assumed to be the frequencies that initiate incomplete and complete tetanus. Based on these assumptions, it is considered that SOL (mostly type I fiber) develops incomplete/ complete tetanus at low frequency ranges (3.9～15.3 Hz), and GC (mostly type II fiber) develops tetanus at high frequency ranges (15.3～40 Hz).

(2) VI (Mixed fiber)

The component ratio of VI is 36% type I, 64% type II (Ariano et al 1973). In addition, it is reported that MMG is a superimposed wave of single type I and type II MMG. Based on this, only type I fiber in the mixed muscle like VI causes incomplete/ complete tetanus at low frequency ranges. Then, the type II fiber also causes tetanus adding to type I fiber with rise in the stimulation frequency. It is considered that, tetanus progression in the mixed muscle reflects the superposition of the individual tetanus progression of type I and type II fibers. In VI stimulated at a frequency under 10Hz (Figure 2), finding incomplete tetanus gently progresses, and rapidly progresses over 10Hz. This shows that type II fiber participated along with type I fiber in incomplete/complete tetanus.

SUMMARY/CONCLUSIONS

Using a laser displacement sensor, displacement MMG of the rat was measured. Then, the tetanus process of VI which was a mixed muscle consisting of type I and type II fibers was examined following the GC (mostly type II fiber) and SOL (mostly type I fiber). As a result, the tetanus process in mixed muscle involved the superposition of the tetanus process in type I fiber in which incomplete/complete tetanus starts at a low frequency ranges on that of type II fiber which starts at high frequency ranges.

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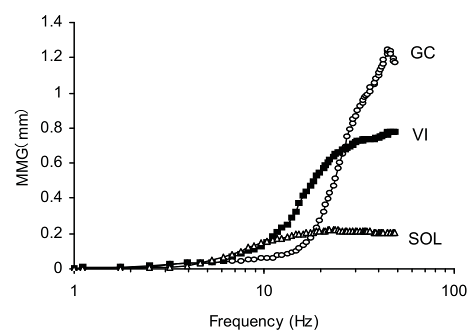

Figure 2: MMG baseline of GC, VI, SOL in tetanus progression by changing the stimulus frequency.

Table 1: MMG max and stimulus frequency of 10, 90% MMG max in tetanus progression.

	GC	SOL	VI
MMGmax (mm)	1.23±0.38	0.22 ±0.12	0.77±0.38
Frequency of 10% MMGmax (Hz)	15.3±4.1	3.9±1.0	7.5±2.0
Frequency of 90% MMGmax (Hz)	40±3.8	15.3±3.9	29±4.5

THE WRISTALYZER: A NEW MYOHAPTIC DEVICE TO INVESTIGATE WRIST FUNCTION IN THE LAB AND IN THE CLINIC

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INTRODUCTION

Evaluation the of metrics of voluntary motion, tremor, spasticity, rigidity or hypotonia in an accurate and standardized way is a challenge (Johnson, 2002; Raethjen et al, 2004). We have built a portable robotic device combining haptic technology with electromyographic assessment (a mechatronic myohaptic device) (Grimaldi et al, 2007). The system, called wristalyzer (Figure 1), allows to assess wrist motion in physiological/pathological conditions by applying loads and mechanical oscillations, taking into account the ergonomy of the joints. It analyses the behavior of the wrist joints and the associated muscle activities during delivery of oscillations at frequencies from 0.1 Hz to 50 Hz. Position, torques and surface EMG activities are analyzed in real time (sampling rate 2048 Hz). The device can characterize the effects of damping on voluntary motion in neurological patients.

METHODS

The characteristic of the wristalyzer (2nd generation) are the following: angular accuracy is 0.35 deg, nominal torque is 6 Nm, maximal rotation velocity is 2000 degrees/sec, with a range of motion of -60 to +60 degrees . We investigated the effects of oscillations delivered by the wristalyzer at 1-15 and 30 Hz, and the effects of biomechanical loading (addition of damping of 1 Ns/m) on tremor, on a visually-guided reaching task (aimed target 0.1 rad) and on alternate movements between 2 targets at 0 and 0.1 rad. Moreover, the relationship between movement times and maximal angular velocity (aimed amplitude 0.1 rad), was studied.

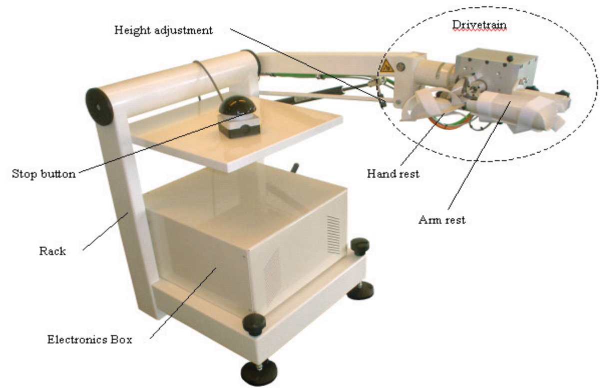

Figure 1: Illustration of the wristalyzer (2nd generation) with its main components.

We investigated 4 healthy subjects, one patient presenting Parkinson's disease with a lateralized rest tremor on right upper limb and 4 cerebellar patients. We investigated the right hand in a seated condition. Surface EMG recordings were performed in right upper limb at the level of the extensor carpi radialis ECR (Delsys Inc., Boston, USA).

RESULTS AND DISCUSSION

FFT of surface EMG signals in control subjects during slow and fast oscillations showed a clear peak in the power spectrum confirming the responsiveness of muscles to low/fast frequency oscillations. The stability of the frequency of the EMG responses over time was confirmed by the Wigner transform. Stretch responses in the cerebellar patients were temporally abnormal. Time-frequency analysis showed diffuse peaks and the sonogram method confirmed the pathological response in the patients (IgorPro 6, Wavemetrics, USA). Figure 2 illustrates the action tremor recording in the Parkinsonian patient. Angular motion, torques and corresponding EMG activities in the ECR muscle are illustrated. Spectral analysis of position, torque and EMG signals showed a peak at 4.37 Hz. Action tremor was markedly reduced by addition of damping of 1 Ns/m. Similar beneficial effect of motion loading was found in the cerebellar patients. Ataxia decreased markedly from 100 % (baseline) to 28.3 +/- 11.7 %.

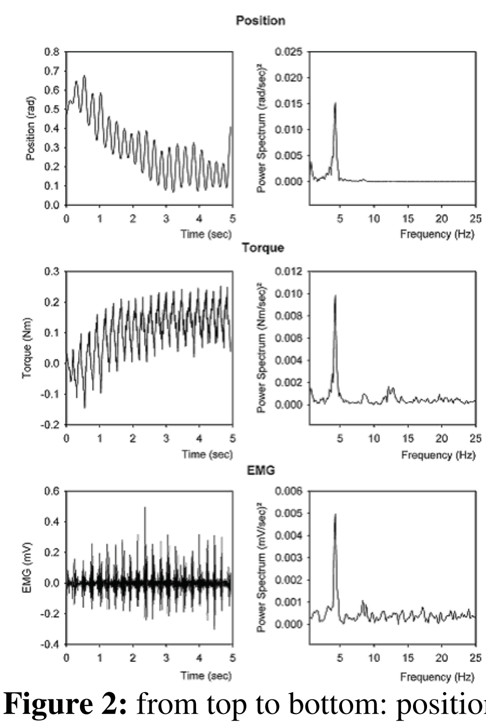

Figure 2: from top to bottom: position, torque and EMG activities of the ECR are illustrated in left panels. Corresponding power spectrums in right panels.

Attempts to reach the aimed targets resulted in overshoots in the cerebellar patients (hypermetria). The inverse linear relationship between movement times and maximal angular velocity in the control groups was lost in case of cerebellar dysfunction (Figure 3).

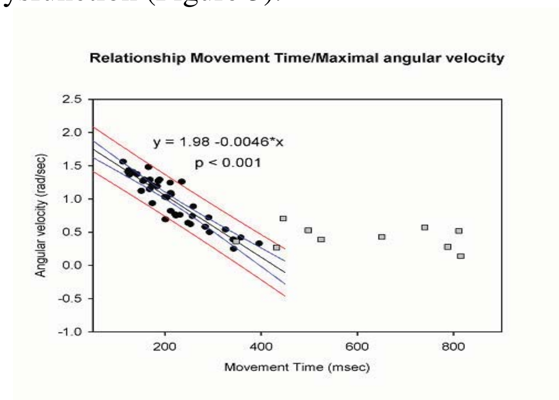

Figure 3: Inverse linear relationship Movement time/Maximal angular velocity in controls (black dots). Blue: 95 % confidence bands; red: 95 % prediction bands. The movement times are increased markedly in a cerebellar patient (grey squares).

SUMMARY/CONCLUSIONS

This new robotic device measures with high accuracy motion of the wrist, for research or clinical purposes. Analysis of tremor or dysmetria are typical applications. We confirm that biomechanical loading can reduce markedly neurological tremor (Manto et al, 2007).

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CHARACTERISATION OF THE HOFFMAN REFLEX USING MECHANOMYOGRAPHY

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INTRODUCTION

Mechanomyography (MMG) is a technique for recording mechanical activity in contracting muscle. The MMG signal is low frequency, typically 5-100Hz (Perry *et al.,* 2001). This MMG 'sound' is produced by lateral oscillations of muscle fibres which occur at the resonant frequency of the muscle (Cole and Barry, 1994). The analysis of MMG signals has allowed examination of various aspects of muscle function such as neuromuscular fatigue, muscle fibre type distributions and neuromuscular disorders.

To date, Electromyography (EMG) has been considered the primary non-invasive technique to record and interpret the physiological properties of contracting muscle. The Hoffmann reflex (H-reflex) is the equivalent of the monosynaptic stretch reflex, elicited by electrical stimulation.

The aim of this investigation was to characterise the Hoffman reflex using an MMG system. The system is based on 2-axis MEMS (Micro Electro-Mechanical System) sensors placed on the soleus muscle.

METHODS

The H-reflex can be elicited most conveniently in adults by the electrical stimulation of the posterior tibial nerve in the popliteal fossa. Stimulation was triggered manually in stepped intensities.

This experiment was conducted on an adult male who had both EMG and MMG sensors placed on the belly of the soleus muscle.

Both the sensor and the electrode were placed close together so as to pick up signals from the same muscle contraction area. The x-axis was orientated to pick up lateral movements and the y-axis proximal movement.

RESULTS AND DISCUSSION

The H Reflex was measured simultaneously using EMG and MMG. A sample recording is shown in figure 1. The EMG trace clearly shows both the H Reflex and an M wave response. Note that there is no stimulus artefact evident in the MMG response; an advantage where EMG techniques are not applicable such as functional electrical stimulation studies.

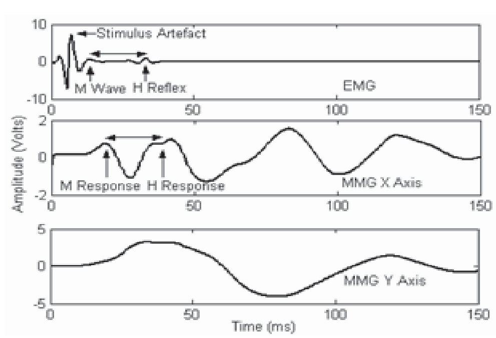

Figure 1: Hoffman response (EMG & MMG).

It is clear that there is a delay between the MMG response and the EMG response. EMG signals are a recording of the motor unit action potentials propagating along the motor units, while the delayed MMG trace is the mechanical response as it builds up as the muscle fibre depolarises (Harba and Chee, 1997).

The latency measured between the H reflex and M wave on the EMG trace is similar to the latency measured between the two responses identified in the MMG x-axis. This is consistent for the duration that events appear in the EMG trace. This would suggest that the EMG and MMG responses share a similar origin.

The acquired EMG recruitment curve is shown in figure 2. While the EMG H reflex starts to roll off due to antidromic activity after 17% stimulation, the MMG responses do not.

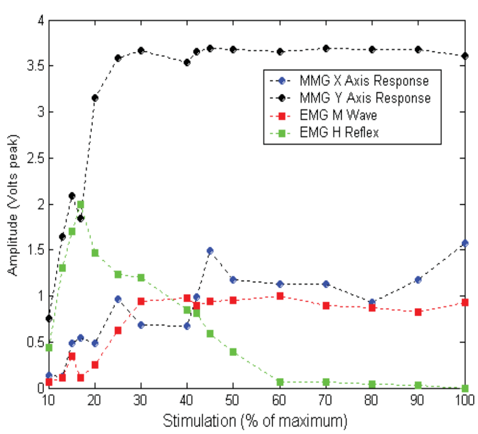

Figure 2: EMG and MMG Recruitment Curve for H Reflex

Note that the MMG x-axis and y-axis trends in the recruitment curve are correlated. The gain on both axes is equal, thus the y-axis demonstrates a response of much greater amplitude. Both MMG responses in figure 2 also appear to correlate with the EMG M response. This is especially evident around 25% stimulation where maximum stimulus is reached.

At lower stimulation intensities those muscle fibres being directly stimulated are furthest from the MMG sensor. It can be observed in figure 3 that as the stimulation increases the latency of the initial MMG response decreases by ~ 34ms. Again, the maximum stimulation is reached at 25%, when all the muscle would appear to have been recruited.

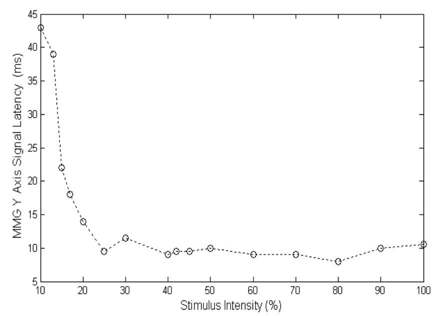

Figure 3: MMG latency compared to Stimulus Intensity

SUMMARY/CONCLUSIONS

This paper outlines a study of the H reflex while recording EMG and MMG signals simultaneously. While the MMG trace is free of electrical artefact there is also evidence that it contains information about the physiological response. The recruitment curve and latency studies provide data that will be used for further studies to investigate their origin.

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PASSIVE FLEXION TORQUE AND EXTENSOR PARESIS CAUSE WRIST DEFORMITY IN CEREBRAL PALSY.

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INTRODUCTION

Wrist flexion deformity is a phenotypic phenomenon in Cerebral Palsy (CP). Mechanically, the deformity is the result of a net torque imbalance which may be passively structural (Lieber et al. 2003, Friden et al. 2003) and/or actively neuro-muscular determined (Gracies 2005 [a,b]).

Treatment of wrist deformities involves orthopedic surgery, neurology and physiotherapy each covering a spectrum of determinants: tendinous and muscular tissue characteristics, paresis and hypertonia.

Identification of the passive and active determinants is needed for a proper choice of treatment in addition to understanding the etiology of wrist deformities and treatment follow-up.

The goal of this study was to discriminate between and quantify passive and active sources resulting in wrist deformities in CP, by means of a non-invasive, integral and biomechanical assessment.

METHODS

A wrist torque robot (Schouten et al. 2005) has been applied to CP patients with mild wrist flexion deformity (n=5, 12.6 yrs ± 2.3, 3m/2f), age-matched controls (N=5, 12.6 yrs ± 2.1, 3m/2f) and adults (N=8, 25.7 yrs ± 2.1, 5m/3f).

Subjects were seated upright, while their forearm was fixated horizontally with approximately 135° of elbow extension. The axis of rotation of the wrist was aligned with the torque motor axis of the wrist robot and the hand was strapped to the handle. The wrist robot was configured as an angular servo and wrist torque was recorded.

Measurements were performed at 7 initial wrist angles, between -45° (flexion) and 45° (extension) and 0^o defined by alignment of the third metacarpal with the distal radius.

Bi-polar surface EMG (Bagnoli-8, Delsys, Inc.) was recorded on the flexor carpi radialis (FCR) and extensor carpi radialis longus (ECR), both assumed to be representative of the total muscle activity in flexion and extension.

The tasks combined passive and active conditions and isometric and isokinetic (dynamic) conditions. The angular profile of the passive and dynamic tasks was an alternating extension-flexion 'ramp and hold' with an sub-reflex threshold angular velocity of $\omega = +/-5.7^o.s^{-1}$ respectively (Fig.1, top panel).

We obtained:

1. *MVC*, recorded in the neutral wrist position.
2. EMG at 20% MVC ($EMG_{.20}$) in the neutral wrist position. This EMG-level was set as target level in the active tasks in wrist flexion and extension positions.
3. *Passive torque (T_{pas})*, the quasi-static average 'ramp and hold' torque at each wrist position (EMG-controlled).
4. *Active flexion (+) and extension (–) torque* (T_{act}) at each wrist position, determined at $EMG_{.20}$

RESULTS AND DISCUSSION

Neutral wrist position is determined by the passive torque balance between the flexion torque in wrist extension and extension torque in wrist flexion. The average neutral wrist angle is predicted by the $T_{pas} = 0$ intercept (Fig.1, lower left panel).

The passive 0-torque intercept in CP-patients is shifted to a smaller (flexion) angle relative to adults and controls (Fig.1, lower left panel). The passive flexion torque in CP-patients is high relative to their controls, but do not exceed individual adult values (not shown).

The active flexion torque T_{act} is significantly lower in CP-patients relative to their controls and adults, especially for the wrist extensors (Fig.1, lower right panel).
Three of five patients could compensate the flexion deviation of the passive torque balance by active extensor forces. Two patients with serious flexion deviations could not actively balance the passive deviation with the $EMG_{.20}$ flexion force equivalent (not shown).

If assumed that the average active extensor torque-angle curve represents the force-length relation of the wrist extensors, normal (control and adult) optimal (minimal) muscle length is observed at positive wrist extension angles. In CP-patients a shift of optimal muscle length towards flexion (left) is observed.

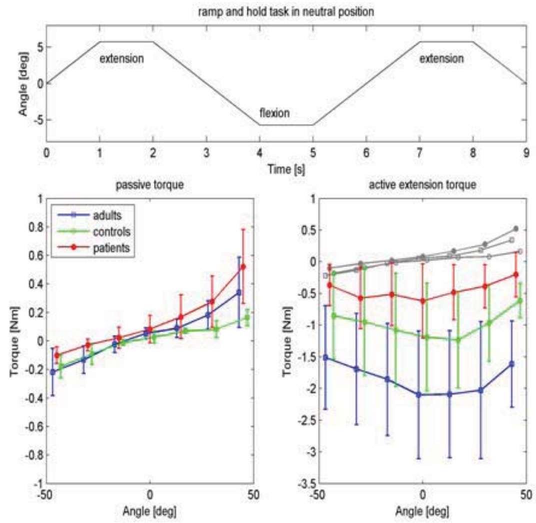

Figure 1: Ramp and hold extension-flexion task (top panel) resulting in passive torques (lower left panel) and active (extension) torques (lower right panel) for patients, age matched controls and adults.

SUMMARY/CONCLUSIONS

A net passive flexion torque in the neutral wrist position causes the deviation of the wrist angle in CP-patients. Additionally, the extensor paresis prevents active compensation to restore the neutral wrist position.

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MRI ANALYSIS OF THE FORCED FLEXION MOVEMENTS OF THE UNILATERAL HIP JOINT IN SUPINE POSITION

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INTRODUCTION

In our previous studies we have confirmed the involvement of movement of the lumbar facet joint, lumbosacral joint, and iliosacral joint with retroversion movement of the pelvis by the analysis of passive maximum flexion movement of the unilateral hip joint by magnetic resonance imaging (MRI). In the present study, the hip joint was weighted in the direction from the maximum flexion to flexion, and flexion movement at a more increased angle of flexion of the hip joint was analyzed.

METHODS

The subjects were 10 healthy women, with a mean age of 21.0 years (range: 20-24 years). Their mean height was 155.5±3.9 cm and their mean body weight was 48.1±4.5 kg. Under the following 5 setting conditions, passive flexion movement of the unilateral hip joint was analyzed in a supine position and in a flexed position of the knee joint by MRI (General Electric Co., SIGNA 1.5 T): Angles of flexion of the hip joint of 0°and 60°; the maximum flexed position; further flexed position by 30-N weight; and further flexed position by 60-N weight. The hip joint was weighted by means of a weighting system (Takei Scientific Instruments Co., Ltd.). The items to be analyzed under each training condition were an angle made by the femoral bone with a line parallel with the body trunk (FH angle), the degree of pelvic inclination of a line made between the posterior upper iliac spine and pubic symphysis from a horizontal plane (PI angle), a lumbosacral angle made by the upper surface of the 1st sacral vertebra (S1) with a horizontal plane (L angle), and a relative angle of inclination to each upper vertebra of the region ranging from the 2nd lumbar vertebra to S1 (Fig.1). The results were assessed by the analysis of variance and multiple comparative test, and differences at $p<0.05$ were considered significant. This study was conducted with approval of the Ethics Committee for Safety of Research, Tokyo Metropolitan University Arakawa campus.

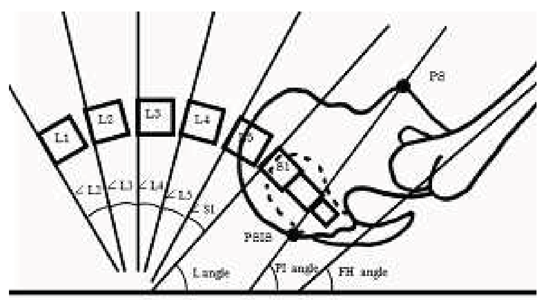

Figure 1: Analyzed items

RESULTS AND DISCUSSION

The mean FH angle under each training condition was 0°at 0°, 56.7°at 60°, 133.5°in the maximum flexed position, 137.9°with 30-N weight, and 138.6°with 60-N weight. The degree of retroversion of the right pelvis was 0°at 0°, 2.6°at 60°, 16.3°in the maximum flexed position, 17.8°with 30-N weight, and 17.8°with 60-N weight. The

degree of retroversion of the left pelvis was 0°at 0°, 0.9°at 60°, 13.4°in the maximum flexed position, 15.2°with 30-N weight, and 15.1°with 60-N weight. The degree of retroversion of the sacral bone was 0°at 0°, 1.4°at 60°, 13.9°in the maximum flexed position, 15.3°with 30-N weight, and 15.3°with 60-N weight. Multiple comparative test revealed that there was a significant difference between the degrees of right and left pelvic retroversion at an FH angle of 56.7°and that there were significant differences between the degrees of right and left pelvic retroversion and between the degree of right pelvic retroversion and the degree of retroversion of the sacral bone at FH angles of 133.5°, 137.9°, and 138.6°. At FH angles of 133.5°in the maximum flexed position, the sacral bone showed nutation(2. 4°) in relation to the right ilium bone. From these observations, movement of the sacroiliac joint was confirmed to be involved in the flexion movement.
The mean degrees of displacement of relative angles of inclination (to each of L1/2, L2/3, L3/4, L4/5, and L5/S1) in the region ranging from the 2nd lumbar vertebra to S1 were as follows: L1/2: 0°, -0.4°, -0.3°, -0.4°, and –0.6°, respectively; L2/3: 0°, -0.1°, 0.4°, 0.4°, and 0.4°, respectively; L3/4: 0°, 0.3°, 2.8°, 2.8°, and 2.6°, respectively; L4/5: 0°, 0.8°, 6.0°, 7.0°, and 7.0°, respectively; L5/S1: 0°, 2.5°, 10.6°, 11.1°, and 11.1°, respectively. Multiple comparative test revealed that the degree of displacement of L5/S1 was significantly high at an FH angle of 56.7°, and there were significant differences among the degrees of displacement of each vertebral body at FH angles higher than 56.7°.

In our previous studies we have confirmed that the degree of retroversion of pelvis on the flexed side is higher than that on the opposite side at 60°in passive flexion movement of the unilateral hip joint and the sacral bone shows a relatively anteversion position to the right coxal bone with increase in the difference in the degree of retroversion in the maximum flexed position and that the retroversion of pelvis is much influenced by movement of lower joints, particularly the lumbosacral joint, in the vicinity of the maximum flexed position. In the present study we analyzed the passive flexion movement at increased angles of flexion of the hip joint. The bilateral pelvis showed retroversion to the same degree as that in the maximum flexed position without change in the condition at FH angles of 137.9°or lower degrees, and there was little or no change in movement of the iliosacral joint. It was subsequently confirmed that any movement of retroversion of pelvis was not recognized at FH angles of 138.6°or lower degrees.

SUMMARY/CONCLUSIONS

On the basis of these observations, it is considered that the bilateral coxal bones and sacral bones show retroversion *en bloc* in the maximum flexed position and under the subsequent conditions, ultimately resulting in the disappearance of movement of pelvic retroversion except that only movement characteristic of the hip joints appears. It was also confirmed that pelvic retroversion is much affected by movement of the lower joints in the maximum flexed position and under the subsequent conditions.

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INFLUENCE OF ANGULAR VELOCITY OF A DYNAMOMETER IN JOINT POSITION SENSE OF KNEE IN NORMAL SUBJECTS

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INTRODUCTION:
The integrity of knee proprioception is essential for a good musculoskeletal control, and for this reason, knowing some physiological mechanisms involved in this process is essential for clearing the complex injury mechanism, favoring the elaboration of more specific and adequate interventions. Among the resources to assess the joint position sense, it's prominent the manual goniometry, continuing passive manipulation, shooting system, isokinetic dynamometry and electrogoniometry, but the isokinetic dynamometry and electrogoniometry are the most reproducible.

OBJECTIVE:
Verify which angular velocity of the dynamometer is the most reproducible for assessing the knee joint position sense in healthy subjects.

METHODS:
It was selected 10 female volunteers, sedentary (without regular physical activity), age between 18 and 21 years old, and without any kind of musculoskeletal injury in lower limbs (hip, knee and ankle joints).It was utilized a Cybex Norm dynamometer and a MIOTEC model miotool 400 of 4 channels with 14 resolution bits, acquisition per channel of 2000 samples per second, 100x, RUÍDO < 2LSB, filter Butterworth hight pass 1 polo 0,1Hz and buterworth low pass 2 polo 500 Hz, spacing between electrodes fixed in 30mm. Surface electrodes of Ag/ClAg, round, pre gelded and auto adhesive from MEDITRACE Kendall 200. The subjects were placed on the dynamometer chair, sitting, back sustained, and fastened by belts according to fabricant regulation. The dynamometer arm passively conducted the dominant limb in a regular velocity of 1º/s, until the proposed angle of 45º of knee flexion for 3 times, standing on this angle during 5 seconds for the perception of the position. After this time, the limb returns to initial position and will be passively conducted by the dynamometer for 25º to 90º knee flexion amplitude, and the subject, with a locking button of the dynamometer, was oriented to stop the movement when realizing to be at 45º. The same procedures were practiced for velocities of 2º/s and 20º/s. During the collection, surface electrodes were placed according to SENIAM on rectus femoris muscles to certify they were relaxed. It was used significance level of 5% and wilcoxon test also (Basmajian & De Luca 1985; Lephart, FU 2000)

RESULTS:
It was observed that there wasn't statistically significant difference, but for angular velocity of 1º/s, the absolute error was 2º, and 1º for angular velocity of 2º/s ($p<0,07$). Nevertheless, for angular velocity of 20º/s, the higher absolute error was 12º ($p<0,001$).

	Angular Velocity 1º/s	**Angular Velocity 2º/s**	**Angular Velocity 20º/**
Absolute error with 45º knee flexion	**2º (p<0,07)**	**1º (p<0,07)**	**12º (p<0,001)**

Table 1. Absolute error of joint position sense of knee at 1º/s, 2º/s and 20º/s.

CONCLUSION:

This study suggests for this sample that the lowest errors for the answer of joint position sense of knee was at angular velocity of 2º/s. This is important to define the one of the parameters of knee proprioception evaluation.

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EVALUATION OF THE MUSCULAR RELATION OF VASTUS MEDIALIS AND VASTUS LATERALIS DURING THE LANDING OF CABRIOLE IN A BALLET DANCER – CASE STUDY

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INTRODUCTION

Among the knee injury mechanisms, the quadríceps performance is an important factor, mainly to perform the technical movements of the classic ballet.
The objective of thi study was to measure the recruitment of the Vastus Medialis (VM) and Vastus Lateralis (VL) muscle fibers during the landing of the *Cabriole* gesture in the classic *ballet*.

METHODS

An electromyographic analysis of the muscle recruitment of the (VM) and (VL) was carried out in a *ballet* dancer, 19 years old, with 15 years of practice, without knee pain was evaluate during the *Cabriole* gesture, which was performed three times with a EMG System of MIOTEC model miotool 400 of 4 channels with 14 resolution bits, acquisition per channel of 2000 samples per second, 100x, RUÍDO < 2LSB, filter Butterworth hight pass 1 polo 0,1Hz and buterworth low pass 2 polo 500 Hz, spacing between electrodes fixed in 30mm. Surface electrodes of Ag/ClAg, round, pre gelded and auto adhesive from MEDITRACE. The VM electrode was placement at 2cm medially from the superior rim of the patella, and for the VL was placed approximately 4 cm above the patella, on an oblique angle just lateral to midline (Cram & Kasman 1998).

An isokinetic Dynamometer (CYBEX NORM) was used to obtain the maximum voluntary contraction of these muscles. (Araujo et al 1995; Basmajian & De Luca 1995).

RESULTS

The first EMG signal detected in the isokinetic dynamometer during the concentric contraction was 124,89µV for VM and 198,78µV for VL, in the eccentric contration the data were 162,3µV for VM and 234,6µV for VL. Whereas during the landing of the gesture the data were 254,87µV for VM and 178,45 µV for VL.

	VM RMS values (µV)	VL RMS values (µV)
CYBEX concentric	124,89	198,78
CYBEX eccentric	162,3	234,6
CABRIOLE	254,87	178,45

Table 1. RMS values of the VM and VL activity in concentric and eccentric contraction at CYBEX and during *cabriole* gesture

DISCUSSION

It was possible to observe that during the *Cabriole* landing the VM was more required than the VL, inverting the muscular relation observed inthe isokinetic, what can certainly cause joint misfit and muscle joint disfunction, accouting for some of the high rates of the knee injury in this sport.

CONCLUSIONS

The research suggests that for the dancer studied sample, during the landing of the *Cabriole* gesture in *ballet*, the VM and VL relation alters, and the VM seems to be more recruited. Other researches with this line, and more subjects could be important to define this biomechanics gesture in ballet dancers.

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ANALYSIS OF FORCES DURING ISOMETRIC MOVEMENT OF THE DELTOID MUSCLE: ELECTROMYOGRAPHIC STUDY

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INTRODUCTION

During the rehabilitation of shoulder impingement syndrome, the individuals present a considerable decrease in force of abduction and external rotation in the injured shoulder, when compared to the healthy limb. These individuals relate pain during the abduction movements and external rotation, possibly caused by the decrease in muscular force. This study is aimed at comparing electromyographic activity of the anterior, middle and posterior portions of the deltoid muscle, as well as the force exerted during abduction, in healthy individuals and in those with shoulder impingement.

METHODS

Fifteen subjects with impingement syndrome (44.1 ± 8.6 years old) and fifteen healthy subjects (43.6 ± 8.1 years old) were paired to those of the first group. Electromyographic signals were collected in static voluntary contraction during traction of a force transducer, while the shoulder remained at 80° abduction, and normalized according to the maximum reference isometric contraction. Mann Whitney's parametric test was used at a significance level of $p < 0.05$ to compare electromyographic (RMS values) of the deltoid muscle pairs, as well as the force between the two groups.

The EMG activity was captured by an electromyograph composed of differential double electrode, a bandpass filter at 20 to 1.0 kHz, and a subsequent amplification of 50 times with a common mode rejection ratio of 120 dB, 16-bit A/D converter and sampled frequency of 2.0 kHz for each channel by (***EMG System do Brasil Ltda*** *)*. A differential double electrode was used, with pre-amplification with 100 times, 25 mm2 contact area and contacts 10 mm apart. Sampling frequency was 2000 Hz. The recommendations from the International Society of Electrophysiology and Kinesiology (ISEK) regarding electromyography applications were followed .

RESULTS AND DISCUSSION

There were no significant differences in electromyographic activity and the force measured in the force transducer between the healthy control group and the impingement syndrome group during the upper limb 80º abduction, in the scapular plane of static voluntary contraction.

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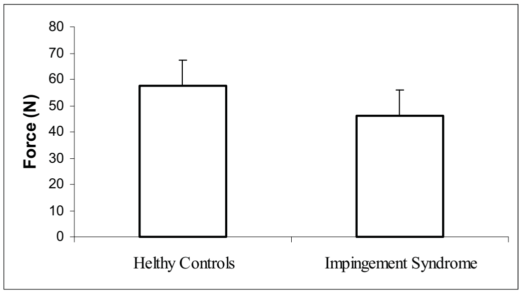

Fig.1. Mean and standard deviation for the maximal force (N) generated in the traction of force transducer during of the upper limb abduction in the scapular plane movement.

SUMMARY/CONCLUSIONS

The study concluded that there was no difference in electromyographic activity of the deltoid muscle or in the force exerted during abduction of the upper limb between healthy controls and individuals with shoulder impingement. It may thus be suggested that deltoid muscle strengthening should not be the main focus during the rehabilitation process.

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Table 1. Mean and standard deviation (SD) of normalized electrical activity readings (as percentage of maximal isometric contraction) of the anterior, middle and posterior portions of the deltoid muscle (Wilcoxon test, *P*< .05).

Muscle	Healthy Syndrome	Impingement Syndrome
Anterior Deltoid:	17.6 ± 2.6 μV	17.1 ± 2.6 μV
Middle Deltoid:	18.6 ± 3.1 μV	20.7 ± 14.3 μV
Posterior Deltoid:	17.4 ± 3.4 μV	18.3 ± 2.8 μV

THE USE OF MECHANOMYOGRAPHY SIGNAL IN THE ASSESSMENT OF MUSCLE FIBER TYPE

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INTRODUCTION

Mechanomyography (MMG) has been used as an alternative approach in muscle contraction studies. Different transducers as, for example, accelerometers, microphones, and piezoelectric contact sensors have been reported in acquiring the MMG signal. Regarding to the mechanical properties of muscle contraction, it has been suggested that most of information derived from the MMG signal seems to be from the lateral oscillations (X direction) of muscle fibers, despite of some authors have also investigated the longitudinal ones (Y direction) from the MMG signal when biaxial accelerometers are used to collect the signal (Matta et al., 2005). As a method that can represent a summation of single muscle fibers activity, some authors have also suggested that the MMG signal features might also contain information concerning to the fiber type (Beck et al., 2007).

This study aimed to evaluate the feasibility of assessment of the muscle fiber type from properties of MMG signals acquired by using biaxial accelerometers. The studied muscles were the *Gastrocnemius* and *Soleus*, both characterized by some authors as mixed and slow, respectively, during different levels of unfatigued isometric contraction.

METHODS

Nineteen healthy and right-handed male volunteers participated on this study. The acquisition system was based on a computer and a 16 bits A/D converter (Spider 8 – HBM, Germany), and range of ± 5V. For the MMG signal acquisition we used a biaxial accelerometer (ADXL202E, Analog Devices, USA) mounted on a small print circuit board, which resulted on a total mass of 1.5 gram. The sampling frequency was set in 960 Hz. The softwares for acquisition of the MMG signals and force were built in LabVIEW 5.0 (National Instruments, USA). A scotch tape was used to fix the accelerometer over the bellies of right *Gastrocnemius* and *Soleus* muscles following the protocol adopted by SENIAM for surface electrodes placement in EMG analysis (Matta et al., 2005).
A mechanical apparatus containing an electro-dynamometer was built for supporting inferior limbs and for collecting the force signal during the tests. After a maximum isometric muscular contraction (MVC) test for each muscle, five arbitrary percentiles were calculated (20%, 40%, 60%, 80% and 100% of MVC) and tested in a randomised order for each volunteer.
The volunteers had to reach each target level by means of visual feedback that was provided through a video screen (computer monitor).

The volunteers sat on the apparatus with the right foot attached to a surface for this purpose, depicted in Figure 1. For the *Soleus* muscle test, the knee joint was kept 90° of flexion (Figure 1). On the other hand, for the *Gastrocnemius* muscle test, the knee joint was maintained in an extended position. In both tests, the volunteers were asked to not move the trunk and arms as well as keep the signal force with a minimum of overshooting regarding to the each target level for six seconds. Despite of this total time just the intermediate period (T) of two seconds was used for analysis purposes.

Figure 1: A volunteer during the *soleus* muscle test performed in the mechanical apparatus specially designed to the research.

The mean frequency (F_{mean}) and the root-mean-square (RMS) value from the MMG signal associated to the Y direction were calculated from each period T and at each level of contraction. Based on these parameters, and after an initial observation of the behaviour of the parameter, we derived a heuristic equation to estimate the percentage of slow muscle fibers, which regarded the values of F_{mean} from 20% to 100% of MVC as well as the normalized RMS values from 20% to 80% of MVC, resulting in a total of nine variables. Thereafter, this equation was used to estimate the percentage of slow and fast fiber types in the muscles analysed.

RESULTS AND DISCUSSION

The mean percentages of each fiber type were calculated for the 19 volunteers. The results are summarized in table 1. They are in agreement with previous studies concerning the composition of the muscle fiber type for *Soleus* and *Gastrocnemius* muscles (Dahmane et al**.**, 2005).

Table 1: Mean percentages of muscle fiber types for each muscle

Muscle	Slow fibers	Fast fibers
Soleus	71.8 ± 10%	28.2 ± 11%
Gastrocnemius	61.3 ± 8%	38.7 ± 9%

The MMG signal can be considered a signal that reflects mechanical properties of a muscle based on the summation of the twitches from the motor units already recruited. As each motor unit is composed by only one type of muscle fiber, it is suggested that most of the fibers in a muscle influences in the compound MMG signal. Other important finding of the present study concerns the fact that many authors reported the use of MMG signals associated to the X accelerations to perform their analysis; instead of signal associated to the Y direction used in the present study, which in our opinion might better represent the muscle twitches and their summation. Therefore, it seems that the parameters derived from the MMG signal in the temporal and frequency domains related to the muscle fibers direction (Y) can contribute to estimate and better understand some mechanical properties of muscle, including fiber type.

SUMMARY/CONCLUSIONS

The results suggest that the analysis of the MMG signal in time and frequency domains can reflect some useful muscle contraction properties that indeed may also help to estimate fiber type characteristics.

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REPEATABILITY OF THE MECHANOMYOGRAPHIC AMPLITUDE VERSUS ISOMETRIC TORQUE PATTERNS OF RESPONSES

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INTRODUCTION

The surface mechanomyogram (MMG) records and quantifies low-frequency lateral oscillations of contracting skeletal muscle (Orizio et al. 2003). The time and frequency domains of the MMG signal may provide unique information regarding motor control strategies (Orizio et al. 2003). Specifically, the time domain (amplitude) is thought to reflect motor unit recruitment (Akataki et al. 2004), whereas the frequency domain may provide a global estimation of motor unit firing rate (Beck et al. 2006).

Since MMG offers a unique non-invasive approach to examine motor control strategies, studies are needed to determine if the MMG amplitude vs. isometric torque patterns of response are similar over multiple testing days. These patterns of responses are traditionally analyzed using statistical techniques (polynomial regression) that are designed to maximize the proportion of variance in the data that can be explained by a particular model (i.e. linear, quadratic, cubic, etc.) (Ryan et al. 2007).

Therefore, the purpose of this study was to determine if the MMG amplitude vs. isometric torque patterns of response were similar across 3 different testing days during submaximal and maximal isometric muscle actions of the vastus lateralis (VL).

METHODS

Nineteen healthy subjects (mean±*SD* age=24±4 yrs; stature= 173±9 cm; mass =75±13 kg) volunteered for this investigation. All participants signed informed consent forms. This study was approved by the University of Oklahoma Institutional Review Board.

Each subject visited the laboratory on 4 occasions: 1 familiarization trial and 3 experimental trials. Each trial was performed at the same time of day (±2 hours) separated by 2 – 5 days.
During each trial, isometric torque of the right leg extensors was measured on a Biodex System 3 dynamometer. Each muscle action was 4-s in duration and performed at a 120º joint angle (between the leg and thigh). Each subject performed isometric maximal voluntary contractions (MVCs) and submaximal isometric leg extensions at 5, 15, 25, 35, 45, 55, 65, 75, 85, and 95% MVC in random order.

MMG signals were recorded with an active miniature accelerometer (EGAS-FS-10-/V05, Measurement Specialties, Inc., Hampton, VA). The accelerometer was taped to the skin over the lateral/anterior portion of the VL muscle. The most stable 1-s epoch of the 4-s torque plateau was selected to calculate the submaximal torque level and MMG ($m \cdot s^{-2}$) amplitude values.

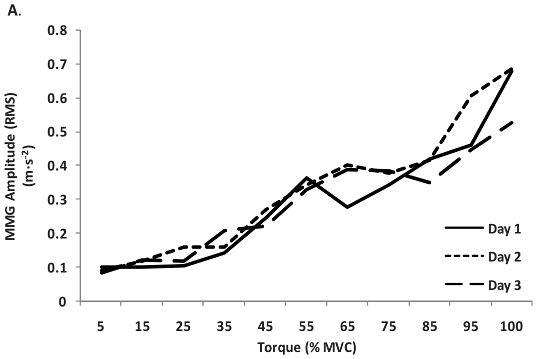
The patterns of responses were examined using polynomial regression to compare the linear, quadratic, or cubic shapes of the relationships. The statistical significance (P

< 0.05) for the increment in the proportion of the variance that would be accounted for by a higher-degree polynomial (i.e., R^2-change) was determined using the F-test described by Pedhazur (1997).

RESULTS AND DISCUSSION

The results from the polynomial regression analyses indicated 6 of 19 subjects were fit with the same model for all 3 days, 10 of 19 were fit with the same model between days 2 and 3, and 2 of 19 were fit with different models for all 3 days. Figure A (subject 13)

A.



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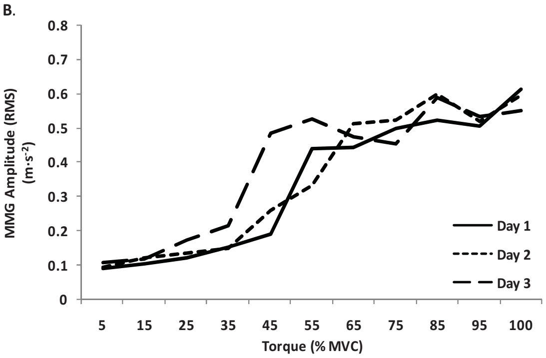

and B (subject 2) are examples of subjects who were fit with the same model all 3 days and fit with a different model for all 3 days, respectively.

SUMMARY/CONCLUSIONS

These findings indicated that although the patterns of responses appeared to be similar across the 3 testing sessions (Figures A & B), the polynomial regression analyses reported 68% (13 of 19) of the subjects were not consistently fit with the same model. Thus, these results highlight the potential drawbacks of using polynomial regression to examine these patterns and suggest that alternative statistical analyses may be necessary to accurately characterize the shape of the MMG amplitude vs. isometric torque relationship.

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Track 05

Motor Units (MU)

QUANTITATIVE ELECTROMYOGRAPHIC EVIDENCE OF CHANGES IN MUP MORPHOLOGY WITH REPETITIVE STRAIN INJURIES

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INTRODUCTION

The wrist extensor muscles have been implicated in a condition often called non-specific arm pain (NSAP) or Work-related upper limb disorder, which, as the names suggest, has an unknown pathophysiology. Patients with NSAP complain of diffuse forearm pain during and after repetitive tasks, and have muscle pain and tenderness on palpation that is not consistent with lateral epicondylitis (LE), a known tendinopathy resulting from repetitive wrist extension. The purpose of this study was to determine: (i) if motor unit morphology and firing statistics recorded from the affected muscle in patients with NSAP are different from individuals with LE and/or control and at-risk subjects, and (ii) if these quantitative EMG parameters suggest that the underlying pathophysiology in NSAP is either myopathic or neuropathic in nature.

METHODS

Patients with signs and symptoms consistent with NSAP or LE, subjects who were at-risk for developing a repetitive strain injury (RSI), and healthy control subjects were recruited. Volunteers underwent a clinical examination to rule out other pathology (such as cervical radiculopathy) and to confirm that their signs and symptoms were appropriate for each experimental group. A quantitative EMG evaluation was then completed. A monopolar surface electrode was placed over the motor point of the extensor carpi radialis brevis (ECRB) muscle on the affected (or more seriously affected) forearm, and a reference electrode was located over the dorsum of the hand. Maximum voluntary electrical (MVE) activation of the ECRB muscle was determined and was used to quantify contraction level throughout the remainder of the testing. A 32 gauge concentric needle electrode was inserted into the ECRB beneath the surface electrode. Each volunteer performed a series of low level wrist extension contractions ranging from 5 to 25% of their MVE; the needle location was changed between contractions. AcquireEMG™ software was used to simultaneously acquire surface and needle EMG data during each trial, and DQEMG™ was used to decompose each data set into surface- and needle- detected motor unit potentials (SMUPs and MUPS respectively). SMUP and MUP morphology and MU firing rates were compared among the four groups using one-way ANOVAs. The α-level was adjusted for multiple comparisons ($\alpha = 0.05/8$), and was therefore set at $\alpha = 0.006$. Post hoc analyses were performed using Tukey's pair-wise comparisons.

RESULTS AND DISCUSSION

Sixteen subjects with NSAP, 11 subjects with LE, 8 subjects at-risk, and 37 control subjects participated. The subjects with NSAP had a mean duration of 27(±32) months since symptom onset, and the subjects with LE had a mean duration of 39(±31) months since symptom onset. There was no difference in this duration between these groups (p=0.33). Significant group differences were found for all MUP (amplitude, duration, number of phases, area-to-amplitude ratio (AAR)) and SMUP (amplitude, area, duration) variables; as well as in mean MU firing rate ($p<0.006$). The patients with NSAP demonstrated smaller MUP and MUP amplitudes compared to the control and LE groups ($p<0.006$), as noted in Table 1. Control subjects' had MUP durations that were significantly shorter than the other three groups ($p<0.006$); SMUP duration was significantly longer in the control group compared to the NSAP group ($p<0.006$). NSAP, LE and at-risk subjects had lower mean MU firing rates than the control subjects (14.98±2.97 Hz).

SUMMARY/CONCLUSIONS

All the surface-detected MUP size-related parameters revealed that the NSAP group had significantly smaller MUPs than the control and LE subjects. Smaller MUPs are often associated with myopathic conditions, and may be indicative of fiber atrophy and/or loss within a motor unit. Evidence of these same changes was found in the at-risk subjects, whose amplitude measures were also smaller than the control subjects, but not smaller than the subjects with NSAP. As such, these findings may reflect the effects of habitual use on muscle structure (all NSAP and at-risk subjects had occupations that required repetitive low-level contractions of the ECR muscles), and not pathology. However, the at-risk subjects and NSAP subjects are not behaving in the exact same way suggesting a possible pathology in the NSAP group and not just a training effect. Interestingly, the group of subjects with LE did not show decreases in MUP size, they actually showed increases in MUP size relative to the control subjects, suggesting that the neuromuscular changes seen in individuals with NSAP are not the same as those seen in individuals with LE. The larger MUPs in the LE group are consistent with the fact that patients with LE are often thought to have cervical radiculopathy. We screened out anyone with clinical signs and symptoms of radiculopathy, but perhaps DQEMG is more sensitive than a clinical exam in detecting this condition. The reduced firing rates seen in all groups relative to the control group may not be clinically relevant as the mean firing rates all remained within a normal range. A prospective study is needed to confirm any causal relationship between smaller MUPs and NSAP as found in this work.

ACKNOWLEDGEMENTS

Financial support for this research was provided by the Workers Safety and Insurance Board of Ontario and the Natural Sciences and Engineering Research Council of Canada and is gratefully acknowledged.

Table 1: MUP and SMUP morphology and mean MU firing rates (* denotes a significant difference from parameters notated with **, † denotes a significant difference from parameters notated with ‡, ▫ denotes a significant difference from parameters notated with ▫▫)

	Control n=37	**At-risk** n=8	**LE** n=11	**NSAP** n=16
	Mean ± SD	Mean ± SD	Mean ± SD	Mean ± SD
Needle-detected MUPs				
Amplitude (μV)	491.1 ± 300.7*	444.1 ± 233.4	519.0 ± 426.6†	420.9 ± 282.4**‡
Duration (ms)	7.64 ± 2.85*	9.76 ± 2.99**‡	9.70 ± 3.16**‡	8.36 ± 2.68**†
Number of Phases	2.55 ± 0.71*	2.72 ± 0.77	2.63 ± 0.81	2.74 ± 0.82**
AAR (ms)	1.27 ± 0.43*	1.43 ± 0.39**▫	1.62 ± 0.59**†▫▫	1.34 ± 0.43**‡
Mean MU Firing rate (Hz)	14.98 ± 2.97*	14.73 ± 3.04	13.86 ± 2.71**	14.53 ± 2.68**
Surface-detected MUPs				
Amplitude (mV)	113.73 ± 90.73*	91.28 ± 49.45**	111.14 ± 109.53†	65.75 ± 41.58**‡
Area (mVms)	593.9 ± 507.4*	424.4 ± 230.1**	502.7 ± 518.7†	273.7 ± 182.6**‡
Duration (ms)	19.76 ± 5.52*	19.69 ± 4.54†	19.21 ± 5.93	17.25 ± 5.21**‡

BAYESIAN AGGREGATION FOR NON-SPECIFIC ARM PAIN CLASSIFICATION

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INTRODUCTION

Bayesian aggregation has been used successfully to characterize a set of motor unit potentials (MUPs) collected together from the same muscle. Previous work, specifically that of Pfeiffer (1999), has shown that Bayesian strategies may be used to produce a whole-muscle characterization based on the probabilistically weighted characterization of a set of quantitatively measured MUPs. We explore the accuracy of Bayesian aggregate classifications based on an assumption of a Gaussian distribution of MUP parameters.

METHODS

Data was obtained from 57 volunteer subjects, 17 of which exhibited symptoms of "non-specific" arm pain (NSAP) with exclusion of lateral epicondylitis. The remaining 40 subjects were normative. Prior to participation, all subjects provided informed written consent. Free running EMG data was collected from the *extensor carpi radialis brevis* muscle using a 32-gauge concentric needle electrode and Ag/AgCl surface electrodes located on the skin overlaying the needle acquisition site. Subjects performed repeated isometric contractions ranging between 5 and 20% of maximum voluntary contraction force, each for 30s in duration. Acquisition was done using Comperio™ clinical EMG amplifiers. Needle and surface data were bandpass filtered at 10Hz-10kHz and 5Hz-5kHz at rates of 31250 and 3125 samples/second respectively. All data was decomposed using the DQEMG program (Stashuk, 2001), producing a table of quantitative features describing each MUP train found. Table 1 shows the quantitative EMG (QEMG) features used, as well as log transformations that were performed to make the feature data more closely follow a Gaussian distribution.

Table 1: QEMG MUP Features

	Feature	**Units**
log	Amplitude	µV
	Duration	µs
	Phases	
	Turns	
log	Area/Ampl. Ratio	ms
log	Macro Ampl.	mV
log	Macro Neg. Pk. Area	mV·ms
log	Macro Neg. Pk. Ampl.	mV
	Macro Neg. Pk. Duration	ms
	IPI mean	ms
	IPI std. dev.	
	IPI covariance	
	Inter-Discharge rate	pps
	Firing Rate	pps
	Firing Rate MCD	pps
	Mean MU Voltage	µV

Three classification treatments were examined. First, each individual MUP was classified independently using a Bayesian normal density discriminant function classifier (Duda *et al* 2001, pp 41). This provides the optimal discriminant function for separation of Gaussian normal distributions, and is based on class mean and covariance values calculated based on training data. These results provide a baseline for comparison with muscle-level characterizations produced using either an application of Bayesian probabilistic aggregation, or a "winner-takes-all" voting approach. In both cases, labels were generated for all MUPs collected from a given subject based on the application of the indicated aggregation technique upon the "MUP level" data. Results may therefore be compared with the independent MUP baseline performance. Probabilities of association with each class, NSAP or No(rmative) required for the Bayesian are based on the same mean and covariance values calculated for the MUP assignments. For each classification, 10-fold cross-validation was performed.

RESULTS AND DISCUSSION

The data in Table 2 shows the summary results for this experiment. Rows describe MUPs by their true label, columns show the label assigned by each classification technique. Perf(ormance) indicates the projected specificity and sensitivity of the classifier based on the cross-validation test.

As this table shows, the accuracy of the "MUP level" classification can be greatly improved through the use of aggregate information applied over the set of MUPs collected from the same muscle.

Unexpectedly, the voting results are comparable to the results of the Bayesian aggregation, in spite of the use of class distribution information in the case of the Bayesian classifier. As discussed in Pfeiffer (1999), this additional information should allow significantly higher accuracies to be obtained; the fact that this is not the case indicates that the information used by the Bayesian technique does not accurately match the true distribution of the QEMG data. This supports the observation that the distribution underlying EMG data may be significantly different from a Gaussian normal distribution, and may be better modeled by more sophisticated techniques.

The high performance numbers seen here are very encouraging, however as the lowest sensitivities or specificities observed are still in excess of 80%; obtaining even higher accuracies will result in extremely accurate classification.

SUMMARY/CONCLUSIONS

Accuracies for Bayesian aggregation and voting classification of MUPs are quite similar, contrary to expectation. Further modeling of the underlying distributions of class parameters will be required to increase classification performance.

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ACKNOWLEDGEMENTS

Data used in this study was collected under a WSIB (Workplace Safety & Insurance Board) grant supported research project.

Table 2: 10-fold cross validation for all three techniques

		Independent MUPs			**Bayesian**			**Vote**		
True/Applied	**Totals**	**No**	**NSAP**	**Perf**	**No**	**NSAP**	**Perf**	**No**	**NSAP**	**Perf**
No(rmative):	1168	900	268	0.771	1099	69	0.941	1008	160	0.863
NSAP:	266	73	193	0.726	30	236	0.887	14	252	0.947
Totals		**973**	**461**		**1129**	**305**		**1022**	**412**	

QUANTITATIVE ELECTROMYOGRAPHIC CHANGES ASSOCIATED WITH MOTOR DEFICITS IN CARPAL TUNNEL SYNDROME

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INTRODUCTION

Carpal tunnel syndrome (CTS) is caused by localized compression of the median nerve within the carpal tunnel. Symptoms associated with CTS include paraesthesias in the sensory distribution of the median nerve, and, in severe cases, progressive weakness of the thenar muscles. Thenar weakness is likely due to chronic nerve compression, which can result in axonal interruption and subsequent degeneration. Collateral sprouting is a compensatory mechanism that may occur in cases of axonal loss whereby a sprout from a nearby motor axon adopts orphaned muscle fibers. Individuals with thenar atrophy associated with carpal tunnel syndrome are expected to exhibit fewer motor units in their abductor pollicis brevis (APB) if they have suffered from axonal loss. Changes in motor unit morphology related to reinnervation of orphaned muscle fibers through collateral sprouts may also exist, however the extent to which this process occurs in this population is not known. The purpose of this study is to determine if, in individuals with CTS, (i) motor unit number estimates (MUNEs) indicate axonal loss and (ii) APB motor unit morphology indicates that collateral sprouting has occurred.

METHODS

Volunteers from healthy and CTS populations underwent a clinical examination to rule out other pathology (such as cervical radiculopathy) as well as nerve conduction tests to confirm their diagnosis of CTS. Electrophysiological testing involved sensory and motor evoked potentials from the median, ulnar and radial nerves across the wrist. A differential setup with an inter-electrode distance of 2 cm was used to record sensory and compound nerve action potentials (SNAPs and CNAPs). Compound muscle action potentials (CMAPs) were recorded using a monopolar configuration with the active surface electrode placed over the motor point of the APB muscle. In both cases the reference electrode was located over the dorsum of the hand. SNAPs, CNAPs and CMAPs were obtained using a stimulus intensity 20% higher than the level that generated a maximal response. Subjects with CTS were included and categorized as mild or severe based on the criteria proposed by Stevens (1997). A quantitative EMG evaluation was then completed as follows. Maximum voluntary electrical activation (MVE) of the APB muscle was determined using the root mean square value over five seconds of EMG data recorded using the monopolar electrode placed over the motor point. A 32 gauge concentric needle electrode was then inserted into the APB beneath the surface electrode. Each volunteer performed submaximal contractions ranging from 5 to 50% MVE until approximately 30 suitable motor unit potentials were obtained. The needle location was changed between contractions. AcquireEMG™ and DQEMG™ software (Stashuk, 2001) was used to generate surface- and needle-detected motor unit potential (SMUPs and MUPs respectively) morphology data sets. The acquired SMUPs and MUPs were ensemble averaged for each individual respectively. MUNEs were determined using the ensemble average technique described by Boe et al. (2004). Satellite potentials were quantified as a percentage of each individual's total number of recorded MUPs. The MUNEs, percentage of satellite potentials and measures of MUP morphology (amplitude, area, duration, number of phases) were compared between the groups using Kruskall-Wallace tests with α adjusted for multiple comparisons and post hoc analysis were performed using Mann-Whitney tests.

RESULTS AND DISCUSSION

Six individuals with severe CTS, eight with mild CTS and nine healthy controls participated in the study. There were no differences in median age between the groups (Severe CTS: 53 (41.25-52.5) years, mild CTS: 46 (41.25-52.5) and control: 43 (30-53.5) years) ($p>0.05$). Significant group differences were found for MUP amplitude and duration ($p<0.017$). No group differences were found in the number of MUP phases nor were there were any group differences in any SMUP parameters (amplitude, area, duration) ($p>0.017$). The peak to peak MUNE and percentage of satellite potentials were significantly different among the groups ($p<0.05$). The severe CTS group demonstrated larger MUP amplitudes compared to the other groups, as well as longer duration MUPs compared to the mild CTS group ($p<0.017$). The severe CTS group exhibited a significantly lower peak to peak MUNE as compared to control group ($p<0.05$). Furthermore, the severe group had a larger percentage of satellite potentials compared to both the control and mild CTS groups ($p<0.05$). See Table 1 for summary data.

SUMMARY/CONCLUSIONS

The combination of MUNE and motor unit potential morphology provide valuable insight into what is occurring at the axonal level in individuals with CTS. The lower MUNE exhibited by the severe CTS group suggests that these patients had a smaller number of functioning motor units in their APB muscles. The severe CTS group also had significantly larger and longer duration MUPs; these values are associated with higher motor unit fiber density and may reflect that the APB in individuals with severe CTS has undergone collateral sprouting, thus increasing their average motor unit size. This theory is supported by the observed increased percentage of satellite potentials exhibited by those with severe CTS. SMUP parameters failed to demonstrate a significant difference between the groups, however fewer SMUPs than MUPs were found on decomposition and included in analysis, which might explain this lack of statistical significance. The number of phases in the MUPs was not different between the groups. This parameter may not be sensitive to detecting morphological differences associated with collateral sprouting. It appears that MUP amplitude and the percentage of satellite potentials may be the most responsive quantitative means of studying the progression of and recovery from peripheral nerve injury such as CTS.

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			Needle-detected MUPs			Surface-detected MUPs		
Group	% Satellite Potentials	Pk-Pk MUNE	Amplitude (µV)	Duration (ms)	No. of Phases	Amplitude (mV)	Area (mVms)	Duration (ms)
Control	0 (0-0)**	147 (110-199)**	442 (384-487)**	5.9 (5.3-7.5)	2.7 (2.3-2.9)	211 (184-294)	993 (746-1279)	25.9 (18.5-30.1)
Mild CTS	0 (0-0)**	77 (70-138)	457 (433-554)**	6.2 (5.5-7.2)**	2.3 (2.2-2.4)	319 (156-383)	1140 (551-1337)	22.9 (21.5-24.0)
Severe CTS	3.8 (1.2-4.6)*	53 (23-85)*	661 (540-774)*	7.9 (7.7-8.6)*	2.7 (2.4-2.9)	379 (200-610)	1626 (837-3048)	26.1 (24.5-29.9)

Table 1: Median MUP and SMUP measures (* denotes a significant difference from parameters notated with **, IQR refers to interquartile range).

RELATIONSHIP BETWEEN BICEPS BRACHII SURFACE EMG CHARACTERISTICS AND MOTOR UNIT FIRING BEHAVIOR

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INTRODUCTION

The surface electromyographic (EMG) signal provides a global representation of muscle activity, while indwelling EMG provides information about the firing patterns of specific motor units. Despite the global nature of the surface EMG signal, its characteristics are often used to make inferences about the firing patterns of individual motor units. However, there is little data to support the relationship between the firing behavior of individual motor units and the characteristics of the interference pattern of surface EMG. The purpose of this study was to investigate the relationship between surface EMG characteristics and motor unit firing behavior across a range of forces in the biceps brachii.

METHODS

Eleven subjects performed isometric contractions of the elbow flexors while surface EMG was monitored from the biceps brachii with a bipolar configuration, and indwelling EMG was monitored with a four-wire needle electrode inserted into the muscle. Each subject performed 10 trials of a ramp contraction from zero to 100% MVC at a rate of 10% MVC per second. Trials were divided into 21 500-ms epochs, starting at the onset of force development. The root mean squared (RMS) amplitude and the mean power frequency (MPF) of the surface EMG signal were calculated over each 500-ms epoch. The mean motor unit firing rate from the indwelling signal was calculated over the same 500 ms epochs. Orthogonal polynomial comparisons were used to examine trends in RMS amplitude, MPF, and mean motor unit firing rate over the contraction time. Linear regression analyses were used to investigate potential relationships between motor unit firing rates and the RMS amplitude and MPF of the surface signal. These regression analyses were performed using i) mean data across all subjects; ii) data from individual trials across all subjects; and iii) individual subject data.

RESULTS/DISCUSSION

A total of 162 motor units were analyzed, with recruitment thresholds ranging from 0.13-96% MVC. Orthogonal polynomial comparisons revealed a linear increase in motor unit firing rate throughout the contraction ($p<.001$). The surface EMG amplitude showed a slow increase initially, followed by a rapid increase, resulting in a statistically significant quadratic trend ($p<.001$). The MPF of the surface EMG showed an initial increase, then leveled off, also resulting in a significant quadratic trend ($p<.001$). Using the average values across all subjects, there was a strong relationship between motor unit firing rate and both RMS amplitude ($r^2=.91$) and MPF ($r^2=.58$). However, when data from individual trials were analyzed the strength of the relationship between motor unit firing rate and both RMS amplitude ($r^2=.19$) and MPF ($r^2=.017$) were much lower. Regression analyses were also performed on data for each individual subject. One subject showed a strong relationship between motor unit firing rate and RMS amplitude (r^2=.59). However, for all other subjects the r^2 coefficients were less than 0.2. In all subjects the relationship between motor unit firing rate and MPF was weak, with r^2 coefficients less than 0.008.

A.

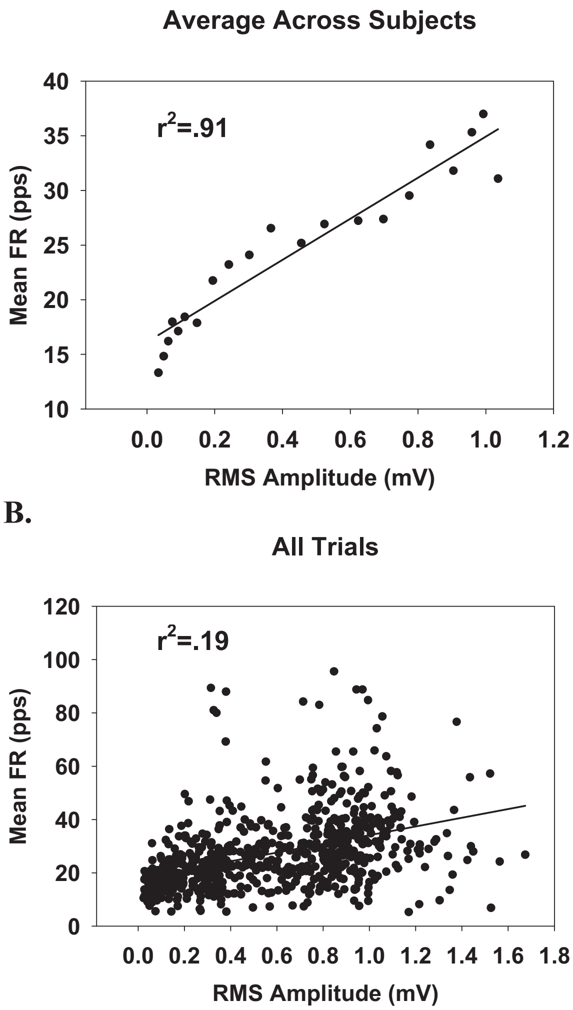

Figure 1. Relationship between RMS amplitude of surface EMG and Mean motor unit firing rate. Using the average values across subjects (A) produced a strong relationship. Using all trials (B) produced a much weaker relationship. Similar results were obtained for the relationship between MPF and motor unit firing rate.

SUMMARY/CONCLUSIONS

These results suggest that composite surface EMG amplitude and frequency characteristics do bear some relationship to motor unit firing rate. However, within an individual subject, neither EMG amplitude nor MPF can be used to predict motor unit firing rate.

COORDINATION OF MOTOR-UNIT FIRING PATTERNS IN ELBOW FLEXORS

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INTRODUCTION

The human biceps brachii (BB), brachialis (B), and brachioradialis (BR) muscles are considered synergists since they all flex the elbow. However, each muscle has a different effect on forearm pronation/supination. Kinesiological EMG studies have shown that they do not always act synergistically and sometimes act reciprocally (Naito, 2004).

One way that intermuscular coordination might manifest itself is through the presence or absence of synchronous modulation of motor-unit (MU) firing rates, referred to as "common drive" (De Luca et al., 1982).

The aim of this study was to investigate the coordination between MU firing patterns in BB, B, and BR during different elbow tasks involving elbow flexion and forearm pronation/supination.

METHODS

Three normal subjects (one male, two female) gave informed consent to participate. A pair of fine wire electrodes was inserted in each muscle (BB, B, and BR). The subjects sat with the shoulder in neutral position, the elbow flexed at 90°, and the forearm unsupported. EMG signals were recorded simultaneously from each wire while the subjects either held a 0.2 kg weight against gravity or slowly rotated the forearm between pronation and supination. The signals were decomposed into MUAP trains using the EMGlab program.

For each motor unit, the instantaneous firing rate (IFR) at each discharge was estimated as the reciprocal of the preceding inter-discharge interval. This was converted to a continuous function by linear interpolation. For the isometric contractions, the IFRs were filtered between 0.25 and 8 Hz and the common drive was estimated by computing the mean cross correlation between IFRs. For the dynamic contractions, recruitment was determined by counting the number of MUs active at each instant.

RESULTS AND DISCUSSION

Firing patterns from an isometric contraction of one subject are shown in Fig. 1. The firing patterns were highly correlated within the same muscle (mean cross-correlation 0.57), and less correlated between muscles

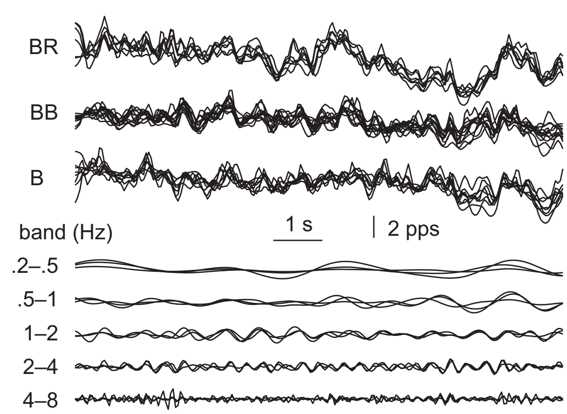

Figure 1. Top: Firing patterns during elbow flexion. The traces for each muscle have been aligned vertically to emphasize the common fluctuations. Bottom: Ensemble averages for each muscle in different frequency bands.

(0.32). Plotting the average pattern of each muscle in different frequency bands showed that some fluctuations were common across two or all three muscles, while others were unique to a specific muscle. Similar results were seen in the other subjects.

The subjects used different strategies for the dynamic contractions. The subject in Fig. 2 recruited B to pronate, BB to supinate, and BR during both transitions. In each muscle, the IFRs exhibited a common modulation pattern which each MU joined and followed soon after it was recruited. (Note that the firing patterns in Fig. 1 have been offset vertically to emphasize the common modulation.) Another subject maintained several B MUs firing at a low rate (6-8 pps) during supination and did not recruit BR at all.

CONCLUSIONS

Our results show that B, BB, and BR are not perfect synergists. During elbow flexion, the MU firing patterns suggest that each muscle receives both common and individual control signals. During forearm rotation, they are activated differently due to their different pronation/supination action.

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ACKNOWLEDGEMENTS

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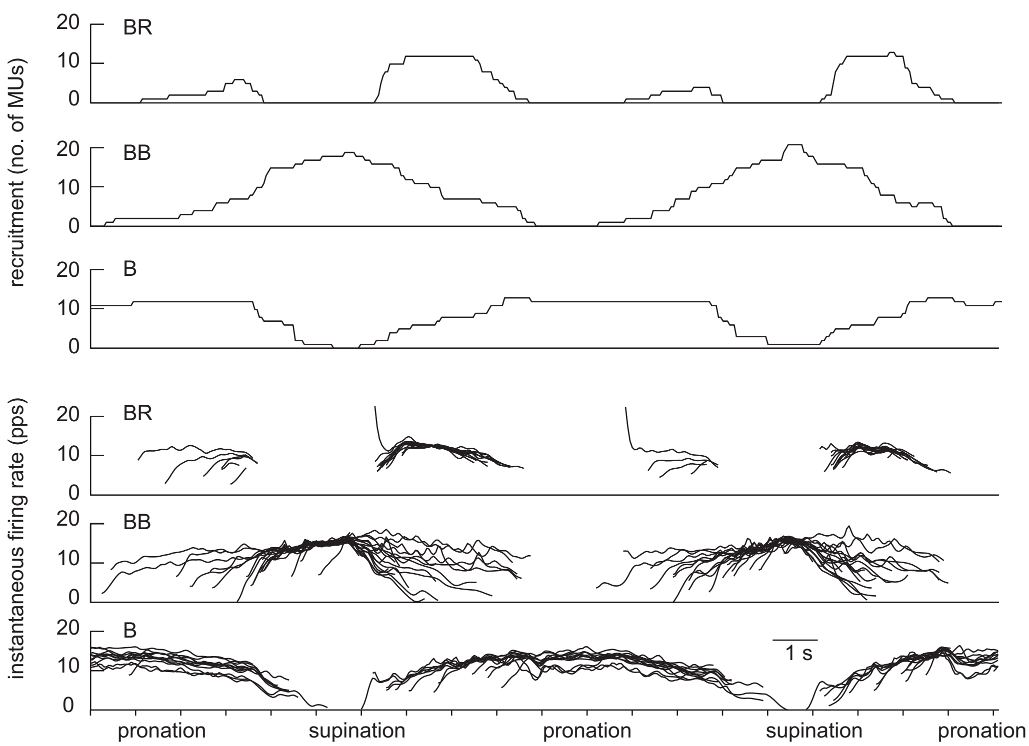

Figure 2. Recruitment and instantaneous firing rate during forearm rotation of one subject. The firing rate traces have been adjusted vertically to emphasize the common modulation pattern in each muscle. One BR MU sometimes recruited with a doublet.

SHORT TERM BED REST REDUCES CONDUCTION VELOCITY OF INDIVIDUAL MOTOR UNITS IN LEG MUSCLES

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INTRODUCTION

Space flight duration will increase with the use of the International Space Station and possible future missions to Mars. Therefore, there is a need to anticipate the medical problems, which could develop during the flight and to prepare astronauts for their return to Earth. Space permanence simulations such as prolonged bed rest (Mulder et al. 2007) can provide study conditions that are more accessible with respect to space flights. A short term bed rest experiment was organized by the German Aerospace Center (DLR), Cologne using the Head Down Tilted Bed Rest at minus 6° model, to simulate the effects of weightlessness for studying the adaptation to this condition. The aim of this study was to assess the consequences of a simulated long duration flight.

METHODS

The study was part of a comprehensive international collaboration: Short Term Bed Rest – Salty Life Study 7 (STBR-SLS7), where specific examination periods were assigned to each study group. Eight healthy, male, sedentary subjects (mean ± SD: age 26.1 ± 3.9 years; body mass index 24.6 ± 0.8 kg/m2) were recruited. Volunteers participated to the study laying in bed for two continuous periods of 14 days separated by 5 months. The participating subjects were divided in 2 groups receiving different diets (low and high sodium content). Surface EMG signals were detected before and after the bed rest periods from Vastus Medialis (VM), Vastus Lateralis (VL) and Tibialis Anterior (TA) muscles. The study protocol was divided in two phases. In the first phase four subjects received a specifically prepared diet with low sodium content (50 mmol/ day) and four subjects received a high sodium content diet (550 mmol/day) during the bed rest. In the second phase the subject groups were switched. EMG recordings were performed 2 days before (pre) and 4 hours after (post) each bed rest. Two experimental set-ups (for knee extensor and ankle flexor muscles) were prepared to test the subjects. For each muscle EMG signals were detected during a 1 minute voluntary contraction at 20% MVC with a linear electrode arrays (8 electrodes, 5 mm IED) positioned between the innervation zone and the distal tendon. The joint angle was at 120°. Subcutaneous tissue layer thickness was measured with an ultrasound scanner. Single motor unit (MU) action potentials were identified from the multi-channel surface EMG signals with the method described by Gazzoni et al. (2004), which tracks single MU action potentials over time with template update in case of progressive shape changes. This technique does not detect superimposed action potentials, thus only an incomplete firing pattern was obtained (Gazzoni et al. 2004). This is not a limitation for the current application. Muscle fiber conduction velocity (CV) was estimated with a multichannel algorithm (Farina and Merletti, 2003) on double differential signals.

RESULTS AND DISCUSSION

Since the same contraction type was repeated on each subject with the electrode arrays in the same positions (for each of the three muscles) in 4 different experimental sessions (Phase 1 pre; Phase 1 post; Phase 2 pre; Phase 2 post), the EMG signals were analysed in order to identify the same motor unit in the 4 sessions (Figure 1).

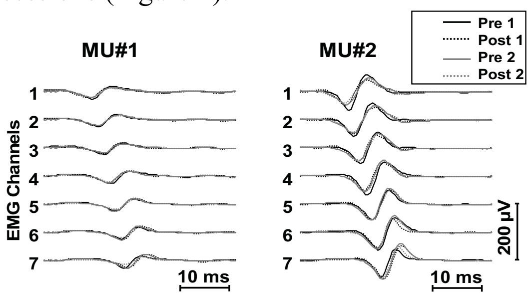

Figure 1: Superpositions of the MUAP tempates of each of two MUs identified in four different sessions on the Vastus Lateralis muscle of a subject.

The overall number of MUs identified in all subjects in the 4 sessions were 12, 12 and 10 for the vastus medialis, vastus lateralis and tibialis anterior muscle respectively.
Figure 2 shows the CV estimates of the identified MUs on the three muscles.

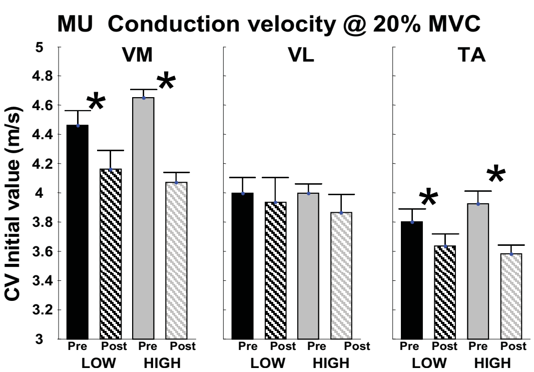

Figure 2: Conduction velocity of identified motor units. Mean ± SE are shown. Stars indicate significant differences.

Matched pairs test (Wilcoxon signed-rank test) showed statistically significant lower motor unit CV after the bed rest period for vastus medialis and tibialis anterior muscles ($p<0.05$). The same results were observed in global CV while the maximal force didn't change after the bed rest period.

No significant change was observed in the subcutaneous tissue layer thickness of each of the three muscles after the bed rest period. No significant variation of ARV and MNF after the bed rest was observed for either diet type.

SUMMARY/CONCLUSIONS

The main results of the present study can be summarized as follows: 1) it was possible to track the same motor unit in different conditions after replacement of the electrodes. 2) no differences were observed in Torque after the bed rest for either diet type; 3) no reduction of subcutaneous tissue layer thickness was observed after the bed rest for either diet type; 4) no variation of ARV and MNF after the bed rest for either diet type; 5) a reduction of global estimate of muscle fiber CV was observed after the bed rest period (around 10% for all the three muscles and all the contraction levels); 6) a reduction of single motor unit CV was observed after the bed rest period (around 10% for VM and TA muscles and around 5% for VL muscle).
As a conclusion the present study showed that conduction velocity of motor units is sensitive to small changes in muscle properties which cannot be detected with sEMG amplitude or spectral parameters.

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ACKNOWLEDGEMENTS

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BAYESIAN MUSCLE CHARACTERIZATION USING QUANTITATIVE ELECTROMYOGRAPHY

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INTRODUCTION

The majority of clinicians that use electromyographic (EMG) signals to help them characterize neuromuscular systems depend upon the qualitative examination of motor unit potentials (MUPs) detected from muscles. However, a recent study showed that faculty and residents using qualitative examinations had an overall 46.9% agreement with the actual diagnosis. Previous work done by Pfeiffer showed that using Bayes rule to determine a muscle characterization by combining the probability of each MUP in a set detected from a muscle under examination proved to be accurate at predicting underlying myopathic or neuropathic conditions. This work will show that using Bayesian aggregation of MUP characterizations estimated using pattern recognition techniques is more accurate than conventional quantitative analysis, (i.e., analysis of MUPs using the means of MUP feature values and counting the number of MUPs that have feature values that are outliers). We will also discuss how Bayesian aggregation uses the same reasoning as the examiner would that uses qualitative analysis but does not depend upon expert opinion that is a difficult skill to acquire.

METHODS

Control and neuropathic MUPs were sampled from the biceps-brachii and first dorsal interosseous muscles. MUPs were sampled from 16 healthy control subjects and 14 patients, including 9 patients with definite amyotrophic lateral sclerosis and 5 patients with Charcot-Marie-Tooth disease type X confirmed via genetic testing.
A disposable concentric needle electrode (Model N53153; Teca Corp., Hawthorne, NY) was used to acquire intramuscular signals using a Neuroscan Comperio (Neuroscan Medical Systems, El Paso,Texas) with a bandpass of 10 Hz–10 kHz at a sampling rate of 31.2 kHz during 30 second voluntary isometric contractions. The EMG data was decomposed into MUP templates using Decomposition-based Quantitative EMG (DQEMG) (Stashuk 1999).

MUPs detected from a muscle under examination were characterized by estimating both the probability that each MUP was detected from a control muscle and an abnormal muscle. MUP characterizations were based on two different pattern recognition techniques for comparison 1) Pattern Discovery (PD) (Pino et al. 2007) and 2) Linear Discriminant analysis (LDA). The MUP characterizations were combined using Bayes rule (Pfeiffer 1999) to estimate the probabilities that the muscle under examination was normal or abnormal – a muscle under examination was characterized as normal or abnormal based on the higher of those two probabilities. These results were compared with a combined quantitative means and outlier analysis. The subset of MUP features, drawn from the following: amplitude, duration, area, thickness, size index, phases and turns, that had the best accuracy for each method was also determined.

RESULTS AND DISCUSSION

Table 1 shows the best feature set as determined by accuracy for each method. B-PD and B-LDA refer to Bayesian aggregation using Pattern Discovery and Linear Discriminant analysis based MUP characterizations respectively. As Table 1 shows both Bayesian methods had better accuracy than the conventional combined method by about eight percent. Also, B-PD had the best balance between sensitivity and specificity as shown by the small difference between sensitivity and specificity.

Despite evidence that quantitative methods are more accurate, qualitative examination is still by far the most widely used method. However, Bayesian muscle characterization is based on the same reasoning as used for qualitative EMG decisions. With regard to MUP characterization, during qualitative EMG examinations the examiner subjectively estimates based on experience and training the conditional probability of a specific MUP being detected given that the muscle from which MUPs are being detected is normal or abnormal. With these estimates in mind, a clinician makes a decision about the muscle under examination by combining the individual MUP characterizations. PD or LDA based MUP characterization provides objective numerical estimates for these conditional probabilities by examination of exemplary training data. These conditional probabilities are combined by multiplication (and then normalized) to achieve an objective muscle characterization.

Experienced qualitative examiners can benefit from using Bayesian aggregation because: 1) the cognitive burden of taking into account multiple feature values simultaneously across a large number of MUPs can be reduced, 2) Bayesian muscle characterization facilitates the determination of “possible”, “probable”, or “definite” levels of diagnosis. Bayesian characterization can be used to support clinical decisions related to initial diagnosis as well as treatment and management over time.

SUMMARY/CONCLUSIONS

As a decision support system, Bayesian aggregation can aid an electromyographer in a single but important step of an EMG examination and can provide an objective record over time that facilitates longitudinal studies. Decisions are then based on facts and not impressions giving electromyography a more reliable role in the diagnosis, management, and treatment of neuromuscular disorders.

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Table 1: The sensitivity, specificity, and accuracy of the best feature set as determined by accuracy.

Method	Best Feature Set	Sens (%)	Spec (%)	Acc (%)
Combined	area/SI	86.0%	75.9%	81.0%
B-PD	dur/area/thick/SI/phases	88.0%	90.4%	89.2%
B-LDA	dur/thick/SI/turns	84.0%	95.2%	89.6%

RESOLVING SUPERIMPOSED MUAPS BY PARTICLE SWARM OPTIMIZATION

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INTRODUCTION

The EMG signal consists of trains of discrete discharges known as motor-unit action potentials (MUAPs). The motor unit (MU) includes a motor neuron, its axon and all of the muscle fibers it innervates.

EMG Decomposition is a method to identify MUAPs and MU firing times in EMG signals. It often happens, even in low level contractions, that MUs sometimes discharge almost simultaneously, resulting in the superimposition of their MUAPs. The process of determining the identities and firing times of the involved MUAPs is known as resolving the superimposition.

The superimposed waveform, *x(t),* can be formulated as below, where *n(t)* is the noise (including activity of MUs not close to the recording electrode), *n* is the maximum number of possible MUAPs involved, w_i is an index indicating whether MUAP i is involved (1) or not (0), and ϕ_i is the time of occurrence of MUAP i.

$$x(t) = n(t) + \sum_{i=1}^{n} w_i \, MUAP_i(t - \phi_i)$$

w_i and ϕ_i are estimated by minimizing the squared difference between x(t) and the sigma term (the "error norm"). This provides a valid estimation as long as the superposition is not too complex.

We have used a continuous-time stochastic population-based optimization method called Particle Swarm Optimization (PSO) (Kennedy, 2001), which is powerful and easy to implement. The results are satisfactory, making this a promising method for decomposition programs.

METHODS

The original PSO was based on the sociological behavior associated with a flock of birds looking for food. In this method a swarm consists of a population of particles, each of which represents a potential solution to the optimization problem.

In our algorithm, the number of particles in the swarm is a function of *n*, the number of possible MUAPs. One of the particles is initialized to the easily computed, but often incorrect, peel-off solution (Merletti, 2004). Half the rest are initialized and regularly updated randomly, and the rest pseudo randomly, using the Sobolian sequence to increase search diversity (Van Den Bergh, 2001). During iteration, the arithmetic crossover operator is used to increase the convergence rate.

A multi-swarm strategy is used to decrease the probability of being stuck in a local minimum. If a swarm fails to make sufficient progress in decreasing the error norm, then another swarm is created to fine-tune this possible solution, while the original swarm is re-initialized to continue exploring the entire search space. The total number of swarms is limited to n+1. The algorithm continues until the global optimum does not change. The location of the swarm with the lowest error norm is taken as the solution.

RESULTS AND DISCUSSION

We simulated superimpositions involving 2-5 MUAPs of different energies. The MUAPs came from real EMG signals sampled at 10 kHz and high-pass filtered at 1 kHz. They were shifted randomly within [-1:1] ms, and added together with random noise. Two cases were considered: the known and unknown constituent problems, in which the w_i were respectively assumed to be known or unknown. 1000 superimpositions were created for each value of *n*, and solved using the PSO algorithm. The error norm was evaluated in the frequency domain, allowing non-integer time shifts. The algorithm's accuracy and the computational complexity (measured by counting the number of error-norm evaluations) are listed in Table 1.

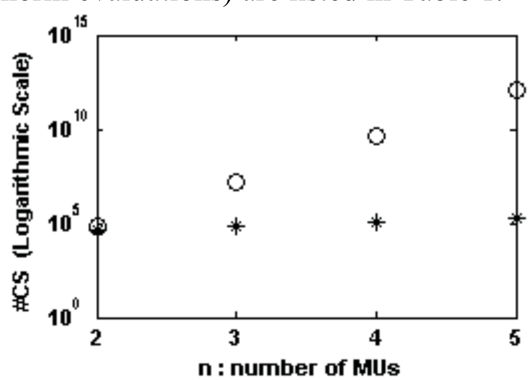

Figure 1: Complexity for PSO (star) and exhaustive search (circle) versus number of MUs involved in superposition.

The complexity is compared with the complexity of using an exhaustive search in Fig. 1. The oversampling factor used for exhaustive search was set to 2 to guarantee 0.05 msec time resolution.

Two versions of PSO were implemented in Matlab and Visual C environments. Using a vectorization package and multi-threading in Visual C resulted in an efficient version of PSO. For the 3-of-3 known-constituent set, the algorithm took an average of 2 sec to find the solution on an Intel Dual-Core 1.83GHz CPU with 2 GB of RAM.

SUMMARY/CONCLUSIONS

In this paper, a new method to resolve superimposed MUAPs is proposed. According to the results, its accuracy and efficiency are satisfactory. By using FFT interpolation, continuous time shifts are handled without explicit oversampling. The multi-swarm strategy can be implemented efficiently in a multi-threaded environment. We are working on a practical version for use in EMG decomposition programs.

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Table 1: The accuracy (in percent) and computational complexity (number of function evaluations) of the PSO algorithm.

Method	**Known Constituent Problem**				**Unknown Constituent Problem**			
	2of2	**3of3**	**4of4**	**5of5**	**2of5**	**3of5**	**4of5**	**5of5**
Accuracy	100	100	100	99	100	100	99	97
Complexity	299033	394637	469802	576692	553560	636079	688934	780601

MOTOR UNIT TRACKING WITH HIGH-DENSITY SURFACE EMG

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INTRODUCTION

Present methods of evaluating motoneuron function at best rely on samples of motor units (MUs). Yet, comparisons of properties of MU samples, drawn at various stages in a motoneuronal disease, can only indirectly determine the survival times and adaptive responses of individual motoneurons. Furthermore, with methods based on (necessarily different) samples, it is difficult to delineate the exact sequence of patho-physiological changes with time. In theory, following (tracking) individual MUs over time can address these issues, as it provides a direct determination of the pathological process in individual MUs (Doherty et al., 1994; Gooch et al., 1997; Chan et al., 1998).

High-density surface EMG provides a characteristic "fingerprint" of individual MUs (Blok et al., 2002). We hypothesize that this fingerprint allows MUs to be re-identified reliably in subsequent sessions, not only in healthy controls, but also in conditions in which MUs change over time.

METHODS

This pilot study assessed the feasibility of the MU tracking approach with high-density sEMG (HD-sEMG) in the median nerve of 10 healthy controls. Monopolar recordings were made with an array of 14x9 electrodes over the abductor pollicis brevis muscle, relative to the second metacarpophalangeal joint. Electrode positions were measured with respect to anatomical landmarks, to allow accurate replacement in subsequent sessions.

Stimulation was started at the distal wrist crease, by increasing stimulus intensity (SI) slowly from below the threshold of the first MU until approximately 5 MUs were active. All responses were recorded, using an ActiveTwo amplifier system (BioSemi Inc, Amsterdam, NL) at 8 kHz per channel. Next, stimulus electrodes were moved proximally, and the procedure was repeated at up to 15 well-defined sites along the course of the median nerve up to the axilla. Then, all electrodes were removed.

In an ensuing pause, data were analysed to determine which 6 of the studied sites were best suitable for tracking, i.e., which showed MU potentials (MUPs) that were clearly distinct from others, either in their threshold or in the characteristics of their fingerprint. We then re-applied the electrodes, stimulated at the 6 selected sites and searched for the same MUPs at that site. The fingerprint, stimulus intensity range of the MU, and the presence of F-waves helped to decide if a MUP was the same as one of the MUPs at that location in the first recording. MUPs that could be recorded again in the second measurement were deemed suitable for tracking.

On a subsequent day (at least two weeks after the initial session), the 6 stimulus sites were each studied twice, with a pause and reapplication of electrodes in between. After each recording, data were exported to Matlab® for full analysis. Composite responses were decomposed into single MU potentials and then identified using a slightly modified version of previously described MUNE software (Van Dijk et al., 2008).

RESULTS AND DISCUSSION

Compared to conventional, single-channel recordings of MUPs, the fingerprints greatly facilitate the detection and repeated recognition of MUPs. This allows more MUs to be tracked than just the ones with the lowest threshold. The fingerprints matched well between sessions, even when these sessions were several months apart (Fig. 1). However, this match is very sensitive to displacement of the recording electrode array. Compensation for shifts or rotations in this array is essential for proper interpretation of changes in pathology.

SUMMARY/CONCLUSIONS

HD-sEMG fingerprints of MUPs contain much more MU-specific information than single-channel recordings. This information aids MU tracking, a scientific approach that can provide insights into disease etiology and pathophysiology that cannot be obtained otherwise.

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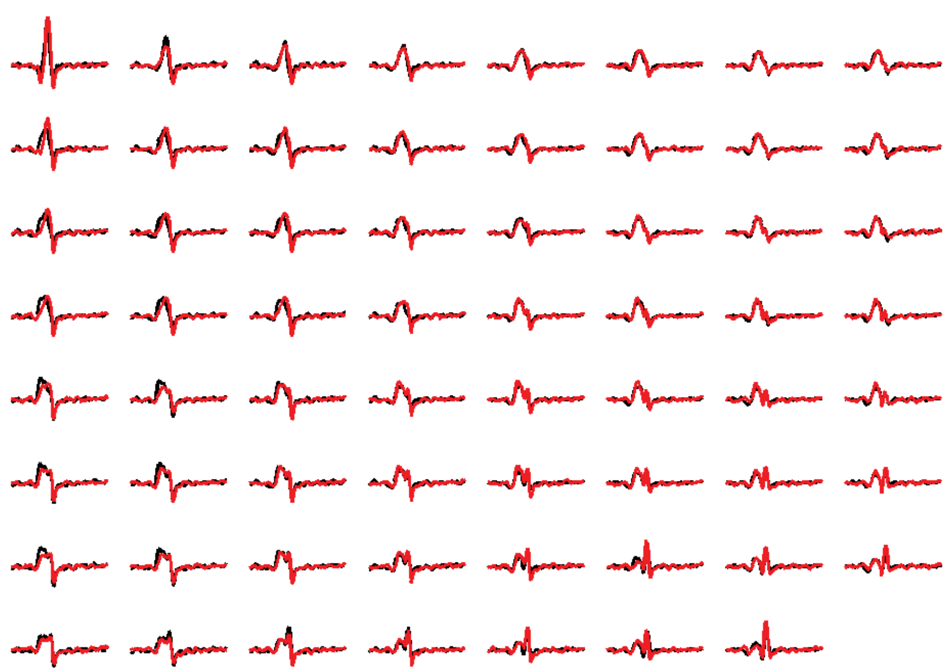

Figure 1: MU fingerprint recorded twice and re-identified in one subject. Red: original recording; black: repeat recording four months later. Each signal represents the MU potential at one of 63 electrodes, shown at the position of that electrode in the array.

DECOMPOSITION OF SURFACE EMG FROM EXTERNAL ANAL SPHINCTER

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INTRODUCTION

External anal sphincter (EAS) is one of the most frequently investigated muscles of the pelvic floor. It voluntarily controls the closing of the anal canal and plays an important role in development of faecal incontinence.

Clinical examinations of neural control of EAS are commonly based on invasive needle electromyography. Although accurate, this methodology prevents the long-term EAS observations. Electric activity of EAS can be observed noninvasively, with surface electrodes applied to mucosa in the anal canal. Recently, multi-electrode probes became available, with circumferential arrays of surface electrodes arranged in circular fashion at several intra anal depths. In contrast to the bipolar surface electrodes, multichannel arrays enable simultaneous acquisition of several surface EMG (sEMG) channels and are currently being used to asses the EAS anatomical features and gender related differences.

The aim of this study was to investigate the feasibility of computer-aided single MU analysis from clinically acquired multichannel sEMG of EAS muscle.

METHODS

Twenty-eight healthy female subjects participated to the study (age, mean ± SD: 35 ± 13 yr; stature 1.67 ± 0.07 m; body mass 62 ± 12 kg). Data was collected in Gynaecological Clinic at University of Tübingen, Germany. The experimental protocol was approved by the local ethics committee and all the subjects signed an informed consent form. Surface EMG was recorded by a 48-channel anal probe (Figure 1). The electrodes (1×10 mm) were arranged in 3 circumferential arrays of 16 contacts each. Inter-electrode and edge-to-edge inter-array distance was 2.7 mm and 5 mm, respectively.

Eleven 10 s long acquisitions were made for each subject. The subject was first asked to lie on the left side, with the anal probe inserted into the anal canal, up to the level, where the electrodes of the external array were just visible at the anal orifice. After a few minutes of rest, one acquisition was made in relaxed condition, followed by three maximum voluntary contractions (MVC) of EAS. Afterwards, another acquisition in relaxed condition was made. Subject was then placed prone and asked to activate the right and left gluteus muscles by lifting the extended right and left leg against the resistance applied by the operator at the ankle. Three lifts of each leg were performed to study crosstalk from the glutei muscles.

The acquired sEMG signals were amplified (gain set to 10000), band-pass filtered (10-500 Hz, 3 dB bandwidth), sampled at 2048 samples/s, and digitized by a 12-bit A/D converter. The gradient Convolution Kernel Compensation (CKC) method (*Holobar and Zazula* 2007) was applied to the acquired sEMG in order to reconstruct discharge patterns of individual MUs.

RESULTS AND DISCUSSION

On average, 6 ± 2 MUs per contraction were identified, regardless of the type of contraction (rest, MVC of EAS, lift of right or left leg). The number of identified MUs was 3-times larger on the external array than on the central or internal arrays ($p<0.01$). MUs identified in the rest condition had discharge rate of 8 ± 2 pulses per second (pps). In all other conditions, MUs discharged with 15 ± 5 pps. Average coefficient of variation of inter-pulse interval was 15 ± 5 %.

Multichannel MU action potentials (MUAPs) were extracted by spike triggered averaging of sEMG by using the identified discharge instants as triggers. They all exhibited stable shape over time (Figure 1), revealing the location of the innervation zone and MUAP angular conduction velocity (CV). In all the contractions, the MUs with "non-propagating MUAPs" (i.e., with CV > 850 rad/s, which corresponds to ~ 6.0 m/s at the surface of the anal probe) were detected. Their number was low, did not differ between the arrays and was significantly lower in rest than in other conditions ($p<0.01$). These MUs were excluded from further processing.

In the rest condition, average CV on the internal and central arrays was 300 ± 140 rad/s (this corresponds to 2.1 ± 1.0 m/s at the surface of the probe, and to 3.6 ± 1.7 m/s at the radial distance of 5 mm from the surface of the probe). On the external array, CV decreased to 230 ± 70 rad/s (1.6 ± 0.5 m/s when measured at the surface of the probe, and 2.8 ± 0.8 m/s at the distance of 5 mm from the surface of the probe). During MVC contractions of EAS and leg lifting, CV on the internal and central arrays increased to 350 ± 150 rad/s (2.5 ± 1.0 m/s at the surface of the probe and 4.2 ± 1.8 m/s at the radial distance of 5 mm from the surface of the probe).

CONCLUSIONS

Non-invasive analysis of single MU in EAS is feasible. The results of this study consider simultaneous co-activation of EAS and gluteus muscles in our protocol and illuminate the problem and difficulties of muscular cross-talk. Low MU CV (analysis limited to the MUs with clearly demonstrated MUAP propagation) and the differences detected by the different electrode arrays may be partially explained by anatomy of anal canal, but physiological explanation is still unsatisfactory.

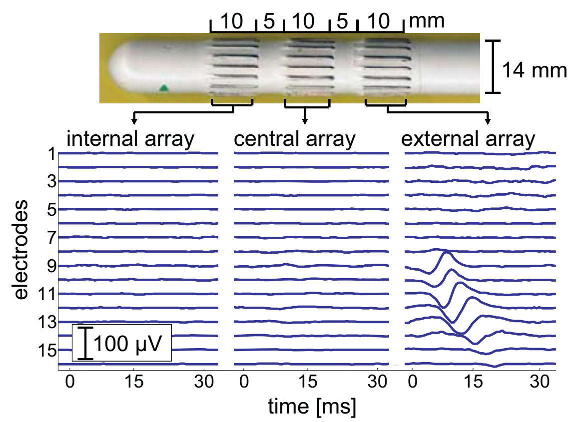

Figure 1: a 48-channel cylindrical anal probe with electrodes arranged in 3 circumferential arrays (*top plot*) and multichannel MUAP of identified MU (*bottom plot*). Surface EMG was acquired during the 10 s long maximum voluntary contraction of the external anal sphincter.

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DEMUSETOOL - A TOOL FOR DECOMPOSITION OF MULTICHANNEL SURFACE ELECTROMYOGRAMS

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INTRODUCTION

Multichannel surface electromyography (sEMG) is becoming a reliable measuring technique, enabling non-invasive, long-term and repetitive monitoring of human neuromuscular system. Recently developed surface acquisition systems simultaneously record up to several hundreds of sEMG channels and provide information about global properties and anatomy of skeletal muscles. On the other hand, there is an evident lack of computer-aided tools for non-invasive and robust assessment of central control strategies and individual motor unit (MU) properties, such as MU activation and discharge patterns, MU conduction velocity, MU fatigability etc.

The aim of the DEMUSE project (*Holobar 2007*) is to develop reliable and easy-to-use tools for decomposition of sEMG signals into constituent motor unit action potential (MUAP) trains, to enable non-invasive extraction of central control strategies, and to promote the clinical use of surface electromyography as a whole.

METHODS

DEMUSEtool was implemented in MATLAB® programme environment. It runs on standard personal computers and enables the user to load, visualize and decompose multichannel sEMG signals, inspect and manually edit automatically reconstructed MU discharge patterns and display the decomposition results. Decomposition is based on gradient Convolution Kernel Compensation (CKC) technique (*Holobar and Zazula 2007*), is fully automatic, and relies minimally on anatomic properties of the investigated muscle. Reconstructed MU discharge patterns are automatically tested against the predefined ranges of physiological variables (i.e., discharge rate, variability of inter-pulse interval, MU conduction velocity etc.) and sorted with respect to the estimated degree of decomposition reliability. Further information on DEMUSEtool is available at http://www.lisin.polito.it/DEMUSE.

RESULTS AND DISCUSSION

DEMUSEtool has already been tested on sEMG recorded from abductor pollicis brevis, biceps brachii, trapezius, vastus lateralis, vastus medialis, abductor hallucis, gastrocnemius, soleus and external anal sphincter muscles. Altogether, more then 100 subjects participated to the aforementioned experiments and more than 3000 MUs were identified. Number of identified MUs per contraction depends on the quality of sEMG signals and typically ranges from 3 to 20 MUs. Extensive tests on synthetic sEMG revealed that 95 % of discharges per identified MUs are automatically reconstructed, on average, with the accuracy of ± 1 ms. Tests on simultaneously recorded surface and intramuscular EMG are currently underway.

CONCLUSIONS

The DEMUSEtool is still under strict clinical validation. However, tests on more than 100 subjects have already shown very promising results and illuminated the potential of surface electromyography as non-invasive monitoring tool.

ACKNOWLEDGEMENT

This research was supported by a Marie Curie Intra-European Fellowship within the 6th European Community Framework Programme (Contract No. 023537).

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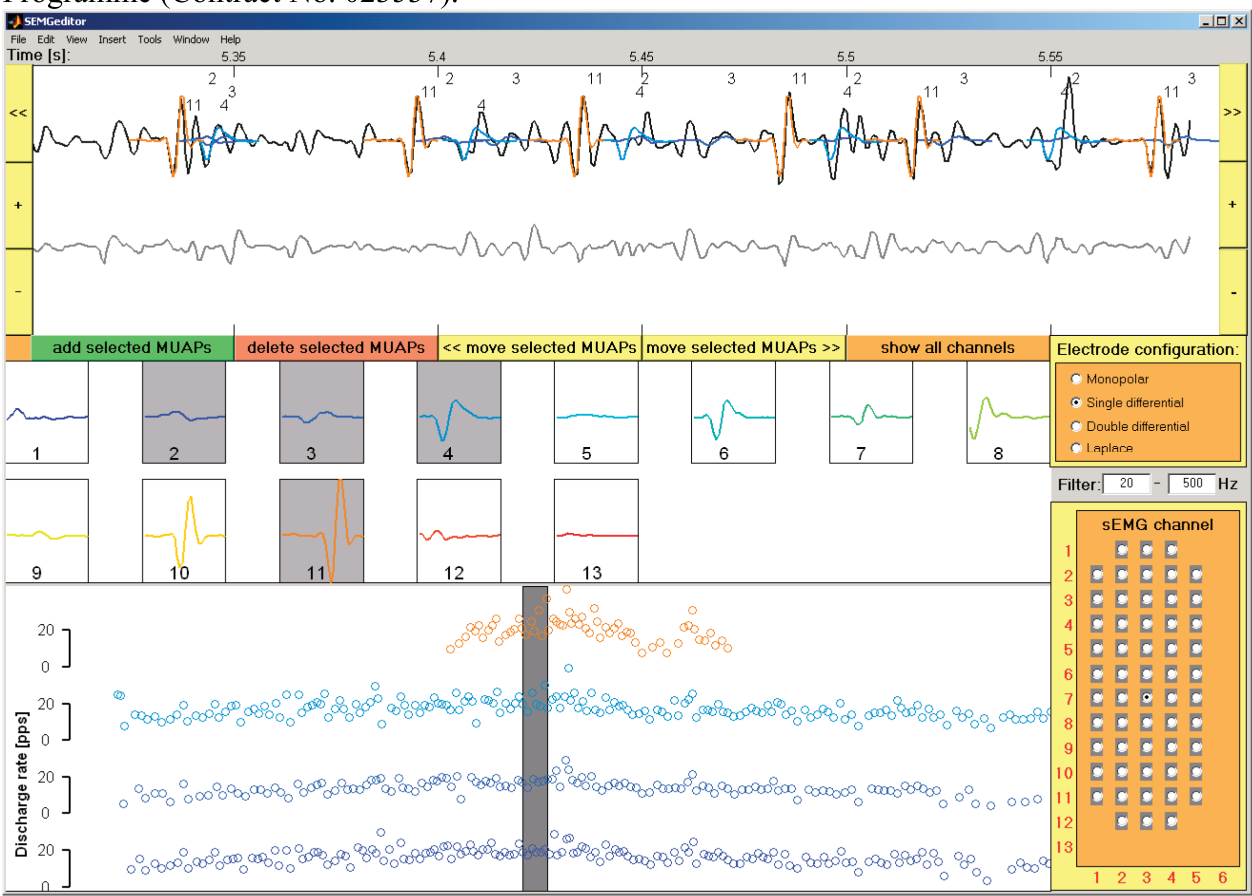

Figure 1: DEMUSEtool window for visualisation and manual editing of surface EMG decomposition results. The *top panel* displays original surface EMG channel (*black line*), and the residual after subtraction of identified MUAPs (*grey line*). The *central panel* displays identified MUAP templates, as detected in the displayed channel. MUAPs can be selected by clicking on corresponding rectangles. Selected MUAPs are depicted in grey rectangles and superimposed on the top of original EMG channel in the *top panel*. The *bottom panel* displays discharge rate plots of selected MUs. The currently displayed channel can be selected by clicking on the Channel Selection Panel (*bottom right corner*). Different electrode configurations and band-pass filtering of the displayed channel are also supported.

MOTOR UNIT RATE CODING AND NEUROMUSCULAR PHASE ADVANCE IN SINUSOIDAL ISOMETRIC CONTRACTIONS

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INTRODUCTION

To produce rapid oscillations in muscular force, the nervous system must overcome the low pass filter characteristics of muscle (Partridge). Experiments in felines demonstrate that one mechanism to overcome such muscle characteristics exists at the spinal level. The alpha motor neuron possesses a property termed 'dynamic sensitivity' in which oscillatory input of greater frequencies (>1 Hz) is amplified prior to its delivery to muscle (Baldissera et al). Additionally, there is a decrease in temporal lead (ms) between changes in motor unit (MU) firing rates and muscular force as oscillation frequencies increase (Baldissera et al., Iyer et al). The purpose of the present experiment was to extend the earlier reports on sinusoidal isometric contractions in humans (Iyer et al.) to a systematic comparison of neural advance across sinusoidal tasks with varied amplitude (x2) and frequency (x3) requirements. Consistent with the earlier work in humans and in felines, we hypothesized that oscillations of greater frequencies would be met with greater phase advance (degrees).

METHODS

Subjects produced isometric index finger abduction forces in sinusoidal force matching tasks. Six force matching conditions centered at 20 %MVC were the combinations of three frequencies (.3, .6, .9 Hz) and two amplitudes (+/- 3, 6 %MVC). In each trial, subjects viewed a plot of the target force curve on a computer monitor. Their task was to trace the target force curve with a real time plot of their own force. Isometric force and motor unit action potentials were recorded simultaneously from a strain gauge force transducer and a four-wire electrode configuration. Among multiple differentially amplified pairs of wires from this electrode, the three pairs bearing the best signal quality were used for spike sorting.

Spike sorting was conducted using custom software that codes like motor unit action potentials based on their amplitude and shape. Each file was then reviewed by an investigator using an interactive program allowing (re)assignment of spike codes and the decomposition of complex waveforms with multiple motor unit action potentials in superposition. All motor unit records analyzed were verified by a second investigator.

Motor unit rate codes (time varying firing rates) were selected for analysis if they were continuous throughout the entire trial. Cross correlation was applied to each motor unit's rate code and the corresponding force recording to obtain lag times (ms) which were then converted to phase angle (degrees). Error bars in the following figures are standard deviations. A two factor repeated measures ANOVA was used to test frequency and amplitude effects with Tukey's post-hoc comparisons as appropriate.

RESULTS AND DISCUSSION

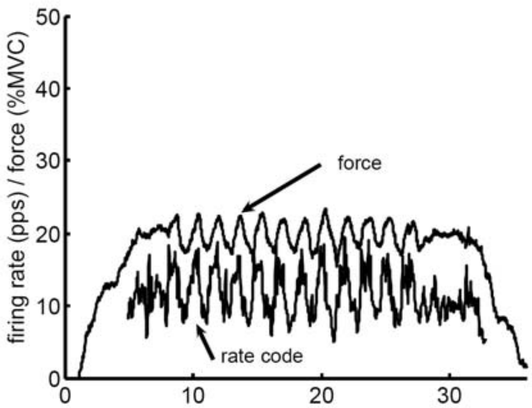

Figure 1. Sample recording of isometric force and the corresponding rate code from one motor unit. Peak correlations between individual MU rate codes and force were consistently high. Correlations (SD) were r=.88 (.1), .84 (.12), and .69 (.15) for the .3, .6 and .9 Hz task frequencies, respectively. This decline in correlations across frequencies was not significant (p=.9).

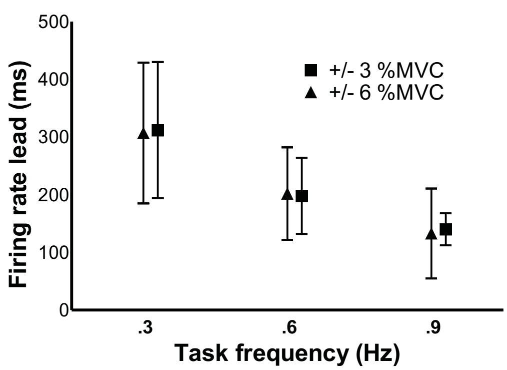

Figure 2. There was greater MU-lead (ms) with respect to force in the .3 Hz condition as compared to the .6 & .9 Hz conditions (p<.01), which were similar (p=.16). There were no effects of task amplitude (p=.32). Also for hand muscles, Iyer et al reported MU-lead values from 37-343 ms for .25 to 5 Hz tasks. The range in the present sample was 20-480 ms. Similar values of firing rate lead have been reported by De Luca et al. for triangular force varying tasks. In feline preparations with faster muscle contractile properties and more direct mechanical recordings, Baldiserra et al reported similar trends across frequencies, yet much shorter MU-lead times (56, 25, and 19 ms for 1, 5 and 10 Hz, respectively).

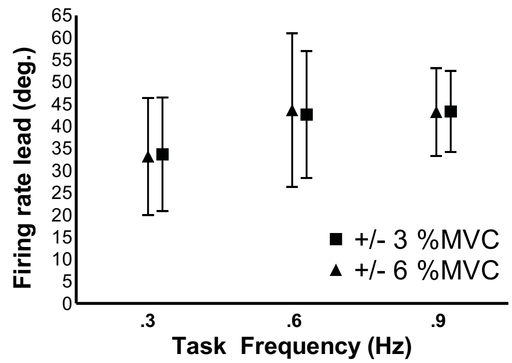

Figure 3. When expressed as phase angles (deg.), there was an increase in MU-lead for the two greater frequencies (p=.01). There were no differences in phase between the .6 and .9 Hz conditions (p=.9) and there were no effects of task amplitude (p=.16). In feline preparations in which the triceps surae were stimulated via the sciatic nerve, the muscular lag similarly ranged from ~30-75 degrees (Partridge).

SUMMARY/CONCLUSIONS

Whereas the temporal measures of neuromuscular advance are informative, phase angles allow comparison across tasks and demonstrate increasing phase advance in faster oscillations.

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EFFECT OF AUTOMATIC DECOMPOSITION ERRORS ON SYNCHRONY AND COMMON DRIVE PARAMETERS

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INTRODUCTION

Automatic decomposition of motor units (MU) often produces errors in recording the actual firing times. There are three possible types of errors while recording the firing times: failing to record or false negative (FN), recording additional firing times or false positive (FP) and the combination of missing the correct MU firing time and assigning it to the wrong MU or false negative positive (FNP). This paper studies the effect of errors on the synchrony (CIS) and the common drive (CD) parameters obtained from MU pairs recorded during twelve isometric contractions.

METHODS

MU firing times were available both from automatic decomposition (‘a’ files) and manually corrected files (‘b’ files). Manually corrected files are considered error free. For each MU in a contraction three different types of errors were randomly introduced on the ‘b’ files. For each type of error, five different rates (1, 2, 6, 10, 20 error/second) were introduced and for each case 25 random realizations were created. CD and CIS values were calculated for every realization and for the original files according to [1] [2]. CD and CIS averages and standard deviations were computed over the 25 realizations for each MU within a contraction.

RESULTS AND DISCUSSION

Synchrony

Table 1 below shows the change in CIS for different error rates with respect to the original ‘b’ file for all 3 error types. CIS values were lower than the original for the FN fault and decreased as the error rate increased. CIS values were higher than original for the FP and FNP faults and increased as the error rate increased. Figure 1 shows the change in CIS values with respect to ‘b’ file values for a rate of 6 errors per 5 seconds for all three false types.

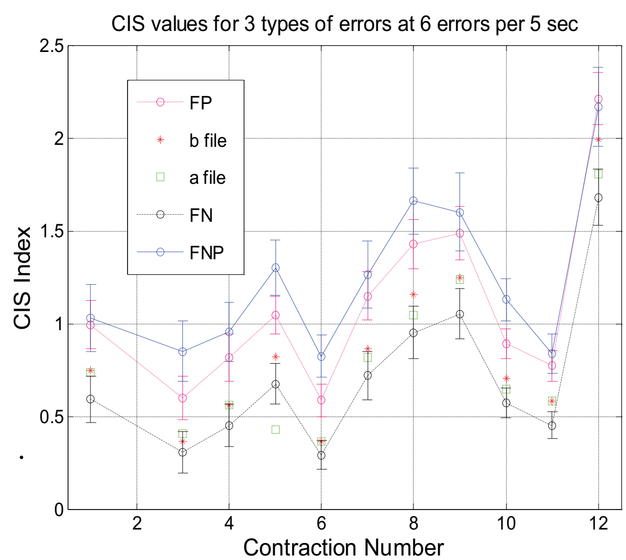

Figure 1: CIS values for original ‘a’ & ‘b’ files and FN, FP & FNP errors. Error bars show the standard deviation over the 25 realizations.

CIS values are different for ‘a’ files and ‘b’ files for most contractions. A CIS increase from an ‘a’ file to a ‘b’ file can be explained by the presence of the FN type of error in the ‘a’ file. Similarly, a CIS decrease can be explained by the presence of FP or FNP faults.

Common Drive

When errors were added common drive values for a MU pair drastically dropped even for a small error rate irrespective of the type of error. Rarely common drive values were higher than original CD values, this happened only for a few realizations of the added random noise while the average CD values for the 25 realizations were always lower than the original CD of the 'b' files. CD values decreased to a small value around 0.25 for all error types as the error rate increases. This can be observed in figure 2.

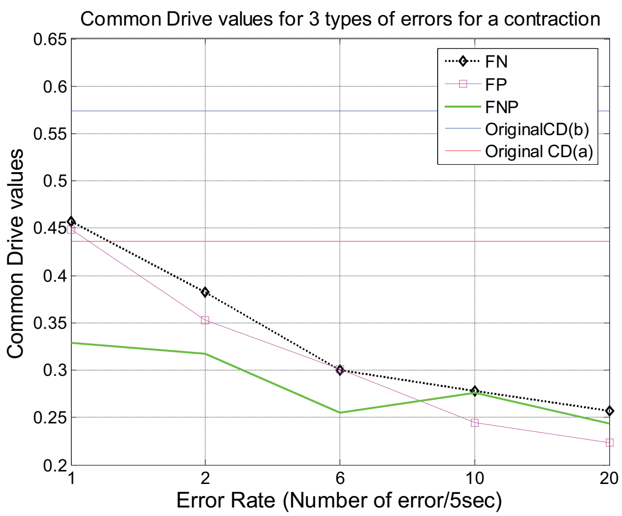

Figure 2: Common drive values obtained for different error rates, accompanied by original CD values for manual decomposition 'b' files (blue line) and automatic decomposition 'a' files (red line) for a MU pair in contraction 12.

It can be observed that the CD from manual decomposition is higher than the CD for the automatic decomposition as expected, 0.57 vs 0.44. Also, the automatic CD value, 0.44, intercepts the FN and FP lines around an error rate of 1 per 5 seconds. This suggests that the automatic decomposition suffered from 1 FN or FP error every 5 seconds.

SUMMARY/CONCLUSIONS

The errors present in the automatically decomposed 'a' files will alter the CIS index as follows. The FN type of fault will decrease the CIS value and the FP & FNP types will increase the CIS value. The change in CIS value is proportional to the amount of error for all three types of error.

The CD index is more sensitive to error than the CIS index especially for low error rates. The CD index always decreases for any of the three types of error studied. As error rate is increased its value tends to converge to a small value.

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ACKNOWLEDGEMENTS

This work has been made possible in part by NSERC grant No. 262282.

Table 1: The values in the table are the difference between the mean of CIS values of the MU pairs in contraction 12 with faults and the original mean CIS values in the 'b' file.

FAULT TYPE	FAULT RATE				
	1/5s	2/5s	6/5s	10/5s	20/5s
FN	-0.053744528	-0.110151	-0.224249	-0.4856316	-0.8554968
FP	0.036526405	0.070844	0.221925	0.3839971	0.8551058
FNP	0.019067583	0.076509	0.175964	0.2948032	0.7091116

DESIGN OF THIN-FILM ELECTRODES FOR MULTI-CHANNEL INTRAMUSCULAR RECORDINGS

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INTRODUCTION

The analysis of motor units *in-vivo* is classically performed on recordings obtained using indwelling intramuscular electrodes. These recordings have high selectivity which makes the separation of motor unit action potentials possible even with relatively high contraction forces. However, the high selectivity limits the number of motor units that can be concurrently investigated which hinders the understanding of the behavior of functionally significant populations of motor units.

The number of detected motor units can be increased by recording from different locations within the muscle (spatial sampling). We thus propose the use of thin-film technology to realize flexible high-density multi-channel intramuscular electrodes. This technology allows the placement of a number of detection sites onto a small and extremely flexible substrate (Yoshida et al. 2000). This study describes the development of two thin-film electrode systems for intramuscular EMG, the test of these systems for mechanical stress and sterilization, and recordings from animal preparations.

METHODS

Design of two thin-film systems

Thin-film systems are developed using microfabrication techniques. A silicon wafer was used as production platform for the polyimide-based electrodes. A base layer, a 5 µm thick layer of polyimide, was spin coated onto the wafer. Connection pads, electrodes, and conductive tracks, that connected the electrodes to the connector, were deposited by platinum sputtering. A second polyimide layer insulated the tracks.

A first prototype was developed for subcutaneous recordings (Farina et al. 2008). The structure of this system is shown in Figure 1.

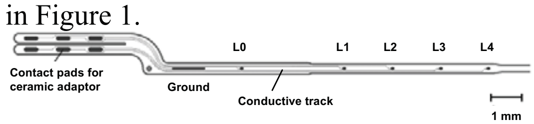

Figure 1: First prototype design showing one side of the system with layout of the electrode, pad and site positions. The system carries a ground electrode (GND), an indifferent recording electrode (L0), and the recording sites (L1-L4).

The device is 220 µm wide, 10 µm thick, 1.5 cm long, and has 8 circular platinum recording sites, each with a diameter of 40 µm distributed along the front and back surfaces, with 1500 µm inter-site spacing. All tracks are 10 µm wide, 300 nm thick, and are made of platinum. The structure was linked to an 80 µm diameter tungsten needle. The placement of the structure into the muscle was performed by inserting the tungsten needle through the epimysium and parymysium, and threading it through the endomysium.

A second prototype with 16 detection sites was developed for insertion through the skin into the muscle with a standard needle, as it is done for classic wire or needle recordings.

This system is based on 10 µm thick polyimide as substrate material with 500 µm inter-site spacing. Platinum electrodes and gold connection lines on the polymer layer are implemented, as in the first prototype. A connection pad at the end of the flexible electrode substrate is connected to a screen printed ceramic with wires or plug connector as interface for the amplier (Figure 2).

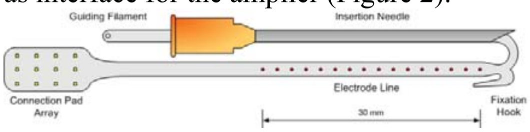

Figure 2: Second prototype, schematic view including the insertion needle. The system has 16 detection sites, located along a line. The guiding filament is placed into a needle to penetrate the muscle tissue.

The single platinum electrode area is oval shaped with a length of 140 µm and a width of 40 µm (total area ~5256 µm^2). The system has a pitch of 2000 µm for stabilization into the muscle and the total length of the electrode line is 30 mm. For insertion into the muscle a guiding filament of thickness 200 µm was designed (Figure 2) which can be positioned into a small needle (e.g. 25G). By inserting the needle into the tissue the polyimide based electrode system follows the insertion direction. When the needle is removed, the fixation hook stabilizes the flexible system inside the muscle. This solution avoids direct insertion of the electrode structure into the needle cannula, thus the thickness of the electrode system does not impose the insertion cannula size.

Test of the systems

After production, the integrity of the thin-film systems was tested under different sterilization protocols, including autoclaving. Moreover, the systems have been tested under different conditions of mechanical stress. The EMG recording quality was tested for the first prototype in 6 experiments where the electrode was implanted into the medial head of the gastrocnemius muscle of rabbits. Asynchronous motor unit activity was induced by eliciting the withdrawal reflex or by sequential crushing of the sciatic nerve.

RESULTS

The systems were not affected by the sterilization procedures and showed optimal tolerance to mechanical stress, such as insertion in rubber material. From the experimental recordings it was possible to detect a total of 67 motor units. Motor unit action potentials were identified with a decomposition algorithm and characterized in terms of amplitude with respect to the noise level. In the bandwidth 200 Hz – 5 kHz, the peak-to-peak amplitude of the action potentials of the detected motor units was 75 ± 12 µV and the root mean square of the noise was 1.6 ± 0.4 µV. The noise level and action potential amplitude were similar for recording periods of up to 40 min.

SUMMARY/CONCLUSIONS

This study proposes two prototypes of thin-film systems for intramuscular EMG recordings. The first 8-channel prototype (Farina et al. 2008) is limited to subcutaneous insertion. The second prototype can be used for intramuscular insertions with standard needles and has 16 recording sites. The guiding filament for insertion allows the placement of thicker systems with the same needle size. The quality of the EMG signal detected with the proposed thin-film systems is similar to that of classic recording systems. In conclusion, the study demonstrates that thin-film is a viable technology for a new generation of high-density, multi-channel intramuscular EMG recordings.

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MOTOR CONTROL OF LOW THRESHOLD TRAPEZIUS MOTOR UNITS IN SUSTAINED CONTRACTIONS OF VARIABLE STRENGTH

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INTRODUCTION

There has been considerable interest in motor control of trapezius low threshold motor units, due to the frequently occurring muscular pain in the shoulder region. Trapezius motor unit firing behavior is also of general interest as an example of postural motor control, as most motor control studies are carried out on extremity muscles. Previous studies (e.g., Westad et al. 2004) have examined motor unit firing behavior at low force levels. This study supplements such insight by describing firing behavior also at high force levels.

METHODS

Four subjects (2 males, 2 females; age ranging from 22 to 60 yrs) out of 9 were successfully recorded. The study protocol was approved by the Ethics Committee of Boston University and carried out according to the declaration of Helsinki.

The intramuscular EMG signal was recorded by a quadrifillar wire electrode, bonding together 4 50-μm nylon coated nickel-chrome alloy wires. The wire bundle was cut transversely, exposing the cross-section of the wires, and inserted into the muscle by use of a hypodermic needle. Force in performing shoulder shrugs was estimated by use of the RMS-detected surface EMG (sEMG) signal, calibrated by the sEMG response at maximal voluntary contraction (% EMG_{max}). Ramp (0-12 % EMG_{max}), triangular and trapezoid force-profile contractions were performed. Peak load ranged from 20 to 100 % EMG_{max}. Single motor units were detected by use of an updated version of the signal decomposition algorithm that has been used in the previous studies (Mambrito and De Luca 1984). High force contractions proved difficult to decompose, but manual inspection and editing ensured reliable trains of firing.

RESULTS AND DISCUSSION

Low-threshold trapezius motor units sustain relatively high firing rates at low force, but show very little further modulation of firing rates with increasing force contractions. Inter-individual differences in firing behavior were also noted. This is illustrated in Figure 1 that shows low-threshold motor unit responses of two subjects (A; slight, sedentary female and B; strong male, active ice hockey player) representing observed extremes in firing behavior.

Firing rates for the female subject increased from ~10 pulses per second (pps) at 3 % EMG_{max} to ~13 pps at 40 % and ~14 % at 60 % EMG_{max}. Corresponding results for the male subject were ~15 pps at 3 % EMG_{max}, ~19 pps at 40 % and ~22 pps at 60 % EMG_{max} (responses at 60 % EMG_{max} not shown). Two further subjects with successful recordings presented intermediate firing behavior, but closer to the behavior of the female subject.

Higher threshold motor units attained higher firing rates than the low threshold motor units in high force contractions for 3 of 4 subjects (not illustrated), trapezius motor

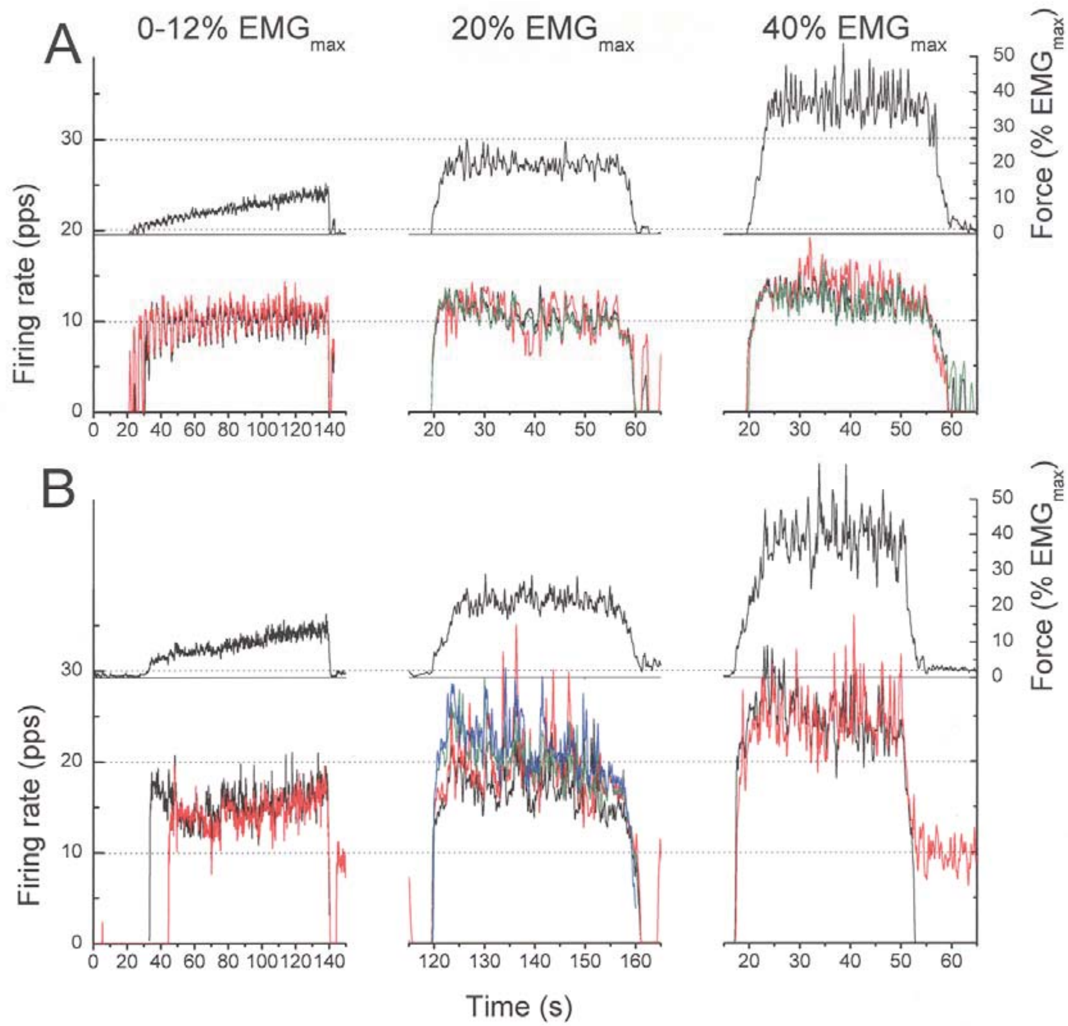

Figure 1: Two subjects (A, B) demonstrating the observed extremes in firing behavior of low-threshold trapezius motor units during low level and moderately high force contractions (top of figure). Note the strong respiratory modulation of motor unit firing in upper ramp contraction (A, left panel) and shift in recruitment threshold (from <1 to 5% EMG_{max}) of red colored motor unit in lower ramp contraction (B, left panel). Firing rates are low-pass filtered at 0.5 Hz.

units thereby deviating from the "onion skin" firing behavior (De Luca and Erim 1994).

SUMMARY

Suppression of low threshold trapezius motor unit firing rates in sustained high force contractions is indicated. Inter-individual differences in firing rates may be due to differences in the distribution of type I and type II motor units.

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MUSCLE ACTIVITY DURING LINEARLY VARYING ISOMETRIC CONTRACTION: ELECTROMYOGRAPHICAL ASPECTS

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INTRODUCTION

The motor unit (MU) activation strategy (MUAS) of human muscle during voluntary contraction has been widely investigated by means of surface electromyogram (EMG) in which the motor unit action potential (MUAP) of each active MU is summated. As a consequence EMG parameters (root mean square (RMS) and mean frequency (MF)) can be used to track the MU recruitment (REC) and the MU firing rate (FR) modulation as a function of the requested muscle tension output (expressed as % of the maximal voluntary contraction, MVC). Indeed the plateauing of MF beyond a given % MVC seems to monitor the end of REC and the use of FR as the sole tool to modulate output force. During voluntary contraction REC takes place according to the Hennemann size principle that gives a specific recruitment threshold to each MU corresponding to a given %MVC. Indeed it seems that MUs have a dynamic threshold too, corresponding to the velocity (%MVC/s) of the output tension changes (2).

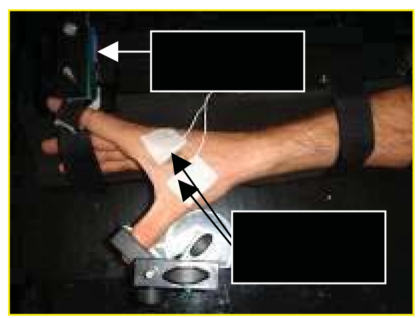
Moreover the motor units deactivation strategy (MUDS) has not been widely investigated, in particular its changes as a function of the velocity of tension decrease have not been described.

On these bases this work is aimed to investigate, by means of EMG-RMS and EMG-MF vs %MVC relationships, possible differences in MUAS and MUDS during on-going (OG) and down-going (DG) isometric ramps and their possible changes with different %MVC/s.

METHODS

According to the principles of the 1964 Helsinki Declaration on humans beings scientific research studies and after fully information about the aim and the experimental procedure 12 male subjects (25-33 years old with no neuromuscular diseases) volunteered to participate in the study. The investigated muscle was the first dorsal interosseus (FDI). The experimental set-up is represented in figure1.

Figure 1.Experimental set-up



The force produced during FDI abduction was recorded by a load cell. The surface EMG was detected by two silver bars (10 x 5 mm) 10 mm spaced. The requested output tension (%MVC) was provided on a PC screen together with the force from the subject for a visual feedback. After MVC determination (highest of three consecutive efforts lasting 3 s and with 1 minute interval in between them) the subject performed: a) three trapezoid isometric contractions: 0-10 -0%, 0-50-0% and 0-100-0 %MVC having 1 %MVC/s, 6.66 %MVC/s and 13.3%MVC/s, respectively (3 minutes between each of them), 10%, 50 and 100% were all hold for 1 s before DG ramp began; b) 10 sustained 6 s long contractions (9 steps, 10 to 90%

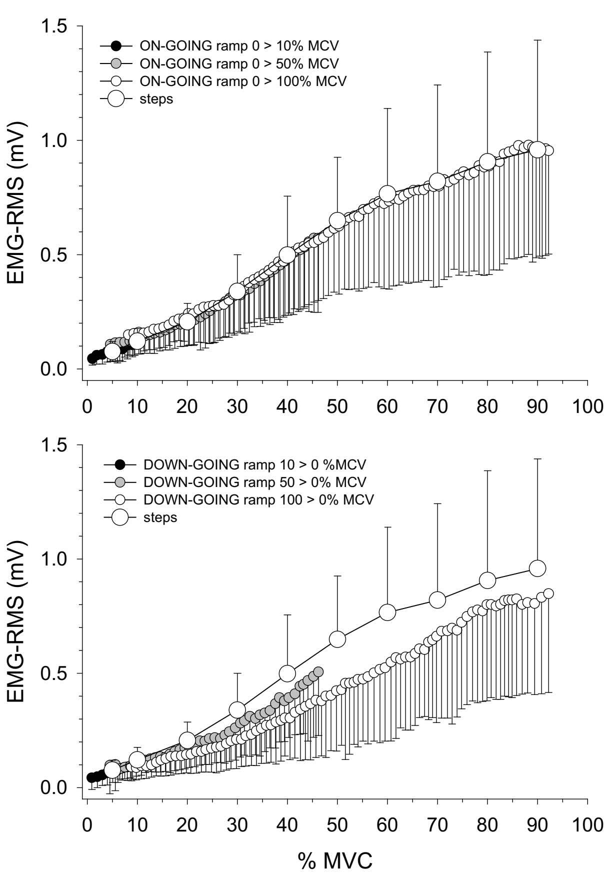
MVC step 10, plus 5% MVC). The EMG was filtered (bandpass 10-500 Hz) stored (1000 Hz) in a PC and off-line processed for RMS and spectral MF calculation. The signal time windows for processing were 500 ms long and centered every 1%MVC during ramps and 1 s long (the most central one) during steps.

RESULTS

From figure 2 it is evident that, out of 10%-0% MVC, the EMG-RMS values throughout DG ramps are different from those of the steps.

From figure 3 it is clear that appreciable differences between steps and ramps EMG-MF-% MVC relationships can be found mostly during DG ramps. The differences depend on the maximum effort of the previous OG ramp (10%, 50% or 100% MVC) as well as on the force decrement velocity (1%, 6.66%, 13.3 %MVC/s).

Figure 2. EMG-RMS vs %MVC relationship during ramps and steps.



DISCUSSION

Our data suggest that the MUAS is not different during steps or ramps. On the contrary MUDS is influenced by the end level of the previous OG ramp and by the de-contraction velocity. The fact that the motor programs during force reduction are not the mirrors of the ones used during OG phase may be due to the MUs mechanical hysteresis (1). This phenomenon provides, at a given FR, more force during DG phase. As a consequence a lower REC, influencing both RMS and MF, may be needed.

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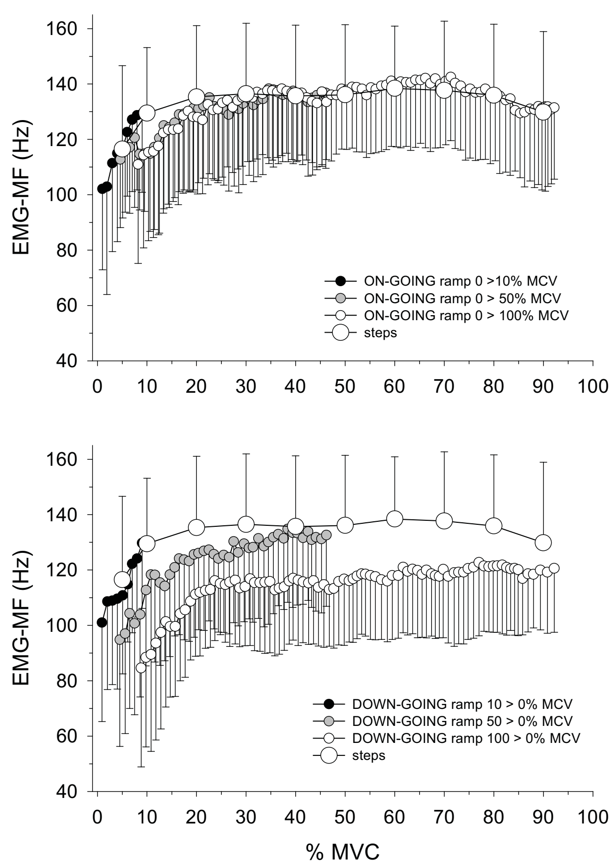
Figure 3. EMG-MF vs %MVC relationship during ramps and steps.



EFFECT OF SMALL MOTOR UNIT POTENTIALS ON THE MOTOR UNIT NUMBER ESTIMATE

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INTRODUCTION

The number of motor units (MUs) in a muscle provides an important piece of information in identifying and monitoring neurogenic diseases. Most motor unit number estimation techniques use a sample of single motor unit potentials (MUPs) to estimate the total population of MUPs in a muscle. Single MUPs are obtained by electrical stimulation of the nerve at low intensity or by voluntary activation of the muscle in combination with needle EMG. A maximal compound muscle action potential (CMAP) is obtained by stimulating the nerve so that all MUs are active simultaneously. The motor unit number estimate (MUNE) is determined by dividing the CMAP by the mean MUP.

The accuracy of the estimate depends on several factors: the number of samples used for the mean MUP, the distribution of amplitudes of the MUPs, the representativity of the sampled mean MUP, and the size of the population the sample is taken from. The influence of samples size has been shown to effect the variability of the MUNE and differs for different MUP amplitude distributions.[1] It has been suggested that small MUPs would have a disproportionate influence on the MUNE. It is therefore suggested that MUPs with a negative peak amplitude of less than 10 μV are omitted from the mean MUP. [2]

Using High-Density surface EMG it is possible to obtain a large sample of single MUPs.[3] By applying many electrode contacts, densely spaced over the muscle, small MUPs can be detected more easily than by using a conventional large electrode.

The main goal of this study was to get more insight in the effects on the MUNE when small MUPs are omitted in healthy subjects and in ALS patients.

METHODS

We obtained single MUPs from 8 healthy subjects and from 7 ALS patients. Data were recorded using a High-Density electrode grid placed transversally over the thenar muscles. All single MUPs were combined to form a simulated muscle's MUP population. It was assumed that by combining all the MUPs obtained from different subjects, a valid distribution of MUPs from an individual muscle could be created. From 8 healthy subjects we retrieved a total of 208 MUPs. A total of 130 MUPs were obtained in the ALS group.

A conventional, single large electrode equivalent (LE) was generated for each MUP by averaging the signals from the electrodes in a 1 cm x 3 cm rectangle. The rectangle was chosen so that the LE-CMAP had the largest amplitude. MUPs with a negative peak amplitude less than 10μV were marked as small MUPs.

Using a random drawing process the MUNE was calculated for different sample sizes with and without the small MUPs. The sample size was varied from 10-50 MUPs and the process was repeated 5000 per sample size.

RESULTS AND DISCUSSION

In the healthy population 27% of all MUPs were small units according to the consensus criteria. As expected variability of the estimate expressed as the coefficient of variation (CoV) decreased with increasing sample size (Fig. 1A). However, MUNE determined without the small MUPs was marginally less variable but introduced a significant error with respect to the true number of motor units. MUNE values dropped about 24% at a sample size of 20 (Fig. 1B).

In ALS, only 12% of the total population of 130 MUPs did not meet the consensus criteria. MUNE dropped about 12% if the small MUPs where omitted from the mean MUP (Fig. 1C,D).

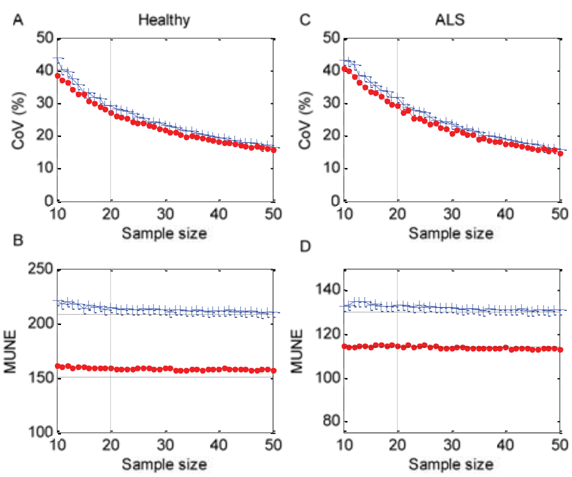

Figure 1: Variability of the estimate expressed as the CoV for the healthy (A) and ALS (C) distribution as a function of the sample size. (B+D) MUNE as function of the sample size for both distributions. Blue line includes the small MUPs, small MUPs are omitted in red line.

By omitting MUPs, the distribution of MUPs can be narrowed somewhat and hence reproducibility can be improved. CoV improved by about 2% at a sample size of 20 (Fig 1).

However, since more small MUPs are present in the healthy muscle, the distinction between ALS and healthy distributions is less if the small MUPs are omitted (Fig. 2).

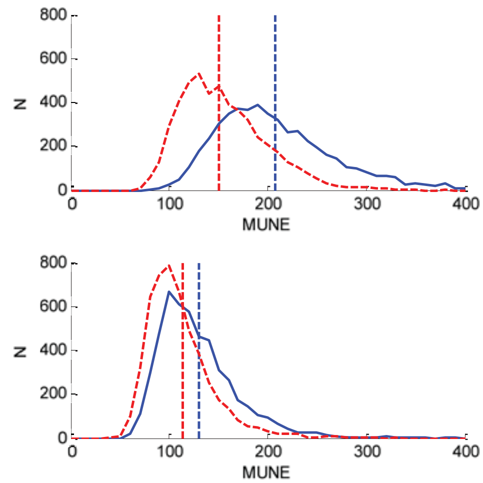

Figure 2: Upper graph, healthy distribution of MUNEs, lower graph ALS distribution of MUNEs with (blue) and without (red) the small MUPs at a sample size of 20. Dashed line indicates the mean value.

SUMMARY/CONCLUSIONS

In conclusion, we showed that omitting small MUPs can improve the reproducibility of MUNE techniques to some extend but that it reduces the accuracy. The exclusion of small MUPs might cause MUNE techniques to become less sensitive to reinnervation in follow-up studies (Fig. 2). Therefore, it is suggested to incorporate small MUPs in the estimate.

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STRATEGIES FOR MOTOR UNIT OPTIMISATION DURING PAIN

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INTRODUCTION

Motor control is altered during pain however the neuromuscular mechanisms that underlie these changes are not clear. Single motor units (SMU) are the smallest functional component of the motor system. The force output of a muscle is dependant on the number, type and firing frequency of these units.

Muscle pain is associated with decreased motor unit firing rate during constant force contractions. Recent data (Tucker et al, 2007) show that in a simple system multiple motor unit recruitment strategies are used to over come this. In a muscle with no synergists, the reduction in firing rate occurs in a select population of SMU's. Firing of other motor units appears to be inhibited. The reduced/inhibited firing is compensated by recruitment of a different population of units. Given the redundancy of the human motor system, very few muscles act without synergists. This study aimed to determine if similar motor control strategies exist in a more complex muscle system.

METHODS

Seven subjects (25.7±3.0 years, mean±SD) with no history of knee pain sat semi-reclined with knee flexed at 30^o (Figure 1a). Fine wire electrodes were inserted into the vastus medialis obliquus (VMO) and vastus lateralis (VL) (Figure 1b).

Subjects isometrically contracted their vastii to produce constant firing of between 5-15 SMU's. Auditory feedback of SMU firing was provided from one of the VMO channels. The firing rate of this SMU was maintained between 6-12 Hz for up to 5 min. Short breaks (~ 10-30s) were provided throughout the trial when the SMU firing was difficult to control or the subject began to feel fatigued. Contractions were repeated before and during induced pain.

Pain was induced via a single bolus injection of hypertonic saline (0.2 mL of 5% NaCl) into the medial portion of the patella fat pad (Bennell et al., 2004) (Figure 1b). Subjects reported their pain on an electronic visual analogue scale (VAS) throughout the pain condition.

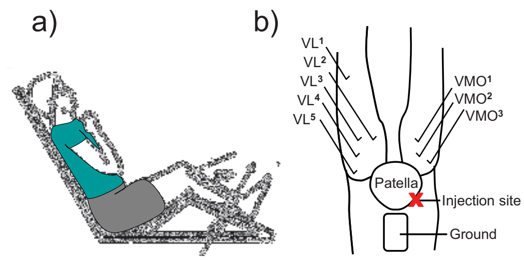

Figure 1: *a) Subjects sat semi reclined with their knees 30^o flexed. Their right leg was fixed just above the ankle to provide resistance during isometric contraction of the vastii muscles. b) Intramuscular EMG was recorded from 8 fine wire electrodes inserted into the vastus lateralis (VL$^{1-5}$) and vastus medialis oblique (VMO$^{1-3}$).*

Analysis: Thirty seconds of data from each trial was selected for analysis based on the best force match between conditions. SMU's were discriminated (using Spike2), and their electrical profile compared to determine if they were present in both conditions.

The number of active SMU's and the firing frequency of *i*) all discriminated units and *ii*) only those units recruited in both conditions were compared. In addition, an estimation of gross muscle activity was determined from the sum of all threshold crossings in each of the discriminated channels. Due to the similar results, data are shown for the VMO and VL combined. In all cases a t-test was used to determine significance between conditions, with $p < 0.05$.

RESULTS AND DISCUSSION

Pain: The sensation following hypertonic saline injection into the infrapatellar fat pad is similar to clinical anterior knee pain (Bennell et al., 2004). In the current study subjects reported pain of 3.5±0.7 /10 (VAS: mean±SD) that did not spread to the muscular tissue surrounding the knee.

Force matching: By selecting only 30 s of data from the complete trace we are able to ensure that the force produced was the same between trials (no pain: 8.5 ±2.0 pain: 8.4±2.1 % MVC).

SMU firing properties: A total of 124 SMU's were discriminated for this study. Of the 124 units, only 38 were present in both the no pain and pain trials. During pain, the firing frequency of these 38 units decreased significantly (Table 1). Twenty-eight units that were recruited during the no pain trial were not recruited during pain, and replaced with 58 new units during pain. In contrast to previous work (Tucker et al, 2007), the firing frequency of the whole motor unit pool was significantly lower during pain than during the no pain task (Table 1). The combination of recruitment of new units, and the reduction in firing frequency of all recruited units resulted in no change to the gross muscle activity (threshold crossings).

Table 1: Population characteristics of SMU's discriminated during 30 s of force matched contraction.

** p < 0.001*	No Pain			Pain	
	#	Mean (SEM)		#	Mean (SEM)
Firing frequency all units (Hz)	66	8.37 (0.30)	*	91	6.77 (0.24)
Firing frequency same units (Hz)	38	8.98 (0.34)	*	38	7.29 (0.37)
Threshold crossings (#)		*2159.1 (304.9)*			2470.9 (305.0)

SUMMARY/CONCLUSIONS

Motor optimization during pain is multi facetted. It involves inhibition which results in the de-recruitment of a sub-population of units and reduced firing frequency of the remaining units. In addition there is a facilitation that results in a shift in the population of units used to produce force. It is likely that this *"motor optimisation"* strategy produces a mechanical advantage that allows force production while limiting aggravation of the painful tissue.

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RELATIONSHIPS BETWEEN MOTOR UNIT SIZE AND RECRUITMENT THRESHOLD IN YOUNG AND OLDER ADULTS

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INTRODUCTION

As individuals age, some muscle fibers lose their innervation. When this occurs, remaining motoneurons may re-innervate these muscle fibers, producing larger motor units. This change in motor unit architecture may make the gradation of precise muscular force difficult in older individuals. However, it is unclear whether re-innervation is similar among muscles in the body. A technique known as macro electromyography (macro EMG) affords the opportunity to assess motor unit size using electrophysiological techniques. The purpose of this study was to compare the macro EMG amplitude of young and older individuals in two different muscles, the first dorsal interosseous (FDI) and the tibialis anterior (TA). This study also examined the relationship between macro EMG amplitude and recruitment threshold in both muscles.

METHODS

Sixteen individuals, eight young (22.4 yrs) and eight older (73.8 yrs) adults participated in the study. In each of the FDI and TA muscles, subjects completed 8-15 30 second isometric contractions requiring peak forces of 10-50% of maximal voluntary contraction (MVC). Motor unit activity was recorded during the contractions using a four-wire needle electrode. Peak-to-peak (p-p) amplitude, area, and recruitment threshold were calculated for each motor unit using spike-triggered averaging.

RESULTS AND DISCUSSION

Motor unit size computed using p-p amplitude correlated quite highly (r=.85) with the estimate obtained macro area analysis, supporting the use of p-p amplitude for further analysis. Macro EMG amplitudes were significantly larger in the FDI than in the TA muscle ($p<.05$), and this was particularly evident in the older adults. Overall, macro EMG amplitudes were significantly greater in the older than in the young subjects ($p < .001$). The correlation between macro EMG amplitude and recruitment threshold was higher in the young than in the older group in both the FDI (r=.66 vs r=.11, respectively) and the TA muscle (r=.61 vs r=.41, respectively).

SUMMARY/CONCLUSIONS

In young adults, macro EMG amplitude is correlated with recruitment threshold, supporting the use of macro EMG to estimate motor unit size and the size principle. However, in older adults, this recruitment threshold vs motor unit size relationship begins to decrease, particularly in the FDI muscle of the hand. It is possible that with advancing age, a variety of factors determine the force at which motor units are recruited

EMGLAB WEBSITE FOR EMG DECOMPOSITION

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INTRODUCTION

The website http://emglab.stanford.com is a forum for sharing software, data, and information related to EMG decomposition. Its goals are to promote decomposition as a research tool, to promote the exchange of EMG data, to encourage attention to accuracy, and to encourage algorithm innovation.

STATUS

Open-source software includes the EMGlab program for viewing and manually decomposing EMG signals (McGill, 2005), an automatic decomposition program (Florestal et al., 2006), and an EMG simulator (Hamilton-Wright and Stashuk, 2005).

The online database contains sample signals from nine institutions illustrating a variety of recording techniques, muscles, and experimental conditions. It includes a set of over 1000 signals from healthy subjects and patients with neuromuscular disorders (Nikolic, 2001). A subset of signals will have "blue ribbon" annotations verified by a panel of experts.

We are developing standards for exchanging EMG decomposition results (annotations), and for estimating and reporting the accuracy of decomposition results.

We tested several existing decomposition methods using the same real EMG signals. All the methods tested were able to reliably identify and extract MUAP trains.

CONCLUSIONS

The EMGlab website is intended to be a resource for the EMG decomposition community. We welcome contributions of software and data, and we hope to encourage advances in the art and practice of EMG decomposition

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P. Bonato, A. Hamilton-Wright, J. Florestal, G. Kamen, Z. Lateva, H. Marateb, P. Mathieu, M. Nikolic, D. Stashuk, Z. Tumbev, J. van Dijk, and A. Velisar have contributed to the website. This work was funded by NIH grant 5-R01-NS051507.

MOTOR UNIT DISCHARGE IN SOLEUS MUSCLE AND CORRELOGRAM WITH CENTER OF PRESSURE PARAMETERS

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INTRODUCTION

During postural sway sEMG activity in soleus muscles is highly correlated to the displacement of CoP (Gatev et al., 1999). Different strategies of motor units (MUs) have been studied during postural contractions (Mochizuki et al., 2005). However, relationships between MUs discharge and CoP behavior have not been deeply studied.

METHODS

Instrumentation: Fine Wire EMG signals were acquired using a pre-amplifier “MA-416, Motion Labs System”. CoP data were acquired in a AMTI plate force.
Protocol: Needle was inserted in soleus muscle. Subjects stood on the plate force with opened eyes first and closed eyes latter (OE and CE stages, 30 s each one).
Signal Processing: EMG was processed in EMGLAB software (McGill et al., 2005). From this software were extracted the prototype template, times of discharge of each template and the plots of instantaneous discharge frequency (IDF). The numbers of each template is randomized. Cross-correlation for discrete signals was made between CoPy parameters (displacement and velocity) with every IDF plot. Spike Triggered Average (STA) was used to study the behavior of CoPy when an occurrence of MU is present (windows of 200 ms). Both parameters were developed in IGOR Pro Wavemetrics 5.05A software.

RESULTS AND DISCUSSION

The global result is that all MUs increase the number of discharges in the CE stage as compared with OE stage. Discharge time plot shows how MUs presents different activity in the control posture (fig. 1). Some MUs were characterized by their continuous discharge (MU1, MU4 in fig. 1, 2a and 2b) in both stages. These MUs presents the greatest quantity of spikes and mean discharge frequency in both stages. There was MUs that only were recruited during CE (MU6, fig. 1). Cross-correlogram of CoPy and IFD plots reveals that a strong correlation exists between these parameters (fig 3a and 3b). In OE stage, MU1 and MU4 showed the best correlations (0.97 and 0.68, respectively). Paradoxically these MUs presents the greatest mean frequency and they are active during almost all OE stage. In this case MUs seems to be grouped according to level of correlation (fig. 3a) with the other MUs showing lowest correlation. In CE stage, MU1, MU4 and MU6 showed the best correlations (0.96, 0.95 and 0.91 respectively). The others MUs progressively decrease their correlation values. In this way MUs form a correlation pattern related with their type of activity (quantity of discharges). Poor correlation was found between CoPy velocity and MUs discharge plots in both stages. STA analysis shows the behavior of CoPy when a spike of MU is present. MUs seems to have more preference for certain CoPy locations (no figures in abstract).

SUMMARY/CONCLUSIONS

MUs have strong correlation with CoPy displacement. The correlation curves followed a pattern that is related to the discharge characters of the UMs. In postural contractions during human balance exists an interplay of different soleus MUs to maintain the CoPy in a specific distribution.

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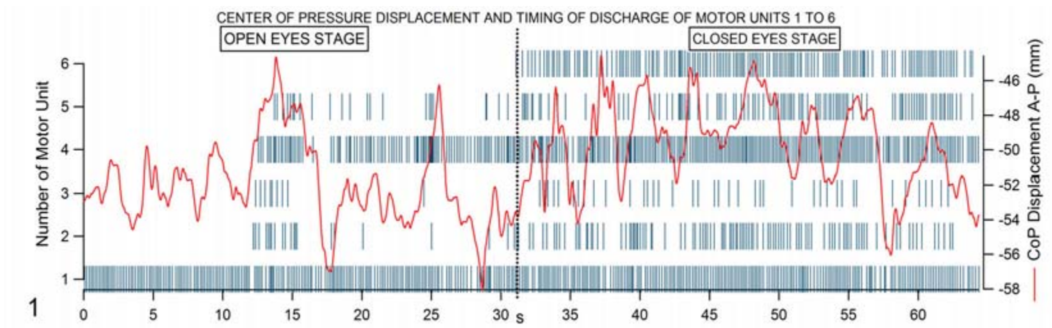

Figure 1: Counts spikes of MUs in OE and CE Stages (blue lines) and behavior of CoPy displacement (red line). MU1 and MU4 presented the greatest quantity of spikes during the test (397 and 348 spikes in OE and CE).

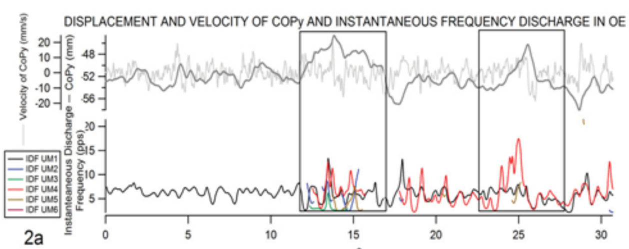

Figure 2a: IFD plots in OE stage. Relation between CoP and MU firings (frames).

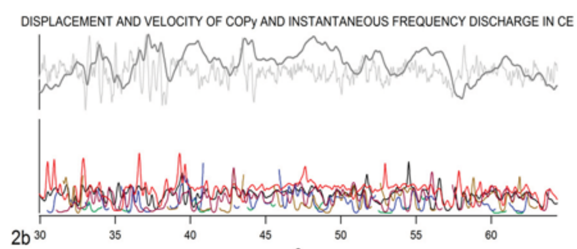

Figure 2b: IFD plots in CE stage. See relations of UM4 (red) with CoPy (gray).

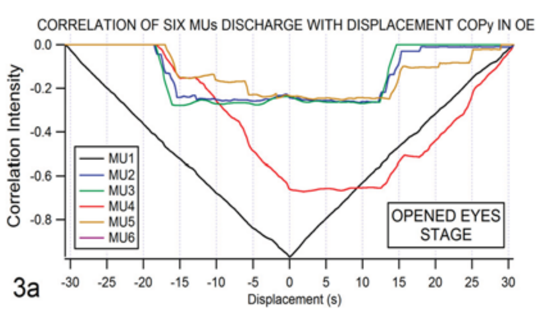

Figure 3a: Correlation of all MUs discharge plots with CoPy displacement. MU1 and MU4 showed the best correlations (0.97 and 0.68). MUs seems to be grouped in this case according to level of correlation.

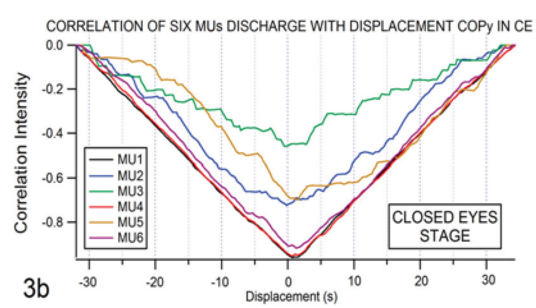

Figure 3b: Correlation of MUs discharge plots with CoP displacement. MU1, MU4 and MU6 showed the best correlations (0.96, 0.95 and 0.91). UMs form a correlation pattern related with their type of activity.

SPATIAL VARIABILITY OF CORTICOMUSCULAR COHERENCE

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INTRODUCTION

Coherence spectrum analysis between signals recorded from the brain (ECoGs, MEG and EEG) and spinal motor-neurons (EMG) is often used to measure the strength and the frequency of the communication between supraspinal and spinal motor systems (Salenius 1997).
Corticomuscular coherence has been studied in healthy subjects and in subjects affected by movement disorders (Grosse 2002). However, the high intra-subject variability of the technique makes the interpretation of the results difficult (Pohja 2005).
The coherence function depends on the cross-spectrum and the autospectra of the two analyzed signals. Therefore the intra-subject variability of coherence may be explained by dependency of the frequency content of the surface EMG signal on the muscle region analyzed.
In this study we assessed the degree of variability of the corticomuscular coherence over different locations on the muscle using join high density multi-channel EEG and EMG recordings.

METHODS

Ten subjects (6 male and 4 female; age range 24-36 yrs) participated in the experiment. Subjects were comfortably seated in a chair. The right ankle was attached with Velcro fasteners to a mechanical force strength measurement system. Subjects were asked to perform isometric dorsi-flexions at forces corresponding to 5 %, 10 % and 15 % of the maximal voluntary contraction (MVC) force. Each force level was maintained during 5 contractions of 1 min duration, separated by 1 min of rest.

Surface EEG and EMG signals were recorded simultaneously from the scalp and the tibialis anterior muscle respectively. EEG activity was recorded using the standard 10-20 electrode placement with 128 tin electrodes (reference right mastoid) but only the ten electrodes placed over the leg motor area were further analyzed. Surface EMG was recorded using a semi-disposable adhesive grid of 64 electrodes (5-mm inter-electrode distance).
During each contraction, both signals were segmented into non-overlapping epochs of 1 s duration. Each EEG signal was filtered, re-referenced to the average of all EEG signals, and visually analyzed to remove epochs with eye/muscle artifacts. Sixty longitudinal rectified bipolar signals were extracted from the grid of EMG electrodes.
Coherence between EEG and full-wave rectified EMG signals was calculated as previously described (Amjad 1989). The confidence limit was provided by the test described in (Amjad 1989). Coherence levels were considered significant when they exceeded the 95% confidence level. The coherence values were calculated between all pairs of 10 EEG electrodes and the 60 differential EMG channels in the frequency range 15-35 Hz. This resulted in 600 EEG-EMG coherence pairs in the selected frequency range for each contraction level. The variability of coherence values was calculated as standard deviation and

coefficient of variation across subjects (inter-subject variability) and across electrode locations (intra-subject variability).

RESULTS AND DISCUSSION

Table 1 reports the coherence magnitude peak and frequency peak values for all subjects during the 5% contraction levels averaged over the selected EEG/EMG channels. The results for the other two contraction levels were consistent with those shown in Table 1. Only significant (95% confidence level) peaks were averaged. The table reports the mean value μ, the standard deviation σ of the magnitude and frequency peak coherence values across the EMG grid, averaged over the EEG and the coefficient of variation CV (expressed in %) of the values. The mean values of the two variables are in the range of previously reported data (Grosse 2002). The results for single subject cases showed that the CV of the magnitude peak over the EMG grid was 25.1 ± 4.3 % and 5.3 ± 1.9 % for the frequency peak values.

SUMMARY/CONCLUSIONS

This study provides data on the variability of the corticomuscular coherence across locations over the tibialis anterior muscle. The results showed a high degree of variability (Table 1) that can partly explain the poor reproducibility of coherence data. The frequency of the peaks of corticomuscular coherence showed lower variability than the peak values.

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ACKNOWLEDGEMENTS

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Table 1: Magnitude and frequency peak values in the range 15-35 Hz (95 % CL) for 5 % MVC contraction level, averaged over the EMG matrix.

Subject	Magnitude Peak			Frequency Peak		
	μ	σ	CV (%)	μ [Hz]	σ [Hz]	CV (%)
1	0.06	0.013	20.8	25.8	1.1	4.1
2	0.08	0.015	19.9	21.1	1.5	7.3
3	0.02	0.006	26.3	26.2	2.8	9.2
4	0.03	0.010	31.1	22.6	0.8	3.6
5	0.10	0.025	25.4	24.5	1.0	4.1
6	0.03	0.010	29.3	24.4	1.2	4.9
7	0.04	0.007	18.6	25.6	1.5	5.8
8	0.05	0.013	27.3	22.6	1.4	6.3
9	0.02	0.006	29.3	26.2	1.1	4.2
10	0.03	0.006	22.9	24.3	0.7	3.0

Track 06

Motor Control (MC)

THE EFFECTS OF STOCHASTIC RESONANCE STIMULATION ON SPINE PROPRIOCEPTION AND POSTURAL CONTROL IN LOW BACK PAIN PATIENTS

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INTRODUCTION

Decreased spine proprioception (Brumagne S. et al. 2000) and larger postural sway have been found in LBP patients (Radebold A. et al. 2001).

Recently, there has been interest into the use of stochastic resonance (SR), a form of white noise, to improve neuro-sensitivity. SR is based on the principle that low-level noise sensitizes sensory systems in a way that reduces the stimulus threshold (Collins J. et al., 1996).

The primary goal of this study was to determine if SR could improve spine proprioception and postural control in chronic LBP patients. An ancillary aim was to determine what level of stimulation results in the best improvement in performance.

METHODS

Eighteen subjects with chronic LBP volunteered for the study.

Electrical stimulation threshold (EST) and vibration perception threshold (VPT) used for SR stimulation were determined using specially designed electrodes (Afferent Corporation, Providence, RI) that were capable of applying both electrical stimulation and mechanical vibration. Two electrode pairs were placed diagonally over the paraspinal muscle belly, above and below L4/L5.

EST was determined by applying current to the electrodes. Band-pass filtered, 0-500 Hz, white noise was used to generate the SR electrical stimulus. Starting with electrical stimulation that the subjects could perceive, the intensity of the signal was then decreased by 10 % until the subject could no longer perceive the stimulus. VPT was determined using a similar approach.

Spine proprioception, measured by subjects' perception of motion, was assessed in three orthopaedic planes (Fig.1). A stepper motor was used to slowly rotate subjects away from the neutral position. Subjects pressed a hand-held button when they first perceived a change in position.

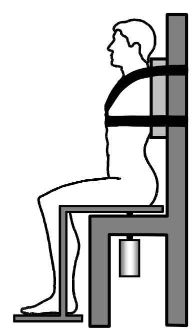

Figure 1: Subject positioned in the spine proprioception testing apparatus in a seated position (transverse plane).

Postural control was assessed using an unstable seat with a hemisphere attached to the bottom (Fig. 2). Subjects balanced with eyes closed on the most challenging size hemisphere they could manage while center-of-pressure (CoP) was recorded with a force plate beneath the seat.

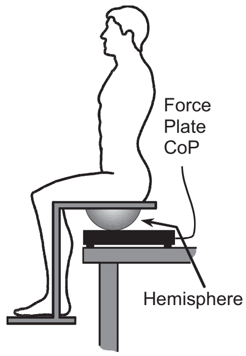

Figure 2: Subject positioned in the unstable sitting apparatus.

Spine proprioception and postural control was assessed with four conditions: device with 0 (not active), 25, 50, and 90 % of EST. When the device was active, mechanical vibration at 90 % of VPT was superimposed on the electrical stimulation.

RESULTS AND DISCUSSION

No significant differences in spine proprioception were observed between SR stimulation levels for any of the three orthopaedic planes (Fig. 3).

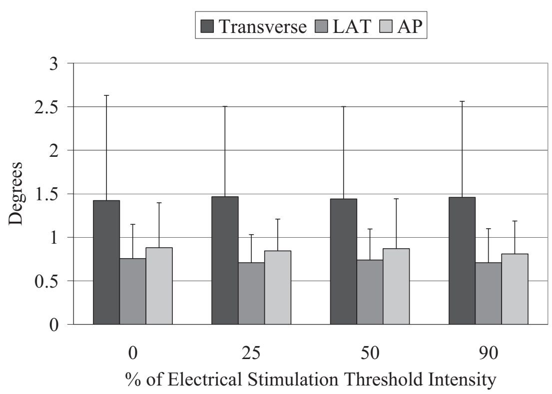

Figure 3: Spine proprioception in the three planes of motion at the four levels of SR stimulation.

SR stimulation significantly improved postural control (Fig. 4). No differences in postural control were observed between stimulation levels 25, 50, and 90 %.

Furthermore, there was no correlation between spine proprioception and postural control.

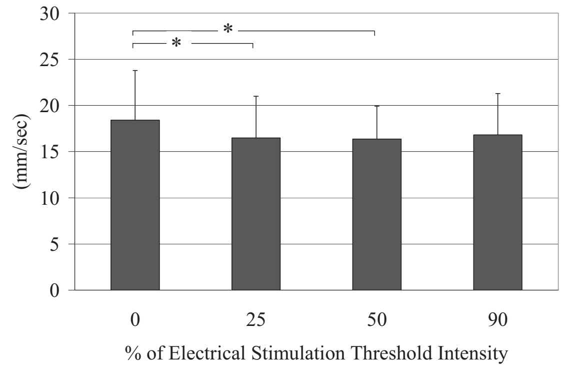

Figure 4: CoP velocity at the four levels of SR stimulation. * represents significance of $p<0.05$.

SUMMARY/CONCLUSIONS

Results suggest that SR stimulation to the paraspinal muscles can improve postural control; however, this improvement cannot be attributed to improved spine proprioception based on the current study. The study does suggest that people with compromised neuromuscular control, such as chronic low back pain patients, could possibly benefit from SR stimulation during postural activities.

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ACKNOWLEDGEMENTS

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THE EFFECT OF ACHILLES TENDON VIBRATION ON ANKLE MUSCLE ACTIVITY IN LOCOMOTION OF STROKE PATIENTS

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INTRODUCTION

Inability to correctly differentiate and respond to conflicting somatosensory inputs for stance control can be observed in stroke patients (Bonan et al. 2004). This deficit has been attributed to altered sensory organization (Rode et al. 1997). Similar problems might also occur in stroke patients for locomotion control. However, related information is scarce.

Mechanical vibration can selectively activate the muscular receptors and elicit conflicting somatosensory inputs (Goodwin et al. 1972). Courtine et al. reported that conflicting somatosensory inputs induced by Achilles tendon vibration had little effect on locomotion control of healthy young adults. The purpose of this study was to investigate the effect of conflicting somatosensory inputs from the unaffected Achilles tendon on the ankle muscular activity during locomotion in stroke patients.

METHODS

Thirty ambulatory stroke patients walked on an 8 meters walkway at their comfortable speeds with and without vibration the unaffected Achilles tendon. The electromyographic (EMG) activity of bilateral tibialis anterior and medial gastrocnemius was recorded using surface electrodes. Mean integrated EMG (IEMG) was calculated through dividing the integration value of processed EMG signal by the duration of the each subphase. Footswitches were applied to bilateral center of heel and big toe to determine the gait phases.

Multivariate repeated measure ANOVA (vibration × gait phase) were conducted. LSD procedure was used as post-hoc tests. Significance level was set at $p < 0.05$.

RESULTS AND DISCUSSION

The IEMG of the ankle muscles showed significant vibration x gait phase interaction ($p=0.041$). During the initial double ($p=0.027$) and single ($p=0.016$) support phases, the IEMG of the affected tibialis anterior was significantly larger in the vibration than in nonvibration condition (Fig. 1a). During the single support phase, larger IEMG in the medial gastrocnemius ($p=0.067$) was also found in vibration than nonvibration condition (Fig. 1).

The ankle muscle activation level changed significantly with conflicting somatosensory inputs during locomotion in stroke patients. This finding contradicts to what was found in healthy adults for stance control, and suggests that altered sensory organization for locomotion control might occur in stroke patients.

SUMMARY/CONCLUSIONS

Additional, conflicting somatosensory inputs from the unaffected Achilles tendon changed the level of muscle activation during locomotion in stroke patients. Such change could be brought about by altered sensory reorganization ability after stroke. It is suggested that sensory organization ability for locomotion control should be taken into consideration in gait training for stroke patients.

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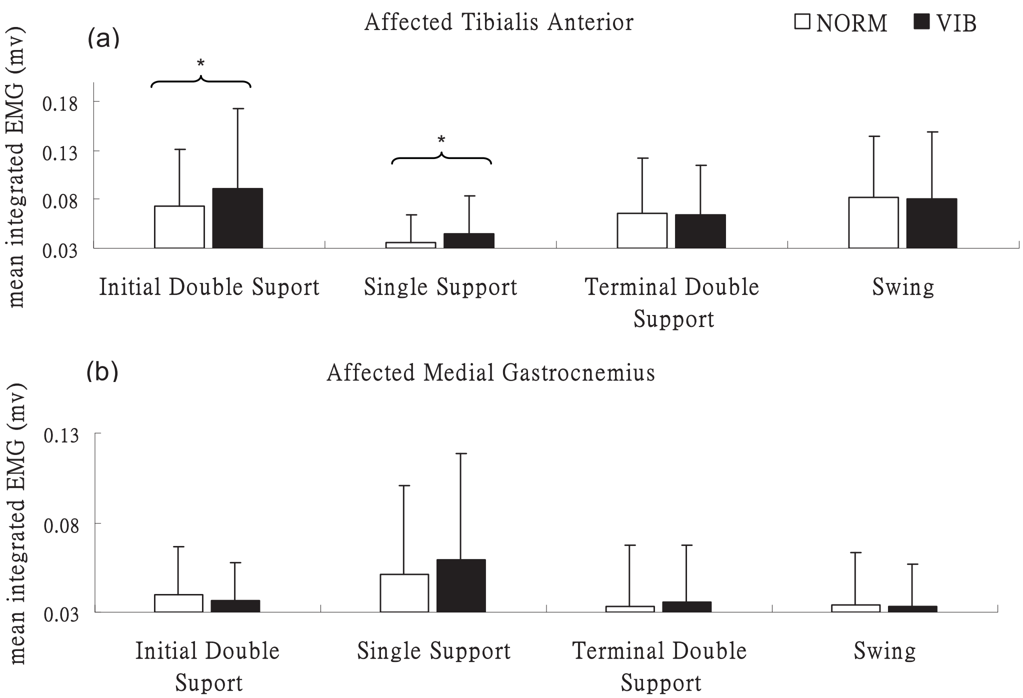

Figure 1: Mean (+ standard deviation) of the EMG of affected tibialis anterior (a) and medial gastrocnemius (b) under walking normally (NORM) and vibration (VIB) condition. *:p<0.05.

SHOULDER JOINT POSITION SENSE IMPROVES WITH ELEVATION ANGLES AND CORRELATES WITH MUSCULAR ACTIVITIES IN SUBJECTS WITH SHOULDER STIFFNESS

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INTRODUCTION

Shoulder joint stability is important for the high level of mobility necessary for performance of daily functional tasks and for more challenging athletic activities. This stability depends on feedback loops from the capsuloligamentous and musculotendinous mechanoreceptors to the central nervous system, termed proprioception (Sherringtonin 1906), to maintain muscle stiffness and coordination about the joint and thus produce smooth movements.

Deformation of capsuloligamentous tissue is believed to stimulate mechanoreceptors and provide the central nervous system with proprioceptive information. In the mid ranges of motion, the capsuloligamentous receptors are relatively inactive and the musculotendinous mechanoreceptors are proposed to be the primary responder to proprioception. However, these effects have not been studied in patients with shoulder tightness. The purposes of this study were twofold: to examine the effect of arm elevation angle on joint position sense (JPS), and to examine the effect of muscular contraction on JPS in subjects with shoulder stiffness. We hypothesized that as the shoulder joint position approached the end range of arm elevation, JPS would be enhanced. We further hypothesized that characteristics of muscle activation were correlated to JPS in the mid range of motion.

METHODS

Twenty subjects with unilateral stiff shoulders (SSs) were analyzed (age=56.2±8.1, 16 frozen shoulders and 4 impingement syndrome). They performed abduction in the scapular plane 6 times by self-selecting an end/mid range position. The electromagnetic motion-capturing system collected kinematic data while surface electromyography (sEMG) collected muscle activities (upper trapezius, lower trapezius, and serratus anterior muscles).

The sEMG data were evaluated by voluntary response index (VRI) including voluntary motor task (magnitude) and EMG distribution across the recorded muscles (similarity index, SI) during movements (Lee et al 2004, Lin et al 2006). The objective of the movements was to move the upper limb to the target position as accurately and similarly as possible without visual guidance. Six trials of each target position were tested to determine acceptable trials to have stabilization of the data, less than 5% of the cumulative mean values for at least three successive trials.

To determine if a significant proprioception difference existed at different elevation angles, t-tests (mid and end ranges) were calculated on the JPS. To determine the effect of muscular activity on JPS, Pearson product-moment coefficients of correlations were used to correlate the errors in shoulder repositioning with the VRI, including magnitude and similarity index.

RESULTS AND DISCUSSION

The data stabilized at the sixth repetition (Figure 1). Measurement values for joint reposition error, similarity index, and EMG magnitude in involved shoulders were demonstrated in Table 1.

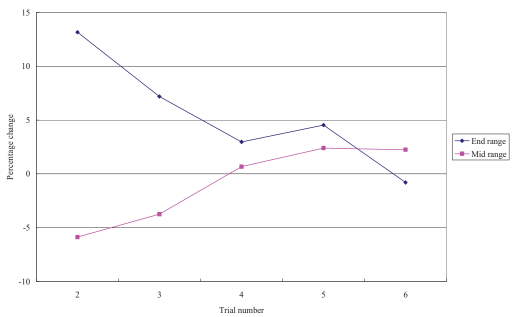

Figure 1: Percentage change in cumulative standard deviation during testing movements. For the end/mid range testing movements, the data stabilized at the fourth repetition. Thus, we used the mean of the fourth, fifth, and sixth repetitions on the JPS, similarity index, and difference magnitude for further analysis.

The magnitude of the repositioning error decreased at end range movements compared to mid range movements (2 degrees, $p<0.05$). The replication accuracy was also enhanced by an increased muscle activation level (magnitude) in the end range of motion ($R=-0.62$, $p<0.05$) and by coordination (SI) among muscle activation in the mid range of motion ($R=-0.87$, $p<0.05$).

SUMMARY/CONCLUSIONS

Our findings demonstrated that enhanced repositioning precision was found as the position approached the end range of movement. The replication accuracy was enhanced by increased muscle activation level in the end range of motion and by muscle coordination in the mid range of motion. Thus, the effect of muscle activation level and muscle coordination involved in shoulder proprioception during unconstrained shoulder movements was demonstrated for the first time. Future research should focus on different samples with and without shoulder dysfunction using JPS and muscle activities in our experimental paradigm.

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ACKNOWLEDGEMENTS

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Table 1. Measurement values for JPS, SI, and EMG magnitude (mean (SD)) in involved shoulders.

	End range	Mid range
JPS (degrees)	5.1(1.2)	7.0(1.4)
SI	0.90(0.05)	0.90(0.08)*
EMG Magnitude (mv)	159.4(19.8)*	145.4(21.1)

*: There was a significant relationship between JPS and VRI (magnitude and SI).

NEURAL CONTROL DURING HYPNOTICOMOTOR TASKS: NEURAL STRATEGIES AS OBSERVED THROUGH ELECTROMYOGRAPHY.

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INTRODUCTION

Classic hypnotic effects often involve alterations in the perception of control of motor movements for example, suggesting to hypnotized subjects that their arm is becoming very stiff and then challenging them to try to bend it. The classic hypnotic response is that when subjects try to bend their arm, they find they cannot. In this study we specifically addressed the problem of characterizing the underlying nature of "stationarity," or the "stiffness" or "immobility" of the arm during the challenge. Given that the neural mechanisms underlying these processes are presently under intensive study, we examined whether processing techniques utilizing cross-correlational analyses of selected muscle pairs would be sensitive to stationarity or lack thereof.

METHODS

Subjects were pre-screened for their hypnotic susceptibility using established techniques (the Water Stanford Scale of Hypnotic Susceptibility, Form C) and graded as either Low-Hypnotizable or High-Hypnotizable. We used a purposeful sample of highly hypnotizable subjects (N=14). Subjects were then asked to attend one session in the Human Movement Laboratory where after a short information session they were instrumented with over the skin sensors for the acquisition of kinematic and electromyographic data. We performed a second series of experiments designed to elucidate the actual motion patterns hypothesized to occur when trying to stiffen the arm. For these experiments we advertised in the university student population for volunteers. These subjects were not asked to submit to any pre-screening procedures nor were they hypnotized. We termed these second series of experiments "Forced Strategies".

EMG acquisition and processing: Muscle activation profiles were acquired using bipolar electromyography (EMG) electrodes attached over the skin of 6 superficial muscles of the upper limb of the subjects. Raw EMG signals for each muscle were acquired from the Upper Trapezius, Pectoralis Major, Anterior Deltoid, Posterior Deltoid, Biceps Brachii and Triceps Brachii muscles. The electrodes were placed over the body of the muscle, which we located by opposing the primary action of the muscle. Two epochs of raw EMG activity were sampled from the complete trial (sampled at 2.5 kHz) to produce two segments of EMG for 10 seconds before the challenge and for the entire time during the challenge to bend the arm (usually 10 seconds). These activation patterns were full-wave rectified and low-pass filtered at 6 Hz thus producing a linear envelope useful for correlation analysis and for showing the relative (phasic) activation of the muscle. To compare between subjects each muscle was normalized to its own resting value, which was established from the original resting trial, collected before the induction of hypnosis. We developed 3 cross-correlations

(Wren et al, 2006) of interest between muscles, specifically Upper Trapezius vs. Pectoralis Major, Anterior Deltoid vs. Posterior Deltoid and Biceps Brachii vs. Triceps Brachii. From these cross-correlations, the mean, maximum, minimum, and correlation at time 0 were calculated. The correlation at time 0 was used for the statistical analysis because it is the time of maximum overlap and has been shown to be the most physiologically relevant measure. While cross-correlation analysis is insensitive to muscle amplitude we were interested in the relative behaviour of the muscle pairs in response to the challenge. Amplitude correlations (Pearson Product) were derived from custom derived software.

RESULTS AND DISCUSSION

In this task subjects and controls were required to retain the arm in a horizontal position against gravity and therefore agonist and antagonist pairs were primarily used for postural purposes and as a consequence highly correlated in time. The best strategy for detecting change was therefore to subtract muscle cross-correlation values over the 10 seconds before the challenge from those during the challenge to bend the arm. A negative value would indicate that the muscle pairs became more "correlated" whilst a positive value would indicate a lowering.

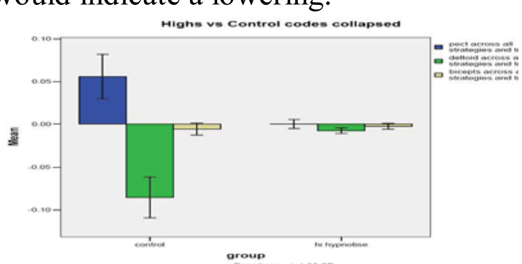

Figure 1: Overall differences in cross correlations (CC) prior to and then during the challenge are shown in for both controls and hypnotized subjects.

In the controls positive difference values indicate that CC was less during the event and their muscle pairs become more displaced in time (see Trap/PecM). Negative values indicate that the muscles become more temporally "correlated" (see A Delt/P Delt). There was very little difference in the Biceps/Triceps behaviour.

In our highly hypnotized subjects there was virtually no change in the behaviour as evidenced by the very low mean differences. These observations were mirrored in the individual strategies we "imposed" on the control subjects.

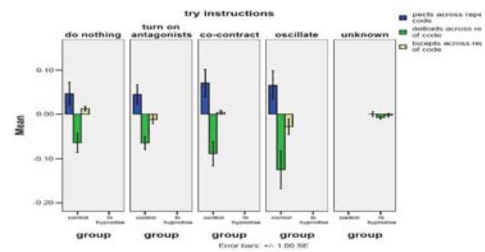

Figure 2: EMG results from the Forced Strategies described by code. The hypnotized subjects showed no evidence of adopting any of the hypothesized strategies.

SUMMARY/CONCLUSIONS

What central nervous system factors oblige the highly hypnotizable subjects to maintain their "stiffened" arm and prevent them from bending their elbow even though they are eminently capable of doing so? Since there was virtually no change in strategy prior to and during the challenge to bend the elbow it is possible that peripheral feedback is being gated however the location remains to be determined.

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COMPENSATING FOR MOTOR DISRUPTION IN PARKINSON'S PATIENTS UPON PERFORMING A DUAL TASK DURING SINGLE PLANE TRACKING

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INTRODUCTION

It has been shown[1-4] that concurrent cognitive tasks may disrupt the coordination of well-trained motions of individuals with Parkinson's disease (PD). Such interferences have been illustrated during walking, serial sequencing of the fingers and with regard to the sinusoidal motion of the distal joints of the upper extremity. Comprehensive upper extremity performance, while gradually varying the task, from a simple rhythmical swinging to a more complex dual task is yet to be investigated.

The primary goal of this study was to determine if ability of individuals with PD differs significantly from that of healthy adults during performance of a dual task during swinging of the arm to a pendulum beat, as means to specify the impaired neural mechanisms.

METHODS

Five healthy individuals (mean age 55) and five patients with PD (mean age 58) at stages 2-3 on Hoehn & Yahr scale, participated in the study. The PD subjects were evaluated by the Unified Parkinson's Disease Rating Scale (UPDRS), range of motion test, Berg balance test (BBS) and Time Up & Go test (TUG). A pendulum oscillating at a frequency of 0.5 Hz and visible peripherally above the subjects' shoulder height was placed parallel to the seated subject. Reflective markers were attached to the pendulum, as well as the ulnar styloid process, lateral epicondyle and acromion of the more affected arm of the PD subjects and the non-dominant arm of the healthy subjects. Surface EMG (sEMG) was recorded from the anterior deltoid (AD), posterior deltoid (PO), and biceps (BI), and triceps (T) of the observed arm.

Protocol. All subjects were instructed to sit on a chair with their non-dominant arm beside the suspended pendulum and, after a short practice session, to perform two types of tracking tasks. The first task included three trials of swinging the arm loosely forward and backward, in rhythm with the swinging pendulum. After five minutes rest, the subjects were asked to repeat the tracking task while counting backwards out loud, from one hundred and down, in steps of 3 (second task). For both tasks, the subjects were instructed to look straight ahead at a target board while viewing peripherally, the bobbing of the pendulum in the anterior part of its path.

Variables. The angular velocities of the total path of the pendulum and the swinging extremity were calculated using a video-based computerized system. Duration and amplitude of the muscle burst at each end of the pendulum motion, anterior (flexion) and posterior (extension), were calculated.

RESULTS AND DISCUSSION

During dual task performance, there was a significant group-by-task interaction ($p< 0.05$), as manifested by the decreased path and angular velocity of the shoulder parameters in subjects with PD. Both the pendulum's path and the angular velocity of the swinging extremity were invariant among the healthy subjects (Figure 1). The elbow kinematics did not exhibit a similar interaction and, in both tasks, the elbow path of the healthy subjects tended to be larger than that of the PD group which presented a

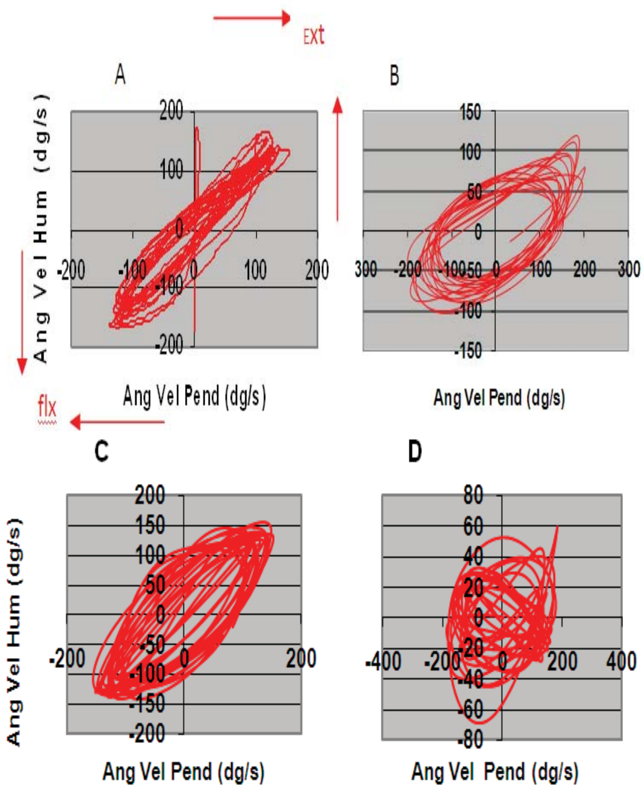

Figure 1: Interaction of the angular velocities of the pendulum and the humerus during pendulum tracking in healthy individuals (A), and PD patients (B), and during DT trials (C,D) respectively.

flexion bias. Moreover, the path and velocity of the shoulder joint correlated with the pendulum's path but not with the elbow. The sEMG in the PD group showed an increase of duration and amplitude in the AD and PO muscles. These muscles were increasingly out of phase in comparison to the healthy. The typical latency for the AD and PO muscles was characterized by shorter bursts peaking at 200 msec on average before the reversal of the motion at each end, more pronounced in extension phase of DT trials.

SUMMARY/CONCLUSIONS

The results indicate that adults with PD respond significantly differently from healthy subjects during dual-task tracking. The systematic reduction in the path and angular velocity may represent a compensatory strategy for improving tracking performance, due to the inherent motor deficit of these individuals. The variation in the shoulder and elbow kinematics suggests that the shoulder is the "primary joint," while the elbow is the "secondary" joint. It is suggested that during the targeted task, PD patients did not regulate the secondary, "subordinate" joint as they did the primary joint. Rather they forsook the secondary joint so as to keep the hands within their visual field and close to the body, as a means of achieving the arm control required to complete the task.

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DEGENERATE ARTICULATION PROFILES OF THE SEGMENTED ELBOW

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INTRODUCTION

In pointing and reaching motions, the hand tracks a highly stereotyped hand path through space, despite an infinitely many possible trajectories [Wann 1988]. Several theories of human motor control are premised on various interpretations of these trajectories, though the mechanisms of this invariant task execution are not yet known. We evaluated the role of the elbow in generating smooth and predictable hand paths across the workspace by analyzing the angular position profiles of the hand in an autonomous 1-D pointing task. We tested the prevailing hypothesis that elbow trajectories are invariantly straight.

METHODS

40 healthy volunteers were seated in the Mechanical Arm Support and Tracker (MAST), supporting the elbow of their dominant arm against gravity, and recording goniometric elbow motion data. Subjects moved smoothly within their maximal range of motion, articulating their elbow in cyclical flexion-extension motions.

Invoking the two-thirds power law relating trajectory curvature to motion velocity [Viviani 1982], we analyzed the subjects' θ(t) profiles for non-linearites, corresponding to regional departures from a trajectory of constant curvature in space (i.e. a non-smooth motion).

A 6-dimensional curve comparison model was devised to categorize these angular profiles as linear or variously non-linear, and trajectory determinacy was evaluated by analysis of the curve type distribution among subjects [Wininger 2007]. Additionally, trajectory types were compared between flexion and extension tasks in order to assess directional dependence.

RESULTS AND DISCUSSION

For a majority of subjects (53%) non-linear trajectory curves were the best-fit models of the greatest proportion of their dataset (*principal trajectory*). Surprisingly, principal trajectory distribution was approximately identical between tasks (Figure 1).

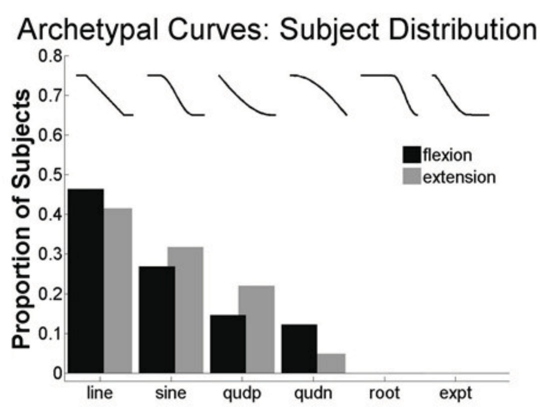

Figure 1: Proportion of subjects exhibiting a given principal trajectory. 6 trajectory models were evaluated, simulating various trajectory curvatures.

The number of *significant trajectories* was determined by a broken-stick scree analysis of model curve distribution within-subjects. A single trajectory model was not sufficient to optimally model the entirety of a typical subject's profile. For movements in both directions, a degenerate set of significant trajectories was found (Table 1).

Elbow activation degeneracy analysis			
Significant Trajectories		Commutativty	
Flexion	Extension	Identical Prin. Traj.	Mean ∩ Sig. Traj.
2.02 ± 0.7	1.94 ± 0.8	0.24	0.40

Table 1: Degeneracy and directional-dependence analyses for flexion and extension tasks. *Left*: number of significant trajectories (mean ± SD, N=40); *Right*: proportion of datasets for which the principal trajectories were the same in flexion and extension; mean intersection for degenerate flexion and extension profiles.

Comparison of principal trajectories across flexion and extension tasks revealed an approximately 1-in-4 prediction between directions. Whereas only 4 of the 6 model types yielded principal trajectories for our subject cohort, this result is considered to equate to chance. Degenerate profiles were analyzed for the average intersection of their significant trajectories between directions. These multiplicitous model sets averaged a 40% overlap, suggesting considerable directional dependence in trajectory formation.

SUMMARY/CONCLUSIONS

Despite the generation of highly concordant and smooth movements in space, flexion and extension activity at the elbow is both unpredictable and non-homogenously curved. In addition to these results, the present work not only introduces the MAST and two-thirds power law as a platform for one-dimensional assessment of the elbow's role in trajectory formation in two dimensions, but also a framework by which trajectory motions can be categorized and their stability quantified.

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TRUNK DYNAMICS ARE ALTERED BY EXPERIMENTAL LOW BACK PAIN

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INTRODUCTION

Pain alters neural control of the trunk. These changes have been shown to increase trunk muscle activity (Dieën et al. 2003) and are interpreted as a possible protective strategy used to enhance stiffness and stability of the spine. Given the redundancy of the system, it is unknown how trunk dynamics alter by a change in activity of measured trunk muscles.

Recent data show that trunk dynamics are altered in people with a history of low back pain (no current pain) compared to healthy controls (Hodges et al. (submitted)). Low back pain patients had *increased* trunk stiffness and *decreased* damping during relaxed upright sitting.

To better understand the development of these neural adaptations, the present study tested whether stiffness and damping of the trunk change with short-term experimental low back pain.

METHODS

Fourteen male subjects (mean±SD age: 24.6 ± 3.2 years) with no history of low back pain sat upright in a semi seated position with their pelvis fixated (Cholewicki et al. 2000) and a harness placed over their shoulder. Weights (~15% of total body mass) were attached via an electromagnet and force transducer to a low friction pulley system that attached to the front and rear (separately) of the trunk harness at the level of the ninth thoracic vertebra. Subjects were instructed to sit upright in a relaxed, neutral posture. At an unpredictable time, either the front or back weight was released 10 times (each) in random order.
Trials were repeated in three conditions; pre-pain, pain and post-pain. During the pain condition subjects were injected with a single bolus of hypertonic saline (5% NaCl, 1.5 ml) into the right erector spinae at the level of the L4 spinous process.

Trunk mass (*M*), damping (*B*) and stiffness (*K*) were estimated when the trunk was perturbed into either a backward (BW) or forward (FW) direction. The parameters were described by a second order linear model (equation 1) and the standard least squares procedure was used to solve the estimation.

$$F(t) = M \cdot \ddot{x}(t) + B \cdot \dot{x}(t) + K \cdot x(t) \quad \text{(equation 1)}$$

The displacement and duration of the perturbation were calculated from the onset until the maximum of the perturbation. All variables were compared with repeated measures ANOVA and Duncan’s multiple range test. Results are shown as mean ± SD.

RESULTS AND DISCUSSION

Due to high inter-subject variability in damping and stiffness (Table 1), values were normalised to the maximum value in either the pre-pain, pain or post pain condition.

With experimental pain, trunk stiffness *decreased* in both perturbation directions (BW: F=6.2, p<0.01; FW: F=4.4, p=0.02; Table 1, Fig. 1). Damping *increased* with FW perturbations (F=5.2, p=0.01, Fig. 1)

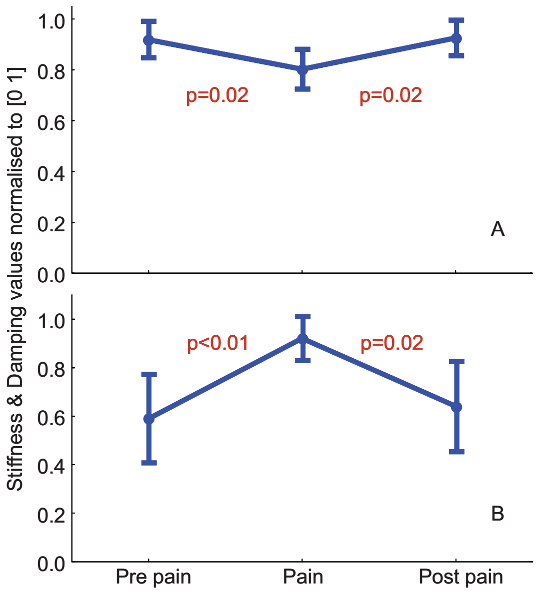

Figure 1: Normalised values for trunk stiffness (A) and damping (B) during pre pain, pain & post pain condition for forward perturbations (post hoc p-values are shown; error bars denote 95% CI).

Both stiffness and damping generate forces that oppose the perturbation. The decrease in stiffness and non significant change in damping with BW perturbation resulted in a change in performance; both displacement and duration increased significantly during experimental pain (F=4.1, p=0.03; F=5.1, p=0.01, respectively, Table 1). With FW perturbation the decrease in stiffness was compensated with an increase in damping, which resulted in similar performance throughout the conditions.

The pain & post-pain estimation of trunk mass (Table 1) with BW perturbations was estimated significantly lower (p=0.01) compared to pre pain. This might be explained by an upward change in axis of rotation of the trunk during the pain & post-pain condition, due to change in trunk muscle recruitment.

In contrast to findings in people with recurrent low back pain, no increase in stiffness was observed during experimental pain. It is possible that an *increase* in damping rather than stiffness represents a more effective strategy to control trunk perturbations in low back pain.

SUMMARY/CONCLUSIONS

Trunk dynamics are changed by a *decrease* in stiffness and an *increase* in damping during pain in forward perturbations, suggesting a specific adaptation of the central nervous system to experimental pain in healthy subjects.

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Table 1. Results with backward (BW) and forward (FW) perturbations

	pre pain		pain		post pain	
	BW	FW	BW	FW	BW	FW
Est. Mass (kg)	14.7±2.7	18.1±3.4	13.0±2.7	16.4±3.3	13.4±3.4	18.1±3.0
Damping (Ns/m)	62.0±52.4	46.6±43.6	77.9±44.7	64.3±42.7	65.6±44.8	46.5±35.6
Stiffness (N/m)	2917±577	2548±522	2486±518	2232±564	2821±702	2604±717
Displacement (cm)	6.4±0.7	7.6±1.4	7.0±0.8	8.1±1.4	6.5±0.9	7.5±1.3
Duration (sec)	0.30±0.02	0.35±0.04	0.33±0.04	0.37±0.05	0.31±0.06	0.36±0.06

COORDINATION OF PELVIS AND THORAX DURING GAIT IN LOW BACK PAIN PATIENTS

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INTRODUCTION

In healthy gait, horizontal plane rotations of pelvis and thorax are mainly *in-phase* at lower velocities, and approach *anti-phase* at higher velocities. In contrast to this, the *in-phase* relation tends to persist at higher velocities in low-back pain (LBP) patients (Lamoth et al. 2002).
In addition LBP patients have altered muscle recruitment patterns that are likely to enhance spinal stability (Dieën et al. 2003) and could decrease range of motion (ROM) between pelvis and thorax during walking through an increase of trunk stiffness.
Such an increase in stiffness during gait will affect the timing of horizontal plane trunk rotations (Kubo et al. 2006).
The aim of this study was to determine if LBP patients and healthy controls demonstrate a difference in i) ROM *of* and relative ROM *between* the pelvis and thorax; and ii) peak velocity of and between pelvis and thorax iii) intersegmental timing of peak rotational velocity *of* and *between* the pelvis and thorax.

METHODS

Fourteen chronic low back pain (LBP) patients (9 female, 5 male) and twelve healthy controls (8 female, 4 male) walked on a treadmill for three minutes at 12 velocities that ranged from 0.6 to 5.0 km/h in increments of 0.4 km/h.
Full-body kinematic data were collected (OPTOTRAK). (Relative) ROM and (relative) rotational velocity of the pelvis and thorax were calculated. Timing of the peak rotational velocity was estimated by fitting a sine wave using a least square fit. The phase of this sine wave was used as an estimate of the timing of the peak rotational velocity, and was expressed as percentage stride cycle after right heel strike (rHS). [0% - 100% = rHs to rHs = one stride cycle]. Data were analyzed with repeated measures ANOVA, with health status as between and velocity as within subjects factor.

RESULTS

Relative phase between pelvis and thorax was affected by velocity as previously reported (F 125.81, $p<0.01$). However, no effect of group or an interaction with group was found.

ROM of the pelvis was affected by velocity ($F=4.44$, $p<0.01$), decreasing (from ~12° to ~8°) for velocities up to 3.8 km/h to then increase again (to 12°). No significant group effect was detected.

The ROM of the thorax decreased with velocity, from ~12° with low velocities to 9° at high velocities ($F=14.64$, $p<0.01$). Although there was no main effect of group, an interaction effect ($F=2.15$, $p=0.02$) indicated that LBP patients decreased their ROM more (to ~7°) at higher velocities than the healthy controls.

The ROM between the pelvis and thorax increased significantly with velocity (from 4° to 15°; F=146.39, p=<0.01), but no significant interaction or difference between the groups was detected.

Peak velocity of the pelvis increased with gait velocity (~10°/sec to ~22°/sec; F=3.43, p<0.01). There was no effect of group nor an interaction with group.

The velocity of the thorax also increased with gait velocity (~10°/sec to ~27°/sec; F=21.4, p<0.01). While no main effect of group was found, an interaction effect (F=4.08, p<0.01) showed that LBP patients limited their peak velocity at ~20°/sec from 2.6 km/h.

As expected, maximum relative velocity between pelvis and thorax increased with velocity (~2.2°/sec to ~ 40°/sec; F=99.88, p<0.01). The maximum relative velocity did not differ between the groups, and no interaction was detected.

Velocity affected the timing of peak rotational velocity of the pelvis (from 10% after rHS to 20% before rHS; F=85.43, p<0.01). No main group effect was found, whereas an interaction indicated that from ~2.2 km/h LBP patients timed their peak rotational velocity ~5% earlier than controls (F=2.70, p<0.01).

Velocity also affected timing of the peak rotational velocity of the thorax (F=3.46, p<0.01). No significant main effect of group was found. However, in patients, timing was approximately constant over all gait velocities (~12% after rHS), while in healthy subjects, timing changed from ~10% after rHS at low velocities to ~16% after rHS at high velocities (F=3.63, p<0.01).

The effect of velocity on timing of relative peak rotational velocity between pelvis and thorax was significant (F=13.60, p<0.01, Fig. 1), and a significant interaction was detected (F=6.28, p=0.02). Furthermore, on average LBP timed their relative peak velocity earlier (F=2.28, p=0.01, Fig 1).

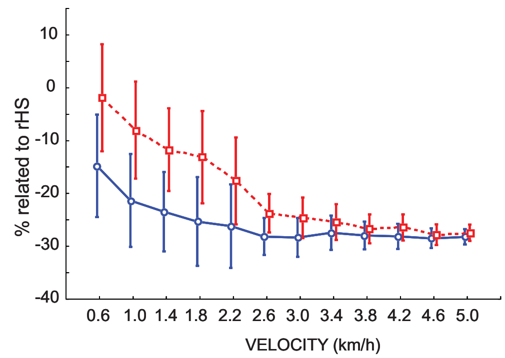

Figure 1: Relative peak rotational velocity timing between pelvis and thorax. Healthy subjects are shown in red, LBP patients in blue. Error bars denote 95% CI.

DISCUSSION

In contrast to previous studies, the LBP patients in the present study had normal *phase* relations between pelvis and thorax rotations. A more detailed analysis of pelvis and thorax kinematics did however reveal differences. Patients showed a lower ROM and peak velocity of the thorax at high walking velocities, possibly explained by an increase in trunk stiffness (Kubo et al. 2006). Timing differences between patients and controls appear to suggest that at lower velocities the patients used a strategy representative for gait at higher velocities in controls, which might reflect more cautious behaviour in patients.

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TRUNK ANTAGONIST CO-ACTIVATION IS ASSOCIATED WITH MORE STRENOUS EXERTIONS AND FATIGUE

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INTRODUCTION

There are a number of studies that have shown the CNS adapts to destabilizing factors that are external, such as unstable force fields (De Serres and Milner, 1991). Therefore, it is also possible that internal factors, such as force variability (neuromuscular noise), could produce compensatory muscle recruitment to maintain stability. For the spine, this most likely would be reflected in terms of antagonist activation. To test this hypothesis, we used two methods believed to increase force variability: strenuous exertions and fatigue. Consequently, the first two hypotheses of this study will examine if force variability increases (1) with more strenuous exertions, and (2) with fatigue. The third hypothesis will examine if antagonist activation also increases with more strenuous exertions and fatigue, suggesting that the CNS is utilizing antagonist activation to maintain spine stability.

METHODS

Twelve healthy subjects (6 females and 6 males) were placed in the testing apparatus used for exerting isometric force in flexion (Figure 1A) and extension (Figure 1B). Force variability (standard deviation of force signal) was assessed for graded isometric trunk exertions (10, 20, 40, 60, 80 % of max) in flexion and extension, and at the start and end of a trunk extensor fatiguing trial. Normalized EMG signals from five trunk muscle pairs (RA – rectus abdominis, EO – external oblique, IO – internal oblique, TE – thoracic erector spinae, and LE – lumbar erector spinae) were collected for each graded exertion and during a fatiguing trial.

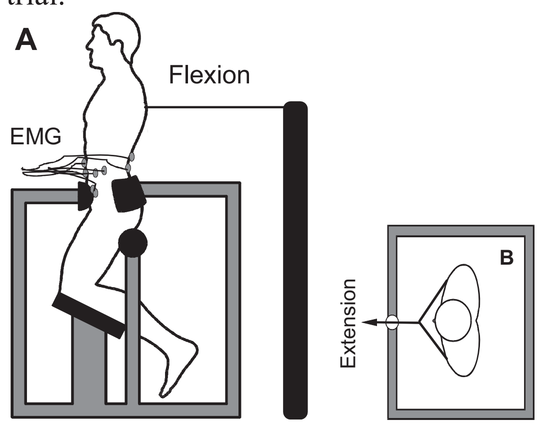

Figure 1: Testing apparatus.

RESULTS AND DISCUSSION

Force variability increased for more strenuous exertions in both flexion ($p<.001$) and extension ($p<.001$, Figure 2), and after extensor fatigue ($p<.014$, Figure 2). In the flexion direction, both antagonist muscles (TE & LE) increased activation for more strenuous exertions ($p<.001$). In the extension direction, all antagonist muscles except RA increased activation for more strenuous exertions ($p<.001$, Figure 3) and following fatigue ($p<.01$). This selective recruitment of abdominal muscles may be functionally related to maintaining spine stability in the presence of internal disturbances.

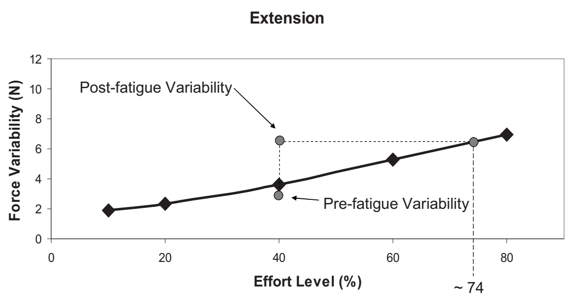

Figure 2: Force variability at various levels of isometric exertions in trunk extension, and before and after fatigue.

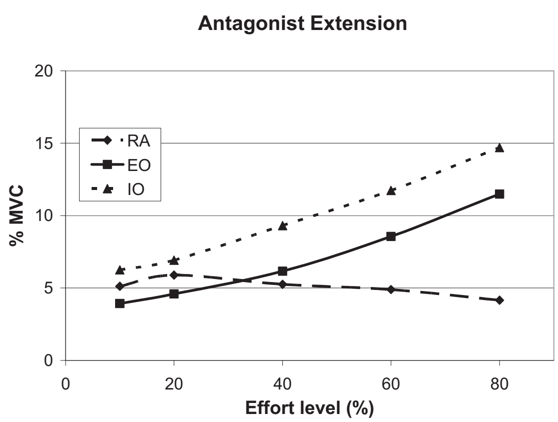

Figure 3: Normalized EMG activity of the trunk flexors at various levels of isometric exertions in trunk extension.

Most likely, some of the antagonist activation is needed to equilibrate the moment produced by the trunk agonist (Thelen et al., 1995;Stokes and Gardner-Morse, 1995), but it is unlikely this would explain the significant increase in antagonist activation with fatigue since moments about the spine are held constant. Therefore, we believe that a large component of the increased antagonist activation with fatigue was related to impaired neuromuscular control. To show how force variability is related to antagonist activation, we predicted the antagonist activation at the end of the fatiguing contraction. Using the Force variability-Effort level graph, the force variability at the end of the fatiguing trial is equivalent to ~ 74 % of Effort level (Figure 2). At the 74 Effort level on the % MVC-Effort level graph for extension (Figure 3), IO and EO would be equivalent to 13.5 and 10.5 % MVC respectively. EMG data collected at the end of the fatiguing trial showed that IO and EO had 12.5 and 9.25 % MVC, which is just slightly below predicted values.

SUMMARY/CONCLUSIONS

There are a number of clinical implications for the findings in the present study. If it is true that impairment in neuromuscular control results in increased antagonist activity, then any scenario in which neuromuscular control is compromised (i.e., fatigue, whole body vibration, prolonged static flexion, to name a few) will most likely increase spinal loading and possibly risk of injury.

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ACKNOWLEDGEMENTS

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TRUNK MUSCLE REACTIONS TO SUDDEN LOADING IN A POSITION WITHOUT VERTICAL POSTURAL DEMAND

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INTRODUCTION

Numerous studies have dealt with the challenge of understanding trunk muscle coordination through perturbations applied either directly to the trunk (e.g. Cresswell et al. 1994) or via disturbances induced by voluntary movements of the limbs (e.g. Hodges and Richardson 1999). Most experiments were performed in standing, which means that the response to the perturbation included also a postural component to keep the trunk upright against gravity. In such experiments, a specific early and direction independent activation has been seen in the deepest abdominal muscle, transversus abdominis. We have recently reported results, which indicate that the coordination between individual abdominal muscles changes when the postural component is eliminated using a hold-release paradigm with the subjects in a side lying position (Eriksson Crommert and Thorstensson 2006).

The aim of the present experiments was to further understand the role of individual trunk muscles in spine stabilization and postural control by investigating trunk muscle coordination in reaction to sudden unexpected loading perturbations administered in a position of minimal postural demand with respect to keeping the trunk upright.

METHODS

Eleven healthy male subjects participated. The subjects lay on their right side on a horizontal swivel-table with the pelvis and lower limbs strapped to an immobile part and the trunk to a movable part of the table. The two parts were connected via a hinge allowing the upper body to move either in flexion or extension (centre of rotation at L3 level). Two 10 kg loads were attached anteriorly and posteriorly to the moveable part of the table via electromagnets. Release of a load caused a sudden pull of the upper body either in the flexion or extension direction. The onset of perturbation was measured with an accelerometer.

The examiner controlled the release of the loads and the subjects had no prior knowledge of when or in what direction the perturbation was going to occur. The order of induced flexion or extension movements was randomized. The instruction to the subjects was to remain relaxed before the loading and then to “catch” the movement.

Muscle activity was measured with intramuscular fine wire EMG electrodes placed under guidance of ultrasound bilaterally in transversus abdominis (TrA), obliquus externus (OE), rectus abdominis (RA) and lumbar erector spinae (ES). Mean EMG amplitudes (RMS) were measured in 10 ms time bins and the onset of activation was analyzed for each muscle in relation to the instant of perturbation (acceleration increase). EMG amplitudes were normalized to a maximal voluntary contraction (% MVC). Preliminary results from the amplitude analysis (200 ms after the start of acceleration) and timing of activation are presented.

RESULTS

EMG onsets. In induced extension, onset of activation of TrA was later compared to OE and RA; onset latencies were 93 ms versus 71 ms and 73 ms, respectively. ES was the last muscle to be activated, onset latency 133 ms. In induced flexion, the mean onset latency for TrA was similar, 93 ms, as in induced extension. Corresponding values for OE and RA were 95 ms and 91 ms, respectively. The onset latency for ES was 100 ms.

EMG amplitudes. OE and RA responded with greater activity in trials with induced extension than flexion, 22 % and 18 % compared to 11 % and 5 %, respectively. No difference in EMG amplitude was present between loading directions for TrA and ES, with mean relative amplitude levels of 15 % and 6 % with induced extension and 13 % and 8 % with induced flexion, respectively.

DISCUSSION

In response to sudden loading of the trunk in side lying, i.e. without vertical postural demand, the onset of the activation of TrA was simultaneous with that of the other abdominal muscles in induced flexion. This is in contrast to standing, where TrA in previous studies have been shown to have an earlier onset than the other abdominal muscles in reaction to sudden flexion loading (Cresswell et al. 1994).

In induced extension, the onset of the TrA response in the present study was delayed compared to that of the other abdominal muscles, which is in accordance with previous results from experiments in a side lying position (Eriksson Crommert and Thorstensson 2006). However, in standing, there was no difference in onset times observed between the abdominal muscles in response to a corresponding extension loading (Cresswell et al. 1994).

The demonstrated insensitivity of TrA amplitude and timing to direction of perturbation strengthens the notion that TrA, as opposed to RA and OE, has a general and unspecific controlling function, e.g. restricting inter-segmental movements of the lumbar spine. However, an earlier onset of TrA compared to other trunk muscles appears to be specific to the maintenance of vertical postural control of the trunk.

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INVESTIGATION OF THRESHOLD FOR RECRUITMENT OF ANTAGONIST CO-CONTRACTION OF THE LUMBO-PELVIS MUSCLES DURING SUBMAXIMAL FLEXION AND EXTENSION EFFORTS

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INTRODUCTION

Antagonist co-contraction of the lumbopelvic muscles has been demonstrated to be important for maintenance of lumbar stability (Cholewicki et al 1997). However, recent data suggest that there is surprisingly little co-contraction with efforts less than 20% maximum voluntary contraction during flexion and extension efforts (McCook et al, 2007).

The first aim of this study was to determine the threshold flexion and extension efforts at which antagonist co-contraction was initiated. Furthermore, consistent with earlier data we predicted that the activity of transversus abdominis (TrA) and the deep fibres of multifidus would increase independently of the direction of force to provide a non-direction specific contribution to spinal control. The second aim of this study was to test this hypothesis.

METHODS

Twelve healthy subjects adopted a semiseated position in a custom-built frame and performed ~4 s isometric efforts into flexion and extension to submaximal target loads of 20, 40, 60, 80 and 100% of their maximal voluntary contraction (MVC) in random order. Surface electromyography (EMG) electrodes were placed over the lumbar and thoracic erector spinae, rectus abdominis, and the upper and lower abdominal muscles. Bipolar fine-wire intramuscular electrodes were inserted into TrA and the superficial (SM) and deep multifidus (DM). RMS EMG amplitudes were calculated and normalized to MVC. Net flexor and extensor EMG activity was calculated by summation of normalized values for the flexor and extensor muscles. EMG amplitude was compared between load levels with a repeated measure ANOVA. Independent t-tests were used to compare whether net EMG increased from zero.

RESULTS AND DISCUSSION

When subjects isometrically flexed or extended their trunk to the target loads the net EMG activity of the antagonist muscles did not increase until the 40% load interval in the extension trials, and the 60% load interval in the flexion trials ($P<0.028$). EMG activity of the lower abdominal muscles (including activity from TrA and obliquus internus abdominis) recorded with surface electrodes and TrA, recorded with intramuscular electrodes, progressively increased with increasing flexion and extension loads ($P<0.022$), and were not different between directions ($P=0.136$ and 0.468, respectively). There was large variability in recruitment strategy between subjects.

An interesting observation was that when subjects matched a 100% MVC target they used a different strategy of muscle activity (i.e. less EMG of the thoracic erector spinae during extension and less activity of the upper abdominal muscle during flexion ($P<0.0016$)) compared to trials in which they were instructed to pull as hard as possible.

SUMMARY/CONCLUSIONS

The results of this study indicate that there is co-contraction of antagonist muscles at 40-60% MVC trunk flexion or extension efforts. However, there was considerable variation between subjects in the specific muscles that co-contracted with the agonists. A novel finding of this study was that despite the generation of identical forces, EMG activity of the agonist trunk muscles was less when subjects aimed to reach a target force than when they were instructed to perform a maximum effort without a target. This highlights the versatility of the trunk to use variable strategies of muscle activation from the redundant trunk muscle system. This study also supports the potential role of the lower abdominal muscles in spinal control with increasing load, irrespective of the task direction.

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CHANGES IN POSTURAL DEMANDS IN SITTING ARE ASSOCIATED WITH DIFFERENTIAL CHANGES IN PARASPINAL MUSCLE ACTIVITY

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INTRODUCTION

Trunk muscles respond rapidly to release of a load with increased or decreased activity. For instance when a load is released from back of the trunk, flexor muscle activity is reduced and extensor activity increases to maintain the position of the trunk (Cholewicki 2005, Radebold 2000, Reeves 2005). This response is predictable based on the ballistic nature of the task. If a posterior load is released slowly the mechanical demands are different and our preliminary observations had identified a gradual reduction in flexor activity. A consistent feature of the task was a burst of extensor activity towards the end of the load release. This burst of extensor activity is consistent with an adjustment of activity to re-establish equilibrium and stability for the new mechanical demands of upright sitting once the load is released. As this adjustment in muscle activity adjusts for changes in stability demands it is likely to provide insight into neural coordination of spine control.

This study aimed to evaluate the response of the lumbar paraspinal muscles during a slow load release, and to compare the response of the deep and superficial paraspinal muscles.

METHODS

Data from 2 sets of isometric loading trials were analyzed for this study. All 14 subjects were healthy male volunteers with no history of low back pain. Subjects were positioned in an upright, semi-seated position in a custom built frame designed to stabilize the pelvis yet permit a natural lordosis during the isometric loading tasks.

The tasks required isometric flexion to target loads against a transducer placed behind the subject and attached by a harness over the shoulders. The first group of subjects (n=7) targeted 50, 100, 150 and 200 N. The second subject group force matched 20% (n=4) or 10% (n=3) increments from 20 or 10% respectively, up to 100% of the individual's maximum voluntary contraction (MVC). The target loads were presented in a random order and 3 repetitions of each task were performed in the trial. Subjects were asked to gradually build up their resistance to match the load target, maintain the force for 3 to 5 s, and then slowly release the load. Rest periods of between 5 and 30 s between loads was employed in the 50 to 200 N trials, and up to 3 minutes between the MVC trials. EMG recordings were made from the paraspinal muscles, including the lumbar erector spinae (LES) with surface electrodes, and the deep multifidus (DM) with intramuscular electrodes. The primary analysis in this study was the timing of onset of activity of the paraspinal muscles relative to the onset of the load release. EMG activity onsets were detected with observer blinded to the concurrent activity of the other trunk muscles and the point of change in load release. A t-test for independent measures was used to compare the onset of the DM and LES. $P<0.05$.

RESULTS AND DISCUSSION

A total of 255 trials were analyzed with the loads ranging from 50 to 850 N. There was considerable variation in timing of the burst of activity of the paraspinal muscles which occurred between 1.12 s before to 4.32 s after the onset of the reduction in flexion force.

A burst of EMG was more likely to occur in DM (84% of trials) than LES (39% of trials). In trials with an EMG burst during the release phase for both muscles, the onset of DM EMG preceded that of LES by 0.980 s (SD 1.22 s; range min -3.82 s, max 0.52 s; $P < 0.0001$).

SUMMARY AND CONCLUSION

These data suggest that the fine-tuning of paraspinal muscle activity after slow release of a load more commonly involves adjustment of DM activity than the LES. Furthermore, the onset of activity of DM preceded that of LES. As this burst of activity is likely to contribute to the readjustment of muscle activity required to meet the demands of the new sitting posture (i.e. without the flexion load) these data suggest a critical role of the deeper paraspinal muscles in this function. The data provide further evidence for differential control of the deep and superficial paraspinal muscles.

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HEALTHY CHILDREN'S EMG PROFILES IN GAIT DIFFER FROM ADULT EMG PROFILES

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INTRODUCTION

In a context of gait analysis in children with cerebral palsy (CP), surface EMG of major lower extremity muscles is often used for additional information on muscle coordination, especially when surgery on the muscle tendon complex is considered [1]. Several studies have reported typical adult EMG patterns during gait, as a reference to assist interpretation of possibly aberrant EMG patterns, eg. [2,3]. The use of these in a paediatric population is not warranted. However, only one study measured children, mainly distal muscles in 12 subjects, which were not reportedcompletely [4]. Moreover when interpreting EMG profiles, it is very important consider walking speed as a co-determinant of EMG patterns [2,4].

METHODS

A total of 30 healthy young children were measured, each in one session. Age was stratified: 10 children aged 6-7 (mean 6.2 years), 10 aged 8-9 (8.4 y) and 10 aged 10-11 (11 y.); 13 boys and 17 girls in total.

Surface EMG was recorded from one leg, of mm. Gluteus Maximus (GMX); Rectus Femoris (RF); Vastus Lateralis (VL); SemiTendinosis (ST); Tibialis Anterior (TA); and Gastrocnemius Medialis (GAM). Bipolar electrodes were used, placement following the SENIAM protocol [5].

Walking was measured at comfortable speed, as well as slow (-30%) and fast (+30%) walking speed. Surface EMG was digitized at 1000 Hz. , and rectified and low pass filtered (@ 3 Hz) to obtain a linear envelop (LE) [3]. LE was ensemble averaged over 10 strides. Walking speed was normalized for leg length ℓ_0 , i.e. Vnorm = $v^*\sqrt{(g \bullet \ell_0)}$ [2].

All profiles were compared with the normalized speed dependent predicted reference profiles given by Hof ea., based on their adult EMG measurements, using their method of gain fitting [2], see figure 1. This resulted in an overall RMS error [2].

RESULTS AND DISCUSSION

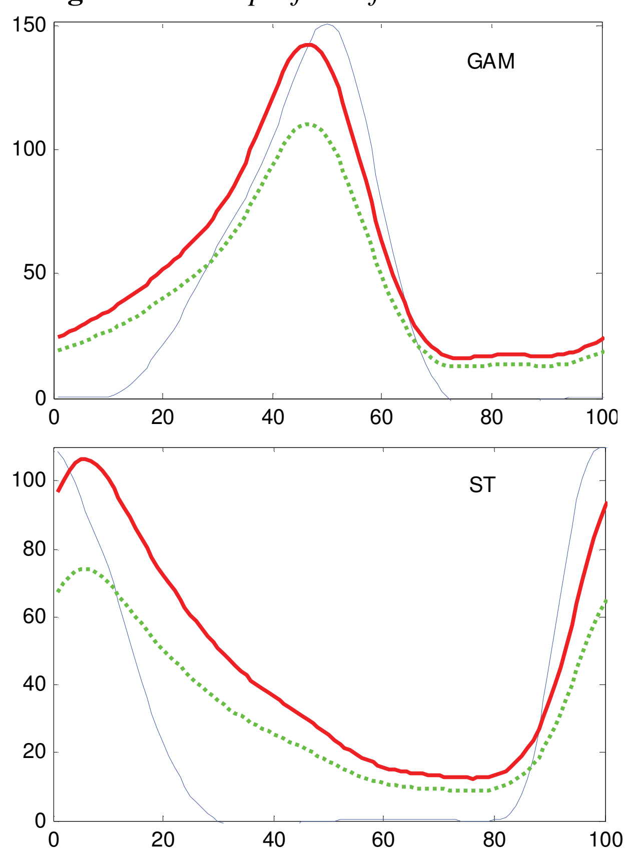
Children show a higher normalised comfortable walking speed. Absolute EMG values in healthy children are lower as in adults, and show an increased sensitivity to speed. After correcting for this by optimal gain fitting [2], EMG in children still show a low predictability from adult values (table I).

High errors (up to 30%) indicate clearly that reference values from an adult population cannot be used in paediatric applications. For two muscles (GAM, showing the lowest error, and ST showing the highest error) , figure 1 shows the mean graphs Generally children show less relaxation between active periods. Electrode placement was identical, and has been shown to be consistent across our laboratories.[6] Our results suggest that age specific reference values should be established.

		MUSCLES					
Condition	**Vnorm**	**GAM**	**TA**	**VL**	**RF**	**ST**	**GMX**
Slow_kids	0.37	13 (10)	24 (13)	8 (16)	12 (26)	23 (30)	6 (22)
Adults [2]	*0.41*	*12 (8)*	*10 (4)*	*4 (5)*	*11 (15)*	*6 (5)*	*3 (7)*
Comf_kids	0.51	20 (13)	38 (16)	13 (17)	20 (28)	32 (29)	7 (17)
Fast_kids	0.65	26 (15)	55 (19)	23 (22)	30 (31)	39 (28)	6 (10)

Table I. RMS error, expressed in µV and (%max) between measured and predicted, adult values from Hof [2]

Figure 1. *Mean profiles of GAM and ST*



Xaxis: gaitcycle (%), Yaxis; µV
Blue solid line: predicted profile [2]
Green dotted line: measured profile
Red bold line: gain corrected profile
Normalized walking speed: 0.51

SUMMARY/CONCLUSIONS

A set of typical EMG patterns, of 6 main muscles during gait in 30 healthy children aged 6-12 were measured for slow, comfortable and fast walking speed. These were compared to adult values reported by Hof ea [1]. EMG profiles during gait from healthy children differ from adult EMG profiles, when compared on the basis of normalized walking speed and gain adjustment, for most muscles. A set of children's EMG profiles is established to serve as a reference, while effects of walking speed are accounted for.

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EMG ANALYSIS OF DEVELOPMENTAL CHANGES IN HUMAN GAIT

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INTRODUCTION

An understanding of the normal development process of learning to walk is enlightening, and crucial for the understanding and treatment of disorders in which the normal pattern does not develop (Sutherland, 1997), as, for example, in cerebral palsy. The purpose of this study was to further investigate the neuromaturation process, by examining the activation patterns of the main muscles involved in walking in conjunction with kinematic and anthropometric data. An indication of the neural control strategies is available through the EMG signals, while musculoskeletal growth may be accounted for through the kinematic and anthropometric data.

METHODS

This was a cross-sectional study of 100 children aged 13 months – 13 years and 9 adults aged 36 ± 9 years (mean ± std dev) to provide reference data. Anthropometric measurements were made and the children completed several walking trials with simultaneous recording of kinematic data from heel, toe and trunk markers at 120 Hz using a six-camera motion analysis system (Vicon) and bipolar surface EMG signals recorded at 1200 Hz (Noraxon Telemyo900) for six muscles of the right leg: gluteus maximus (GM), long head of biceps femoris (BF), rectus femoris (RF), vastus lateralis (VL), lateral head of gastrocnemius (LG) and tibialis anterior (TA).

The kinematic recordings were analyzed using the foot velocity algorithm (O'Connor *et al.,* 2007) to determine the timings of heel strike and toe off for each step. Velocity, V, was computed and scaled to account for size and determine dimensionless velocity, $\beta = V \big/ \sqrt{g \times LL}$, where g is gravitational acceleration and LL is leg length (Vaughan *et al.,* 2003). EMG signals were high pass filtered (30 Hz) to remove motion artifact, and activation patterns determined by full wave rectification and low pass filtering (6 Hz). The activation patterns were normalized in time for each gait cycle and in amplitude to the peak activation values. The average of at least 5 steps was computed for each child along with the ensemble average of all the adults.

Cross-correlation between the activation patterns of the different muscles at different ages with the corresponding adult patterns was calculated. The relative activation amplitudes of each muscle for each section of the gait cycle were analysed to identify significant trends with age.

RESULTS AND DISCUSSION

While velocity increased steadily with age, dimensionless velocity increased initially with age but leveled off around the adult value, β_f, at approximately 5 years. It was fitted with a growth curve: $\beta(t) = \beta_f(1 - e^{-t/\tau})$, with a time constant, τ, of 1.5 years. As the changes in physical size are accounted for by the non-dimensional scaling, it was proposed that the changes in dimensionless velocity are related to the process of neuromaturation – the development of mature neural control strategies for locomotion (Vaughan *et al.,* 2003).

Correlation between each child's activation patterns and the adult patterns indicated that the children's patterns converged on the adult pattern with age, following a similar pattern to that of the dimensionless velocity, Figure 1. The superimposed curves are all plotted using the form: $\alpha = \alpha_f(1 - e^{-t/\tau})$, with appropriate values of α_f and the original value of τ which fitted the dimensionless velocity. The correlation between the EMG activation patterns can be seen as a measure of how close the neural control is to the adult pattern, thus an objective measure of the neuromaturation for locomotion. It appears probable that the changes in walking strategy gradually increase the dynamic similarity with the mature gait pattern, which is the characteristic being captured with the measure of dimensionless velocity.

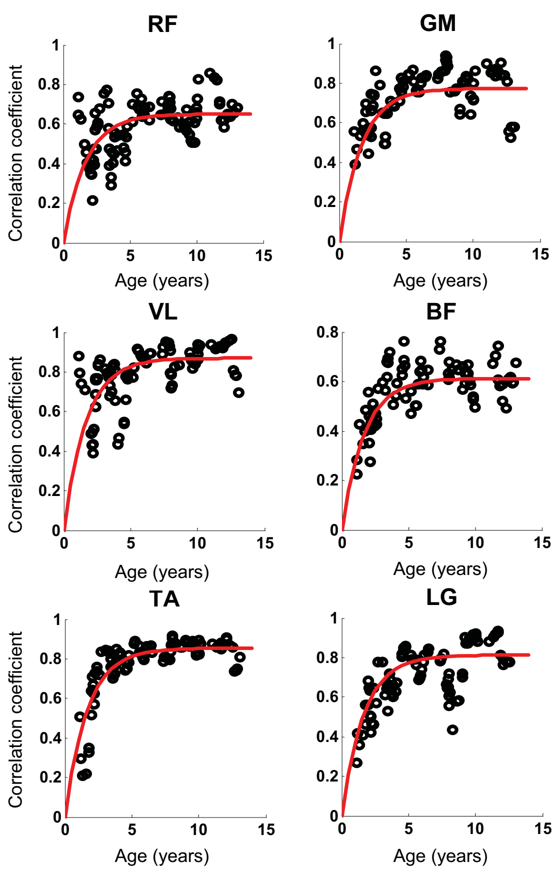

Figure 1: Correlation between the activation patterns of the different muscles at different ages with the corresponding adult patterns.

Noteworthy changes in activation patterns were quantified. For example, in the early part of stance, the relative activation of LG decreased at the same rate as dimensionless velocity. This change is part of the emergence of a clearly defined reciprocal pattern for an agonist-antagonist muscles pair as TA is at peak activation at this stage.

SUMMARY

The most interesting finding is that both the overall development of the muscle activation patterns and the rate of change in key developments are well approximated by the dimensionless velocity. This link suggests that the extent to which the original patterns of locomotion have altered may be quantified with an index of neuromaturation. This will be of benefit in further research into the processes of neuromaturation and may have clinical significance in the assessment of pathological gait.

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SEEING THE WRONG ARM: THE EFFECTS OF A MISMATCH BETWEEN VISUAL AND PROPRIOCEPTIVE INFORMATION ON EMG IN TYPICALLY DEVELOPED CHILDREN

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INTRODUCTION

In previous research we investigated the effects of the 'mirror box' on upper limb kinematics in typically developed children and children with spastic hemiparetic cerebral palsy (SHCP). A 'mirror box' creates a visual illusion, which gives rise to a visual perception of a zero lag symmetric movement between the upper limbs. The 'mirror box' allows the examination of the effect of different perceptual conditions (normal; visual occlusion of 1 arm; visual illusion of 1 arm) on bimanual coordination.

Our previous research with children showed that upper limb movement variability during a circular symmetrical bimanual task increased when visual information was occluded from 1 arm. More importantly, no differences in movement variability were found when participants received visual information from both arms or visual information of one arm and a superimposed mirror reflection of the contralateral arm ('mirror box'). It could be reasoned that the increased variability was a result of more co-contraction, which occurred in the absence of visual feedback.

To date, only a few studies have recorded electromyography (EMG) to investigate age-related differences in typically developed children. It was found that higher levels of EMG activity occurred in the lower limb muscles in younger children compared to older children.

Therefore, the aim of this study was to investigate upper limb EMG activity in younger (5-10 years) and older (12-18 years) typically developed children when performing the circular symmetrical bimanual task in the presence of the 'mirror box'. It was predicted that the younger age group would have higher levels of EMG activity in the arm muscles during the required bimanual tasks. Additionally, it was expected that levels of muscular activation in typically developed children would be highest when visual information of only one arm would be available.

METHODS

A divide was securely placed between the arms of younger aged (n = 11) and older aged (n = 10) children, while seated at a table. The divide could be a transparent screen (glass condition), an opaque screen (screen condition) or a mirror (mirror condition). Accordingly, participants could see: both arms, 1 arm or 1 arm and its mirror reflection. In each divide condition, the participant's head was positioned towards both their dominant or non-dominant hand side. In each hand, participants grasped a handle from an arm ergometer, which was attached to a wooden disc that spun freely through 360^{o} around a vertical axis. The discs were both rotated continuously by the participant in an inward symmetrically coordination pattern at a self selected but constant pace. The protocol consisted of six conditions (3 divides x 2 head positions).

Surface EMG recordings were made from: flexor digitorum superficialis (FDS), extensor carpi radialis brevis (ECRB), biceps brachii brevis (BBB), tricpes brachii longus, (TBL), deltoideus pars anterior (DPA) and deltoideus pars posterior (DPP). To quantify the muscle activity, the signals were normalized to the highest muscle activity obtained from maximum voluntary contraction measurements.

A mixed ANOVA was used to compare the younger with the older children on mean EMG activity with three repeated factors: hand (2 levels), head position (2 levels) and divide (3 levels). Post hoc pair-wise comparisons were performed on main and interaction effects. Alpha was set at 0.05.

RESULTS AND DISCUSSION

Results showed that for the 'younger' age group, mean EMG activity in all of the muscles was consistently higher during the task compared to the 'older' age group. The difference in mean EMG activity between the two age groups reached significance for the ECRB, BBB, DPA and DPP muscles. The expected differences in mean EMG activity between the two age groups might indicate that the younger children had a higher EMG-to-force ratio compared to the older age group, possibly caused by developmental differences in neural activation and muscle physiology. Alternatively, the higher levels of muscle activation for the younger children suggest that they required higher levels of co-contraction to stabilize the joints involved in the bimanual movement. The increased co-contraction together with the increased variability of the movement found in a previous study would imply that the younger children can complete the required bimanual task but are less skilled in the required arm coordination.

In spite of differences found in previous research for the variability of bimanual movements between the divide conditions, no significant differences were found for the amount of muscular activity between the three divides for the typically developed children. This can be explained by the existence of a non-linear relationship between the neuromuscular activation and movement variability. Another possibility is that the novel task and lab setting might have caused the children to increase there muscle activation, which would mask any differences between the divides.

These novel findings provide a baseline for ongoing research on the effects of the 'mirror box' on upper limb coordination in SHCP children. This is important because it helps to understand how improvements in movement coordination of SHCP children in the 'mirror box' condition are related to neuromuscular activation.

SUMMARY/CONCLUSIONS

This study showed that during a circular bimanual symmetrical movement the younger children had higher levels of muscle activity compared to the older typically developed children. However, no differences in mean muscle activity were found when the children placed their upper limbs either side of the opaque screen compared to when they placed their arms either side of the glass and mirror condition.

PATTERNS OF SURFACE ELECTROMYOGRAPHIC ACTIVITY RECORDED DURING THE STANCE PHASE IN TRANSFEMORAL AMPUTEES WITH OSSEOINTEGRATED PROSTHESES

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INTRODUCTION

Prosthetic limbs have shown rapid improvements in the last 100 years. Two recent innovations are the micro-computer controlled prosthesis and the osseointegrated prosthesis. However, despite current developments, approximately one quarter of transfemoral (TF) amputees still remain dissatisfied with their prosthesis. One limiting factor of the lower limb prosthesis is that the neural link between the residual limb and the prosthesis is absent. A communication link between the residual limb and the prosthesis has been developed in upper limb amputees using surface electromyographic (sEMG) signals as a myoprocessor (Cavallaro et al. 2006). However, it has not yet been commercially incorporated into lower limb prostheses largely because of the high degree of variability in sEMG of individual muscles between successive gait cycles. Factors influencing the variability include the presence of noise in the signal and the changing activation patterns of motor neurone pools. The latter factor is related to the observation that there are an indeterminate number of ways in which the muscles can achieve a desired movement (Winter 2004). A central tenet in this study is the concept of the central pattern generator (CPG). The presence of a CPG will produce a regular underlying pattern of signals which is a necessary prerequisite for a sensor to classify different phases of locomotion. Whereas CPG and locomotor control mechanisms have been extensively studied in spinal cord injured subjects (Dimitrijevic et al. 2005), little research has been undertaken on TF amputees. The effect that the grossly diminished afferent input to the spinal cord and the reduced effectors in the residual limb has on the CPG in these subjects is unknown. The objective of the study is to investigate patterns of sEMG activity during the stance phase of locomotion with the aim of attaining an improved knowledge of locomotor control in TF amputees. The clinical significance of this study is to assess the viability of using sEMG as a natural sensor during the stance phase of gait.

METHODS

The study group consisted of five male TF amputees with osseointegrated prostheses. sEMG was measured from five hip muscles of the residual limb. The subjects were identified as A1 to A5. The muscles selected were gluteus maximus (GMX), gluteus medius (GMD), rectus femoris (RF), adductor magnus (AM) and biceps femoris (BF). The signal was collected using the Biometrics DataLINK DLK800 system (Biometrics Ltd, Gwent, UK) with surface pre amplifier type SX230 at a sampling rate of 1000Hz. The site selected for GMX and GMD was in the location recommended by SENIAM (Freriks & Hermens 1999). The electrode placement for RF, AM and BF was determined by palpation and resistive testing. Each subject was asked to walk at their normal pace ten times along a 10m long walkway containing dual 3.3m forceplates in the centre. The data was analysed in MATLAB (The MathWorks, Inc.). Kinematic data was also recorded on the first measuring occasion (MacReflex, Qualisys AB, Sweden). Repeat measurements were taken of two of the amputees 11 months later with the subject identifier having the postscript *rep*. The signal was divided into stance and swing sections. Only the stance phases of gait were further analysed which were time normalised to 30 points. Four types of variables were calculated through application of a linear envelope filter, short time Fourier analysis, adaptive Choi-Williams distribution and discrete wavelet transform. The between stance and between measurement day variability was determined by calculating the coefficient of multiple correlation (CMC). Principle component analysis (PCA) was applied to the variables to determine co-activation of muscles and to reduce the feature set. Cluster analysis was then applied to the reduced feature set to identify groupings within the stance phase.

RESULTS AND DISCUSSION

All muscles displayed increased activity during both isometric and dynamic muscle contraction. Repeatability was low for between-stance sEMG with mean CMCs for the 5 muscles ranging from 0.40 to 0.61. The CMC was highest for the amplitude derived variables. PCA revealed major co-contraction of muscles with data from the 5 muscles being accounted for by one principal component (PC) for certain subjects and stances.

Table 1. Results of cluster analysis applied to the PC matrices derived from the time-dependent parameters. Cluster members described in stance timepoints.

DATASET	Cluster Group 1	2	3	4
A1	1-6	7-16	17-23	24-30
A2	1-11	12-17	18-22	23-30
A3	1-10	11-21	22-30	
A3rep	1-9	10-22	23-30	
A4	1-5	6-15	17-22	23-30
A4rep	1-5	7-12	13-23	24-30
A5	1-6	6-15,27-30	16-19	20-26

Cluster analysis of multi-muscle matrices demonstrated significant cluster patterns for the stance phase (see Table 1). The cluster patterns were composed of 3 or 4 groups for the stance phase. Cluster patterns showed little change for the 2 amputees who had repeat measurements taken eleven months later.

CONCLUSIONS

Poor inter-stance repeatability was displayed emphasising that the locomotor control system does not regulate muscles individually but collectively as a functional unit. Further evidence of functional activation can be observed from the PCA which revealed significant co-contraction of muscles. There appears to be 3 or 4 different control modes during the stance phase as determined by cluster analysis. The precise timing of the control modes is individual to the subject. However, the control mode pattern remains constant for the 2 individuals investigated. The findings suggest that sEMG activity can potentially be harnessed to activate a micro-computer controlled prosthetic joint during locomotion. Further research must be undertaken involving a greater number of subjects, analysing different variables and employing different pattern classifiers.

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EFFECT OF TRANSCRANIAL DIRECT CURRENT STIMULATION (tDCS) ON MOTOR CORTEX EXCITABILITY

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INTRODUCTION

Transcranial direct current stimulation (tDCS) feeds a very weak electrical current through the skull (≤ 1 mA). Cathodal tDCS, where the cathode is placed over the primary motor cortex and the anode above the contralateral eyebrow, leads to a reduction of the excitability of the motor cortex. If tDCS is applied for several minutes, the changes can outlast the stimulation by up to one hour. Stimulating up to 15 minutes with an intensity of 1 mA causes no severe side-effects. So, a tDCS protocol with more than 3 minutes duration, up to 15 minutes, can be used.

The present study was undertaken to confirm the reported influence of tDCS on the motor cortex excitability, which is expressed in routine TMS induced EMG response variables. Intracortical-inhibition (ICI) and -facilitation (ICF) are added to the set of response variables. Reported results on the latter parameters are contradictory.

METHODS

Twelve healthy subjects participated in this study. The study protocol includes three experimental sessions separated by one week. We studied cathodal tDCS since the primary further step in our study is to study therapeutical effects by reducing cortical excitability in patient groups (e. g. in ALS).

Transcranial direct current stimulation
In the subsequent sessions, participants receive cathodal tDCS (1 mA) for 7, 10 or 15 minutes in subsequent sessions. tDCS was delivered using a constant-current stimulator (Eldith, neuroConn GmbH, Ilmenau, Germany) via two conductive rubber electrodes (35 cm^2) inside saline-soaked sponges placed on the scalp (Figure 1). Prior to positioning the electrodes, the skin was prepared to reduce the skin impedance. The target skin impedance was < 35 kΩ.

Figure 1: Set up of a tDCS

Transcranial magnetic stimulation (TMS):
For the evaluation of the impact of tDCS on the excitability of the motor cortex, TMS indices were measured: before tDCS (pre), 5 and 20 minutes after the tDCS (post 1 and 2). We measured three TMS induced EMG responses (MEPs): (i) resting motor threshold of activation (RMT) expressed as changed intensity in TMS output needed to elicit a 0.5mV MEP, (ii) the change in a single pulse motor evoked potential with TMS output needed to evoke a 1mV MEP in the pre condition, (iii) paired-pulse TMS induced short-latency ICI and ICF, caused by a conditioning stimulus with a variable ISI (2,3,10,12 ms) before the test stimulus. EMG activity of the ADM, FDI and APB

muscles was recorded. Visual feedback of the EMG activity of the ADM was given to ensure relaxation. The peak-to-peak amplitude of each MEP (in mV) was calculated.

RESULTS AND DISCUSSION

Two subjects were excluded from the analysis because they did not complete all the measurements or sessions. The results for the RMT (figure 2) and the single MEP amplitude (figure 3) are according to expectations and were significant apart from figure 3 (post2). The bars indicate 95% confidence intervals. The three different tDCS durations did not lead to significant differences.

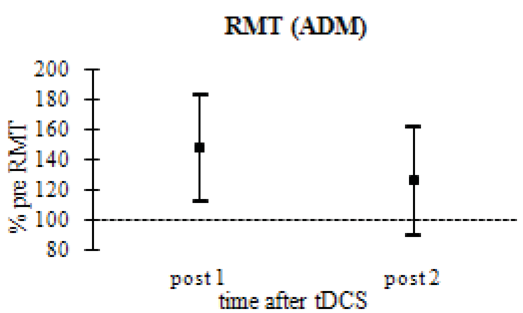

Figure 2: Resting motor threshold in % change of TMS output needed (pre = 100%, vertical bars indicate 95% confidence intervals).

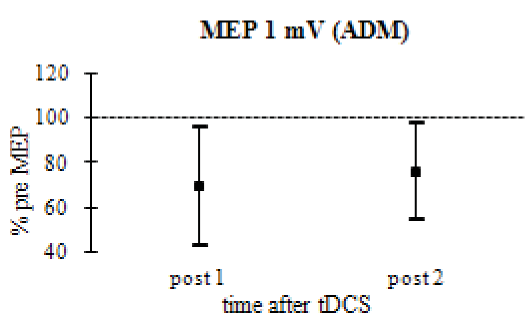

Figure 3: Change in MEP amplitude compared to the pre-session (vertical bars indicate 95% confidence intervals).

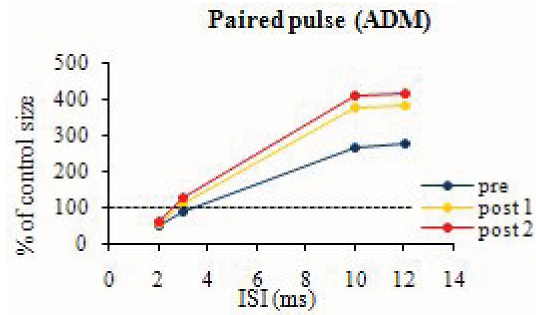

Figure 4: Change in MEP amplitude under inhibitory (ISI = 2,3 ms ISI) and facilitatory (ISI = 10, 12ms) conditions (100% = unconditioned MEP).

One can speculate on the mechanisms underlying tDCS. Induced changes in the resting potential of cortical neurons must be primarily responsible. The paired pulse results (figure 4) show an exaggerated ICF effect (ISI 10, 12 ms). This is counterintuitive, but may contribute to a better understanding of tDCS. Possibly, mainly inhibitory interneurons are depressed by the tDCS in the ICF condition.

SUMMARY/CONCLUSIONS

The effects of tDCS were found to be in line with what was found by others. However, the effects on intracortical interaction were opposite to what we expected. Before clinical applications are envisaged, more basic research towards a better understanding of tDCS is needed.

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SELECTIVE ACTIVATION OF NEUROMUSCULAR COMPARTMENTS IN THE TRAPEZIUS MUSCLE

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INTRODUCTION

For decades, the regions of the trapezius muscle have been considered as functionally independent anatomical subdivisions (Inman et al 1944), termed neuromuscular compartments. However, only relative activity differences between the trapezius subdivisions are observed in different tasks (Johnson & Pandyan 2005; Mathiassen & Aminoff 1997). Therefore, it is unknown whether the anatomical subdivisions of the trapezius muscle can be independently activated by voluntary command.

The neurophysiologic prerequisite for an independent activation of the trapezius subdivisions is the ability to selectively innervate the fine cranial and main branch to the spinal accessory nerve of the upper and lower subdivisions of the human trapezius muscle (Kierner et al 2002).

Biofeedback of muscle activity by electromyography (EMG) is well suited to study voluntary motor control (Basmaijan et al 1963). Therefore, biofeedback from the different anatomical subdivisions of the trapezius muscle was applied to investigate selective activation of the trapezius subdivisions by voluntary command.

METHODS

Six females and 9 males participated in the study. Bipolar surface EMG electrodes were placed at the four anatomical subdivisions (clavicular, descending, transverse, and ascending) of the dominant trapezius muscle. Throughout the entire experiment, the subjects lay prone on a bench with appropriate head support and arms along the side of the body, receiving visual biofeedback of muscle activity from each anatomical subdivision. The subjects performed three isometric maximal voluntary contractions (MVC) of arm abduction with solid bands around the body and arms. Then, the subjects received instructions to selectively activate the anatomical subdivisions without moving arms, shoulders or head.

The EMG root-mean-square amplitude was calculated with a moving window (1 s duration and 100 ms steps) throughout the experiment. The maximal RMS for each anatomical subdivision during the MVCs was used to normalize the EMG recording of the respective anatomical subdivisions. The threshold for "active" and "rest" for each trapezius subdivision was set to >12% and <1.5% of the highest EMG amplitude recorded during the MVCs.

RESULTS AND DISCUSSION

The first task involved activation of the lower (transverse and ascending) subdivisions while keeping the two superior regions at rest. Within a few minutes, all subjects attained a selective activation of the lower subdivisions by voluntary command. The second task was selective activation of the two lower subdivisions separately. Two-thirds of the subjects were able to selectively activate the ascending subdivision, while none of the subjects managed to selectively activate the transverse subdivision. The third task was to activate the upper (clavicular and descending) subdivisions in isolation from the lower subdivisions. Seven subjects succeeded to activate both upper subdivisions in isolation from the two lower subdivisions. Moreover, two subjects attained isolated activation of the clavicular subdivision, while another subject managed isolated activation of the descending subdivision. The final task involved selective activation in absence of EMG biofeedback. All subjects were able to activate the lower (transverse and ascending) subdivisions in isolation from the upper subdivisions. No differences in skill were observed between the genders.

SUMMARY/CONCLUSIONS

The selective activation of the trapezius subdivisions demonstrates the existence of neuromuscular compartments in human muscles. The independent activation of the anatomical subdivisions designates an ability of the trapezius muscle to impose delicate actions on the scapula. The demonstrated voluntary activation of neuromuscular compartments provides new insight about motor control characteristics and function of the trapezius muscle that may be of relevance for pathophysiology and rehabilitation of trapezius myalgia.

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PRACTICE OF SKILLED TRUNK MUSCLE ACTIVATION CHANGES COORDINATION IN AN UNTRAINED TASK

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INTRODUCTION

Exercise management of low back pain commonly involves attempts to change the coordination of the trunk muscles in order to optimize the control of movement and stability of the spine and pelvis (Richardson et al, 1999). A range of strategies is used clinically to train trunk muscle coordination and there is considerable debate regarding the most ideal approach. One approach includes repeated skilled activation of specific muscles that are found to have altered activation in people with recurrent episodes of low back pain, such as the lumbar multifidus. This is followed by more comprehensive functional rehabilitation. It remains unclear whether the repeated skilled activation of the trunk muscles leads to transfer of changes in the recruitment of this muscle in an unpracticed task. The first aim of this study was to determine whether practice of skilled activation of the multifidus muscle induces changes in muscle coordination in an untrained task in people with low back pain. The second aim was to determine whether similar changes could be induced by identical activation of the multifidus muscle, but during a simple unskilled trunk extension task that activates many of the trunk extensors.

METHODS

Twenty volunteers with unilateral LBP were randomly assigned to perform either skilled activation of deep multifidus in a manner that was isolated from the other paraspinal muscles, or extension training to activate deep multifidus at the same intensity but in a non-isolated manner. To test transfer of changes in activation to an untrained task we investigated the activation of the trunk muscles during a small amplitude trunk flexion and extension around neutral upright sitting. Previous work has argued that the minimum activity around this position is an indication of the activity required to maintain stability of the spine (Cholewicki et a, 1997). Subjects slowly flexed and extended the trunk through an arc of approximately 20 degrees before and after the session of back muscle training. Electromyographic (EMG) activity of the erector spinae (thoracic and lumbar), latissimus dorsi, rectus abdominis, and obliquus externus and internus abdominis was recorded using surface electrodes and activity of the deep and superficial fibres of multifidus were recorded with intramuscular fine-wire electrodes. All recordings were made bilaterally. For analysis of the EMG activity during the trunk movement task root mean square (RMS) amplitude of the 12 surface EMG recordings was calculated and the time of minimum activity identified. The EMG amplitude of each individual muscle was also identified at this time point. A post-:pre-training ratio was calculated for the surface EMG RMS minimum activity. RMS EMG activity recorded for 1 s during the exercise tasks was calculated. Pre-:post-training minimum EMG ratios and RMS EMG during the training tasks were compared between training groups with repeated measures ANOVAs. EMG activity of each of the trunk muscles at the time of the EMG minimum was also compared between pre- and post-training values for each group.

RESULTS AND DISCUSSION

Following skilled practice of activation of mutifidus, the EMG activity of the superficial abdominal and paraspinal muscles was reduced in the neutral upright sitting posture when moving between flexion and extension (all: $P<0.05$). In contrast, EMG activity of deep multifidus on the painful side was increased ($P<0.05$). These changes were not observed after non-isolated training. The ratio of the minimum RMS EMG amplitude between post-:pre-training was significantly lower for the skilled training group than the extension training group ($P<0.04$) (Figure 1).

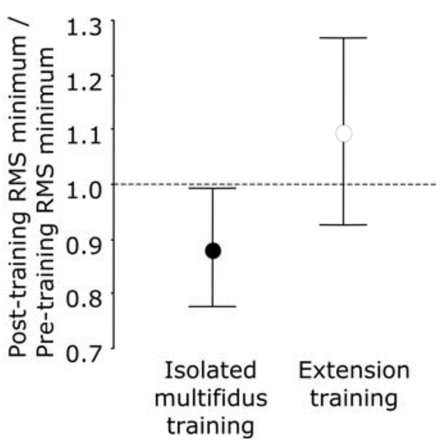

Figure 1. Ratio of minimum RMS EMG amplitude between post-:pre-training for the surface EMG recordings.

Despite the different intentions of the two exercise interventions, and the aim of the skilled training to activate the deep multidus relatively independently from the other trunk muscles, there was no difference in activation of any muscle between training protocols except the superficial multifidus which was recruited to a greater amount in the extension training group ($P<0.03$).

SUMMARY/CO NCLUSIONS

These data show that activation of the trunk muscles can be changed in an untrained task by practice of skilled activation of multifidus. A novel finding was that similar changes could not be induced by extension training, despite identical overall patterns of activity of the superficial trunk muscles. This finding indicates that the difference in neural organization of the two tasks (conscious attention to a specific muscle versus no specific attention to any muscle) leads to different outcomes. This is consistent with greater changes in cortical representation following skill training than strength training (Remple et al. 2001).

We argue that the reduced activation of superficial trunk muscles observed following skilled training may result in reduced stability of the spine. However this is likely to be beneficial as people with LBP often excessively stabilise the spine, with a resultant increase in spinal load and modified damping. The results may help to explain the efficacy of training interventions that involve isolated voluntary contractions.

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INTERACTION OF VISUAL AND PROPRIOCEPTIVE INPUTS FOR THE CONTROL OF OBSTACLE CROSSING

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INTRODUCTION

Various types of sensory inputs, such as visual and proprioceptive, can be used for the control of obstacle crossing during walking. Dynamic visual inputs obtained while approaching the obstacle have been found to be essential for the feedforward control of limb motion (Patla, 2006). Proprioceptive inputs from the leg, on the other hand, can provide online monitoring and feedback control of limb motion.

Specifically, proprioceptive inputs about postural stability arose from the trailing limb might be of particular importance. Conflicting proprioceptive inputs elicited by mechanical vibration to the leg muscles have been shown to alter the neuromuscular control of locomotion in blindfolded healthy adults (Verschueren, 2003). However, how such proprioceptive interference would interact with dynamic visual inputs to affect the control of obstacle crossing is poorly understood.

The purpose of this study was to determine how changes in the availability and accuracy in visual and proprioceptive inputs from the trailing limb affected the control of obstacle crossing.

METHODS

Eight healthy young adults walked on a 10-meter walkway at their comfortable speeds and crossed a low obstacle at the 7th step under two visual conditions, normal and occluded. In the occluded conditions, visual inputs about the obstacle would be blocked by a pair of eye goggles four steps ahead of crossing. For both visual conditions, subjects walked with or without mechanical vibration to the Achilles tendon of the trailing limb.

For the trailing limb, surface electromyography was used to record the activation of the tibialis anterior (TA) and medial gastrocnemius (MG), and the Vicon Motion Analysis System was used to capture the foot motion in order to determine the single support and swing phases.

EMG signals were band-pass filtered (10~300Hz) and rectified. Mean integrated (IEMG) amplitude value was calculated through dividing the integration value of the processed EMG by the duration of the two gait phases respectively. Due to small sample size, only descriptive analysis was conducted.

RESULTS AND DISCUSSION

Vibration did not appear to have apparent effects on muscle activation under normal visual condition (Fig. 1). Under the occluded visual condition, the changes in IEMG amplitude with vibration seemed to be different for the two muscles. In the single stance phase, there was a tendency of greater MG IEMG amplitude with vibration (Fig. 2). In the swing phase, vibration appeared to lead to smaller TA IEMG.

The results showed that conflicting proprioceptive inputs from the ankle had no effect on muscle activation under sufficient visual inputs. Such conflicting sensory information only altered the muscle activation of the trailing limb while dynamic visual inputs were insufficient for obstacle crossing. Thus, it seems that insufficient visual inputs during obstacle crossing may partially disable feedforward control of the limb motion and allow feedback control to play a greater role.

Furthermore, it is likely that proprioceptive inputs containing information about joint angle could be of great significance. Vibration to the Achilles tendon can lead to illusory sensation of plantarflexing. In visual occluded condition, the vibration-induced increased activation in the GM during the stance phase and in the TA during the swing phase could serve to compensate for the illusory ankle plantarflexion.

SUMMARY/CONCLUSIONS

Availability and accuracy of visual and proprioceptive inputs affected the neuromuscular control of the trailing limb during obstacle crossing. Specifically, in the absence of dynamic visual inputs, the role of proprioceptive inputs became more dominant.

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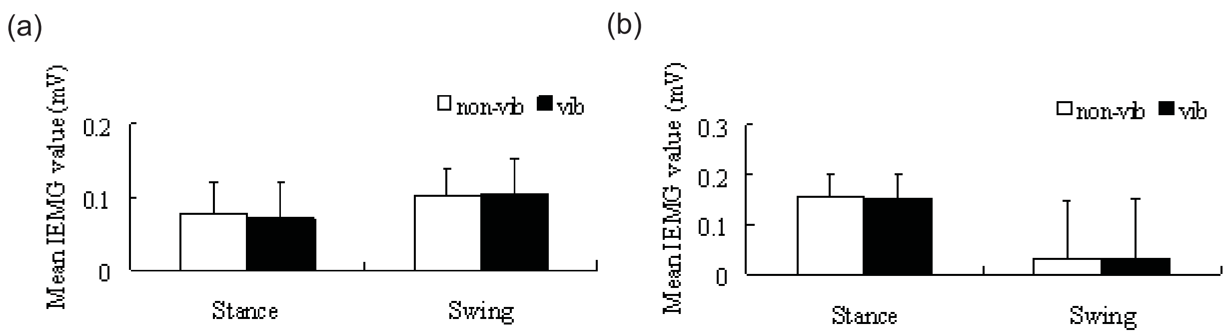

Figure 1: Means and standard deviations of the mean IEMG (integrated amplitude) value of tibialis anterior (a) and medial gastrocnemius (b) of the trailing limb in swing and single support phases of obstacle crossing under normal visual condition, with (vib) and without vibration (non-vib). The effect of vibration appeared to be minimal.

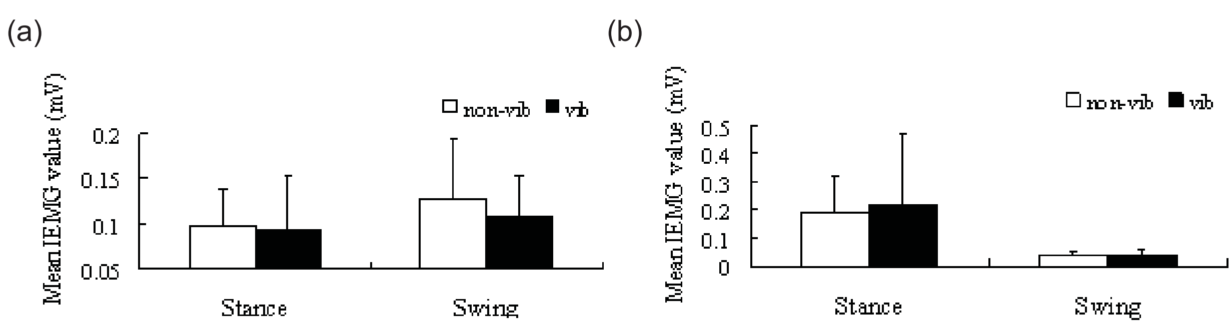

Figure 2: Means and standard deviations of the IEMG value of tibialis anterior (a) and medial gastrocnemius (b) of the trailing limb in swing and single support phases under occluded visual condition, with and without vibration.

EXPERIMENTAL MUSCLE PAIN CHANGES THE SPATIAL DISTRIBUTION OF TRAPEZIUS MUSCLE ACTIVITY DURING DYNAMIC TASKS

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INTRODUCTION

The economic cost of work-related muscle pain in the neck and shoulder region necessitates research in the area.
In particular, trapezius myalgia is a common complaint with increasing incidence (Juul-Kristensen et al., 2006). Most studies investigating the interaction between pain and trapezius muscle activity have been limited to isometric contractions. However, dynamic contractions are more frequent in everyday life and repetitive tasks are considered one of the major risk factors for neck and shoulder pain (Miranda et al., 2008 add Malchaire et al. 2001).
Recent studies using high-density surface electromyography (EMG) (Madeleine et al., 2006; Falla et al, 2008) have shown a change in the spatial distribution of trapezius muscle activity during sustained isometric contractions following noxious stimulation of the upper trapezius muscle. This finding suggests that during painful contractions of the trapezius, the distribution of activity within the muscle is altered with respect to normal activation. This altered motor strategy may have implications for the development of chronic work-related neck / shoulder pain.
The aim of this study was to evaluate whether experimentally induced upper trapezius muscle pain resulted in a reorganization of the distribution of trapezius muscle activity during a repetitive dynamic task consistent with previous observations in isometric contractions.

METHODS

Ten healthy male volunteers (age 26.2 ± 3.1 yrs) participated in the experiment. Subjects were required to lift a 1 kg box between shelves positioned at hip (P1) and shoulder height (P2) with a cycle time of 3 s (Figure 1) for 50 cycles.

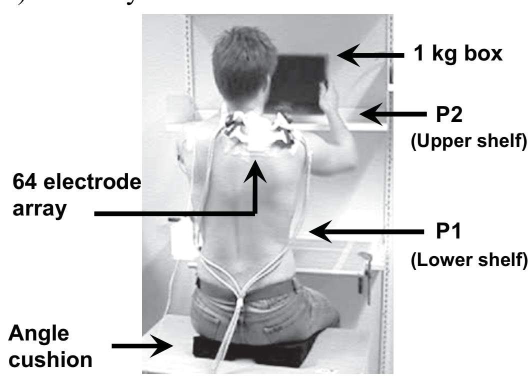

Figure 1: Experimental set-up.

Experimental muscle pain was induced by injection of 0.4 ml of sterile hypertonic saline (5.8%) into the right upper trapezius along the line between the spinous process of C7 and the acromion, 1 cm lateral with respect to the main innervation zone. Isotonic saline (0.5 ml, 0.9%) was injected at a similar location as a control injection. The order of the two injections was randomized. The box lifting task was performed in 4 conditions: baseline,

immediately after injection of isotonic and hypertonic saline and 20 min after the last injection (recovery).
EMG signals were detected from the right trapezius with a 13 × 5 grid of electrodes (Figure 1). Two uniaxial accelerometers were mounted on the box to obtain the start and end points of the cyclic movement. Surface EMG average rectified value (ARV) averaged across the grid of electrodes was computed in a 500 ms interval starting from the instant when the box was lifted from position P1. The centroid of the ARV map was calculated to quantify the spatial distribution of muscle activity. Regression lines of ARV barycentre positions and average values were computed across the 50 values extracted from the cycles.

RESULTS AND DISCUSSION

One way repeated measures analysis of variance (ANOVA) showed a significant decrease of the initial value of average ARV during the painful condition ($P<0.01$). A significant shift of the centroid of the ARV map towards the caudal region of the trapezius ($P<0.05$) was also observed (Figure 2).

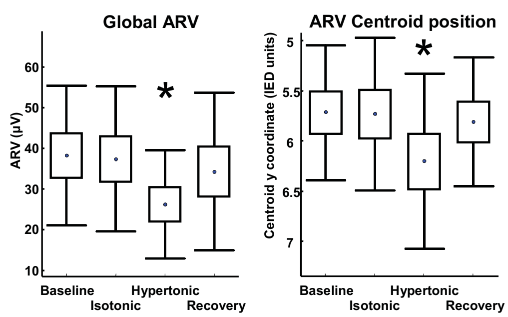

Figure 2: Initial values of ARV averaged across the map (left) and centroid of the ARV map (right). Mean±SD, SE are shown.

The shift of the centroid implies a relatively larger decrease in activity in the cranial with respect to the caudal region of the muscle which is in agreement with previous observations during isometric shoulder abduction (Madeleine et al. 2006).

SUMMARY/CONCLUSIONS

This study shows that it is possible to extract maps of EMG parameters and to track the position of the barycentre of these maps during a prolonged dynamic task.
We confirmed that local excitation of nociceptive afferents selectively inhibits motor neurons innervating muscle fibers with specific spatial localization in the muscle and we showed that the activation strategy to perform a repetitive task in a controlled environment is reorganized in a similar way as during static contractions.

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GENDER-SPECIFIC ADAPTATIONS OF UPPER TRAPEZIUS MUSCLE ACTIVITY TO ACUTE NOCICEPTIVE STIMULATION

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INTRODUCTION

Neck pain affects approximately 60% of the population at some stage of their life with a greater incidence for women than men (Cote et al. 2004). Furthermore, women are more likely to suffer from persistent neck pain (Cote et al. 2004) and female gender is a prognostic factor for poor outcome following a whiplash injury (e.g. Harder et al. 1998).

The reason for gender differences in the prevalence of neck-related disorders and the higher risk for the development of chronic symptoms in women are not fully understood. A factor of influence may be a gender-specific effect of pain on motor control strategies.

In this study it is hypothesized that the mechanisms of adaptation of the relative activity of regions of the upper trapezius during sustained contractions are altered by muscle pain in a different way in men and women. Therefore, the purpose of the study was to examine gender differences in the change of upper trapezius muscle activity distribution in response to experimental muscle pain.

METHODS

Nine men (age: 26.0 ± 4.3 yr; height: 1.7 ± 0.9 m; weight: 74.1 ± 13.6 kg) and nine women (age: 28.2 ± 10.0 yr; height: 1.7 ± 0.7 m; weight: 66.1 ± 7.2 kg) participated in the study. Surface electromyographic (EMG) signals were recorded from multiple locations over the upper trapezius muscle with a 10×5 grid of electrodes during 90° shoulder abduction sustained for 60 s. Measurements were performed before and after injection of 0.4 ml hypertonic (painful) and isotonic (control) saline into the cranial region of the upper trapezius muscle.

To characterize the spatial distribution of muscle activity, the following variables were extracted from the 39 bipolar signals: root mean square (RMS) and mean frequency (MF) averaged over the 39 signals, and the two coordinates of the centroid of the RMS map (*x* and *y*-axis coordinates for the medial-lateral and cranial-caudal direction, respectively).

RESULTS AND DISCUSSION

The peak pain intensity following the injection of hypertonic saline was greater for women (numerical rating scale 0-10: women 6.0 ± 2.1, men 4.2 ± 0.9; $P < 0.01$). For both genders, upper trapezius RMS averaged across the grid decreased following injection of hypertonic saline ($P < 0.0001$). Moreover, there was a relatively larger pain-induced decrease in RMS in the cranial region compared to the caudal region of the muscle for both genders. During the non-painful sustained contractions, the EMG RMS progressively increased more in the cranial than caudal region, for both men and women, due to fatigue. This was observed as a decrease of the *y*-axis coordinate (cranial-caudal) of the RMS map across the duration of the contraction (ANOVA: F = 57.1, P < 0.000001). This mechanism was maintained in men but not in women during the painful condition (ANOVA: F = 4.5, P < 0.05).

The results demonstrate that muscle pain impairs the normal adaptation of upper trapezius muscle activity to fatigue in women but not in men. The change in relative muscle activity during the non-painful contractions is believed to be an efficient strategy to maintain the force by distributing the activity across different regions of the muscle rather than overloading a specific muscle region (Falla and Farina 2007). This mechanism has the functional importance of prolonging the endurance time (Farina et al. 2007) and is considered beneficial for the muscle due to reduced overload of the muscle regions active at the beginning of the task. Assuming that this strategy is the most efficient in the presence of fatigue, since it is the strategy observed in non-painful conditions, the response observed for women during the painful condition may be considered as suboptimal for the muscle control.

It has been hypothesized that continuous muscle activity and overexertion of low-threshold motor units may induce or perpetuate muscle pain (Hagg 1991). Furthermore this hypothesis suggests that prolonged overuse of muscle regions may result in additional peripheral adaptations within the muscle. The results of this study indicate that women performed sustained painful contractions of the upper trapezius with the same regions of the muscle active over the duration of the contraction. This finding suggests prolonged overuse of similar muscle compartments during fatigue and is in accordance with the presence of histological and biochemical alterations of the upper trapezius muscle in women with trapezius myalgia (e.g. Larsson et al. 1990)

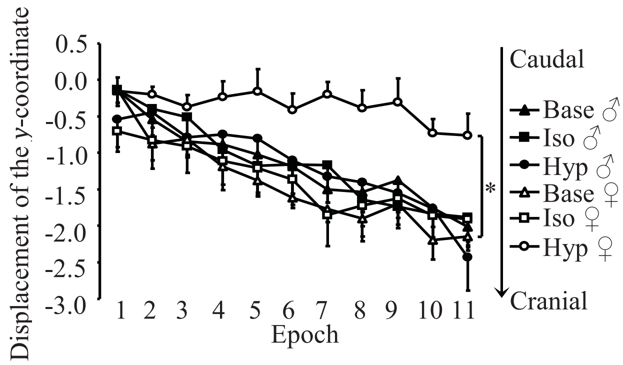

Figure 1: Mean (± SE) of the shift of the *y*-axis coordinate of the RMS map across the sustained contraction relative to the initial position before (base) and following injection of isotonic (iso) and hypertonic (hyp) saline. * P < 0.05

SUMMARY/CONCLUSIONS

Experimentally-induced upper trapezius muscle pain impairs the normal adaptation of muscle activity to fatigue in women but not in men. This differential response may have relevance for the greater likelihood of development and chronification of neck pain in women.

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CONTROL OF PARTIAL SPECTRAL POWER IN DIFFERENT FREQUENCY REGIONS OF THE SURFACE EMG SIGNAL OF A SINGLE MUSCLE

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INTRODUCTION

Subjects can control single motor units using needle EMG recordings (Basmajian, 1963, Basmajian & De Luca, 1985), and electrode arrays show the spatial activation of different regions of a muscle in surface EMG (sEMG) recordings (Grassme et al., 2003). These results have shown the capabilities of humans to voluntarily modify the EMG signal. Based on these and related studies (Scholle et al., 2005), we quantified the capabilities of a subject to simultaneously control the partial spectral power of two different frequency bands in the surface EMG signal of the Auricularis Superior (AS) muscle. Such control over partial spectral power requires a combination of many factors including motor unit recruitment and the precise coordination of their firing rate. Control of different bands of the power spectrum could eventually be used to generate independent control signals that could provide spinal cord injured individuals with a multi-channel interface to assistive technologies.

METHODS

To demonstrate the partial control of different frequency bands of the sEMG signals simultaneously and to determine the frequency and bandwidth of those regions, two approaches were used: a computer simulation and a case study on a single subject.

Simulation: The action potential train from a single motor unit (MUAPt) was modeled as a train of doublet functions, with constant amplitude and firing rate. Hundreds of MUs (each with their own randomly generated amplitude and firing rate) were added in time to simulate an EMG signal. The firing rates of the different MUs were then optimized to find a combination that could simultaneously generate specific spectral power levels within two pre-selected frequency regions. Each MU contributed power across the entire spectral range, thus hundreds of iterations were needed to optimize the firing rates.

Case Study: The sEMG signals of the AS muscle of a single subject were acquired by standard methods using non-disposable surface electrodes (bipolar configuration), and bandpass filtered to obtain signals that contained only one of the two frequency bands required.

Biofeedback using a computer cursor position task was used to train the subject to manipulate the partial spectral power of two different frequency regions. The spectral powers generated in each frequency band proportionally changed either the X or Y position of a cursor on the computer screen.

RESULTS AND DISCUSSION

Results from the simulations proved that mathematical solutions were feasible for frequency bands narrower than 50 Hz, since the powers in these two bands were weakly correlated. Band widths of 20 Hz were used for the case study. Experimental data showed a weak correlation ($R<0.1$) between the 20-40 Hz and 60-80 Hz bands, which were thus selected for these studies.

Figure 1 presents the results obtained in two different cursor-to-target activities (X=10,Y=2 & X=2,Y=10) for the case study. The results show the respective average values and standard deviations (200 contractions) of the X and Y positions generated after the subject completed a training protocol. When the target was placed at the position X=10;Y=2, the power in the 60-80 Hz band was smaller and statistically different than that of the band at 20-40 Hz.

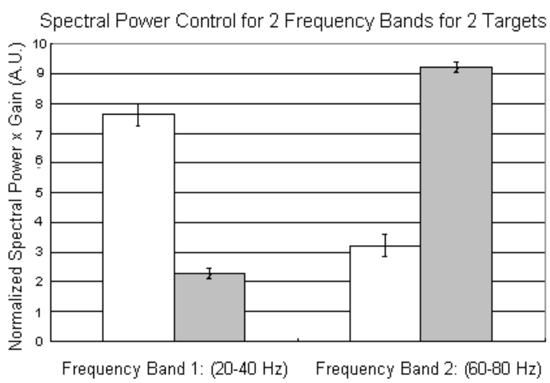

Figure 1: Average power and std. dev. in each frequency band for two different target positions (200 attempts). White bars correspond to the target at X=10;Y=2. The gray bars correspond to the target at X=2;Y=10.

When the target was placed X=2;Y=10, the power in the 60-80 Hz band was more than 4 times greater than that in the 20-40 Hz band. These results also show that the subject was able to voluntarily enhance or suppress absolute power levels within each band, as required by the target position.

SUMMARY AND CONCLUSIONS

Electrophysiological data support simulations showing that human subjects can independently modify the spectral power of two different frequency bands in the sEMG signal of a single muscle, requiring the central nervous system to recruit and coordinate the firing of many motor units. Although many other studies have demonstrated the control capabilities of sEMG signals, the results presented here demonstrate the multiple channel control potential of these signals with obvious implications for developing non-invasive and accessible interfaces for many tasks such as mobility enhancement and computer operation, using a vestigial muscle.

Biofeedback training was key to the results because it allowed the subject to learn to modify the contractions of a single muscle, in order to generate a significant difference ($p<0.001$) in the spectral power of two 20 Hz wide frequency bands. Moreover, this suggests that humans have the capability of voluntarily coordinating the activation of different groups of motor units to generate the differences seen in the partial spectral power of the sEMG signal.

Further studies are needed on a statistically significant group of subjects to determine the extent of their control abilities over different frequency regions of the sEMG. Such control abilities can then be tested on simple external devices such as a computer mouse.

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IMPAIRMENT OF TORQUE-TIME AND EMG-TIME PARAMETERS IN FEMALE WORKERS WITH CHRONIC NECK MUSCLE PAIN

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INTRODUCTION

Neck pain – and especially pain from the upper trapezius muscle - is frequent in employed female office workers (1;2). Maximal muscle strength and neural activation is generally impaired in many different types of neck pain. While these parameters are determined as static peak values, most daily living activities occur during situations of fluctuating force levels. Changes of these parameters in the time domain – e.g. the rate of torque development - may provide more useful functional information than static peak values alone. The present study investigated torque-time and EMG-time parameters during maximal static contraction among workers with and without chronic neck muscle pain

METHODS

41 female office workers with chronic neck muscle pain defined as trapezius myalgia (MYA) and 20 healthy matched controls (CON) participated. Maximal static shoulder abductions were performed at 115° in a Biodex dynamometer. The subjects were instructed to abduct the shoulder as rapid and forcefully as possible. Three maximal contractions of 3-4 sec were performed. EMG was measured simultaneously in the painful upper trapezius and in the pain free deltoideus muscle with a bipolar surface EMG configuration. Signals were sampled at 1000 Hz and filtered using a linear EMG envelope. Peak torque (PT) and peak EMG amplitude (PEMG) were determined as the highest maximal value of the three trials. The rate of torque development (RTD) and rate of EMG rise (RER) was determined as the greatest slope over 100 ms of the rising part of the torque-time and EMG-time curve, respectively. For RTD and RER the trial with the highest RTD value was used. A recording of torque and EMG are shown in Fig.1, with illustrations of PT, PEMG, RTD and RER.

RESULTS

PT was lowered 18 % in MYA compared to CON ($P<0.0001$). Likewise PEMG was lowered 29% in the painful trapezius ($P<0.01$) but not significantly in the pain free deltoideus (14 %, $P=0.11$). RTD was lowered 54 % ($P<0.0001$), and RER was lowered 50% and 44% for the trapezius and deltoideus, respectively ($P<0.0001$), in MYA compared to CON.

CONCLUSION

The present study showed that in females with chronic neck pain torque-time and EMG-time parameters of muscle function are impaired to a larger extent than static peak values. Inhibition of PEMG was specific for the painful trapezius muscle, whereas the RER was lowered in both muscles. This suggests that specific inhibitory feedback from nociceptors inhibited PEMG in the trapezius, while a more common mechanism – e.g. fear of pain – may have inhibited RER in both muscles during the initial phase of contraction.

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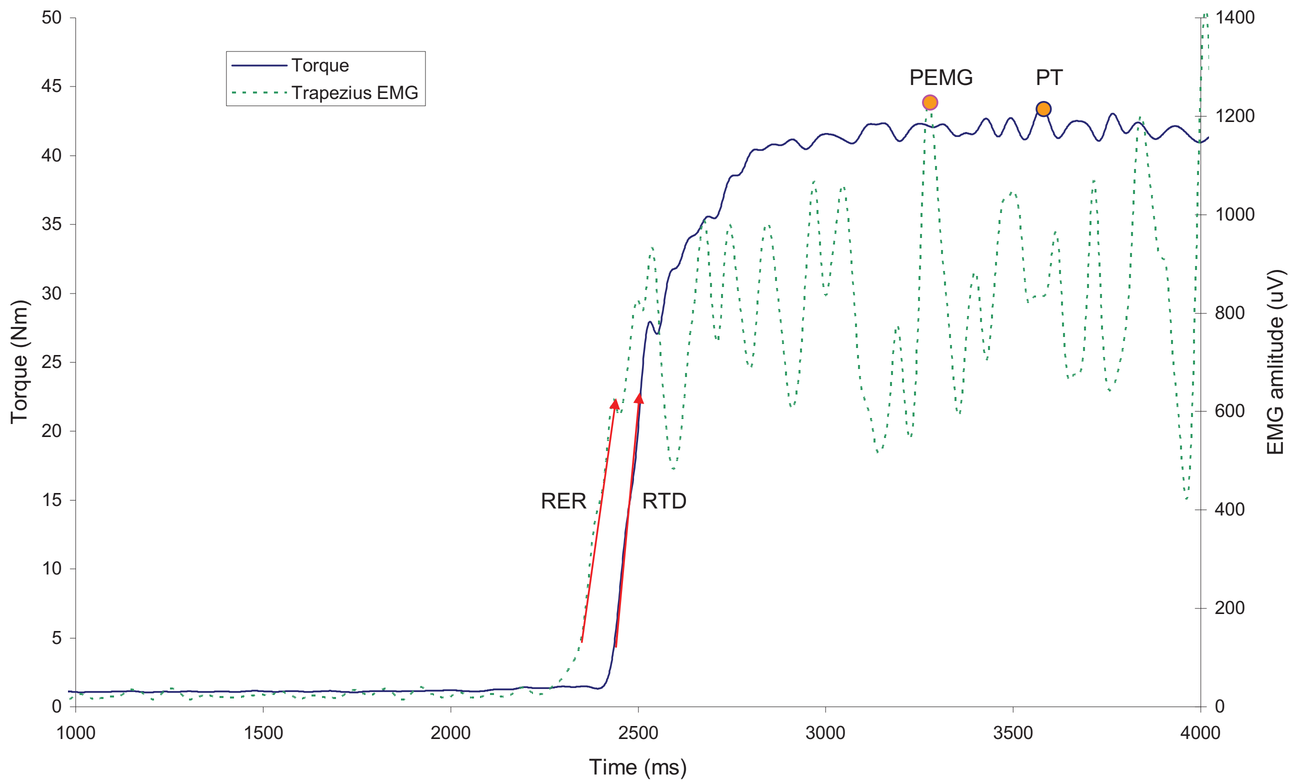

Figure 1. A recording of torque and EMG, with illustrations of peak EMG amplitude (PEMG), peak torque (PT), rate of torque development (RTD) and rate of EMG rise (RER).

HUMAN KINAESTHETIC SENSITIVITY: MEASURING JUST NOTICEABLE DIFFERENCE THRESHOLDS FOR PASSIVE FOREARM MOTION

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INTRODUCTION

Kinaesthesia, the perception of one's body and limbs, is essential for intact motor control. Impairments in kinaesthesia commonly arise after peripheral nerve damage, and recent research indicates that kinaesthetic acuity is reduced in patients a dysfunction of the cerebro-basal ganglia loop (Maschke et al., 2003). This study is part of a series of investigations to systematically study the kinaesthetic acuity in healthy controls and patients with basal ganglia or cerebellar disease. Although the basic neurophysiology of limb perception is understood today, there is a paucity of psychophysical data on the kinaesthetic sensitivity of humans across the lifespan. The aim of this study was to gain normative data from healthy subjects on their sensitivity to distinguish between two passive limb motions. Knowledge about such *just noticeable difference thresholds* for passive limb motion sense can be used to assess the degree of perceptual dysfunction in patients with peripheral and central nervous system disease.

We employed a custom-built passive motion apparatus and used established psychophysical methods to determine the *just noticeable difference thresholds* of passive forearm motion.

METHODS

Subjects. 30 right-handed dominant healthy adults (21 females, 9 males) with an age range of 18-30 years were recruited. Only the right hand was tested.

Apparatus. A subject's forearm was passively moved by a passive motion apparatus (for details see: Konczak et al., 2007). Subjects rested their arms on the splint which was a padded metal rectangle bar (9cm x 60cm). A precision electric stepper motor was used to rotate the splint, effectively producing either elbow flexion or extension. The splint was moved at constant angular velocities between 0.075 ~ 1.65 °/s. The apparatus did not produce any audible noise or vibrations to be used as cues by the subject to detect the onset of motion.

Procedures. Subjects sat on a chair which was adjusted according to their sitting height. Starting position of the splint corresponded to a 90° elbow angle. To insure that participants relied on joint proprioception to motion perception vision was blocked by opaque glasses, and headphones masked any auditory distractions. Each trial presented a pair of angular velocities separated by a 3 second inter-stimulus interval. Using a forced-choice paradigm subjects indicated at the end of the trial, whether the angular velocities of the two stimuli were the "same" or "different

Experimental design. Each pair of stimuli consisted of one standard velocity and one comparison velocity. We tested two standard velocities (1.15 °/s - *fast* condition; 0.60 °/s - *slow* condition). Two fixed staircase procedures with a step size of 0.1°/s were employed (ascending, descending). The range of comparison velocities was 0.65 ~ 1.65 °/s for *fast* condition and 0.1 ~ 1.1 °/s for *slow* condition. Both staircase procedures were intertwined, that is, the ascending and descending staircases were performed in alternating order.

The independent variables were Δ stim, which is the difference of angular velocity between standard and comparison velocity in each trial, and movement direction (flexion/ extension). Angular velocities were presented using a pseudorandom order. A total of 50 trials were administered (25 trials for each condition).

RESULTS AND DISCUSSION

We found that ability to discriminate between two angular velocities increased almost linearly in the *slow* condition (Fig. 1). The just noticeable difference threshold was determined to be the Δ stim at which a probability of 75% correct response was achieved. In slow condition, the thresholds were 0.39 °/s for flexion and 0.38 °/s for extension. Subjects reached 50% correct response at a Δ stim of 0.14°/s for flexion and 0.19°/s for extension.

In *fast* condition, 60% of our subjects reached 75% correct response at a Δ stim of 0.4 ~ 0.5 °/s; however, the mean of all subjects did not reach 75% correct response in the range of Δ stim. Subjects reached 50% correct response at a Δ stim of 0.20°/s for flexion and 0.19°/s for extension.

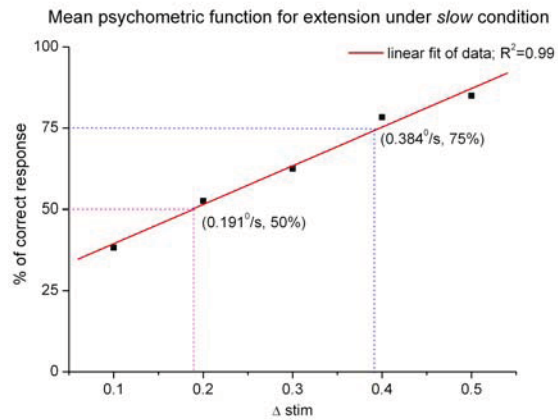

Figure 1: Just noticeable difference thresholds at 50% and 75% correct response rate.

SUMMARY/CONCLUSIONS

The direction of arm movement had no effect on passive motion discrimination thresholds. The sensitivity to discriminate between angular velocities was velocity dependent. For standard velocities below 1°/s the acuity to discriminate to limb velocities was approximately 0.4°/s. For standard velocities above 1°/s acuity will decrease beyond 0.5°/s.

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UNCHANGEABILITY OF THE RELATIVE TIMING OF GAIT

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INTRODUCTION

To improve gait rhythms in Parkinson's disease and cerebrovascular disorder, causing rhythm-formation failure, we perform gait rehabilitation in patients with such disorders, in which they walk while pacing themselves to sounds and vocal rhythms. However, the time ratio of the stance to the swing phase and the muscular activity ratio of these phases at different rhythms during gait have not been clarified. In this study, we examined the time ratio of the stance to the swing phase and the muscular activity ratio of the anterior tibial and gastrocnemial muscles by changing the gait pattern synchronized with rhythms.

METHODS

Eight healthy subjects (age: 21±0.9) with no history of major to their lower extremity participated in the study. The study was adequately explained to all subjects, and performed after obtaining informed consent.

The subjects were asked to walk to 18 rhythmic sounds of 30, 40, 50, 60, 70, 80, 90, 100, 110, 120, 130, 140, 150, 160, 170, 180, 190, and 200 beats/min made using an electronic metronome. The activities of the anterior tibial and gastrocnemial muscles in the right lower limb during gait were recorded at a sampling frequency of 1.5 kHz by surface electromyography. Simultaneously, the stance and swing phases were identified using switches fixed on the right plantar (heel region and the first metatarsal base). Analysis was performed using 15 stable gait cycles. After full-wave rectification of electromyographic signals, the RMS value of each gait cycle was calculated, and the relative muscular activity (iEMG) (%) in the stance and swing phases was determined by defining 1 gait cycle as 100%. The time ratio of the stance to the swing phase (%) was also determined by defining 1 gait cycle as 100%. Statistical analysis was performed using Stat View 5.0 (SAS Inc.) by one-way analysis of variance, the multiple comparison test, and unpaired t-test, and a p-value < 0.05 was defined as significant.

RESULTS AND DISCUSSION

The time ratio was examined by the multiple comparison test, and, based on the results, the subjects were classified into 2 groups: one was group A, in which the gait was divided at a pace of 150 beats/min into 2 patterns, and the other was group B, in which it was divided at paces of 70 and 150 beats/min into 3 patterns. The iEMG ratios in the anterior tibial and gastrocnemial muscles were also classified into groups A and B. The time and iEMG ratios in group A were examined by the unpaired t-test, and there were significant differences in these ratios between the paces of lower and higher than 150 beats/min. In group B, the time and iEMG ratios were examined by the multiple comparison test, and there were significant differences in these ratios between the gait patterns, but the difference in the iEMG ratio in the anterior tibial muscle alone between the paces of lower and higher than 150 beats/min was not significant.

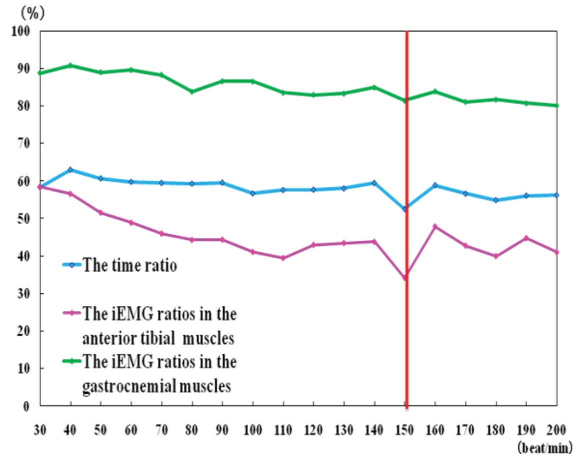

Figure 1: Group A, The gait was divided at a pace of 150 beats/min into 2 patterns.

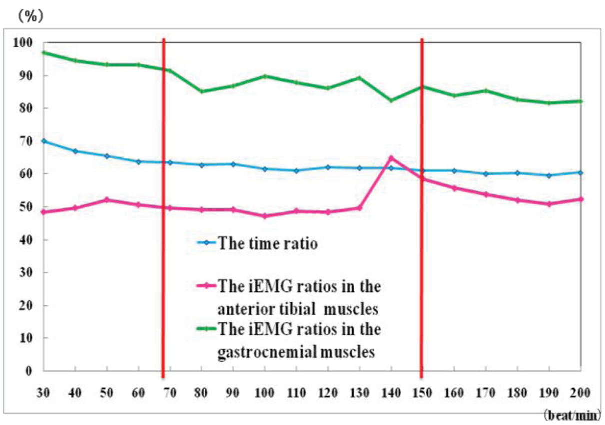

Figure 2: Group B, The gait was divided at a pace of 70 and 150 beats/min into 3 patterns.

When the subjects walked to rhythms, the time and iEMG ratios showed several patterns, and there were significant differences between these patterns. However, gait movements were relatively stable in the patterns. The gait patterns at all paces used in this study were not the same as produced by a single gait program, but several patterns were observed, suggesting that there was a special relative timing of gait in each pattern. Clinically, gait rehabilitation to rhythms is often performed, but it was suggested that it is necessary to perform gait rehabilitation to not only a single but also different rhythms.

CONCLUSIONS

The gait patterns at 30～200(beat/min)paces used in this study were not the same as produced by a single gait program, but several patterns were observed, suggesting that there was a special relative timing of gait in each pattern.

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THE INFLUENCE OF GENDER ON THE ELECTROMYOGRAPHIC SIGNAL CHARACTERISTICS OF QUADRICEPS FEMORIS

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INTRODUCTION

It has been related that muscle fiber diameter and composition of men are different from women's and that these variables influence the electromyographic signal, as for time as for frequency domains (BILODEAU et al., 2003). In general, the highest median frequency values are found in muscles with high percentage of type II fibers or with larger diameter (GERDLE et al., 2000). Besides, some studies have shown that within quadriceps femoris there is a variation in the fiber composition on the muscles that compound it (BILODEAU et al., 2003). In view of the foregoing, the aim of this study was to evaluate the influence of gender on the electromyographic signal characteristics (RMS and median frequency) of the different components of quadriceps femoris.

METHODS

Twenty-three subjects [14 women (22.3 ± 2.1 years) and 9 men (23.2 ± 2.2 years)], all of them healthy, sedentary and with no history of systemic or orthopedic diseases, gave written informed consent to participate in this study. Approval for the project was obtained from the Ethics Committee on Human Research of the institution where the study was developed (protocol number 65/05). Initially, the electrical activity of the vastus lateralis (VL), vastus medialis (VM) and rectus femoris (RF) muscles of the dominant member was obtained by a signal acquisition module, model EMG1000 (Lynx®), with impedance 10^9 Ohms, resolution of 16 bits entry band of ± 5V, and interface with a Pentium III microcomputer. Data acquisition and storage was carried out with Aqdados software (Lynx®), version 7.02 for Windows®.

The subjects remained seated on the Bonet table, with the hip joint at 90° and the knee joint at full extension. Before the electrodes were put into place, the skin was prepared, trichotomized and cleaned with 70% alcohol. To capture the action potentials, 3 simple differential, active surface bipolar electrodes (Lynx®) (20 times gain, IRMC > 100 dB, signal noise rate < 3 μV RMS and inter-electrode distance of 10mm) were used and placed according to SENIAM specifications. The channels were adjusted for a total gain of 1000 times, with band-pass filter of 20-1000 Hz (Butterworth type) and sampling frequency of 2000 Hz. The reference electrode was attached to the anterior tibial tuberosity of the analyzed leg. To measure the force (Kgf) of leg extension a properly calibrated load cell model MM-100 (Kratos®) was used. Signals were collected simultaneously in the 3 electrodes and in the load cell during maximum voluntary isometric contraction (MVIC) for 5 seconds, repeated 3 times and with an interval of 1 minute between them. After collection, the signals were processed in routines specified for the analysis of the Root Mean Square (RMS) in μV and median frequency (MF) in Hz (FFT of 512 points, with *hanning type* window of 256ms) in the *software* Matlab® 6.5.1.

Normality test was performed by K-S test and all data presented normal distribution. For the comparisons among the 3 quadriceps femoris muscles in the same group, the one-way ANOVA and Tukey's post-hoc test was used. For gender comparisons the statistical test used was independent student's *t* test Level of significance was set at 5%.

RESULTS AND DISCUSSION

We verify in figure 1 that male group showed statistically higher leg extension force than female group (p=0.0023). As for RMS (µV) of quadriceps femoris compounding muscles, we verify in table 1 that for female group there was no significant difference among the 3 muscles; for male group, RMS of VM was significantly higher than the values found for VL. In gender comparisons, men values of RMS were significantly higher than women's for all the 3 analyzed muscles (VL, VM and RF). As for median frequency (Hz) analysis, the female group showed higher values for RF compared to VM and VL; for male group there was no significant difference among the 3 muscles. In gender comparisons, there was no significant difference for the 3 muscles (Table 1).

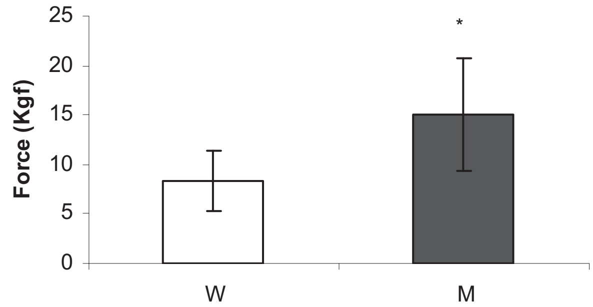

Figure 1: Leg extension force (Kgf) during MIVC of female (W) and male (M) groups, n=23. *p<0.05 in relation to female group. Values expressed as mean ± standard deviation (SD).

CONCLUSIONS

Based on results obtained from this study, we can conclude that force and RMS index of quadriceps femoris of men are higher than women's and that median frequency is not affected by gender differences.

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Table 1: RMS index (µV) and median frequency (Hz) of vastus lateralis (VL), vastus medialis (VM) and rectus femoris (RF) muscles of female (W) and male (M) groups. *p<0.05 in relation to the respective muscle of female group; $^{\#}$p<0.05 in relation to VL of the same group, $^{\psi}$p<0.05 in relation to VM of the same group, n=23. Values expressed as mean ± SD.

	RMS (µV)		Median Frequency (Hz)	
	W	**M**	**W**	**M**
VL	81.91 ± 28.12	154.6 ± 66.47 *	79.52 ± 7.97	81.45 ± 17.34
VM	77.48 ± 45.19	219.61 ± 125.3$^{\#*}$	74.99 ± 13.26	82.14 ± 23.65
RF	78.18 ± 25.8	186.68 ± 74.22 *	86.24 ± 10.62 $^{\#\psi}$	86.98 ± 17.19

DIFFERENCES IN THE MOMENTUM DEVELOPMENT DURING SIT TO STAND TEST BETWEEN FALL AND NO FALL ELDERLY.

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INTRODUCTION.

Rising from a chair (sit - to - stand, STS) is one of the most common tasks of daily life (Cahill et al, 1999). Aging has been shown to produce a progressive decline in the ability to stand up from a chair (Bohannan et al. 2006). The loss of the ability to execute STS, implies an important loss in functionality, independence and quality of life, besides being related with falls, one of most relevant problems in elderly individuals. The momentum developed on the superior body; head, arms and trunk (HAT) in the execution of STS could be related with a loss of balance under dynamic conditions determining higher risk of falls. Therefore, the momentum developed by the HAT would be a good indicator of the ability to execute this task. The aim of this study is to quantify the differences in momentum development during STS in a sample of "fall" and "non-fall" elders.

METHODS.

The sample consisted of twenty three voluntary elderly subjects (n=23), divided into two groups. The first group had 7 elderly adults (five women, two men) with a history of frequent falls (two or more within a year period) and the other group of 16 (six female and ten male) without a record of frequent falls. The following criteria for exclusion were used: i) Dementia (minimental <10). ii); Neurological illnesses; ii) Musculoskeletal disabling pathologies; iii) Vestibular alterations; iv) Uncorrected visual pathologies; v) Obesity or malnutrition.

The kinematics of the upper body (HAT) during sit-to-stand was registered through a motion analysis system (APAS, Ariel Dynamics, Inc. San Diego, USA) with cameras sampling at 60fps. Starting from the kinematic and anthropometric data, we calculate the vertical (P_V) and horizontal (P_H) momentum of the center of mass of the HAT (COM_{HAT}), as well as the angular momentum (*L*) of the HAT segment, in the following manner:

$$P_H = m\dot{S}_x COM_{HAT}$$
$$P_V = m\dot{S}_Z COM_{HAT}$$
$$L = I\dot{\theta}_{HAT}$$

Where, $\dot{S}_z COM_{HAT}$ and $\dot{S}_x COM_{HAT}$ are the velocity of the COM_{HAT} in the vertical and horizontal directions; $\dot{\theta}_{HAT}$ it is the angular velocity of the HAT; m is the mass, and I the moment of inertia of the HAT. As analysis variables we determined the maximum values of the vertical ($P_V M$) and horizontal ($P_H M$) lineal momenta, the minimum (L_{Max}) and maximum (L_{Min}) values of the angular momentum. All the analysis variables were subjected to a non-parametric test (Mann-Whitney) to determine the possible differences among the groups. The level of significance used was of 95%.

RESULTS AND DISCUSSION

No differences in $P_H M$, L_{Max} and L_{Min} (p values >0.05) were observed between both groups. However, a significant difference was found for the $P_V M$ variable (p values = 0.03) between both groups (see figure 1).

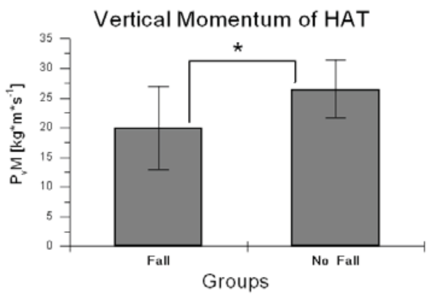

Figure 1: Averages of vertical momentum in fall and no-fall groups. (*p value = 0.03).

Aging results in a progressive loss in the ability to execute tasks. A possible cause of this deterioration could be the inability of generating appropriate magnitudes of force and power in lower limbs (Scarborough et al. 1999). The main discovery of this work was the poor capacity of elderly subjects with a frequent fall antecedent for the production of vertical momentum on the HAT during STS, in comparison with the group with no antecedent of falls. Our results showed that the fall group presents more deterioration in the generation of forces that apply a greater vertical velocity to the HAT. This difference was not observed in the horizontal direction, so that the inability of applying higher velocity to the HAT in the group of fall subjects becomes evident when this segment moves against gravity. This highlights that those subjects with an antecedent of frequent falls present a loss in the capacity to generate appropriate levels of power in the lower extremities, which are the main structures responsible for the change in vertical position that the HAT experiences during STS. Falls are without a doubt, one of the most relevant problems in elderly adults (Reyes-Ortiz et al. 2005). For this reason, it is necessary to design a predictor test for the fall-prone condition. In this direction, Bernardi et al, demonstrated the existence of a relationship between the force of the knee extensors and the capacity to generate velocity at the center of mass during STS in a group of adults with musculoskeletal pathologies. Our results showed that the low momentum production on HAT is a characteristic of the "fall" subjects. Therefore, the values of HAT momenta may be considered a better predictor of falls in elderly subjects. On the other hand, the relevance of fall episodes requires the design of therapeutic training aiming to diminish the risk of falling in elderly adults. Bernardi showed that training improves the ability to generate velocity at the center of mass. Future research should be conducted to determine whether improving the capacity to generate momentum at the HAT can diminish the risk of falls in elderly adults.

CONCLUSIONS

We can conclude that, for the sample studied, the frequent-fall condition relates to a smaller capacity to develop vertical momentum at the upper body.

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DOES WEARING A COMPRESSION STOCKING AFFECT STANDING BALANCE CONTROL/POSTURAL STABILITY?

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INTRODUCTION

Compression stockings (CS) stimulate the wearer's somatosensory system. Pressure on the lower extremities is heightened by wearing CS, and it is considered that static and dynamic postural control might be affected.

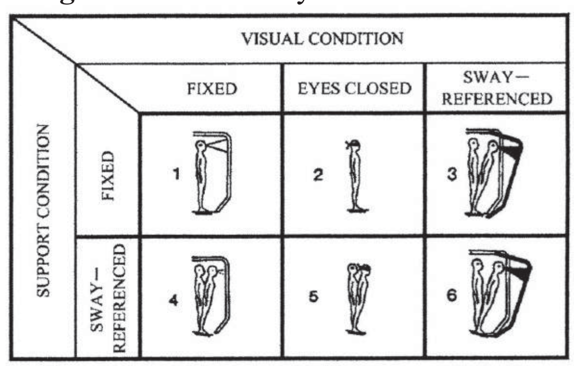
As a precursor to investigating any such affect in the elderly, we carried out a study on the effect of wearing CS on both static and dynamic standing balance control of healthy young adults.

METHODS

Subjects comprised 24 young adults (14 male, 10 female) with no history of disease affecting control over balance. Their average age was 19.2±0.5 years, average height 167.7±9.6 ㎝, average weight 60.1±8.6kg, and average BMI 21.3±1.7. All participants gave informed consent and the study adhered to the research ethics of Ryotokuji University.

Subjects were fitted with knee-high around 25 mmHg CS (Toko Inc.). Static and dynamic balance control and postural stability were measured under 2 treatments, bare feet (BF) and CS, by EquiTest (NeuroCom International, Inc.). Two assessment protocols were used, namely the Sensory Organization Test (SOT) which quantifies postural stability/centre of gravity sway, under 6 sensory conditions (see Figure 1), and the Motor Control Test (MCT) which measures the latency to recover balance under the conditions of medium/large forward /backward simulated horizontal movement of the ground. Order effects were avoided by use of a random number table to determine sequence of SOT and MCT and BF and CS.

Figure 1: Six sensory conditions of SOT.



For statistical analysis, differences between BF and CS scores from the SOT and MCT were analyzed using a paired t test and/or the Wilcoxon test. Statistical significance was defined as $p < .05$.

RESULTS

The average equilibrium scores of BF and CS under the 6 conditions of SOT are shown in table 1. No significant difference between BF and CS was found under any condition

The average balance recovery latency scores of BF and CS under the 4 conditions of MCT are shown in table 2. No significant difference between BF and CS was found under any condition

Table 1: Sensory Organization Test

	BF n=24	**CS n=24**
Condition	Mean equilibrium score	
1	94.0±1.8	93.9±2.1
2	90.9±3.8	91.4±2.9
3	92.1±3.0	91.8±3.5
4	82.8±7.1	80.2±11.0
5	65.9±9.2	65.7±12.0
6	69.8±11.5	68.5±14.0

Table 2: Motor Control Test

	BF n=24	**CS n=24**
Condition	Mean latency (msec)	
Medium backward	124.8±10.8	124.6±13.0
Medium forward	125.0±12.2	124.6±11.3
Large backward	130.2±13.5	131.5±11.9
Large forward	129.0±13.5	130.0±7.8

DISCUSSION

In previous research, it has been reported that intention tremor of ataxia patients is inhibited by touch stimulus and/or pressure applied to the surface of the joint of a limb, and also that cutaneous stimulation of the ipsilateral or contralateral lower extremity increased the quadriceps femoris muscle excitability and the reflex response.

The CS was designed in order to apply graduated compression with maximum pressure at the ankle and pressure gradually decreasing up the leg. In this study, CS of around 25 mmHg were used. However, no facilitatory effect occurred for the proprioceptors of the crural muscles or the mechanoreceptors of plantar surfaces of the foot and ankle joint. As for the level of CS compression, we consider investigation with different compressions is necessary in future studies because only 1 size (around 25mmHg) was examined in this study.

The CS used in this study was knee-high sock types covering the lower leg from the tip of the foot. This type results in a sensation that the whole foot is pressed in from both the right and the left. A large number of mechanoreceptors exist on the plantar surfaces of the feet receiving afferent information necessary for static and dynamic postural control. Therefore, it is considered that CS of a shape or design which does not impede the activity of mechanoreceptors on the plantar surfaces of the feet may facilitate static and dynamic postural control.

Such facilitation was not found in this study. However, we consider continued research using different/improved research protocols is warranted. We will also consider the development of a CS of a new shape.

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MECHANICAL ENERGY AT THE BEGGINING OF GAIT IN YOUNG HEALTHY SUBJECTS

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INTRODUCTION

One of the main activities developed by the human being, and which has captured the interest of movement specialists, has been the biped locomotion. The analytic study about the generation of mechanical energy is the key in the study of the human gait. This analysis shows the principle of conservation of the energy in the motor activity of locomotion. This would be related with motor control, but this has not still been established. The characteristic of mechanical energy generation over corporal segments can be an indicator of the indemnity or deterioration of musculoskeletal and neurological systems. In this study the mechanical energy generation rate was described during the beginning of gait, also establishing the differences between males and females.

METHODS

Nine subjects were measured, (n=9, women: 4, men: 5) with an average weight of 66.2±13.4kg who accepted to participate in this project by means of informed consent. The exclusion criteria consisted in: antecedents of neurological, muscular or skeletal pathology, surgery history in limbs or spine and people who habitually practice some sport discipline. They were placed 4 passive markers located bilaterally on the external border of the acromion process apex in the scapula and on the greater trochanter apex of the femur. With these markers, trunk segment was configured. The volunteers walked 5 times, which was located inside the volume of calibration of the motion analysis system used for the acquisition of this data (APAS System, Ariel Dynamics Inc. San Diego, USA). They realized 2 gaits of adaptation and later on 3 gaits that were registered, being executed to a comfortable velocity selected in natural form by each subject, beginning all with the same foot, which was chosen by the subject in study. For the later analysis of the data in relation to the gait cycle was considered the progression of the same extremity. During the gait analysis, the COM displacement of the trunk segment was measured during the three first steps of the limb with which the subject began the test.
The first step was considered from the heel off (HO), until the heel contact (HC) of the same foot. The following two gait cycles went from the second HC through to third HC of the same limb. The first cycle was considered, for the data analysis, like a 50% of a normal cycle. Each following step corresponds to 100% of a normal gait cycle. This gave a total of 250% of a normal gait cycle.
The total energy (E_T) was calculated in the following manner:

$$E_T = E_{ug} + E_k$$

Where, E_k is kinetic energy, and E_{ug}, the gravitational potential energy.

The lineal regression was calculated between the peaks of total energy and the percentage of the gait cycle (%GC), this way the slope describe the mechanical energy generation rate. To the obtained results they were applied a statistical non-parametric Mann-Whitney test through a statistical analysis software (Minitab, Minitab Inc. company Pennsylvania State, USA), with a significance level of 95% (α=0.05).

RESULTS

The studied subjects presented average slopes of 37.96±11.92 (J/%GC). The average slope for men was 41.71±14.87 (J/%GC), with a mean one of 40.33 (J/%GC). The women presented an average slope 33.28±5.63 (J/%GC) with a mean of 33.79, with a maximum of 38.78 and minimum of 26.77(J/%GC). The subjects didn't present any significant differences as for slopes differentiation between sex ($p>0.05$), according to Mann-Whitney test.
.

DISCUSSION

The quantification of energy slopes and their correlation grade gives precise information about the mechanical energy generation rate. This rate can be useful to future investigations to value the energy generation in pathologies that affect the neurological system as well the musculoskeletal system. The relevance of evaluating the energy slopes during the first three steps, resides in that this gives the information of the energy change from a previous state to the beginning of gait, where the translational velocity of COM is zero, toward a state in which the displacement velocity of COM increases. In many pathologies the rate of energy generation is increased, but isn't evaluated the fact that they vary in the increment from the steady state until later gait cycles, beyond the first three cycles. For the other hand, for the total energy system calculation -in this case, the trunk- although it was considered *Ek* and the *Eug*, it haven't been considered the elastic potential energy neither the energy rotational component.
Previous investigations have described the COM excursion in relation to the vertical axis during continuous gait cycles, giving the position changes variations and COM velocity, of body and their segments. In this research they weren't significant variations as for sex, neither they have related slopes of many energy peaks, in spite of the fact that evidence shows an increase in the energy generation during normal gait cycles (Miller and Verstraere, 1999; Gard et al, 2004). However, this research shows an increase in energy peaks, establishing positive slopes among them.

CONCLUSIONS

i.-The total energy generation rates of trunk segment increased during the first three steps of gait cycle. ii.-They weren't significant differences as for the slopes normalized in the time, between men and women.

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Do we need feedback information to control isometric net joint torque?

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Introduction

Previously, we found that accuracy to keep a certain level of isometric torque is affected by feedback information and fatigue. However, visual inspection of mean isometric torque-time profiles suggests that how the subjects controls the net joint torque at a certain sub maximal torque level is different whether they are aware about how much torque it is applying. In order to verify if torque accuracy strategy depends on feedback information, the objective of this study is to analyze the muscle activation during a torque accuracy task.

Method

Subjects

Thirteen subjects (20-28 years old) participated in this study. They had no locomotors or musculoskeletal pain disorders. They gave informed consent according to the procedures approved by the local Ethical Committee.

Apparatus

The measurement of submaximal isometric torque at the knee was provided by a isokinetic dinamometer (Biodex System 3, USA). We recorded the eletromyographic signals of m. vastus medialis obliquo, VMO; m. rectus femoris, RF; m. vastus lateralis, VL; m. biceps femoris, BFL; m. gastrocnemius laterallis, GL; m. semitendinosus, ST; and m. tibialis anterioris, TA. Both systems were connected to an acquisition system (Noraxon, Myosystem 1400, USA), controlled by software (Myoresearch 103.04, Noraxon, USA). The sampling frequency was 1 kHz.

Protocol

To warm-up, the subject walked on a treadmill for 5 min, and performed 20 knee flexion-extension at the dynamometer at 120°/s. We applied SENIAM (Surface EMG for a Non-Invasive Assessment of Muscle) (Hermens et al., 2000) recommendations for EMG. The subject sat at the dynamometer system and flexed its right knee at 60° fixed to the machine. The first task was to perform two maximal voluntary isometric contractions (MVIC) for 10 s each. After 20 min, the subject performed the accuracy test: to execute 3 sets of 2 submaximal MVIC. At the first subMVIC, it should look at the computer monitor to see how torque it was applying to the system. After 10s rest, it should perform the second subMVIC without the visual feedback information. The torque levels (20, 40, 60, and 80% of MVIC) were randomly ordered. To induce fatigue, we asked to the subject to hold the 80% MVIC as long as it could support. After reaching the fatigue, the subject runs the accuracy test again.

Data processing and statistics

EMG signal was low-pass filtered (400 Hz, 4th order Butterworth filter). The filtered EMG was rectified and low-pass filtered (200 Hz, 4th order Butterworth filter). And we calculated the EMG and torque standard deviations during the beggining [0.75, 1.25]s, the middle [3.74, 4.25]s, and the end [6.75, 7.25]s time-windows.
The standard deviation was analyzed across conditions. We ran 4-way ANOVA to test the effect of feedback information (with or without visual feedback), the effect of fatigue (before and after fatigue exercise); torque levels (20, 40, 60, and 80% of MIVC); and instant of contraction (beginning, middle, and end). We set the significance level at $p < 0.05$. For post hoc analysis, we run Tukey test.

Results

We ran ANOVA to test the effect of fatigue, torque level, instant, and feedback information on variability of muscle activation and isometric knee torque. All muscles were affected by torque level ($F_{(576,3)}>64$, $p<0.0001$) and visual feedback information ($F_{(3,576)}>6.6$, $p<0.01$). The post hoc test showed that the variability of muscle activation increases as the target load enhances ($p<0.0001$) (but for TA and GL at 20 and 40% MIVC) and increases without feedback ($p<0.01$).
VL and ST muscles were affected by fatigue ($F_{(1,576)}>11.7$, $p<0.001$). For both muscles, variability increases after fatigue ($p<0.001$).
Torque level ($F_{(3,576)}=21.8$, $p<0.0001$) and visual feedback information ($F_{(1,576)}=6.1$, $p=0.01$) affected the torque variability. Torque variability increases as the level of sub maximal torque increases (except 20 and 40%) and increases without feedback.
We did not find any effect of instant on the variability of muscle activation and torque.

Discussion

Although our initial expectation, the torque and EMG variability did not change along the isometric task because we did not find any effect of instant on those variabilities. On the other hand, all muscles presented more variability with increasing isometric loads and without visual feedback information. The increasing EMG variability suggests that flexible combination of muscular activations may provide the expected level of net joint torque. It is important to stress that not only knee extensors participate into this flexible combination, but also knee flexor muscles and ankle muscles.
Only ST and VL showed more variability after fatigue. It suggests that after fatigue the muscle coordination changes.
Torque variability grows as the level of sub maximal torque increases (except 20 and 40%) and it increases without visual feedback information. The higher variability during stronger isometric torque or without feedback information shows that torque accuracy depends on how much force someone needs to apply.

Conclusion

The decrease in torque accuracy occurs with increasing loads. As net joint torque increases, more variable is the EMG activation. Feedback information about the net joint torque is important to avoid the decrease in torque accuracy.

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SPONTANEOUSLY ACCELERATIVE MOTOR BEHAVIOR: MAPPINGS OF THE SEGMENTED ELBOW SpAce

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INTRODUCTION

Human motion is generally described in terms of scalar kinematic parameters, such as speed, accuracy, excursion, and smoothness (or its opposite: jerk). While these parameters provide useful comparative indices, they are opaque to the underlying complex series of neuromuscular events that cannot be compressed into a single number. The goal of this report is to develop a new parameter that broadens our view of motion, to include its spatiotemporal variability.

Herein we present an assessment of motion smoothness, based on accelerative transients in the trajectory record. Spontaneous accelerations (SpAce) are mapped in a high-resolution, spatially distributed profile indicating position and magnitude of the locally unsmooth aspects of a movement. We compared a set of volunteers with no known motor impairments, and a cohort of chronic stroke patients.

METHODS

8 volunteers (4 healthy subjects, 4 chronic stroke patients) were seated with their right arms in the Mechanical Arm Support and Tracker (MAST), supporting the elbow of their dominant arm against gravity, and recording goniometric elbow motion data. Stroke patients were all right-affected. Subjects moved smoothly within their maximal range of motion, articulating their elbow in cyclical flexion-extension motions.

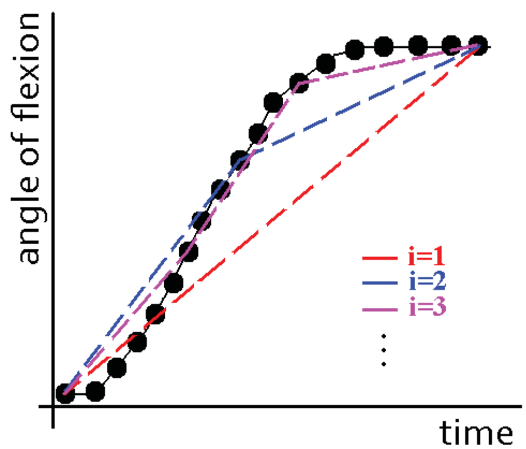

Figure 1: Schematic depiction of the SpAce map: overlapping intervals obtain motion data. Each time point (circles) is compared against linear segments spanning iteratively finer portions of the recorded motion as *i* increases.

Several repetitions were committed over the course of a single visit.

Angular trajectory motions were compared against progressively finer resolution piece-wise linear functions for identification of regionally non-linear trajectory traces (Figure 1).

For each iteration, *j,* linear segments of approximately equal (and successively smaller) length were matched to the motion record and RMS values were computed over each interval as piecewise linear functions. Heat maps of transient departure from these linear segments were generated from the linearly combined RMS scores (Figure 2).

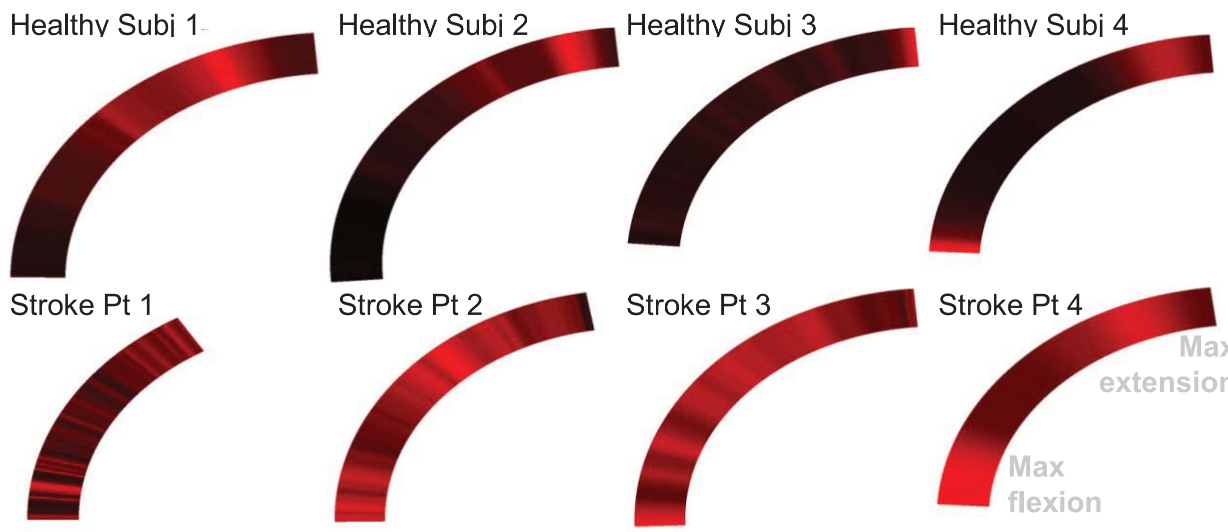

Figure 2 Spasticity heat maps across healthy (*Top*) and chronic stroke (*Bottom*) subjects. Traces span subjects' ranges of motion, represent ensemble averages of OSI, normalized scale. *Note compromised ability to extend about the elbow in stroke patients 1 and 2.* Traces reflect a top-view of hand path through space; red=greatest spasticity, black=least spasticity.

$$E(\varphi) = \frac{i}{N^2 - N} \sum_{i=1}^{N-1} \sum_{k=\beta}^{\beta+\frac{N}{i}} \left(\left(\theta_\beta + \frac{\theta_{\beta+\frac{N}{i}} - \theta_\beta}{N/i} \left(k - (\beta - 1)\right) \right) - \theta_k \right)^2 ; \qquad \beta = \min\{\alpha - \varphi \geq 0\}$$

The mean error value of a given location φ within a subject's range of motion due to accelerative transients is given by $E(\varphi)$: where $\alpha = \{1, \frac{2N}{i}, \frac{3N}{i}, \cdots, N\}$ $\quad i \in \Re < N$.

RESULTS AND DISCUSSION

SpAce maps of control subjects were relatively black and hence smooth, except for a single departure from constancy in the distal regimes of the workspace. As seen in **Figure 2**, chronic stroke patients exhibited typically noisier maps, indicating jerkiness at various regions of their elbow activation profile (based on ensemble averages of a minimum of 10 repetitions in a single session).

SUMMARY/CONCLUSIONS

SpAce maps, over the range of an individual's joint workspace yield a high-resolution spatial mapping of speed variation across the range of motion.

Preliminary analysis of joint activation profiles suggests that hemi-paretic individuals are variable across a much greater proportion of the workspace compared with non-neurologically impaired persons.

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EFFECT OF EMG BIOFEEDBACK ON FORCE PRODEUCTION CONTROL

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INTRODUCTION

Biofeedback is augmented information about the state of different body parts such as heart, muscle, brain, etc. Electromyography (EMG) is a typical biofeedback method that provides information about the tension and activity level of muscles (Blumenstein, Bar-Eli, Tenenbaum, 2002). This type of biofeedback is used as an effective intervention for muscular relaxation or strength increment (Horowitz, 2006, Voreman, 2004, Voreman,vollenbroek,-Hutten ,Hermens, 2006). Some authors have shown that EMG biofeedback is useful for posture stability or decrease of misalignment (Lee, Wong, Tang, 1996, Bogdanov, Nikolaeva, Mikhailenok, 1990, Metherall, Dymond, Gravill, 1996, Nouwen, 1983, Shafizadeh, 2007). But the role of this method for parameterization or motor response programming is not well-documented. Therefore, the aim of present investigation was to study the role of EMG biofeedback in control of force production. Previous studies have demonstrated that augmented feedback about the outcome of movement such as amount of produced force can facilitates the response planning (Kohl, Guadagnoli,2001) because it adds to intrinsic feedback (sensory information) and increases the recall schema that is necessary for response planning (Schmidth, Wrisberg, 2004).

METHODS

40 right-handed university students were participated voluntarily and divided into four groups; feedback, biofeedback, mixed and biofeedback-criterion. They were naïve about the task. Hand grip dynamometer (Takei, Japan) was used for force-production task. Wave Rider, 2 cx, is used for measuring surface EMG and providing biofeedback information.
The experiment was consisted of 25 trials in acquisition and 5 trials in transfer phases. All participants should to estimate their force outcome after 3 seconds of each trials completion. After that, the experimenter provided them augmented feedback. In addition the two biofeedback conditions and mixed groups have seen their muscular tension of flexor carpi radialis muscle through biofeedback device and by graph that is produced in monitor. So, those groups could check their muscular contraction during movement, but the feedback group only received the terminal information. The biofeedback-criterion group also received feedback about the proper muscular tension via produced force before acquisition trials in 5 times to check the EMG results with produced force of dynamometer.

Force control was computed by absolute constant error (ACE). Analysis of variance is used for comparison between groups and least significant differences (LSD) post hoc test is used for follow-up analysis.

RESULTS AND DISCUSSION

The mixed ANOVA results have shown that there was significant main effect for intervention method ($F_{3,37}$= 2.9, P<0.05) . The LSD follow-up has shown that the ACE for biofeedback groups was higher than feedback and mixed groups (P<0.05).But, there was no significant difference between feedback and mixed groups (p>0.05) in two phases.

These results demonstrated that the merely biofeedback information was not effective for response plan same as knowledge of result. But, combining it with the acquired results is beneficial for response correction. Thus it is recommended that for better results, the practitioners should to set the biofeedback information according to desired outcome but not separately, because of it guides the performer's attention to ultimate goal of intervention that is correct movement or better interpretation of EMG activity and it is useful for produce the correct response through the control of muscles contraction.

CONCLUSION

Biofeedback besides its tension reduction and muscular force increment also is useful for response planning or force control if it combines with outcome information. This role is helpful for motor control of disabled individuals or athletes for succession in their training period.

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INFLUENCE OF CADENCE OF DYNAMIC CONTRACTIONS ON SURFACE EMG

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INTRODUCTION

Surface electromyography (sEMG) is a very useful method that helps to evaluate muscle force gradation strategies during different tasks (De Luca, 1997). However, most of the studies are based on isometric contractions and non-fatigued conditions, which are not always well related to daily activities. Farina (2006) discusses that the interpretation of sEMG in dynamic contractions becomes a difficult task because besides the mechanical issues (ex: muscle-length tension; moment of arm; external forces; and moment of inertia), one must deal with the relative shift of electrodes to the myoelectric activity source, which can disrupt this signal. Therefore, regarding to all those variables that can interfere in the acquisition and analysis of sEMG signal, it seems that a useful interpretation of sEMG demands some sophisticated methods that are not available in commercial EMG systems.

Thus, the purpose of this study was to evaluate the effects of dynamic contractions of three different cadencies on the sEMG signal of *biceps brachii* during full elbow flexion-extension movement, regarding frequency and temporal domain analysis.

METHODS

Fifteen male subjects (22.8 ± 3.6 years), all right-handed, performed dynamics contractions at 20% of maximum voluntary contraction (MVC). The acquisition system consisted by an electromyography (3000P - Mega Electronics, Ltd, Finland) and a computer (Pentium II). The sampling frequency and gain were set at 1 kHz and 1000, respectively. Surface electrodes (Ag/AgCl - 3M Korea Ltd., Seoul, Korea; 1 cm of diameter) were placed over the subject dominant *biceps brachii* muscle following SENIAM protocol (Hermens et al., 2000). A metronome (GMT-200P, Groovin USA Design) was used to control cadence during the tasks in three different and arbitrarily selected cadences (30, 45 and 60 bpm), which were randomized prior to sEMG signal acquisition.

The subjects performed full elbow flexion-extension movement while standing and with the left hand resting on the hip (Figure 1).

Figure 1: Position adopted by the subjects while performing the three tests.

The subjects were instructed to reach the limits of elbow range of movement (flexion and extension – FE cycle) while synchronized by the metronome. The subjects performed seven full FE cycles in each cadence, and the 4th one was considered for the analysis. A set of execices was conducted prior to data collection for familiarization purpose.

The median frequency (F_{med}) and the root-mean-square (RMS) value were calculated from each 4th full FE cycle. One-way ANOVA was applied for analysis and the level of significance was set at 0.05.

RESULTS AND DISCUSSION

An increasing behavior of RMS value was observed as the cadency increased (Figure 2), although no statistical significant difference was detected ($F_{(2,42)}$=1.002; p=0.376).

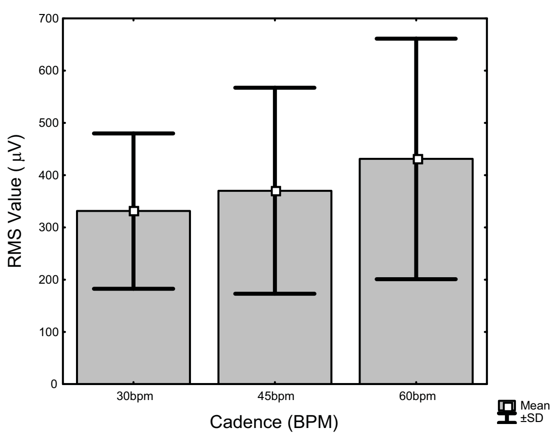

Figure 2: The RMS values (mean ± SD) observed for the three cadences.

Considering that the high variance of the data would explain the lack of statistical singificance found, one might interpret that higher cadence, which means increasing in velocity of movement, would demand higher levels of motor unit recruitment and firing rates. Therefore, the summation of more motor unit action potentials would conduct to the increasing of sEMG signal (Beck et al., 2006).

Regarding to F_{med}, no trend was observed and there was no statistical significant difference ($F_{(2,42)}$ = 0.21065; p = 0.81091) among the three cadences.

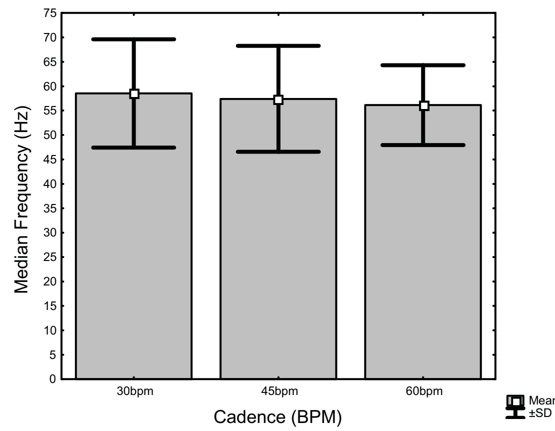

Figure 3: The F_{med} values (mean ± SD) observed for the three cadences.

This result suggests that it was not possible to obtain any interpretation of muscle gradation force strategies by means of sEMG frequency analysis related to the task. Some authors (Hermens et al., 2000) suggest that spectral analysis is not feasible in cases when the signal is disrupted by non-stationarities as can be seen in anisometric contractions.

SUMMARY/CONCLUSIONS

Despite the absence of significant statistical differences, the results reinforce the RMS value as a useful parameter in dynamic contractions, allowing one to establish a suitable interpretation about muscle effort or power gradation properties.

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STUDY OF THE OCCLUSAL SPLINT FOR SLEEP BRUXISM: AN ELECTROMYOGRAPHIC ASSOCIATED TO HELKIMO

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INTRODUCTION

The sleep polissonographic studies indicated that SB occurs generally during the fragmental periods of sleep, with the increase of electroencephalographic (EEG) and electromyographic (EMG) activities, as well as the increase of cardio respiratory frequency.
Medications for symptoms control, muscle exercises, physiotherapy, occlusal splints and occlusal rehabilitation are some of the described treatments for bruxism.
Occlusal splints are frequently used in SB, in order to protect teeth from damages resulting from the contraction force of mandibular muscles or to reduce the orofacial pain by relaxing masticatory muscles.

METHODS

The sample was composed by 15 subjects(14 female and 1 male) ranging from 19 to 29 years old (average of 22,13 $\pm$ 2,72 years). All subjects were undergraduate students in the city of São José dos Campos, São Paulo, Brazil.
The inclusion criteria adopted demanded that patients were all undergraduate students, bearers of SB, with presence of signs and symptoms of temporomandibular disorder and that they have never been treated with occlusal splints.
Clinical diagnosis of SB was made under world standards of diagnosis, based upon patient history and orofacial examination . Teeth's wearing was evaluated with a dental mirror and adequate light. Upper and lower casts were made to analyze teeth wear degree· Diagnose of muscular hypertrophy was also taken considering patient age and dental facial morphology.·
This study was previously approved by the ethical committee from UNIVAP – University of Vale do Paraíba..
The EMG study used a system of electromyographic signs registration (*EMG System do Brasil Ltda*., Brazil) of 8 channels of analogical input with amplification of a thousand times, filter with frequency range of 20 and 500 Hz and digital analogical converter of 16 bits of resolution. The frequency sampling was of 2000 Hz by channel. The EMG registration were realized using bipolar surface electrodes of 10 mm of diameter, after realizing muscle function proof of masseter and temporal [25]. The cleansing of skin was done with ethylic alcohol 70% to reduce the impedance.

RESULTS AND DISCUSSION

The descriptive data of electromyographic variations (rest/ isometry) from RMS (Root Mean Square) values in the masseter and temporal muscles, left and right sides, are shown on Table 1. and values from

Kolmogorov-Smirnov Tests are available which demonstrate normal distribution of electromyographic variables.
According to the values of T-pared Test, with a confident interval of 95%, there was no statistically significant difference on the muscles studied before and after wearing occlusal splints for 60 days.
The difference between the averages of EMG values of both right and left masseter muscles (rest/isometry) was not significant.
It was not found significant difference on the averages of RMS (rest/ isometry) before and after the occlusal splint wearing both on right and left temporal muscles.

SUMMARY/CONCLUSIONS

Occlusal splint used for 60 days by SB bearers presented a significant reduction in the clinical signs and symptoms of TMD. However, the electromyographic evaluation of the surfaces of masseter and temporal muscles in bearers of SB under therapy with occlusal splint for 60 days did not present significant difference before and after treatment.

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Table 01: Descriptive data of electromyographic variations (rest/ isometry) from RMS values in masseter and temporal muscles, left and right sides

Group	Before splint using		After splint using	
Side	**Right**	**Left**	**Right**	**Left**
Temporalis	142.20± 5.8 μV	66.45± 2.2 μV	181.77± 9.3 μV	57.20± 1.6 μV
Masseter:	58.60± 2.6 μV	42.79± 6 μV	45.68± 1.5 μV	54.37± 4.3 μV

BIOMECHANICAL ANALYSIS OF MOUTH'S ORBICULAR MUSCLE IN INDIVIDUALS CLASSE II:STUDY ELECTROMYOGRAPHICS

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INTRODUCTION

Bucal respiration is believed to bring serious effects in cranio facial development and occlusion, when there is no correlation among the internal and external forces of bucal musculature. This way, for a precocious diagnosis and the formulation of a suitable tratment plan it is fundamental to know whether the individuals' peribucal musculature has suffered ambiental influences, to the point of altering its physiology. This electromyographic study's purpose was to compare the medial superior region of mouth's orbicular muscle in two groups : G1 (predominantly nasal respiratory pattern) e G2 (predominantly bucal respiratory pattern).

METHODS

50 brazilian children from 6 to 9 years old were evaluated, 25 boys and 25 girls with Angle's Class II division 1 malocclusion. Signals were acquired by an electromyograph composed of differential double surface electrodes with 20 times pre-amplification, amplifier with 1000 times gain, 20 to 1.0 kHz band pass filters, commom mode rejection of 120 dB, 16 bits A/D converter and sampling frequency of 2.0 kHz per each channel by (**EMG System do Brasil Ltda**). The electrodes were placed bilaterally over the mouth's orbicular muscle. All applicable recomendations from the International Society of Electrophysiology and Kinesiology (ISEK) regarding electromyography's applications were followed in all EMG signal's procedures. Signal handling consisted in full wave rectification, linear envelope through 4th order Butterworth filter, with 5 Hz cut off frequency, normalized in time base and amplitude, this one through average value. EMG signal's intensity variability was calculated through the variability coeficient (CV). The comparison between the EMG signals from the various muscles was made with the t-test , with significance level of 0.05

RESULTS AND DISCUSSION

Variations of around 20% were found among samples during the pronunciation of letters A and F, not found at rest.

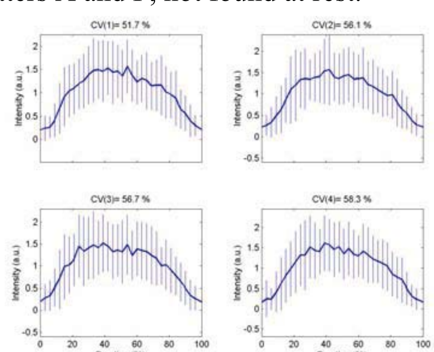

Figure 1: EMG signal's intensity variability calculated through the variability coeficient (CV).

SUMMARY/CONCLUSIONS

Other new observations to confirm these findings should be performed, specially with older children, to detect the time frame in which occurs a differentiation (the time frame in which, for instance, the bucal habits are installed, and the time frame in which ortodontists or phono must interfere), resulting in labial incompetence. The deep knowledge of muscular dynamics gives the basis to the correct terapy. In this context, electromyography becomes a vast exploration field, with valuable contributions not only to Ortodonty, but also to Physioteraphy and Phonoaudiology.

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Table 1. Mean and standard deviation (SD) of normalized electrical activity readings of the orbicular muscle (T Student, $P< .05$).

Muscle	predominantly nasal respiratory	predominantly bucal respiratory
orbicular muscle:	54.6 ± 2.6 μV	79.1 ± 2.6 μV

THE EFFECT OF MUSCULAR VIBRATION ON ANKLE MUSCLE ACTIVITY IN LOCOMOTION OF STROKE PATIENTS WITH IMPAIRED ANKLE POSITION SENSE

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INTRODUCTION

Continuous sensory integration from bilateral lower extremities is crucial for locomotion control (Dietz et al. 1992). Impaired joint position sense after stroke might affect sensory control of locomotion. Therefore, sensory integration during locomotion might be altered in patients with such sensory dysfunction.

Mechanical vibration can activate primary muscular afferents and cause additional and often conflicting sensory inputs (Goodwin et al. 1972). It has been reported that additional, conflicting proprioceptive inputs could alter the neuromuscular control of locomotion in healthy adults and sensory integration for stance control after stroke (Bonan et al. 2004; Verschueren et al. 2003). However, the effect of such conflicting proprioceptive inputs of stroke patients with impaired joint position sense is still unknown. The purpose of this study was to investigate the effect of Achilles tendon vibration on ankle muscle activation in locomotion of stroke patients with impaired ankle joint position sense.

METHODS

Thirty stroke patients were recruited. A joint position matching test that involved matching the ankle joint angles of the two legs was used to determine the joint position sense. Seventeen stroke patients were found to have impaired ankle joint position sense and participated in the study.

Subjects walked on an 8 meters walkway at their comfortable speeds first without and then with the affected and unaffected Achilles tendon vibrated. Muscle activities of the affected tibialis anterior and medial gastrocnemius were obtained by surface electromyography (EMG). Mean integrated EMG value was calculated through dividing the integration value of processed EMG signal by phase duration. Footswitches were used to detect the gait phases.

A multivariate repeated measure ANOVA (vibration × gait phases) was conducted, and the LSD procedure was used as post-hoc tests. Significance level was set at $p<0.05$.

RESULTS AND DISCUSSION

The result of MANOVA showed that the EMG amplitude of the 4 muscles had significant main vibration effect ($p=0.002$). However, no interaction with gait phases was found ($p=0.525$). Post-hoc tests showed that the EMG amplitude of the affected tibialis anterior was significantly larger in the unaffected vibration than the non-vibration ($p=0.024$) and affected vibration ($p=0.014$) conditions (Fig. 1). There was no significant difference between affected and non-vibration conditions (Fig. 1).

The results show that conflicting proprioceptive inputs from the affected versus the unaffected Achilles tendon had significantly different effect on the neuromuscular control of stroke patients during locomotion. First, as expected, sensory inputs from the unaffected side, even transmitting information conflicting to other inputs, were used for locomotion control in stroke patients. Second, and most interestingly, such conflicting inputs from the impaired joint, i.e. the affected ankle, did not alter the neuromuscular control of locomotion of stroke patients. This finding contradicts to what was found in healthy adults for locomotion control under the condition of sensory conflict (Verschueren et al. 2003). Therefore, it might indicate the afferent information from sensory impaired joint might not be used for the sensory integration during locomotion in stroke patients.

SUMMARY/CONCLUSIONS

Conflicting proprioceptive inputs from the affected Achilles tendon vibration did not change the level of muscle activation during locomotion in stroke patients with ankle joint position sense impairment, while those to the unaffected Achilles tendon did. These findings imply that the sensory integration might be altered due to sensory dysfunction and the sensory information from sensory impaired joint might not contribute to locomotion control of stroke patients.

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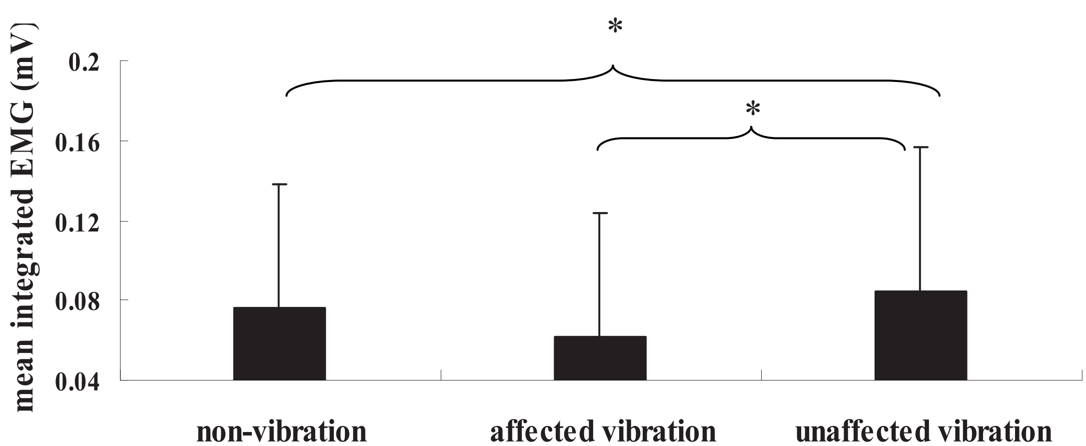

Figure 1: Mean integrated EMG of the affected tibialis anterior. *: statistically significant differences.

A COMPARISON OF AMPLITUDE PROBABILITY DISTRIBUTION FUNCTION MEASURES OF MUSCLE ACTIVITY IN SYMPTOMATIC OFFICE WORKERS

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INTRODUCTION

Past studies have commonly reported on the Amplitude Probability Distribution Function (APDF) as measures of muscle activity amplitudes during different occupational or functional tasks. The levels of APDF that have commonly been reported are 10^{th}% APDF, 50^{th}%APDF (median) and 90^{th}% APDF. The difference between the 10^{th}% and 90^{th}% APDF has also been used as an indicator of the extent of variation of muscle activity amplitude (APDF range). Increased median muscle activities in the trapezius have been reported in several studies as a manifestation of altered motor control – an important mechanism contributing to the development of Work-related Musculoskeletal Disorders (WMSD)(Szeto et al, 2005).

The present study aimed to examine whether such altered motor control was also present in other parameters of the APDF – the 10^{th}%, 90^{th}% and range values.

METHODS

Female office workers who preformed a minimum of 4 hours of daily computer work were recruited. They were divided into Cases (n=21) and Control (n=18) Groups, based on their past and present history of WMSD if any.

Surface electromyography (sEMG) was measured using the Noraxon Telemyo system (Noraxon, U.S.A. Inc., U.S.A.) with a sampling frequency of 1000Hz and band-width of 10-500Hz. Bipolar Ag-AgCl (3M™ Infant Red Dot™) surface electrodes of 15mm diameter (3M Hong Kong Ltd, Hong Kong) were placed on the muscles with an inter- electrode distance fixed at 20mm. The muscles of cervical erector spinae (CES) and upper trapezii (UT) bilaterally were examined and standard normalization procedures were performed with 3 trials of maximum voluntary contractions (MVC) of 5 sec each.

Each subject performed 3 tasks (typing, mousing and type-and-mouse) of 20 mins each. EMG data was captured for 30 sec repeatedly for 5 trials during each task. The tasks were mainly designed to involve the simple actions of typing and/or mousing, without any stressful mental demands.

The EMG data were processed with a high-pass filter at 20Hz, a low-pass filter at 200Hz and notch filters at 50Hz and 60Hz to reduce the noise levels. Signals were demeaned, rectified, down-sampled to 10Hz root mean square (RMS) values and normalised to compute the %Maximum EMG (MEMG). The dependent variables of 10^{th}%, 90^{th}% and the APDF range were compared between the 2 groups, among the 3 tasks.

RESULTS AND DISCUSSION

Case subjects had significantly greater 10th%APDF and 90th% APDF than control subjects for CES, with a similar non-significant trend for UT. For example, for right CES during typing, Case 10th% APDF (mean =11.43%MEMG$\pm$6.05) and 90th% (17.64%$\pm$9.08) were significantly greater than Control 10th% (mean=7.98$\pm$4.56; $F_{1,37}$=4.621, p =.038) and 90th% (mean=12.42$\pm$5.92; $F_{1,37}$=6.305, p =.017). The right UT muscle showed similar trends but the group comparison was not statistically significant (10th%: $F_{1,37}$=2.159, p =.150; 90th%: $F_{1,37}$=2.472, p =.124). The 10th% group means for right UT were 9.55% MEMG $\pm$8.41(Case) and 7.37%$\pm$9.43 (Control), and the 90th%APDF showed 14.59%$\pm$12.05 (Case) and 11.37%$\pm$13.96 (Control).

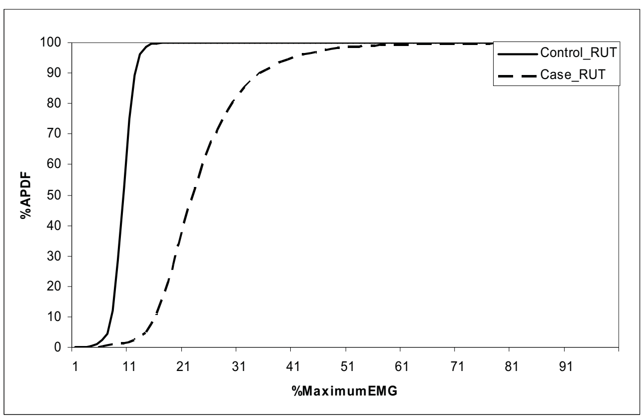

Figure 1: Illustrations of typical APDF curves in right Upper Trapezius for Case and Control Group subjects

SUMMARY/CONCLUSIONS

The APDF range is a new variable being examined and it can be considered as an indicator of the slope of the APDF line (see Fig. 1). The results showed that the Case Group had significantly greater APDF range compared to Control Group across all 3 tasks for CES, with a similar non significant trend for UT. For example, right CES Case range was 6.21$\pm$3.42 compared to Control range 4.44$\pm$1.92. This implies that the muscles were working with a greater extent of variation from lower to higher activities, and this was consistently found in both the bilateral hand tasks (typing) as well as unilateral tasks (mousing).

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INTER INDIVIDUAL VARIABILITY OF EMG PATTERNS AND PEDAL FORCE PROFILES IN CYCLISTS

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INTRODUCTION

The variability in human movement has been the focus of numerous studies across multiple disciplines within the movement sciences. In fact, it is well documented that the nervous system has multiple ways of accomplishing a given motor task. This phenomenon, named "motor equivalence" by Lashley (1933), results from the fact that the number of muscles is substantially larger than the number of joints. It allows the central nervous system (CNS) to adopt a personal control strategy and makes it possible that different muscles activation patterns can result in the same end-effector trajectory and force pattern.

Cycling task represents a typical multi-joint movement characterized by several degrees of freedom. In contrast with other movements, the constant circular trajectory of the pedal constrains lower extremity displacement. Despite that, some studies have reported a high variability of EMG patterns even when in trained cyclists (Hug et al., 2004; Ryan and Gregor, 1992). However, there is a lack of information concerning the inter-subject variability of the the pedal force application patterns. To our knowledge, no previous study has focused on the putative inter-individual differences in pedal force components (effective and total force) as well as on the EMG patterns of the main lower limb muscles in the same population

Thus, the purpose of the present study was to determine whether the relatively high inter-individual variability in EMG patterns during pedaling is accompanied by variability in the pedal force application patterns. It was hypothesized that, in a population of trained cyclists, forces profiles would exhibit a very lower inter-subject variability compared to EMG patterns.

METHODS

Eleven cyclists were tested at a sub maximal power output (i.e. 150 W) on an electronically braked cycle ergometer (Excalibur Sport, Lode®, Netherlands) equipped with instrumented pedals specifically design for pedaling load measurements by VélUS group (Department of Mechanical Engineering, Sherbrooke University). Total pedal force and its effective component (effective force) were measured continuously and were then synchronized with surface electromyography signals measured in ten muscles of the right lower limb: gluteus maximus (GMax), semimembranosus (SM), biceps femoris (BF), vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), gastrocnemius medialis (GM) and lateralis (GL), soleus (SOL) and tibialis anterior (TA). Individual EMG patterns and mechanical profiles were obtained from averaging data across 30 consecutive pedaling cycle, normalizing to the mean value calculated over the complete pedaling cycle.

The inter-subject variability of EMG and mechanical patterns was assessed using standard deviation (SD), coefficient of variation (CV), variance ratio (VR) and coefficient of cross-correlation (R_0, with lag time=0).

RESULTS AND DISCUSSION

The results demonstrated a high inter-subject variability of EMG patterns for bi-articular muscles as a whole (and especially for GL and RF) and for one mono-articular muscle (TA; Figure 1). CV values are depicted in Figure 2.

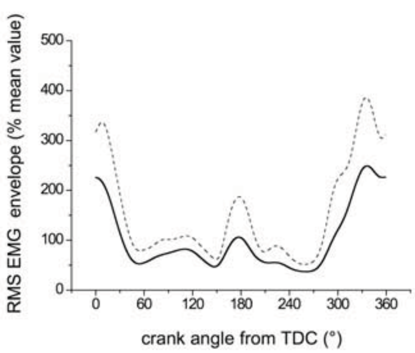

Figure 1: RMS EMG envelope for TA muscle obtained during pedaling. Solid line indicates mean and broken line indicates standard deviation. TDC, Top Dead Center.

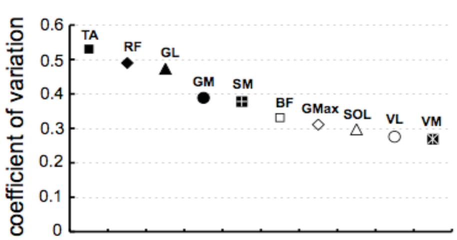

Figure : Coefficient of variation of complete cycle EMG RMS patterns for the ten muscles.

However, this heterogeneity of EMG patterns is not accompanied by a so high inter-subject variability in pedal force application patterns (Figure 3). A very low variability in the two mechanical profiles (effective force and total force), was obtained in the propulsive downstroke phase although a greater variability in these mechanical patterns was found during upstroke and around the top dead center.

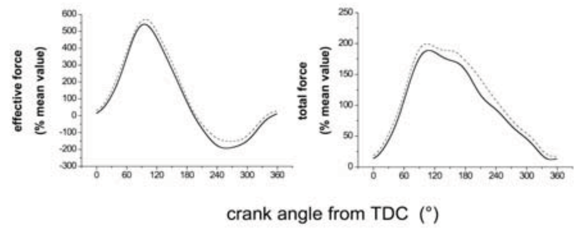

Figure : Inter-individual variability of effective force and total force profiles. Solid line indicates mean and broken line indicates standard deviation. TDC, Top Dead Center.

SUMMARY/CONCLUSIONS

This study shows high inter-subject variability of EMG, especially for bi-articular muscles. It suggests that despite their high and homogeneous level of expertise, cyclists adopt a personal muscle activation strategy during pedaling. However, this heterogeneity of EMG patterns is not accompanied by a so high inter-subject variability in pedal force application patterns. These results provide additional evidence for redundancy in the nervous system: the nervous system has multiple ways of accomplishing a given motor task as has been suggested previously by the "motor equivalence" concept.

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USE OF ELECTROMIOGRAPHY DURING FEEDING TERM AND PRETERM BABIES: LITERATURE REVIEW.

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INTRODUCTION

To achieve growth and proper craniofacial development, not only is normal genetic outworking necessary but also external stimuli such as breathing, sucking (breastfeeding), swallowing and chewing. Movements of baby suckle help harmonic facial growth when maxillary muscles are stimulated to rise in the correct direction (Lawrence et al., 2007).

Electromyography (EMG) can be used to evaluate feeding methods of suckling babies. Considering the scarcity of studies in this area demonstrating differences between sucking muscles in infants, this study was intended as a review of literature about the use of EMG during the sucking of term and preterm babies.

METHODS

Studies were searched for in the following databases: Medline (Jan 1966 to Dec 2007), EMBASE (Jan 1985 to Dec 2007), Scielo (Jan 1997 to Dec 2007), CINAHL (Jan 1982 to Dec 2007), Ebsco and Ovid. The search strategy consisted of the following keywords: electromyography, suction, breastfeeding, bottle-feeding, cup-feeding, facial muscles and premature.

The authors screened search results for potentially eligible studies. When titles and abstracts suggested a study was potentially eligible for inclusion, a full copy of the study was obtained.

RESULTS AND DISCUSSION

Five studies were found about term infants and two studies related to preterm babies.

Inoue et al. (1995) verified differences between breastfed (n = 12) and bottle-fed babies (n = 12). The muscle studied was the masseter; the authors found that the activity of this muscle was significantly reduced in bottle-fed babies when compared with the group of breast-fed babies.

Tamura et al. (1996) investigated the activities of the temporal, masseter, orbicular and suprahyoid muscles of 25 infants during sucking. The results pointed out that the muscular activity of temporal, masseter and orbicular muscles were most active when the sucking pressure and the suprahyoid showed highest activity in the negative-pressure phase.

Sakashita et al. (1996) investigated temporal, masseter, orbicular oris and suprahyoid activities during the bottle-feeding of babies (n = 12). The authors observed that the masseter muscle activity of babies who were fed with chewing type bottle teats was similar to that of breast fed babies (control group) qualitatively and quantitatively. Therefore, muscular activity in these babies was visibly different than in babies who fed with sucking-type bottle teats ($P < 0.05$).

Tamura et al. (1998) examined EMG activity of the perioral muscles during breastfeeding in fifty-six infants. The activity of the suprahyoid increased significantly with age, while there was no appreciable increase in the activity of the temporal, masseter and orbicular muscles in either the cross-sectional study or the follow-up. Moreover, the active tongue-and jaw-lowering movement may play a primary role in increasing sucking strength during the suckle-feeding period in infants.

Gomes et al. (2006) studied masseter, temporal and buccinator muscles in three groups of babies (n = 60) (during breastfeeding, bottle-feeding and cup-feeding). This study was the only one related to cup-feeding babies. Authors concluded that the similarities between muscle activity in the breastfeeding and in the cup-feeding groups suggest that cup-feeding can be used as an alternative infant feeding method, preferable to bottle-feeding.

Two studies evaluated muscular activity of preterm babies. Daniëls et al. (1986), investigated digastric and mylohyoid muscles of 18 preterm infants. The authors discussed that differences in feeding efficiency could not be related to differences in sucking rate. Long bursts of sucking and high milk intake during the movements analyzed were related to quick, efficient drinking. On the other hand, slow inefficient drinking was characterized by short sucking bursts and a small amount of milk intake.

Nyqvist et al. (2001) investigated the orbicularis oris muscle of twenty-six preterm infants during breastfeeding. Besides EMG analysis, the authors assessed the agreement between direct observation of sucking and EMG data by two raters (considered high). In conclusion, the authors provided evidence of early sucking competence in preterm infants during breastfeeding through EMG signals.

CONCLUSION

Studies found in this literature review showed the importance of using the EMG as a method of feeding assessment, not only in term babies but also in preterm babies. This is because of the difficulties that the preterm babies have in dealing with cup-feeding. Studies are necessary in this area, especially because of the hospitalization requirements of these babies. Yet, more research, using EMG data with direct observation, to understand sucking patterns in breast, bottle and cup-fed babies are necessary and must be considered by health professionals.

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SELECTIVE ACTIVATION OF NEUROMUSCULAR COMPARTMENTS IN THE SERRATUS ANTERIOR MUSCLE

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INTRODUCTION

The human serratus anterior muscle consists of nine digitations arising from the upper nine ribs inserting into almost the whole length of the ventral vertebral border and the apex of the scapula (Gray 2000).The muscle acts together with the trapezius muscle to rotate and stabilize the scapula (Büll et al 1989).

Both the serratus anterior and the trapezius muscle are commonly implicated in the development of neck and shoulder pathologies (Cools et al 2007). Restoring of normal function in these muscles is therefore a common objective in treatment of in several types of shoulder and neck pathologies (Cools et al 2007). Numerous studies have investigated the motor control of the trapezius muscle (e.g. Westgaard and De Luca 2001). However, the motor control of the serratus anterior muscle is only recently examined (Alexander et al 2007).

The human serratus anterior muscle is, like the trapezius muscle suggested to be divided into separate functional parts. Already in 1944, Inman and colleagues described how the lower digitations of the serratus anterior, together with the inferior part of the trapezius, constitute the lower component of the scapular rotary force couple. As well, the upper digitations of the serratus anterior and the upper trapezius constitute a functional coupling whose activities are essentially the same (Inman et al 1944).

Recently, we reported a selective activation of the different parts of the human trapezius muscle by voluntary command with biofeedback guidance (Holtermann et al 2008). Thus, if the opinion of a functional coupling between the upper and lower parts of the serratus anterior and the trapezius muscle (Inman et al 1944) holds, the different parts of the serratus anterior can be expected to be selectively activated by voluntary command as well. Therefore, the main aim of this study was to examine whether the different parts of the serratus anterior can be selectively activated by voluntary command with biofeedback guidance.

METHODS

Four males participated in the study. Bipolar surface EMG electrodes were placed at the 5^{th} to 8^{th} digitations of the dominant serratus anterior arising from the 5^{th} to 8^{th} rib. The electrodes were placed in line with the fiber directions above the thickest muscle part of the digitations. Throughout the entire experiment, the subjects lay prone on a bench with appropriate head support and arms along the side of the body, receiving visual biofeedback of muscle activity from each digitation. The subjects performed isometric maximal voluntary contractions

(MVCs) of scapular protraction with the shoulder at 90º flexion, and during upward scapula rotation with flexed shoulder (Ekstrom et al 2005). The MVCs were performed lying on a bench pushing against solid bands. Then, the subjects received instructions to selectively activate the different digitations of the serratus anterior without moving arms, shoulders or head.

The EMG root-mean-square amplitude was calculated with a moving window (1 s duration and 100 ms steps) throughout the experiment. The maximal RMS for each digitation during the MVCs was used to normalize the EMG recording of the respective digitation. The threshold for "active" and "rest" for each serratus anterior digitation was set to >12% and <1.5% of the highest EMG amplitude recorded during the MVCs.

RESULTS AND DISCUSSION

After a few minutes of practice, all subjects were able to activate the lower digitations (7^{th}, 8^{th}) of the serratus anterior by voluntary command while keeping the two upper digitations at rest. Moreover, two subjects succeeded to selectively activate the lowest (8^{th}) digitation in isolation from the other digitations. None of the subjects managed to selectively activate the two upper digitations (5^{th}, 6^{th}) in isolation from the lower digitations.

SUMMARY/CONCLUSIONS

The selective activation of the lower digitations of the serratus anterior in isolation from the upper digitations indicates that at least a part of the serratus anterior can be independently controlled by voluntary command.

From a motor control point of view, this finding supports the possibility of a functional coupling between the upper parts of the serratus anterior and the trapezius muscles like described by Inman (1944). Since the serratus anterior and the trapezius muscle may work as synergists but also in an uncoupled fashion, the muscles do not have a direct reflex connection (Alexander et al 2007). Therefore, a functional coupling between the serratus anterior and the trapezius muscles is likely to require a selective activation of the different parts of the muscles by voluntary command.

The different parts of the serratus anterior muscle are considered to be key contributors for normal scapular motion and control (Inman et al 1944). Therefore, the selective activation of the parts of the serratus anterior and the trapezius muscles by voluntary command may be relevant for rehabilitation of neck and shoulder pathologies (Cools et al 2007).

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THE EFFECT OF PRECUES ON THE STAGE OF RESPONSE PROGRAMMING

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INTRODUCTION

The parameter precuing is a technique for studying motor programming. In this technique, partial or complete information about the response are presented in advance of a stimulus. The numerous experiments using this technique have shown that the precues reduce the reaction time (RT) (e.g., Anson et al. 2004; Eversheim & Bock, 2002), the mechanisms by which this reduction is achieved, however, remain disputed.

The effect of precue on different stages of sensorimotor processing was identified (e.g., Eversheim & Bock, 2002). However, according to cortical cell assembly theory of motor programming (Wickens et al., 1994) RT is a function of degree of overlap of muscle representations in the motor cortex, and in previous studies this factor was not controlled.

Therefore, the present study independently varied the number and nature of precued parameters, degree of overlap of cell assemblies, spatial extent of precues, and number of choices under 6 conditions, with the purpose of investigating the mechanisms of the effect of precue on RT in the stage of response programming of sensorimotor processing.

METHODS

Thirty-four (17 male, 17 female) non-athlete right-handed students of Islamic Azad University in age range 18-24 years selected randomly and participated in two experiments.

The task of Exp. A was isometric force production immediately after appearing target. They performed the task by parameter precuing apparatus under the following 5 precue and 1 control conditions (Fig. 1):

1. Two -choice, large spatial extent, high overlap (known direction);
2. Two-choice, small spatial extent, high overlap (known direction);
3. Two -choice, small spatial extent, low overlap (unknown direction);
4. Three-choice, large spatial extent, high overlap (known direction);
5. Two -choice, small spatial extent, low overlap, mutually conditional combination of direction and force precues; and
6. Non-precue.

In Exp. B, participants performed key pressing task under the precue conditions same as Exp. A in order to eliminate the relationship between precues and response parameters. The participants repeated each condition 30 times with random order after performing 200 training trials.

RT of different conditions in each experiment was measured and analyzed by 6 (condition) * 30 (repetition) ANOVA with repeated measures and Bonferroni post hoc test.

RESULTS AND DISCUSSION

The results of Exp. A indicated effects of number of alternatives, precued parameters, and degree of overlap of cell assemblies ($p<.01$); but the main effect of repetition and interaction of 2 factors were not significant ($p>.05$)(Fig.2).These results were consistent with the findings of Anson et al. (2000, 2004). They observed the effect of number of alternatives and precued parameters on reaction time in an aiming task.

According to the results of similar statistical analysis for Exp. B, all significant differences in Exp. A; except the differences between control and other conditions, were abolished ($p>.05$)(Fig.2). The findings of Exp. B were consistent with Eversheim and Bock (2002).

SUMMARY/CONCLUSIONS

The results of this study indicated that the precues reduce RT by involving of response selection and motor preparation stages of sensorimotor processing.

Furthermore, the results related to the effect of degree of overlap of neuronal representations of muscles on RT were consistent with Anson et al. (2004) and supported the cortical cell assemblies theory (Wickens et al., 1994).

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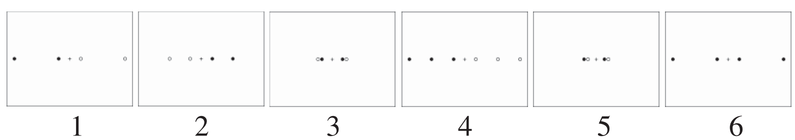

Figure 1: Different precue and control conditions.

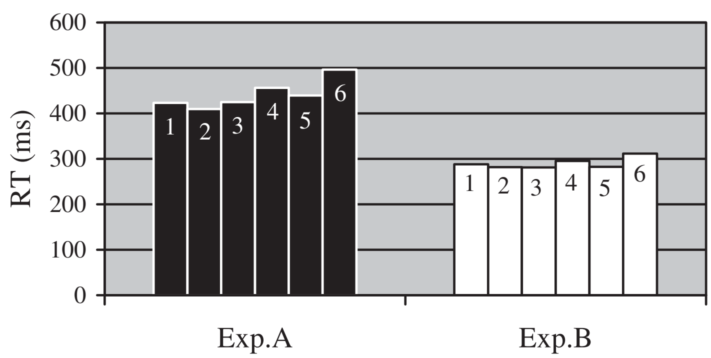

Figure 1: Mean and standard deviation of reaction time of different precue conditions.

MYOELECTRIC ACTIVITY OF CONTRALATERAL BICEPS BRACHII MUSCLE DURING SUBMAXIMAL ISOMETRIC CONTRACTIONS OF ITS HOMOLOGOUS

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INTRODUCTION

Some studies have revealed that during high levels of muscular contractions, it is possible to detect an associated myoelectrical (EMG) activity of some other muscles from the same limb as well as contra lateral to that one. Despite being a normal phenomenon during the first years of childhood, which gradually decreases in normal conditions of development of the Central Nervous System (CNS), some physical therapists have proposed the use of this mechanism in the rehabilitation program of neurological patients (Kautz and Patten, 2004). Its is suggested, for instance, that the contra lateral activity, which can be elicited by means of some exercises, may help to improve the motor control over the hemiplegic side.

Researches have been conducted to better understand how and at which level of the CNS control this “leakage” of descending commands reaches the alpha motoneurons of other muscles not explicitly related to the elicited movement. However, most of them have been attempted while subjects perform non-fatigued maximal voluntary contractions - MVC - (Matta et al., 2007) and few have discussed how fatigue is also related to this phenomenon (Todd et al., 2003).

Therefore, the aim of this study was to evaluate the pattern of motor unit recruitment of both *biceps brachii* muscles by means of surface EMG (sEMG), one as the dominant one submitted to fatigue and its homologous, while kept under low level of contraction but at the same position.

METHODS

Fifteen male subjects (22.8 ± 3.6 years), all right-handed, performed isometric contractions at 80% of maximum voluntary contraction (MVC). The acquisition system was constituted by an electromyography (3000P - Mega Electronics, Ltd, Finland) and a computer (Pentium II). The sampling frequency and gain were set at 1 kHz and 1000, respectively. Surface electrodes (Ag/AgCl - 3M Korea Ltd., Seoul, Korea; 1 cm of diameter) were placed over dominant *biceps brachii* muscle following SENIAM protocol (Hermens et al., 1999).

The subjects performed isometric contraction of right elbow flexors (D_L) at 90° of flexion until exhaustion, while maintaining the contra lateral limb (ND_L) at the same position (Figure 1). They were seated with arms supported by the back support of the chair.

Figure 1: Position adopted by the subjects while performed the fatigue test.

From the sEMG signal the median frequency (MF), the mean power frequency (MPF) and the RMS values were computed for consecutive epochs of 1s length, assembling three time series. From these, the coefficient of linear regression was calculated taking the time as the independent variable. Comparisons between the results of right and left data were performed through Mann-Whitney, and for this purpose the percentage of the rate was taken instead of the absolute value.

RESULTS AND DISCUSSION

There was a clear trend of increasing in RMS values and a decrease in the spectral variables on both sides. The mean (SD) rate of the MF was –35.9 (-17.2) Hz, -40.4 (17.1) Hz for MPF, and 195% (223%) for the RMS. From these tests, 7 from all subjects presented results which would be considered as attending to the criteria for a fatigued muscle i.e., a mean value of –14.8 (10.1) Hz for MF, -22.2 (17.13) Hz for MPF and 180% (203%) for RMS (Figure 2).

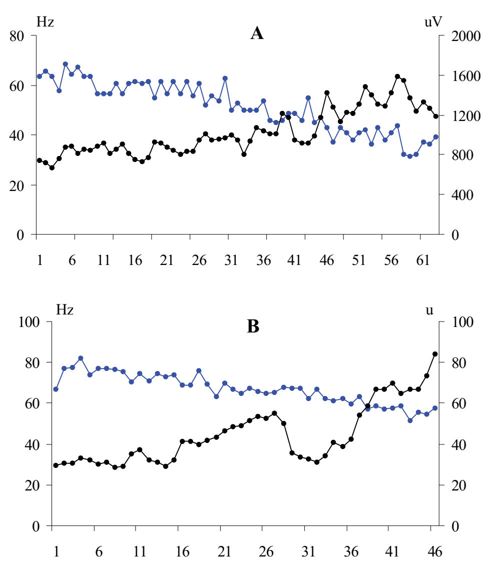

Figure 2: Results of MPF (blue) and RMS (black) values computed from the sEMG of the ipsilateral (A) and contralateral (B) muscles during the test of one subject.

Further statistical analysis revealed the rate values obtained at the right hand side to be different from those at left only for the MF ($p= 0.046$; $p=0.057$ for FPM, and $p=0.830$ for RMS).

Zijdewind et al. (2006) mapped the origin of the action potentials that led to the ipsilateral muscle contraction using Trancranial Magnetic Stimulation (TMS). Their results suggest that both the interhemispheric connection via Corpus Callosum and the ipsilateral motor pathway are responsable for this motor potential “leakage” to the ipsilateral musculature. Some studies suggest an association between ipsilateral muscle activity and mirror neurons, and although that would be a relatively easy experimental design, such study has not been conducted yet. In our study the results suggest that even a minimum level of contraction is enhanced by the MVC in the contra lateral muscle.

SUMMARY/CONCLUSIONS

Although this may appear to have fascinating repercutions on rehabilitation and physical training, more studies such as the effect of MVC on the right arm while the left one performs active movements are still lacking.

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Track 07

Posture and Balance (PB)

EXERCISE-INDUCED REMODELING OF MUSCLE ACTIVATION PATTERNS FOLLOWING STROKE

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INTRODUCTION

A recent review of the recovery of standing balance and gait after stroke provided evidence in favour of a task-oriented exercise approach, albeit the effects were largely restricted to tasks that were directly trained in the exercise program (Van Peppen et al. 2004). Strength exercises after stroke, however, did not lead to better improvement of motor function and gait as compared to conventional therapy (Moreland et al. 2003). These findings may be explained by the fact that neither gait nor balance require muscle strength (large force contractions) but rather require rapid coordinated muscle activation. Consistent with this thinking, Marigold et al. (2005) found that exercise focusing on agility (fast-paced dynamic movements) resulted in faster step reaction times and earlier muscle onset latencies to force platform perturbations than stretching and weight shifting exercises. The purpose of this study is to examine whether a single session of exercise retraining with individuals following stroke will result in a change of muscle activation patterns in response to a standing balance perturbation.

METHODS

Fifteen subjects one month post-stroke and 10 age- and sex-matched controls were tested. Sensation, motor recovery, and functional balance, along with location and side of the stroke, were recorded to determine whether these parameters are associated with the ability of subjects to retrain the muscle activation patterns.

Physiological balance assessment was performed before, immediately after and 15 minutes after completion of the exercise retraining. Each subject was fitted into a safety harness and stood with each foot on a separate force platform. Activity of bilateral hamstrings, quadriceps femoris, and soleus and tibialis anterior muscles was recorded using surface EMG electrodes. Postural responses were studied during a unilateral rapid forward arm flexion of the non-paretic arm (or dominant arm for control subjects), measured with an accelerometer taped to the hand. Subjects performed 10 trials with rest periods of 3-5 s between trials.

An exercise retraining protocol was designed to influence the pattern and the speed of muscle activation. Two exercises were performed that required fast, but not strong, contractions. Subjects were given rest breaks as required. In one exercise, subjects stood with each foot on a force platform and performed 50 squats to 30 degrees of hip and knee flexion, with each squat performed as fast as possible. A second exercise involved 50 small steps with each leg, in which the subject stepped forward quickly on the force platform and stopped abruptly. The speed of the exercises was measured with accelerometers taped to the patella of each leg. Surface EMG, ground reaction forces and acceleration were monitored. Subjects were given feedback in the rest breaks about their performance.

RESULTS AND DISCUSSION

During a unilateral arm flexion perturbation, healthy individuals demonstrate anticipatory postural adjustments with early activation of the ipsilateral hamstrings muscle followed by the contralateral hamstrings, whereas subjects following stroke rarely exhibit these sharp bursts of coordinated muscle activity (Garland et al. 2007).

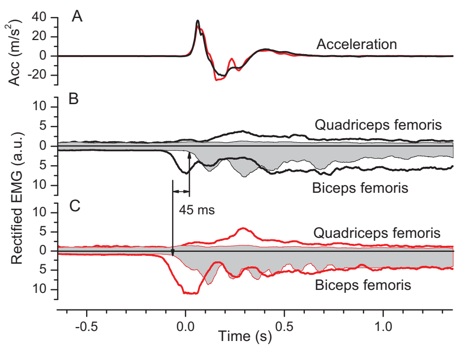

Figure 1: Muscle activation in response to unilateral arm flexion perturbation in a subject post-stroke. All traces are averages of 10 trials aligned to the beginning of acceleration (A). Non-paretic (solid line) and paretic (shaded) muscles are presented before (B; black) and after (C; red) the exercises, with quadriceps pointing up and hamstrings pointing down. Note the 45 ms decrease in latency in the paretic hamstrings.

Healthy age- and sex-matched subjects did not demonstrate a change in EMG or center of pressure (CP) on the balance test after exercise, whereas subjects following stroke exhibited a decrease in the EMG burst latency (revealing earlier activation of the paretic hamstrings) and an increase in the EMG burst slope (Table 1; Figure 1). CP displacement shows a significant increase following the exercise (without a change in arm acceleration – data not shown). One explanation is that subjects demonstrated smaller CP excursions than the controls before exercise and the increase is reflective of a more normal response.

SUMMARY/CONCLUSIONS

Exercise retraining has the capacity to induce change but more research is required to determine if the short-term learning of muscle activation patterns can be retained for a longer period of time.

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ACKNOWLEDGEMENTS

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Table 1. EMG for contralateral (age- and sex-matched healthy subjects) or paretic hamstrings and CP during the arm flexion balance test before and after the exercises

Parameter	Healthy		Stroke		
	Before	**After**	**Before**	**After**	**Retention**
Burst Latency (ms)	12.9 ± 17.4	12.1 ± 11.1	38.1 ± 9.2	19.6 ± 7.9†	22.4 ± 7.0
Burst Slope (a.u./s)	17.4 ± 9.9	17.8 ± 9.1	4.4 ± 2.9	6.4 ± 4.5*	5.9 ± 4.4*
CP AP Displacement (cm)	2.8 ± 1.0	3.0 ± 0.9	2.2 ± 0.5	3.3 ± 1.2*	3.1 ± 0.8*

Data are mean ± SD. * $p< 0.05$, † $p< 0.1$; with respect to before training

CORRELATION BETWEEN HEAD AND SHOULDERS POSTURE ALIGNMENT AND MASTICATORY MUSCLES SURFACE ELECTROMYOGRAPHIC ACTIVITY IN TEMPOROMANDIBULAR DISORDER SUBJECTS.

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INTRODUCTION

Postural alterations and muscular hyperactivity are important generating and perpetuating factors in temporomandibular dysfunctions (TMD) (Braun B.L., 1991; Huggare J.A., Raustia A.N., 1992). Several studies relates forward head posture and round shoulders to TMD (Zonnenberg A.J.J *et al*., 1996; Sonnesen L. *et al*., 2001; Ghessa G. *et al*., 2002), while others establish relationships between electromyographic alterations and this syndrome (Liu Z.J. *et al*, 1999; Pinho J.C. *et al*., 2000).

The aim of this study was to investigate the correlation between head and shoulders posture, and anterior temporalis and masseter muscles electromyographic activity, and its relationship with the temporomandibular disorder (TMD).

METHODS

This study applied questionnaires, electromyography (EMG) and photogrametry as analysis tools. There were selected 32 volunteers shared in TMD and Control groups, according to *Research Diagnostic Criteria* (RDC) (Dworkin SF, LeResche L, 1992).

The EMG signal was registered for three days in the same week with silver/silver chloride bipolar surface electrodes (EMG System Ltda.) joined to preamplifiers of 20 times gain. The electrodes were positioned in the most prominent point during the maximal contraction of the superficial masseter and anterior temporalis muscles along the muscle fibers with an interelectrode distance of 10 mm. The signal acquisition was made by 12 channels of simultaneous recording equipment (Myosystem I / Datahominis Tec. Co.). The analog EMG signal recorded were digitized using 12 bit A/D converter at a sampling rate of 2KHz. The tasks were bilateral mastication and maximal voluntary clenching. The EMG recordings were processed and normalized by MATLAB, and maximum instant (IMAX) and active period (ON) were calculated and analyzed for each muscle during mastication cycle (Nagae M.H., 2005).

Frontal and lateral photographs were taken in orthostatic posture. The following anatomical landmarks were palpated and marked: both acromiones, temporomandibular joints and ear lobes. The frontal and sagital angles were formed to assess head tilt, shoulders lift, round shoulders and forward head posture.

Spearman correlation coefficient was used to examine the linear correlation between posture alignment (angles measures) and masticatory muscles activity (EMG values). The statistical significance level was set at $p<0,05$.

RESULTS AND DISCUSSION

Significative correlations were found between smaller shoulders lift and a higher right anterior temporalis muscle IMAX in TMD group ($p=0,009$) and right masseter muscle IMAX in Control group ($p=0,0306$).

The head and shoulders posture alterations can influence in the postural position of the jaw (Sollow B., Sandham A., 2002) and to provoke an increase in the electric activity of the masticatory muscles and consequently damage some functions as the mastication for instance (Steenks M.H., DeWijer A., 1996). Therefore, a small posture disalignment does not justify the largest activity of the anterior temporalis muscles and masseter muscles, what it was found in this study. Probably it happened because of the presence of the TMD, that might have provoked functional unbalance in these muscles in TMD group, affecting the activity of the temporalis muscle that in the attempt to stabilize the jaw (Bakke M., 1993), it would be overloaded.

Forward head posture and right anterior temporalis IMAX had positive correlations in TMD group (p=0, 05). A common posture alteration is the forward head posture that can cause alterations in the muscle activity pattern between masseter and temporalis muscles (Gadotti I.C. *et al*., 2005). Researchers verified that during head's extension there is an increase of the EMG muscular activity of the anterior temporalis, decreasing the activity of the masseter and digastric muscles (Forsberg C.M. *et al*., 1985; Boyd C.H. *et al*., 1987). So, it is expected that the extension of the upper cervical spine, where there is a forward head posture, the individual is subject to a larger activity of the anterior portion of the temporalis muscle.

SUMMARY/CONCLUSIONS

The results suggest that:

1) Forward head posture can provoke an increase of anterior temporalis electromyographic activity and;

2) TMD can influence in masticatory muscles electromyographic activity.

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ACTIVITY OF COMPONENTS OF THE PARASPINAL AND ABDOMINAL MUSCLE GROUPS DIFFERS WITH SPINAL CURVES IN SITTING

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INTRODUCTION

The relationship between lumbar spinal curves and regional muscle activity is important for study of spinal neuromuscular control and workplace ergonomics, but evidence is scarce.

Kyphotic (*slump*) lumbar sitting postures often require less muscle activity than upright postures (Floyd, 1955; O'Sullivan et al., 2006; O'Sullivan et al., 2002), and surface EMG has shown differential activation of iliocostalis and multifidus muscles when sitting with the *long lordosis* and *short lordosis* spinal curves (O'Sullivan et al., 2006), but surface EMG cannot differentiate activity between superficial and deep fibres of a muscle group, or between neighbouring muscles that are below subcutaneous fat (Solomonow et al., 1994). Hence, this study used a combination of fine-wire and surface EMG to determine how spinal curves in four sitting postures affect activity of the deep and superficial paraspinal and abdominal muscles.

METHODS

14 healthy males with a mean (SD) age of 22 (8) years, height of 178 (8) cm, and weight of 71 (10) kg participated in this study. This study examined muscle activity in four combinations of thoraco-lumbar and lumbar spinal curves shown in Fig 1

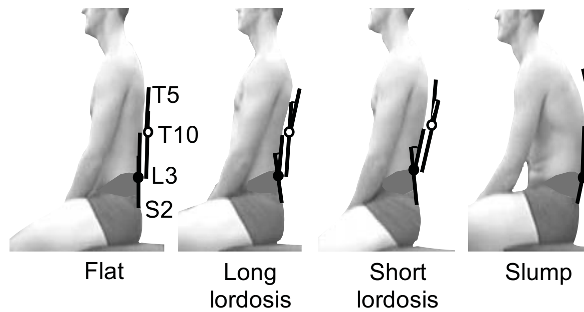

Figure 1: Postures and spinal curve measures. i) *flat* - at both regions ii) *long lordosis* - lordotic at both regions iii) *short lordosis* - thoracic kyphosis and lumbar lordosis, and iv) *slump* - kyphosed at both regions.

To quantify spinal curves in the sagittal plane, 3-D tracking systems were used to record position data (n = 11 with electromagnetic tracking, n = 3 with optical tracking). Sagittal angles representing surface spinal curves at thoraco-lumbar and lumbar regions of the spine were measured between segments connecting T5-T10 and T10-L3 (thoraco-lumbar angle), T10-L3 and L3-S2 (lumbar angle), (Figure 1)(Claus et al., 2007).

EMG activity of deep multifidus, superficial multifidus, iliocostalis lateral to L2, iliocostalis lateral to T11, longissimus thoracis lateral to T11 and transverses abdominis were recorded with bipolar fine-wire electrodes (Teflon-coated stainless steel wire, 75 μm diameter, 1 mm Teflon removed from the cut ends bent back to form hooks at ~ 1 mm and 2.5 mm). Fine-wire electrodes were inserted via hypodermic needle, with ultrasound guidance for needle placement. Superficial abdominal muscle activity was recorded with either fine-wire or surface EMG electrodes (obliquus externus, obliquus internus and rectus abdominis).

Data were high-pass filtered at 50 Hz, 5 s RMS amplitudes were calculated and normalised to peak activity across the sitting postures. Although verbal feedback and manual facilitation were provided, some trials had to be excluded when subjects failed to maintain appropriate spinal curves. A linear mixed model was used for statistical comparison of muscle activity between postures for each of the nine muscles.

RESULTS AND DISCUSSION

The results showed that alteration of the direction of thoraco-lumbar and lumbar spinal curves influenced regional activity of the extensor and abdominal muscles in sitting. Deep and superficial fibres of multifidus as well as obliquus internus abdominis (Figure 2) most consistently altered muscle activity with changes in spinal curves. Activity of deep and superficial multifidus differed between all postures ($p < 0.05$) except between the *flat* and *slump* postures ($p > 0.59$). Activity of obliquus internus abdominis differed between all postures ($p < 0.05$) except between the *flat* and *long lordosis* postures ($p = 0.33$).

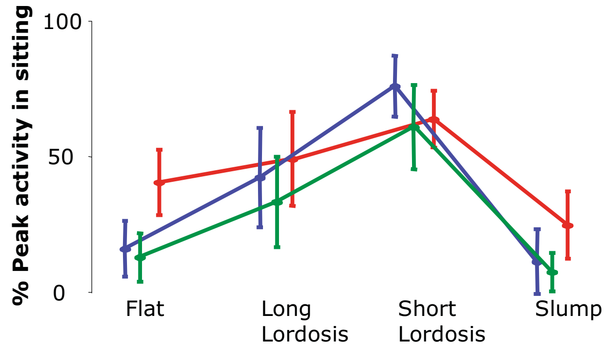

Figure 2: EMG results, error bars – 95 % CI, deep fibres of multifidus – blue, superficial fibres of multifidus – green, obliquus internus abdominis – red.

CONCLUSIONS

Fine adjustment of muscle activity occurs at distinct regions of spinal extensor and abdominal muscles to alter spinal curves in sitting.

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THE EFFECT OF FATIGUE ON POSTURAL CONTROL DURING ONE-LEG STANCE IN FEMALE ELITE HANDBALL PLAYERS

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INTRODUCTION

A high incidence of Anterior Cruciate Ligament injuries are observed in female sports like handball (Myklebust et al. 1998). It has been shown that fatigue may alter the movement strategy (Wojtys EM et al., 1996), and that the majority injuries occur in the late stage of a match (Gabett TJ, 2000; Pinto M et al., 1999). Alternated motor control strategies have been identified as a potential risk factor for an ACL injury (Malinzak RM et al., 2001). Disorders involving any of the movement control systems, such as proprioception, may lead to increased postural sway and potentially the loss of balance. The purpose of this study therefore was to investigate if muscle fatigue induced by a simulated handball match would result in altered leg muscle motor patterns during postural sway in female elite team handball players.

METHODS

29 elite female handball players were tested for postural sway pre and post a simulated handball match. Centre of pressure (CoP) of the ground reaction force was measured during a standardized single leg stance maneuver (30 sec with eyes open) using a force platform. EMG activity (gluteus medius, vastus lateralis and medialis, rectus femoris, biceps femoris, semitendinosus, gastrocnemius lateralis and medialis) was recorded synchronously at 1,000 Hz. All EMG signals were highpass filtered (5 Hz cutoff) and smoothed by a moving RMS filter (30 ms time constant). The simulated handball match consisted of a series of intermittent exercises (side steps, cross over steps, jumps, high and low intensity running and sprinting) mimicking handball match activity (50 min).
Fig.1 illustrates a typical example of CoP excursion, pre and post fatigue.

RESULTS

Postural sway increased as shown by an increased CoP excursion (16% greater confidence ellipse area, $P<0.05$) after the simulated handball match. EMG activity decreased in all muscles examined (19-43%, $P<0.05$) except for the gluteus medius, where no change was observed. The decrease in EMG activity of the examined muscles was most pronounced in the semitendinosus muscle (43%, $P<0.001$).

CONCLUSION

This study indicates that acute fatigue induced by handball related exercises, involving substantial eccentric and rotational forces, impairs postural control in female elite handball players, which may potentially increase the risk for an injury during team match play. Furthermore, contraction of the medial hamstring muscle (semitendinosus) is important to medially compress the knee joint and thereby limit the risk of excessive valgus movement. The observed pronounced decrease in semitendinosus EMG activity may increase the risk of dynamic valgus; hence increase the potential risk of ACL injury.

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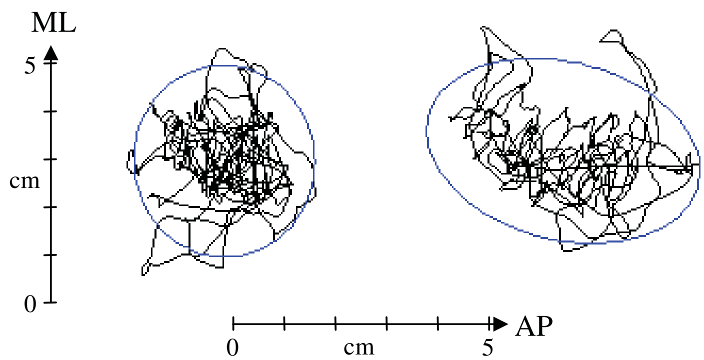

Figure 1. A typical recording of postural sway pre (left) and post (right) fatigue. ML and AP represent the medio-lateral and anterior-posterior excursion, respectively.

EVIDENCE OF PLATEAU POTENCIAL IN MOTONEURONS DRIVING FOOT MUSCLES DURING QUIET STANCE

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INTRODUCTION

Evidences of a self-sustained muscle activation following a brief electrical stimulation have been reported (Collins *et al.,* 2002, Nickolls *et al.,* 2004). The main cause seems to be the genesis of motoneuron plateau potentials in response to a train of postsynaptic potentials. Theses self-sustained muscle activity could be useful during posture. Nozaki *et al.* (2003) showed self-sustained phenomena in the *soleus* muscle in subjects in a supine position. Regarding posture, muscles of the foot have also been shown to have at least an auxiliary role. The purpose of this study is to demonstrate that train stimulation induces a sustained muscle contraction that outlasts the stimulation period at the muscle *Flexor Digitorum Brevis* (FDB) during quiet stance.

METHODS

Eight healthy subjects were requested to stand upright during 40 s. Surface EMG electrodes were placed on the following muscles: FDB, *Soleus* and *Tibialis Anterior* of the right leg. After 20 s of background muscle activity (BGA) acquisition, a 50 Hz train of stimuli (2 s train duration, with 1 ms pulses) was applied to the tibial nerve at the popliteal fossa. Each subject normally performed three trials each, in two visual condition (CE: closed eyes and OE: open eyes). In order to avoid fatigue, each condition was performed after a 2 min resting interval. The BGA and the self-sustained muscle contraction (after the electrical stimuli) were quantified by the root mean square value (RMS). The ANOVA two-way repeated measure was used to compare both situations –visual and muscles (P<0.05).

RESULTS AND DISCUSSION

A sustained electrical muscle activity in the FDB after the train stimulation was turned off was found 87.5% of the sample (Figure 1).

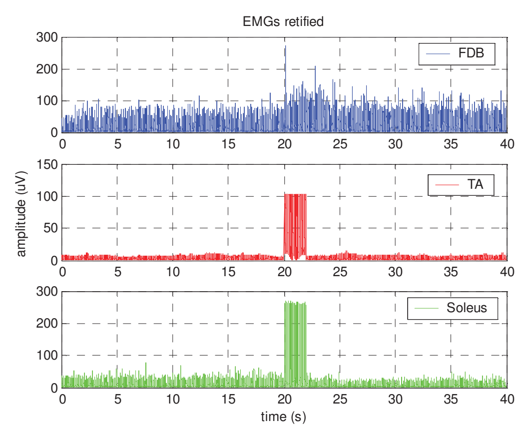

Figure 1: This graph illustrates a typical rectified EMG. Blue line: FDB, Red line: Tibialis Anterior, Green Line: Soleus. The peaks at 20 sec correspond to train of stimuli (2s train duration).

In the population, there was a 13% average increase in RMS amplitude of FDB EMG with respect to that of BGA. At the same time, EMG activity of the *Soleus* was 9% decreased and *Tibialis Anterior* did not change post-stimuli, with respect to their BGAs.

Previous work hat found PP evidence in humans was made in supine position and considers the Soleus muscle (Nozaki *et al.,* 2003; Kiehn & Eken, 1997, Collins *et al.*, 2001; 2002). Our results showed a similar activity at a foot muscle that is innervated by a ramification of tibial nerve. This remaining activity of FDB after train stimulation in upright posture can be a cue that this phenomenon is not Soleus muscle exclusivity.

Self-sustained firing of motoneurons as this may reduce the need for prolonged synaptic input for postural tone (Gorassini *et al.*, 1998). Considering this functional importance of PP in humans (Nozaki *et al.* 2003, Collins *et al.*, 2001; 2002) is clear that this phenomenon must be associated with the *Soleus* muscle at quiet stance. Our data suggest it may happen in the FDB, which is an auxiliary muscle to posture control.

SUMMARY/CONCLUSIONS

In summary, we have presented evidence that to some extent, there is a prolonged activity of FDB after a train of stimuli. This raises the possibility that the motoneurons of this muscle, together with those of the *Soleus* exhibit plateau potentials, which are useful for postural maintenance.

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A STUDY ON THE STRIDE RATE VARIABILITY USING TREADMILL ON DEMAND

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INTRODUCTION

Since the stride-to-stride variability is the small amount of fluctuation during the gait cycle which explains the response and control of neuromusculoskeletal system, there is a possibility that walking with fixed speed treadmill condition might influence the gait variability.

Thus to study the effect of the fixed speed of the treadmill on the variability, this study tried to investigate the difference in the stride rate variability between two treadmill conditions; traditional treadmill and treadmill on demand whose speed can be adjusted automatically by subject's walking speed. The hypotheses of this study are as follows; first, the amount of fluctuation during treadmill on demand walking will be greater than that of the traditional treadmill walking. Second, there will be a parameter or parameters to have a difference in the variability.

METHODS

Eight university male students (25.1 years, 172.7 cm, 66.6 kg) who don't have any musculoskeletal diseases were participated in treadmill walking experiment. The treadmill (RX9200S, Tobeone Co.) whose speed can be adjusted automatically by subject's walking speed was used in this study. Preferred walking speed (PWS) of each subject was determined by 10 min. walking on this treadmill. Each subject performed walking experiment with fixed PWS condition and with free PWS condition for 10 minutes. In this study, free condition means that the speed of treadmill is adjusted automatically. Fixed condition means that the speed of treadmill is fixed with predetermined PWS of each subject. 3D motion capture system (Motion analysis Corp., USA) with 6 cameras was used to collect motion data with sampling frequency of 120Hz.

Temporal (stance time, swing time, stride time, step time and double support time) and spatial (stride length and step length) variables were calculated. To remove outliers, the method proposed by Owings et al. (2004) was used. The coefficient of variance (CV) and the slope of detrended fluctuation analysis (DFA) were used to quantify the amount and structure of the variability, respectively (Hausdorff 2005). MATLAB^{TM} 7.0 (Mathworks Inc.) was used for the computation of all variables and SPSS^{TM} 12.0k (SPSS Inc.) was used for the statistical analysis.

RESULTS AND DISCUSSION

The mean, CV and DFA of selected variables and statistics of each variable are given in Table 1 and 2. Average value of each variable is almost same for two different conditions. However, Table showed that the amount (CV) of variability during free PWS condition was greater than that of fixed PWS condition (for step length, $p<.01$). For temporal variables, the result of our free condition walking agreed with published data that CV of general ground walking was around 3% (Hausdorff et al. 1996). However, for spatial variable, the result of this study showed relatively larger values than the published data (Jordan et al. 2007). The structure of variability during free PWS condition was statistically different from that of fixed PWS condition (for step length, and double support time p<.01, for stance time, p<.05). This means that the range of fluctuation of each variable was reduced as a result of the fixed speed of treadmill.

SUMMARY/CONCLUSIONS

To study the effect of the fixed speed of the treadmill on the stride-to-stride variability, we used both traditional treadmill and treadmill on demand whose speed can be adjusted automatically by subject's walking speed. The amount (CV) and structure (DFA) of variability during free PWS condition was generally different from that of fixed PWS condition. From these results, it is possible that traditional treadmill study might give incorrect conclusion about gait variability study. Further study is necessary to clarify these matters by considering the number of subjects, experimental time, and gait variables for the study of stride-to-stride variability.

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Table 1: Mean, CV and DFA of the variables used in this study (*: p<.05, **: p<.01)

	Mean		CV		DFA	
	Free	Fixed	Free	Fixed	Free	Fixed
Stance time	0.76 sec	0.75 sec	2.51	1.90	0.88	0.72*
Swing time	0.41 sec	0.41 sec	2.10	1.94	0.78	0.70
Stride time	1.17 sec	1.16 sec	1.92	1.51	0.90	0.82
Step time	0.58 sec	0.58 sec	2.60	2.21	0.84	0.74
Double support time	0.17 sec	0.17 sec	6.73	5.45	0.79	0.57**
Stride length	1.17 m	1.30 m	5.52	4.75	1.04	1.23
Step length	0.56 m	0.56 m	4.47	2.11**	0.99	0.74**

Table 2: Statistics of the mean, CV and DFA of the variables (*: p<.05, **: p<.01)

	Mean		CV		DFA	
	t	p-value	t	p-value	t	p-value
Stance time	0.093	0.927	1.914	0.076	2.863	0.013*
Swing time	0.672	0.513	0.490	0.632	1.615	0.129
Stride time	0.252	0.804	1.306	0.213	1.391	0.186
Step time	0.167	0.870	0.991	0.338	1.417	0.178
Double support time	0.000	1.000	2.068	0.058	5.597	0.000**
Stride length	-1.417	0.178	0.840	0.415	-2.022	0.063
Step length	0.059	0.954	3.575	0.003**	3.488	0.004**

BIOMECHANICAL ANALYSIS OF DEFENSE TECHNIQUES IN TAI CHI PUSH HANDS

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INTRODUCTION

Tai Chi (Tai Chi Chuan) is a kind of physical exercise usually described by the following aspects: slow, focus, breathing, and relaxing (Dennis & Chu, 2004). Although Tai Chi is developed from traditional Chinese martial arts, it has become a popular exercise all over the world.

Tai Chi exercise has been shown to be beneficial for the elderly in preventing falls. Xu et al. (2004) indicated that older people practicing Tai Chi had better proprioception than the swimming and jogging groups. Li et al. (2005) investigated 256 older people with six months of Tai Chi exercise and found their falling probability declined about 55%.

Biomechanical analysis revealed that center of gravity is always low and joints are well coordinated during Tai Chi push movements (Chan et al., 2003). Compared to normal gait, Tai chi gait has more double-support duration and the frequency of changing motion direction is faster (Mao et al., 2006).

Tai Chi exercise actually consists of practicing forms and Push Hands. To date, researches focused extensively on the analysis and the effects of practicing forms. Studies on Tai Chi Push Hands are scarce and with only two-dimensional (2D) analysis. Thus the present study employs 3D analysis with two force plates to attain load on each foot and characterize in more detail the Push Hands techniques.

METHODS

A Tai Chi master (age 69; height 1.60 m; weight 70 kg) participated in the study after given written informed consent. He has been practicing Tai Chi form (Cheng Tzu's style) and Push Hands for 40 and 30 years, respectively. He was asked to defend pushing by another person who exerted maximum effort and had no previous experience in Tai Chi for three trials.

Eight Eagle video cameras (Motion Analysis Corporation) at 200 Hz, two Kistler Type9281B force plates at 1000 Hz, and a MA-300 EMG System (Motion Lab Systems, Inc.) at 1000 Hz were synchronized during data recording. Helen Hays Marker Set (with 29 markers) was used to indicate anatomical landmarks. Two force plates obtained kinetic data on each foot of the master when resisting the push movement. The surface electrodes were placed on the left and right sides of the large muscle groups of whole body including the deltoid, triceps, latissimus dorsi, erector spinae, rectus femoris, semitendinosus, and the medial head of gastrocnemius. All EMG data were filtered by the Butterworth fourth order bandpass filter of 10~400 Hz and bandstop filter of 60 Hz.

RESULTS AND DISCUSSION

EMG measurements of different muscle groups are illustrated as time functions (Fig. 1). Among the muscle groups examined, the lowest EMG activities occurred in the medial hamstrings and the right (R) medial head of gastrocnemius, with a higher value in the erector spinae muscle, deltoid and the highest by far in the left (L) rectus femoris muscles.

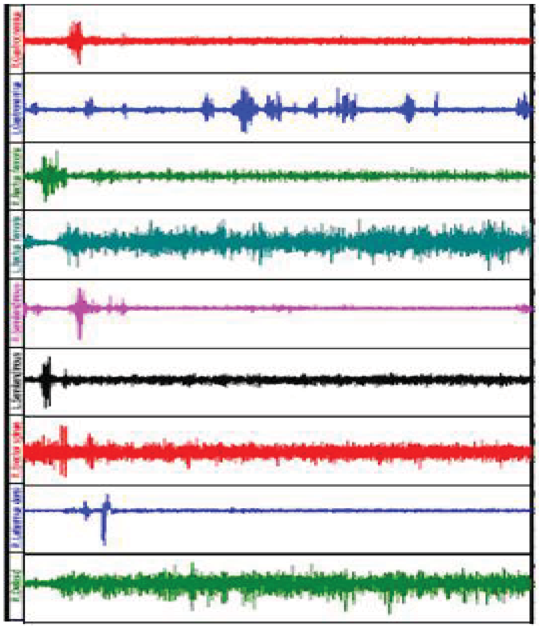

Figure 1: Variations in muscle activities of upper and lower limbs during the defense. (From top to down: R & L gastrocnemius, R & L rectus femoris, R & L semitendinosus, R erector spinae, R latissimus dorsi, and R deltoid.)

Changes in vertical ground reaction forces (GRF) are shown using two force plates (Figure 2). When the master began to defend, higher value shifted from the right (forward) to left (backward) foot. The highest left GRF value is 677.74 N, opposed to the concurrent right foot value of 32.80 N.

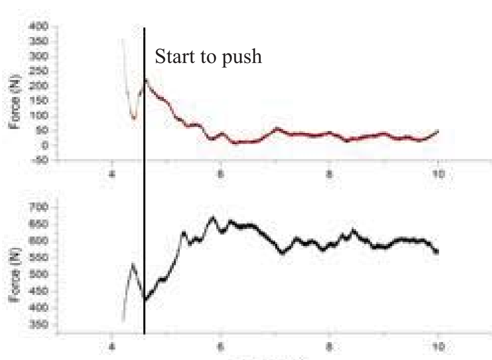

Figure 2: Ensemble of vertical GRF during defense. The right foot is on the first force plate (top) and the left foot is on the second force plate (bottom).

SUMMARY/CONCLUSIONS

The kinematics, kinetics, and EMG characteristics of a fundamental defense of Tai Chi Push Hands were examined. Both concentric and eccentric muscle contractions occurred in lower limb muscles, and the eccentric contraction occurred mainly in the rectus femoris and the medial head of gastrocnemius, especially of the backward foot. When the attacker started to push the master, peak GRF occurred immediately in the front foot and then quickly shifted to back foot. This was accompanied by higher EMG value in the right erector spinae and the left rectus femoris muscles.

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CIRCUMVENTION OF A SUDDENLY APPEARING OBSTACLE

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INTRODUCTION

The need to circumvent an obstacle, which is noticed only shortly before collision, could be a cause of falls and injury, especially in older adults, either because obstacle avoidance is unsuccessful, or because the sudden sideward movement perturbs balance. Previous studies on obstacle circumvention (Vallis & McFadyen, 2003; Gérin-Lajoie et al., 2006; Lowrey et al., 2006) do not provide insight in avoidance of suddenly appearing obstacles, as the obstacle was visible well in advance of a potential collision. Sudden externally applied lateral perturbations of stance have shown different balance recovery strategies, most importantly sidestepping and crossover stepping (Maki et al., 1996; Maki et al., 2000; Rogers & Mille, 2003; Mille et al., 2005), In the present study, we tested the hypothesis that similar strategies would be used when circumventing a suddenly appearing obstacle during gait. Moreover, we aimed to characterize these responses in terms of muscular activity and their kinetics and to determine the effect of available response time (ART) on avoidance strategies.

METHODS

Twenty-one young adults walked down a 12-m platform at 1.2 m/s, while in 16 (out of 96) trials an obstacle (horizontal poles at ankle and shoulder height) appeared half-way, blocking their passage (Fig. 1). Ground reaction forces (Fgr) and kinematics of both legs as well as EMG activity of the bilateral rectus femoris (RF), biceps femoris (BF), gluteus medius (GM), adductor magnus (AM) were recorded. Obstacle appearance was timed based on-line kinematic data, to provide 850 or 700 ms ART, such that right heel strike (rHS; Fig. 1A and 2A) occurred just in front of the obstacle on a force plate. Peak horizontal ground reaction forces and horizontal impulses were determined. In addition, mean EMG activity over 300 ms following right heel strike and left heel strike (lHS; Fig 1B and 2C) minus the corresponding value for normal walking were determined.

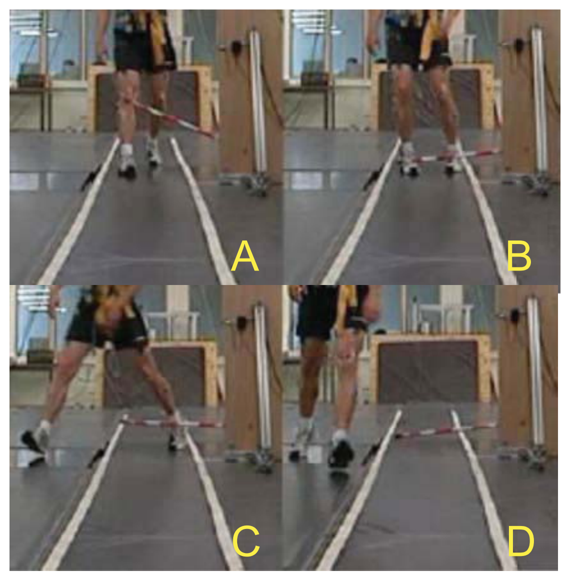

Figure 1. Subject performing a sidestep to circumvent the obstacle.

Logistic GEE regression analysis was used to test for effects of ART on strategy choice. Linear GEE regression was used to test for effects of strategy and ART on EMG and kinetic variables.

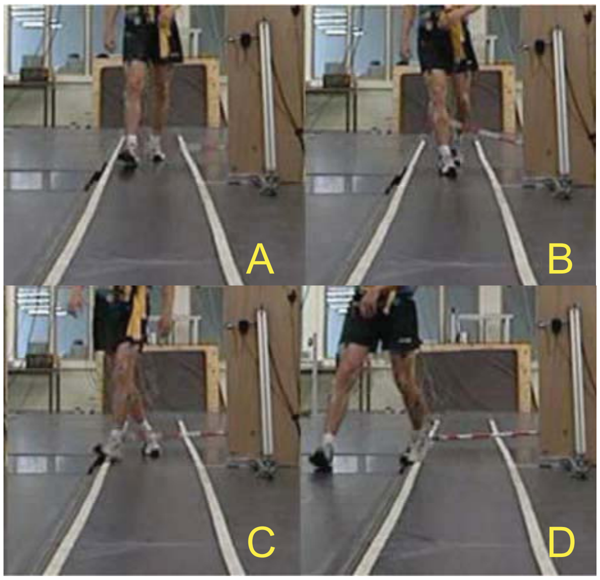

Figure 2: Subject performing a crossover step, to avoid the obstacle.

RESULTS

Avoidance strategies could be classified as either sidestepping (Fig. 1) or crossover stepping (Fig. 2). rHS which occurred shortly after obstacle appearance (~229 or 115 ms, depending on ART; Fig. 1B), was followed by a pronounced backward Fgr, braking the ongoing gate. Subsequent push-off was in most cases minimal or absent. lHS occurred ~334 ms and ~576 ms after rHS in sidesteps and crossover steps respectively. In some cases, also after lHS a pronounced backward Fgr was measured. Sidewards impulse was mainly created after lHS.

Strategy choice was significantly affected by ART, with 61 and 18% of trials being a crossover step with long and short ART, respectively ($p < 0.001$).

The backward impulse and peak Fgr as well as the sideward impulse and peak Fgr were higher in sidestepping than in crossover stepping (all $p < 0.01$). The independent effect of ART was significant only for the horizontal impulse ($p < 0.02$).

EMG activity after rHS and lHS was higher than in normal gait (all $p < 0.05$), in particular in RF and GM. Right and left RF activity and left GM activity were more pronounced in sidestepping than in crossover stepping, while right BF showed the opposite effect (all $p < 0.01$).

DISCUSSION

As hypothesized, sidestepping and crossover stepping strategies could be discerned in sudden obstacle circumvention, the former being more common as ART was shorter. A burst of RF activity around right heel strike in front of the obstacle appeared to be the cause of the strong backward Fgr. GM was active at the same time, presumably to bring the upper body over the right stance leg. After lHS, the left GM provided the sideward push-off. Sidestepping appears more demanding than crossover stepping, while it provides larger obstacle clearance and probably is more stable.

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TRADE-OFF AND COACTIVATION BETWEEN GASTROCNEMII DURING A QUIET STANDING TEST: PRELIMINARY RESULTS

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INTRODUCTION

Recently, body sways during quiet standing has been conceived as the result of corrective shortenings of calf muscles (Loram et al., 2005).

Instead of being stretched as springs when the center of gravity (COG) sways forward, calf muscles are actively shortened, suggesting a predictive mechanism to stabilize COG position (Loram et al., 2005). The assumption of a predictive controller is enhanced by evidences of anticipated increase of calf muscles surface electromyographic (sEMG) activity with respect to center of pressure (COP) sway (Mello et al., 2007) or COG velocity (Loram et al., 2005). On the other hand, the observed bipolar sEMG signal accounts for muscle activity within a limited volume under the skin, unrepresentative of the large and heterogeneous triceps surae muscles.

For the first time, we use a high density multi-channel sEMG system to assess the relationship between gastrocnemii activity and body sway during quiet standing. We show that both medial (MG) and lateral (LG) gastrocnemii are not constrained to be stereotyped mechanical impulse generators.

METHODS

Stabilometric and sEMG data were synchronously recorded from a single subject (24 years, 85kg and 1.9m), standing 40s in upright position, with arms along the body, feet parallel with a comfortable width and eyes open.

Monopolar sEMG signals were acquired at 2048 samples/s from a matrix of 120 electrodes (8x15, eyelet electrodes with 2mm diameter and 10mm interelectrode distance) using a “driven right leg” circuit to attenuate 50Hz interference, a gain of 5k and a 15-350Hz filter. COP data was measured using a piezoelectric forceplate (9286AA Kistler, Milan, Italy). Both COP and rectified sEMG data were low pass filtered using a bidirectional 4th order butterworth filter with 5Hz cutoff frequency and down sampled to 64Hz.

COG time series was estimated from COP data and decomposed into individual unidirectional forward sways. The maximum value of the cross-covariance function between each sEMG channel envelope and COP data was evaluated for each forward sway.

To overcome problems with matrix positioning, the boundaries of MG and LG and their junction, were detected using an ultrasound scanner (Fukuda Denshi, UF 4000, 7.5 MHz linear probe) and marked over the skin. Thus, the matrix was placed to cover the largest area of both muscles.

RESULTS AND DISCUSSION

22 unidirectional forward sways of COG were identified for the 40s standing test,

with size and duration ranging from 0.06mm and 0.17s to 13.7mm and 2.85s respectively. MG and LG did not behave as a single unit, providing an impulsive torque burst to correct forward sways (Figure 1a). While MG sEMG increased with COP displacement, which reflects ankle torque, LG activity decreased.

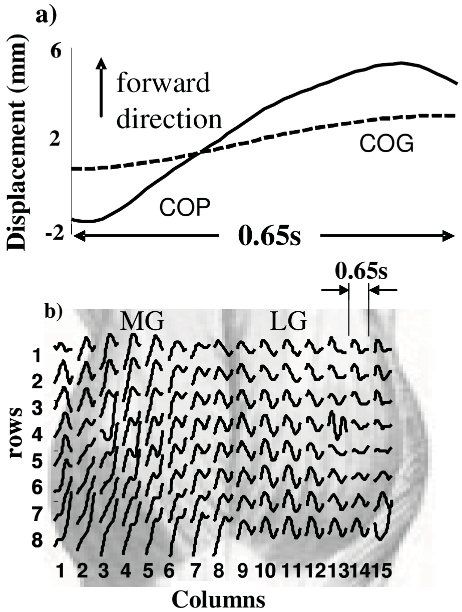

Figure 1: A single forward sway (a) and the corresponding sEMG envelope of each channel (b). Note the rate of change in sEMG envelope for MG and LG

The sEMG pattern depicted in Figure 1b was the same for both gastrocnemius across all forward sway events. MG and LG sEMG exhibited correlations with COP of either the same or opposite sign (Figure 2), corresponding to co-activation and trade-off global strategies, respectively (McLean and Goudy, 2004).

Such patterns of activation reveal a flexible mechanism for opposing the forward pull of gravity. It would be implausible to respond to forward sways, which span a wide range along the antero-posterior axis (0.06 to 13.7mm), with a stereotyped corrective torque.

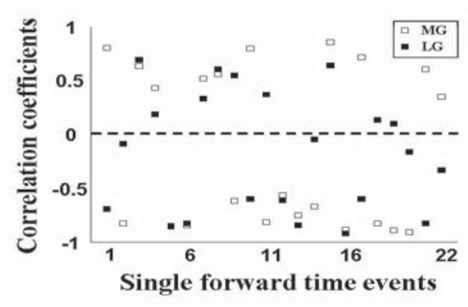

Figure 2: Mean correlation coefficients for each forward sway between COP and sEMG envelopes of 18 out of 120 channels from MG (□) and LG (■), chosen according to SENIAM recommendations.

CONCLUSIONS

MG and LG bursts of torque to correct forward sways are not always in phase. Opposite fluctuations of sEMG activity in the two muscles suggest a trade-off strategy to minimize fatigue during quiet standing. Although shown for a single subject, these results were observed for a sample of 8 subjects.

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ASSESSMENT OF GASTROCNEMIUS HETEROGENEITY USING A HIGH DENSITY SEMG SYSTEM

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INTRODUCTION

So far, the role of calf muscles to oppose the gravity toppling torque during quiet standing is not established, although innovative attempts to address this question emerged recently (Loram et al., 2005).

Using ultrasound images, Loram et al. (2005) observed that gastrocnemius medial and soleus behave simultaneously as a single actuator, shortening during every forward sway to compensate for the Achilles tendon compliance. On the other hand, invasive and non invasive assessment of calf muscles EMG revealed an inhomogeneous pattern of activation during isometric contractions, depending on effort duration, knee joint angle and force level (Kenedy and Cresswell, 2001; McLean and Goudy, 2004). Changes of loading sharing among triceps surae muscles and coactivation are two global neuromuscular strategies for minimizing fatigue and sustaining force level target, respectively (McLean and Goudy, 2004).

This study aims to assess heterogeneity of EMG amplitude distribution between gastrocnemii muscles during isometric contractions at two different knee joint angles and with torque levels similar to those required for quiet standing.

METHODS

Six healthy male subjects (19–36 years), participated to the experiments after giving informed consent. Each subject sustained plantar flexion with ankle in neutral position for 10s at torque levels corresponding to the averaged minimum (low) and maximum (high) ankle torque values measured from three quiet standing trials, using a forceplate (Kistler, Milan, Italy) and providing that ankle axis of rotation was parallel to forceplate lateral axis. Subjects exerted isometric contractions a) in prone position (knee extended KE) and b) on hands and knees, with hip, knee and shoulder joints 90° flexed (knee flexed KF). Plantar flexion MVC torque scores for KE and KF position were evaluated as well.

Both gastrocnemii sEMG were synchronously recorded with torque signals using a matrix of 120 electrodes (8x15, eyelet electrodes with 2 mm diameter and 10 mm interelectrode distance). The matrix covered the largest portion of both muscles, as observed with ultrasound scanning (Fukuda Denshi, UF 4000, 7.5 MHz linear probe).

Monopolar sEMG signals were band pass filtered (15-350 Hz) and RMS amplitude was estimated for each signal with non overlapping epochs of 250 ms. The barycenter (centroid) of RMS spatial distribution and the global RMS value were calculated for each epoch and subject.

RESULTS AND DISCUSSION

Minimum and maximum standing torque levels ranged from 10 to 40% of MVC in KE position and up to 54% of MVC in KF position. A significant shift of sEMG amplitude toward lateral gastrocnemius was observed from KE to KF position for all subjects and from low to high torque level in most cases (Table 1). A trend of activity was observed along the longitudinal axis toward either proximal or distal muscle portion.

In two subjects, the sEMG amplitude increased from KE to KF position to maintain the same low or high torque levels (Table 1). An opposite behavior or interaction between knee angle and torque level factors (2x2 ANOVA, $p<0.05$) was observed for RMS amplitude in the remaining subjects. The shift of RMS distribution between gastrocnemii muscles and the increased sEMG amplitude are clearly observed in Figure 1.

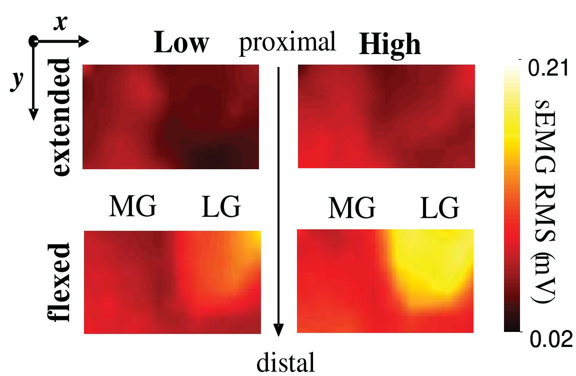

Figure 1: RMS spatial distribution for subject BoAn, according to knee joint position extended and flexed, and isometric torque level low and high.

According to Kennedy and Cresswell (2002), larger MUs of medial gastrocnemius are recruited when the muscle is shortened and some force threshold is exceeded (≈31% of MVC). The results observed in this study suggest that calf muscles are operating close to individual force threshold when maintaining quiet standing torque levels.

CONCLUSIONS

Even at torque levels similar to that produced in quiet standing, gastrocnemius activity is heterogeneous. Calf muscles may not behave as a single actuator to oppose gravity during quiet standing.

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Table 1: Barycenter coordinates along longitudinal (BarY) and lateral axis (BarX), its RMS (BarRMS) mean(SD) values and absolute and relative (%MVC) isometric torque levels. N=40 epochs.

		BarX (mm)		BarY (mm)		BarRMS (µV)		Torque	%MVC	
Subject		**KE**	**KF**	**KE**	**KF**	**KE**	**KF**	**Nm**	**KE**	**KF**
BoAn	**Low**	63(1.6)†*	75(1.5)	35(0.6)	34(0.5)	64,4(6,6)†	81,6(10,0)	45	32	41
	High	68(1.8)‡	70(2.9)	35(0.6)	35(0.7)	65,3(6,4)‡	87,9(13,2)	60	43	54
DaEd	**Low**	57(1.1)†*	73(1.9)	37(0.5)	37(0.6)	37,5(3,4)*	38,0(4,9)	23	18	27
	High	62(1.8)‡	75(0.9)	36(0.5)	36(0.3)	67,1(6,9)‡	72,6(6,5)	40	32	47

* torque level additive effect ($p<0.05$). † additive effect of knee angle ($p<0.05$). ‡ interaction between torque level and knee angle ($p<0.05$).

ASSESSMENT OF POSTURAL SWAY DURING MULTIPLE LOAD AND VISUAL CONDITIONS

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INTRODUCTION

This study investigated the effect of variations in applied load (weight and size) and visual condition on quiet-stance postural sway. Occupations such as structural firefighting are often performed in conditions with poor lighting and smoke. A self-contained breathing apparatus (SCBA) that includes a face mask and shoulder pack with air bottle is used in these situations. There is controversy as to what type of air bottle should be used based on weight, size, and cost. Previous studies have examined the effect on balance of wearing personal protective equipment (PPE) with SCBA packs for firefighter (Punakallio 2003) and hazardous material workers (Kincl 2002); however, no systematic assessment has been done on multiple load carrying bottle configurations and visual conditions and their effect on postural stability. Therefore, in this study, we investigated how modest changes in applied load weight and size would affect quiet-stance postural sway under eyes open and eyes closed conditions, while wearing PPE used by firefighters.

METHODS

Four air bottles were tested: An aluminum bottle (AL) representing current low-budget, low pressure, heavy and large designs: 9.1kg, 53.3cm (L), 17.2cm (dia). A carbon fiber bottle (CF) representing current expensive, high pressure, light and small designs: 5.4 kg, 47.0 cm (L), 14.0 cm (dia). To assess the effect of weight, a fiberglass bottle (FG) with similar size as CF but weight of AL was constructed: 9.1kg, 49.5cm (L), 14.0cm (dia). To examine center of mass location effect, a novel redesigned bottle (RD) was constructed, by cutting a 60-min CF bottle in half, to provide a light and short design that was lower on the back: 5.4 kg, 31.8cm (L), 19.0cm (dia).

Twenty-one male firefighters (age 28±5 yrs, height 177±8 cm, and mass 89±21 kg) participated in this study. Each wore his own PPE (bunker coat, pants, and boots). Helmet and SCBA pack (Scott Air-Pak Fifty 4.5) were provided.

For each bottle configuration, three 60 s trials of quiet stance were conducted for either of two visual conditions (eyes open and looking ahead at a target, or eyes closed). The subject stood on a force plate (AMTI, BP600900) with arms crossed at the chest. For all trials, subjects stood within foot tracings that were created during the first trial of each bottle condition. (No significant differences in stance width were noted between bottle conditions.) The presentation order of bottles and visual conditions were randomized between subjects. Center of pressure (COP) data were sampled at 1000 Hz.

Average COP measures were determined from data based on three trials per condition. Traditional parameters (Prieto 1996) included Angular Deviation (*AngDev*) from the anterioposterior (AP) axis; and AP, mediolateral (ML), and radial (RAD) components of Maximum Distance (*MaxDist*), Displacement Standard Deviation (*SD*), Range (*Range*), and Mean Frequency (*MeanFreq*). Two-way repeated measures ANOVA tests examined whether bottle configuration and visual condition affected these parameters. The level of significance was set to $\alpha = 0.05$. Statistical analyses were run on SPSS (SPSS Inc., Chicago, IL; v15).

RESULTS AND DISCUSSION

In general, body sway decreased as different bottle configurations were used in the order of AL, FG, CF, and RD (i.e., heavy and large to light and small); however, only ML components were found to be statistically significant differences. There were significant main effects for bottle configuration on $MaxDist_{ML}$ (p=0.025), SD_{ML} (p=0.006), $Range_{ML}$ (p=0.015), and $MeanFreq_{ML}$ (p=0.006). Post hoc tests showed heavy bottles (AL, FG) had significantly larger ML values than light bottles (CF, RD), e.g., Figure 1. This suggests that weight significantly affected sway response.

There were no significant differences between AL and FG or CF and RD, suggesting that size and COM location did not significantly affect postural sway.

Significant main effects for visual condition were found in all parameters except $MeanFreq_{ML}$ and *AngDev*; however there was a significant bottle × vision interaction for $MeanFreq_{ML}$. No other interaction effects were noted. When eyes were open, the values of parameters decreased significantly.

We found that postural sway increased significantly in the ML direction with weight, but not load size, and lack of vision. Schiffman (2006) found that path length in both AP and ML direction increased as carrying weight increased with soldiers. Punakallio (2003) found that sway velocity increased when PPE was used, compared to sportswear, and between eyes open to eyes closed conditions.

SUMMARY/CONCLUSIONS

From the results above, we can conclude that modest increases in posterior load weight significantly increases ML postural sway, but not necessarily AP sway. Changing load location appears to have minimal effect. Visual condition also significantly affected postural sway of individuals wearing added load.

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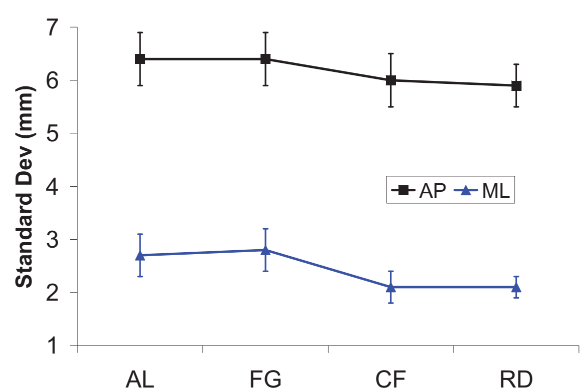

Figure 1. Mean (standard error) for standard deviation (*SD*) in AP and ML directions.

HUMAN ANKLE ANGLE EXCURSIONS LINKED TO DETECTION OF SMALL HORIZONTAL SINUSOIDAL PLATFORM TRANSLATIONS

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INTRODUCTION

Humans sway when standing. Such sway is generally imperceptible. We undertook a series of psychophysical studies to determine the lowest or threshold amplitude at which a subject can detect a sinusoidal postural oscillation imposed upon a normal sway pattern. We search for biomechanical or physiological response(s) that nearly always appear above detection threshold and rarely below it. Such a marker could serve as an important new clinical screening tool for postural stability.

METHODS

Ten healthy young adults participated in this study under an IRB approved protocol. They stood on our Sliding Linear Investigative Platform for Assessing Lower Limb Stability with Synced Tracking, EMG and Pressure measurements [SLIP-FALLS-STEPm, Robinson et al., 1998], which was sinusoidally oscillated for and aft in the horizontal plane. We collected peri-move physiological, psychophysical and biomechanical time series data at 1000 Hz using LabVIEW routines. Center of Pressure measures were collected. Leg surface EMGs were amplified via a Delsys 16 chnl system, foot pressure distribution changes via Tekscan's HRTekMat, and motion capture data via a Vicon-Peak 6-camera system running at 250 Hz. Excepting marker data, other data are converted to engineering units on-the-fly. All time series were decimated to 125 Hz to best compare marker, EMG, and other data.

In a random 20 of the 30 trials of a run, a 0.5 Hz oscillatory burst was applied to the platform for 3.5 cycles (Fig. 1). Ten null trials were interspersed per a modified Single-Interval Adjustment Matrix (mSIAM) Protocol [Kaernbach, 1990]. This protocol uses the correctness of the subject's report of whether (s)he felt the move to determine detection threshold by titrating stimulus amplitudes in sequential trials. Responses were categorized into 'Hits', 'Misses', 'Correct Rejections' and 'False Alarms.' For reference, a HIT occurred when subject correctly signaled that a stimulus did occur. Data from a subject's HIT and MISS trials were separately point-by-point averaged.

Motion capture data provided an independent measure of platform movement, two measures of ankle angle (with respect to knee as well as head), head movement, and other angles and lengths. Sine amplitudes were kept below normal sway path length. Ankle angles were referenced from vertical.

RESULTS

The first observation was that the mSIAM protocol resulted in most subjects iterating to a stable detection threshold within a 30 trial run. The second was that ankle changes were usually phase-locked to the stimulus, often at below-threshold values. Differences were seen in a number of subjects in ankle angle offset and total excursion between HIT and MISS trials. Fig. 1 shows the averaged ankle angle time series for HIT and for MISS trials.

These ankle angles are with respect to the knee, but those measured with respect to the head are similar. The average stimulus was smaller in MISS vs. HIT trials, as would be expected psychophysically. An unexpected result was that angle offsets from vertical differed in these two cases. Fig. 2 (top) compares phase plane plots of ankle angle vs. stimulus amplitude. The HIT angles oscillated around 7.575° with a 0.15° total excursion. For MISS, the excursion was identical (0.15°), but the offset was different (7.425°). Also, the MISS lock-in profile for this 22 yr old male was apparent but more random. Other subjects showed similar differences in offset. A 20 yr old female had a 0.3° cyclic swing around 10.25° for HITS; and 0.2° around 9.9° for MISSES (Fig 2 bottom).

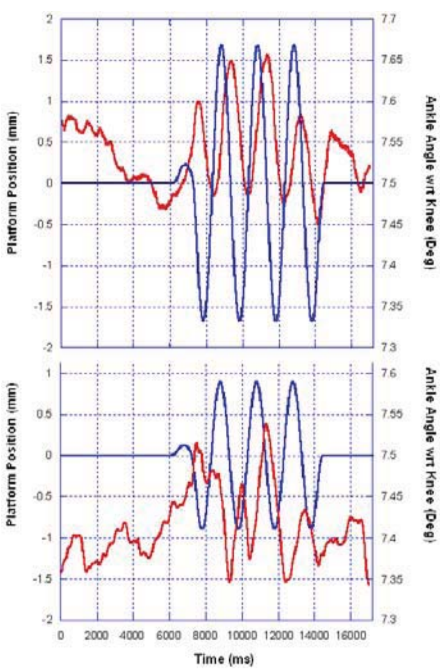

Figure 1: Averaged Platform Position (blue) and Ankle Angle (red) for HIT (upper) and MISS trials (lower) for a 22 y.o. male undergoing 0.5 Hz platform perturbations.

DISCUSSION /FUTURE WORK

It is worth further pursuit in our other subjects to see if indeed the vertical ankle angle offset is a predictor of the ability to correctly detect a near threshold oscillation. In both the HIT and MISS cases, the range of lock-in ankle angle oscillation is similar and small (~0.2°). So it cannot be the oscillation length that is providing the cue. The presence/absence of oscillatory patterns in the TA and GS surface EMGs are being examined to see if they might be predictors.

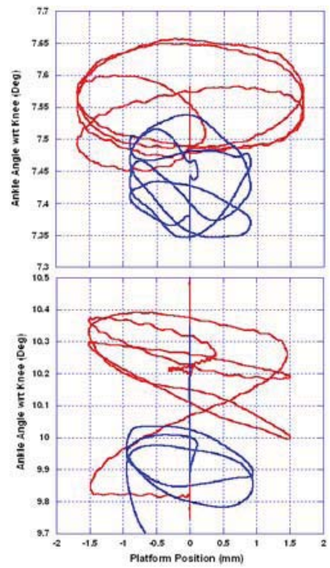

Figure 2: Phase plane plots for averaged HIT (Red) or MISS (Blue) data for a 22 y.o. male (top) and a 20 y.o. female (bottom).

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ACKNOWLEDGEMENTS

Initial data collection was supported by a VA Senior Rehabilitation Research Career Scientist Award to CJR. Later data collection and analysis were supported by NIH R01AG026553 and by a Coulter Foundation endowment to Clarkson University.

TA ACTIVATION OCCURS ONLY WITH CORRECTLY DETECTED, SHORT PLATFORM TRANSLATIONS MADE AT AND ABOVE DETECTION THRESHOLD

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INTRODUCTION

We are interested in determining what physiological inputs are used by individuals to detect short translational displacements on a platform upon which they stand, when the moves lie within a subject's quiet-standing sway length and are near the subject's perceptual detection threshold. This method of studying balance control mechanisms differs markedly from those that use stronger and longer perturbations, and allows us to address control mechanisms of normal sway.

METHODS

We use an ultra-low vibration, translating platform (Robinson et al, 1998) and adaptive psychophysical techniques (Richerson et al, 2006) to sequentially iterate stimulus levels during a run of 30 trials until a stable detection threshold is reached. A 16mm anterior, low jerk translation occurs in 1 of 2 sequential intervals and a subject is required to signal in which interval that he/she perceived that a move occurred (hence, the term *2-Alternative Forced Choice* task or 2AFC). With displacement fixed, the acceleration of the platform is adaptively adjusted from trial to trial, and usually converges to threshold.

We collect peri-move physiological, psychological and biomechanical data using our SLIP-FALLS system (for CoP measures), a Delsys 16 chnl EMG system, an HR TekMat overlay (for foot pressure distribution changes), and a Vicon-Peak 4- or 6-camera marker system (to detect 0.1 ° ankle angle changes). Data collected at 1KHz are converted to engineering units. With means subtracted, raw surface EMG signals from L&R TA and GS are filtered and converted to RMS values using a 21 pt filter. For this paper, the left and right side EMG RMS time series are added together for each muscle group.

Since the 16mm perturbations are made at different accelerations near threshold and moves are made in one of two sequential intervals, it is necessary to pick a reference point to align each trial made within a single run. Thus all data is referenced to the midpoint of each move (the point of maximum jerk) and raster plots constructed. Trials are sorted such that moves made at the highest acceleration are placed at the top and at the lowest, at the bottom. Trials correctly detected are indicated by solid lines. Those made at threshold are plotted in black.

Tests were conducted at the Shreveport VA Medical Center, under a protocol approved by its IRB. Ten healthy adults (21 to 59 y.o.) participated after giving informed consent.

RESULTS

In four subjects, there was clear evidence of evoked TA EMG activity for most moves made above threshold, and little if any activity for below-threshold moves (Fig. 1). In contrast, their GS EMG activity was not so linked. When questioned after a full run, subjects in general often said that the made the interval choices based on the feeling that "their leg muscles work harder."

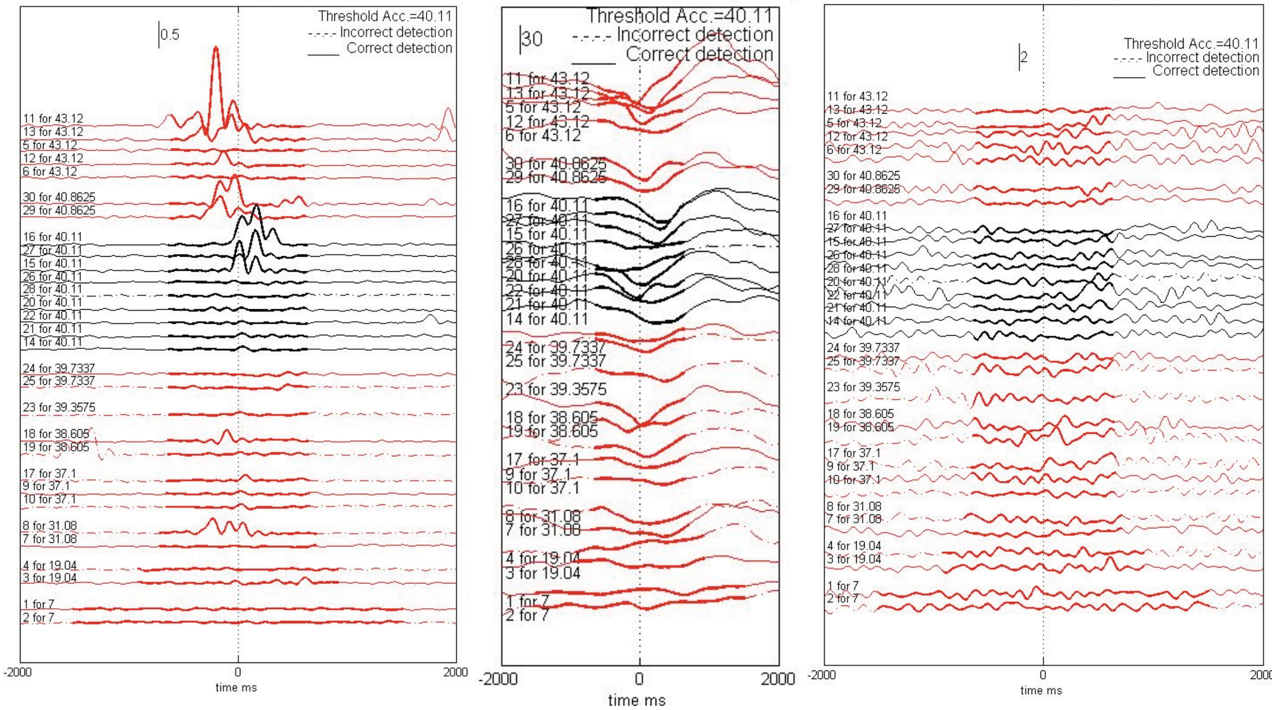

Fig 1: Raster plots of combined L&R TA (A) and GS (C) surface EMG activity and AP COP (B) for 16mm moves for a 59 y.o. subject, sorted by the test acceleration applied. Larger energy EMGs are higher up within each accel. set. Black = threshold. Dotted lines = incorrect detection.

Similar findings like Fig. 1 were seen for other subjects at 16mm. This patterning was not normally seen for 1 and 4 mm moves. Plots similar to Fig. 1, but centered on where a move would have been in a non-stimulated interval, in general show no activation. Note that a strong deterministic AP COP response generally occurs above threshold (Fig. 1B), but seldom below. Threshold or superthreshold forward perturbations cause a rearward shift in COP that apparently generates compensatory Tibialis Anterior activation. In contrast, the later forward COP shift does not seem to be the result of Gastrocnemius activation. In further studies, we will also be monitoring bilateral Soleus EMG activity.

DISCUSSION/ CONCLUSION

Fig. 1 shows that most, but not all threshold or supra-threshold moves result in TA activation. In keeping with accepted practice, we define psychophysical threshold as the acceleration at which a subject can detect a move 75% of the time, and our test procedures titrate the peak acceleration value from trial to trial until this value is reached. If the TA signal is indeed one of the ones used by a subject to help make a force choice decision, it should be expected that some perithreshold trials might not see such activaion. In some ways, based on these findings and on the subjects' comments, we might have found a way to measure a kinesthetic sense.

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ACKNOWLEDGEMENTS

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SHOULDER MUSCLE CONTRIBUTION TO RECOVERY FROM A TRIP

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INTRODUCTION

Recovery responses after tripping have extensively been studied in order to gain insight into reasons why some elderly fall more often than others (van Dieën et al., 2005; Grabiner et al., 2007). Those recovery responses involve multiple limbs. The support leg generates a vigorous push-off and the recovery leg is placed as far as possible forward. Both responses have been shown to be effective in breaking the forward angular momentum of the body (Pavol et al., 2001; Pijnappels et al., 2004). Furthermore, it has been shown that both arms are elevated after tripping (Roos et al., 2007). However, as of yet it is unclear whether arm elevation serves to brake an impeding fall or whether those movements contribute to balance recovery. The purpose of the present study was to uncover to what extent arm movements are initiated actively after tripping and what role the arms serve in the recovery after tripping.

METHODS

Ten healthy young adults (6 males and 4 females) walked 70 times over a 12 x 2.5 m. platform at a self-selected velocity.
In about 10 trials subjects were tripped. On-line kinematic data were used to select which of 14 hidden obstacles should pop up100 ms prior to collision in order to catch the left leg at mid swing (Pijnappels et al., 2005). Subjects wore a safety harness connected to a ceiling-mounted rail in order to prevent injury in case of a fall.
Surface EMG was recorded of the left and right main shoulder muscles: m. pectoralis major (Pe), m. deltoideus pars clavicularis (Dc), m. deltoideus pars acromialis (Da), m. biceps brachii (Bi), and m. triceps brachii (Tr). EMG signals were whitened (fifth order), Hilbert transformed, and low-pass filtered with a fifth order (frame size 21) Savitzky-Golay filter (see Pijnappels et al., 2005). Muscle response onset was determined according to Staude and Wolf (1999) after subtraction of an average of 2 normal walking strides. Response magnitude was calculated as the average amplitude deviation from normal walking in the first 200 ms after trip initiation.
Full body 3D kinematics were recorded at a sample rate of 100 Hz (Optotrak, Northern Digital) and body angular momentum and angular rotation were calculated in all 3 planes of motion. The contribution of the arms to recovery was quantified by calculating how the body would rotate between trip initiation and recovery foot landing if the arms had not been there. Only the first trip was analyzed for the present study.

RESULTS AND DISCUSSION

Averaged over subjects, EMG data showed deviations from normal walking activity within 100 ms after trip initiation in all muscles (Figure 1). Only small differences in onset times were seen between muscles and between the left and right sides. Response magnitudes were asymmetric in that they were significantly larger in the retroflexor (Tr) in the left arm and in the antefexors (Dc and Bi) in the right arm. This was consistent with arm kinematics, which showed that sideward elevation was combined with retroflexion in the left arm and anteflexion in the right arm.

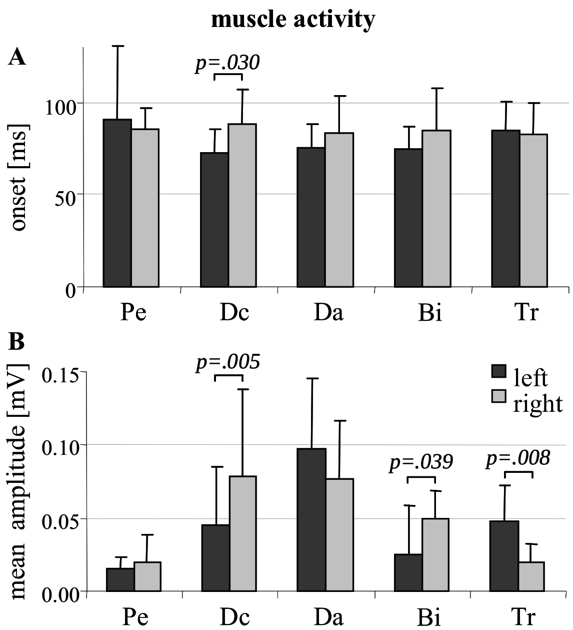

Figure 1: Onset and average amplitude of shoulder muscle surface EMG within 200 ms after tripping. Note that normal walking EMG was subtracted.

The asymmetry of the shoulder EMG responses and of the resulting arm motions indicate that the arm motions after tripping are not intended to brake a possible fall. Analysis of the angular momentum and of the resulting body rotation at the instant of recovery foot landing showed that the arm motions have a minor (about 3 ^{o}) contribution to the reduction of forward body rotation and a major (over 20^{o}) contribution to the body rotation in the transverse plane. Specifically, arm motions caused the body to rotate more to the right around the vertical axis. This rotation helped to place the tripped left leg further forwards during landing. Forward positioning of the recovery leg has been shown previously to be important for braking the forward body rotation and prevent falling (Pijnappels et al., 2004).

SUMMARY/CONCLUSIONS

EMG and kinematics were measured in order to uncover the role of the arms in attempted recovery from a trip during gait. Asymmetric shoulder muscle EMG responses and asymmetric arm movements were found, indicating that the arms were not just elevated forward to brake a possible fall. The arm movements appear to affect body rotation in such a way that recovery is supported. This effect is relatively small in the sagittal plane and large in the transverse plane.

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POSTURE-MOVEMENT STRATEGIES TO ADAPT TO REPETITIVE MOTION-INDUCED ARM FATIGUE

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INTRODUCTION

Upper limb repetitive motions are a regular occurrence in everyday life and in many occupational settings. Recent work has shown that repetitive motion-induced fatigue not only impairs the arm motion characteristics but also provokes complex changes across the body (Côté et al. 2002, Nussbaum et al. 2001, Sparto et al. 1997). These observations could reflect either an altered ability to stabilize one's posture resulting from central fatigue or rather the development of voluntary whole-body compensation strategies as fatigue develops in the arm. These hypotheses have never been addressed using detailed 3-dimensional posture and movement analysis.

METHODS

Healthy subjects (N = 14, 8 males and 6 females) stood and performed a task with the dominant arm consisting of continuously reaching between two targets with the index finger placed at shoulder height, at 30 and 100% of arm's length, in front of the subject's midline. They executed one reach/s until reporting a perceived level of exertion of 8 on the Borg CR-10 scale for the shoulder region or the movement frequency could no longer be maintained. Subjects were required to keep their arm at or above shoulder height. This was ensured using a mesh barrier placed just below the shoulder height that spanned the arm trajectory. During the last 30 s of each minute the task was performed, we recorded the activity of 16 muscles (Noraxon©), whole body kinematics (VICON-Peak©) and forces under the feet (AMTI©). Whole-body center of mass (COM) was computed from segment kinematics and standard anthropometric tables. The effect of arm fatigue on posture and movement characteristics while performing the reaching task was assessed every minute but only the first and last one minute intervals have been analyzed to date to represent No-fatigue and Fatigue conditions, respectively. Immediately before and after the reaching task, subjects performed maximal voluntary isometric contractions (MVICs) for shoulder flexion and elevation and elbow flexion and extension while we recorded the activity of the main shoulder and arm muscle agonists (anterior deltoid, trapezius, biceps, triceps). Force output during MVICs was also measured using an upper limb dynamometry system (BTE©). Statistical comparisons were performed using Student's paired T-tests with appropriate Bonferroni corrections.

RESULTS AND DISCUSSION

Subjects performed the reaching task for a mean time of 7.9 ± 4.0 minutes. The reason for stoppage of the reaching task for 13 out 14 subjects was reporting a Borg rating of 8 or above for the shoulder region. Force output for the shoulder elevation MVIC decreased by a mean of $4.9 \pm 8.2\%$ ($p<0.05$) providing evidence that subjects were affected by the prolonged reaching task at a functional level. When reaching, average

COM and center of pressure (COP) positions were both shifted more laterally toward the non-reaching side for the Fatigue condition in 13 of 14 subjects (shifts of 11.6 ± 9.7mm, $p<0.001$ and 11.1 ± 9.9mm, $p<0.001$ for COM and COP respectively, see Figure 1). Average vertical ground reaction forces under the non-reaching side foot showed a significant increase of 27.5 ± 22.6N ($p<0.001$). Whole-body postural changes were accompanied by an increase in the reaching arm's shoulder joint elevation (6.8±7.8mm, $p<0.01$) with respect to the thorax and a decrease in average shoulder joint abduction angle (-8.3 ± 4.4º, $p<0.001$). No changes were observed in elbow or wrist angles or endpoint trajectory. Finally, only the COM and COP medio-lateral average velocities showed increased inter-trial variability with fatigue (2.1 ± 0.9mm/s vs. 4.0 ± 1.9mm/s, $p<0.001$ and 3.8 ± 1.5mm/s vs. 6.3 ± 2.9mm/s, $p<0.01$, for COM and COP, No-Fatigue vs. Fatigue, respectively). It is believed that the observed kinematic adaptations to fatigue may reflect voluntary strategies aimed at reducing the load on the trapezius muscle with the goal of prolonging task duration. Decreasing the shoulder abduction angle would thus decrease the moment arm of the reaching arm segment. It is hypothesized that the strategy of shifting one's COM to the non-reaching side and elevating the shoulder joint is to allow the subject to keep the arm above the barrier height and maintain the necessary endpoint trajectory even with reduced shoulder abduction. In turn, this suggests that even when fatigued, subjects are able to recruit additional degrees of freedom to maintain task performance. The observation that there was little variability in motor patterns with fatigue reinforces the hypothesis that the motor system is able to find new, stable patterns of motion despite the presence of fatigue.

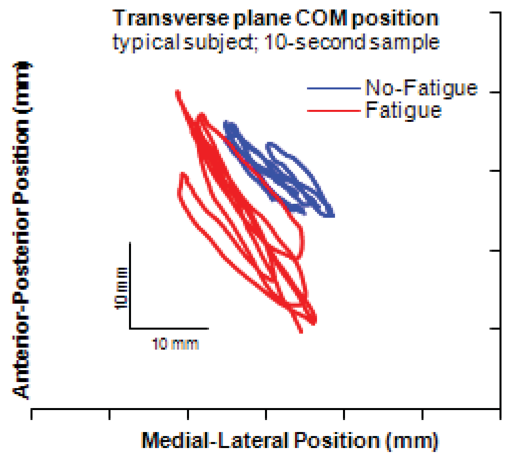

Figure 1: COM trajectory for No-Fatigue and Fatigue conditions. COM average shifts laterally towards the left with fatigue.

SUMMARY/CONCLUSIONS

Performing a prolonged repetitive reaching task with the dominant arm results in localized changes in the predominant area of exertion (shoulder region) as well as whole-body changes. Although fatigue related changes have previously been associated with postural instability, we propose that these changes may in fact be an attempt to adopt a strategy that will allow for prolonged task duration. It is hypothesized that similar strategies may be utilized during many other high-repetition tasks.

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ASSOCIATIONS AMONG BALANCE, ELECTROMYOGRAPHY AND LOWER EXTREMITY INJURY IN BALLET DANCERS: A PRELIMINARY STUDY

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INTRODUCTION

Ballet is a vulnerable expertise because of the high skilled performance, demanding physical activity and specific academic dance technique. The ankle and foot injuries constitute 35~55% of the ballet injuries. The contributing factors to ankle and foot injuries may include poor control of ankle stability in ankle plantarflexion and faulty landing. Five basic positions of classical ballet are also thought to contribute to the prevalence of ankle and foot injuries due to the forced motion of lower extremities to achieve the parallel feet.

Ballet dancers require high level of balance control during the dancing performance. Previous study had found that professional dancers had better balance control than that in normal subjects in single-legged standing. This result, however, may not represent the balance control for ballet dancers because ballet dancers may be good at unilateral balance tests but exhibit unsatisfactory balance control in ballet choreography. In addition, a significant decrease in muscle activity of peroneus longus was observed in the injured side compared with the uninjured side. This result indicated a possible link between injury and muscle activity. Thus, the purpose of this study was to determine whether the balance control and muscle activation were correlated in single-legged standing and basic position 1 and 5 and how they affect the injury.

METHODS

Seven professional ballet dancers, three in injured group and four in uninjured group, and six normal healthy age matched subjects as a control group were recruited for this study at current stage. The injury criteria were that dancers suffered from one or more injuries in the lower extremity related to the ballet in the past year and it stopped her from participating dance for at least twenty-four hours. All of the ballet subjects filled out the self-reported questionnaire about the history of dance injuries and training history. The tasks were one-legged standing with eye open (EO) and close (EC) of each leg, first (B1) and fifth (B5) basic position of classical ballet.

A Kistler force plate (Type 9281B, Kistler Instrument Corp., Switzerland) was used to record the three-dimensional resultant ground reaction forces. The standard deviation of COP position in anterior-posterior (SD_{AP}) and medial-lateral (SD_{ML}) direction represented the variation in the distribution of the COP position. The total trajectory (COP_{TRA}) of the COP represented the total distance that COP traveled. The COP outcome measures were normalized to corresponding foot length. The surface electrodes (MA-300-10 EMG system) were applied over the peroneus longus (PL), tibialis anterior (TA), medial gastrocnemius (MG) of both legs, and the vastus medialis (VM), vastus lateralis (VL), hip adductors (ADD), hamstrings (HAM) of dominant leg. The EMG signals were pre-amplified with a bandwidth of 20-2,000Hz and sampled at

1000 Hz using data acquisition software (EVaRT4.4). The root mean square (RMS) value represents the square root of the average power for a given interval which was set for 0.1 second. The EMG normalization was done by normalizing to the peak magnitude recorded of each muscle during each trial.

RESULTS AND DISCUSSION

The injured group had greater SD_{AP}, SD_{ML}, and COP_{TRA} in EO and EC trials compared with the uninjured group which may indicate better balance control in uninjured group than that of injured group. Ballet dancers had similar COP results in B1 and B5 trials in either group.

The EMG results showed individual variation in all muscles. We didn't find the correlation between COP variation and EMG activity in EO and EC conditions at current stage. However, the muscle activity of PL and MG in uninjured dancers seems greater than that in injured dancers during B1 and B5. On the contrary, the muscle activities of VM in uninjured dancers were less than that in injured dancers in the same trials (Fig.1). This may indicate that different strategies used in balance control between injured and uninjured dancers. Good balance control during dancing requires sufficient muscle strength and proprioception. The muscles are demanded to frequently plantarflexed the ankle in dancers. Therefore, sufficient ankle muscle activity during ballet dancing was important, particularly the PL which is to against the inversion instability and to prevent injury.

SUMMARY/CONCLUSIONS

The balance strategy used in uninjured dancers was different from that in injured dancers based on our preliminary findings. Thus, ankle muscle training should be advised to dancers and dance educators to prevent further dance injury. Correlation among EMG, balance and injury will be determined and added to this study after collecting more subjects.

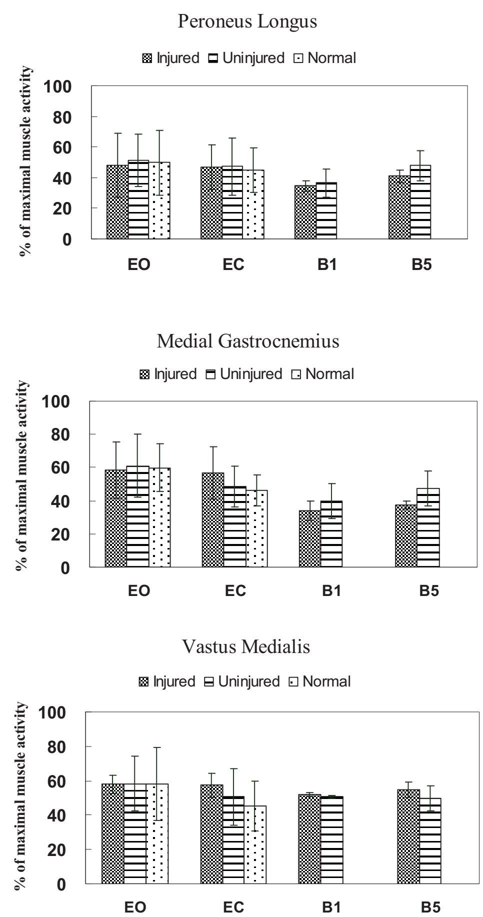

Fig.1: Mean RMS EMG of peroneus longus, medial gastrocnemius, and vastus medialis.

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EFECTS OF SENSORY STIMULI ON POSTURAL CONTROL PROSPECTIVE CASE STUDY.

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INTRODUCTION

The postural control systems are constantly confronted with sensory challenges. For example, (W. Van Asten et al 1988) have shown that postural oscillations increase when an individual observes rotating visual scenes. Postural balance can also be influenced by automatically perceived information coming from the ankles and soles of the feet. When this information is available, postural stability increases, whereas reduced information, due to peripheral neuropathies, results in diminished postural control. The challenge to integrate peripheral information and generate adequate motor response is even greater when the spinal cord is damaged.

This is one of the problems to be solved by rehabilitation medicine; understanding which sensory mechanisms facilitate better postural control in individuals with spinal cord injury. The objective was analyze the specific manual sensory stimulus (SMSS) that generate changes in postural control in a subject who has suffered an incomplete spinal cord section at fifth cervical vertebra (C5) level.

METHODS

A prospective case study. Tetraplegic patient, 21 years old, with incomplete spinal cord section at C5 level. According to ASIA index, he has 33 motor-score and 10 sensory-score. The subject gave written consent approved by the institutional review board of the Universidad de Talca. The patient was evaluated in static posturographic platform in sitting position five minutes before and five minutes after the intervention with pressure stimuli (Vojta Therapy) since day number 465 after spinal cord section once a month for ten months. The evaluation consisted of three phases. In the first phase, the subject was asked to maintain the center of pressure (CoP) in the support base for 30 seconds while a visual feedback of the center of pressure was provided through a computer screen. The second phase was similar to the first phase but the subject had to keep his eyes looking to the front without visual feedback, and the last phase was similar to the previous one but with eyes closed. Signals obtained from the displacement of center of pressure during the three tests were analyzed by the Wavelet Transform in order to determine the intensity of each band (4Hz, 2Hz, 1Hz, 1/2Hz, 1/4Hz 1/8 Hz and 1/16Hz). Area and peak-velocity of pressure center displacement (DCoP) was calculated for each evaluation.

RESULTS

We analyzed only five evaluations. Lower levels of RTE were obtained in all evaluations when the pre-intervention evaluation were compared with post-intervention evaluation with eyes open 0.0537 ± 0.06 to 0.0264 ± 0.04 J (see figure1) and eyes closed 0.1300 ± 0.17 to 0.1061 ± 0.16 J phases (see figure2). Lower DCoP area was also obtained after intervention of both phases

16473.75± 25003.03 to 10183.07± 17315.04 mm^2 and 77011,42± 60822,84 to 68377,03± 107073,25 mm^2 respectively (see figure 3). Greater decrease of DCoP peak-velocity was found in both phases 268.345± 64.87 to 243.3± 46.97 mm/s and 325.26± 135.04 to 308.1±145.77 mm /s respectively (see figure 4).

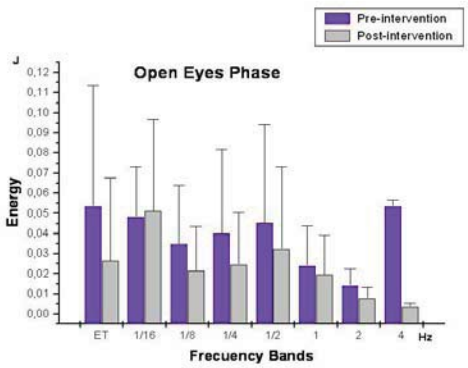

Figure 1: Shows the distribution of frequency bands in open eyes phase (mean ± SD).

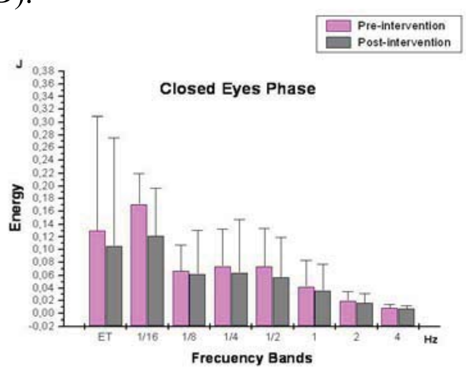

Figure 2: Shows the distribution of frequency bands in closed eyes phase (mean ± SD).

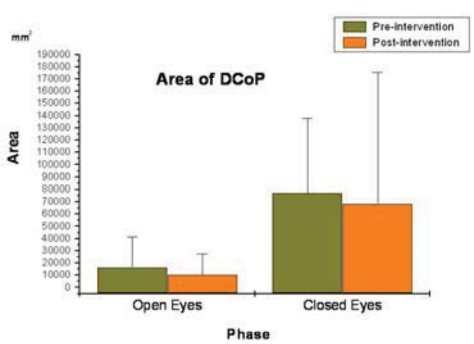

Figure 3: Shows Area of displacement centre of pressure (mean ± SD).

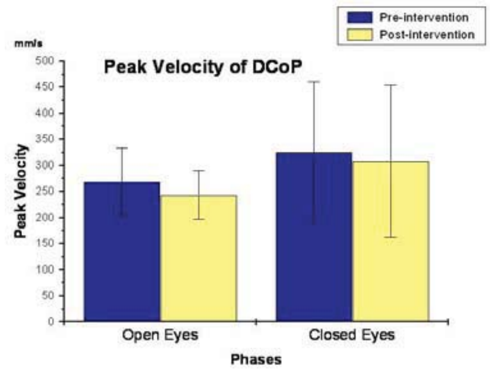

Figure 4: Shows peak-velocity of displacement centre of pressure (mean ± SD).

DISCUSSION

Lower levels of ETR obtained with eyes open as well as eyes closed could be understood as training or improvement in maintaining balance at reflex/automatic levels. This would be due to lower levels of energy in low frequency bands (predominantly in the 1/16 Hz) and the lower DCoP and the decrease in the peak-velocity area post-stimulation. Therefore SMSS could indicate an improvement in the postural control in a Tetraplegic through the activation of receptors of low threshold (mechanic and propioceptive) which act at involuntary level firing the activation of diverse descending unharmed circuits (or internal patterns) that would producing an increase in the stiffness in the undamaged muscles in the upper body. As it is known any change in the stiffness joints would be affect the control of balance (D.A Winter et al 2003), in this case maybe an improvement in the postural control in space.

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OCCLUSION STATUS EVALUATED DURING MEASUREMENTS OF DYNAMIC BALANCE BY MEASURING MASSETER ACTIVITY USING EMG SYSTEMS.

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INTRODUCTION

Factors involved in the ability to balance include equilibrium function, visual acuity, postural reflexes, and muscle strength. Causal factors of falls have been reported to include reduced muscle strength of the lower legs, reduced visual acuity, stumbling, and slipping. However, there are no reports considering occlusions as an additional causal factor. Though some reports have indicated an relationship between occlusion and balancing ability with general motor function. Some investigators have suggested that occlusion and head position affect the sway of the center of gravity. Therefore, a relationship may exist between occlusion and balancing ability, and thus, between occlusion and the risk of a fall.

METHODS

Thirty healthy adolescents (15 males and 15 females, average age, 20.3 SD 1.6 years) without any equilibrium or stomatognathic function abnormalities were included in the study. All 30 subjects were studied under conditions of with or without occlusion. The average subject body height (plus standard deviation) was 160.6 SD 7.4 cm. Informed consent was obtained from all subjects following a thorough explanation of the objectives and procedures, and before the commencement of the study. The study was approved by the ethics committee of our university.

Occlusion is a term meaning “jaw clenching”. Masseter responses were measured in real-time by attaching electrodes with a built-in amplifier, part of a portable EMG measurement system (SX230, Biometrics Ltd, Gwent,UK), on the skin over the bilateral masseter muscles. Muscular discharge was captured at a sampling frequency of 1 KHz with a low cut-off frequency of 5 Hz and a high cut-off frequency of 500 Hz. The normalized root mean square (nRMS) of the muscular potential obtained from the bilateral masseter was averaged (nRMSavg). Right and left maximum voluntary contractions (MVC) were also measured five times for the upper and lower second molars of each subject while in the standing position, and the average was regarded as 100% MVC. The level of 50% MVC against 100% MVC was then shown on a PC display. Subjects were asked to try and produce 50% MVC with their jaws, doing this 10 times. The levels produced were confirmed by visual biofeedback. At least 20% MVC was regarded as “with occlusion”. Balancing ability was measured while urging the subjects to produce 50% MVC. Since subjects were required to pay attention to the visual surroundings during measurement of balancing ability, they could not provide visual biofeedback; thus, the investigator checked in real-time that the nRMSavg value corresponded to 50% MVC. In cases with values below 20% MVC, re-measurements were conducted. For the "without occlusion" condition, subjects were urged not to clench their teeth, and in this case, re-measurements were carried out when the nRMSavg was 10% MVC or higher.

The data obtained were analyzed with a significance level of $p < 0.05$. SPSS version 13.0 (SPSS Inc., Chicago, Illinois, USA) was used for all statistical analyses.

RESULTS

During measurements of dynamic balance, the nRMSavg of the masseter muscular potential in the presence of occlusion was 61.3, SD 29.4% MVC, with a maximum of 95.2% MVC and a minimum of 29.4% MVC. On the other hand, nRMSavg in the absence of occlusion was 4.3, SD 1.7% MVC, with a maximum of 6.9% MVC and a minimum of 2.1% MVC. No cut-off value was observed for occlusion status during measurement of balancing ability. No re-measurements were required.

DISCUSSION

The EMG data obtained during the balancing ability measurements were higher than the predetermined values in the presence and absence of occlusion. In the presence of occlusion, although an nRMSavg of 50% MVC was prompted, the obtained value was 61.3, SD 29.4% MVC, and the minimum 29.4% MVC. The nRMSavg value in the absence of occlusion, for which subjects were prompted not to clench, was 4.3, SD 1.7%MVC with a maximum of 6.9% MVC. The reason for these higher values in % MVC compared to the predetermined values was thought to be due to unconscious clenching in an effort to maintain balance when an external disturbance was applied. However, occlusion status was considered to have been adequately satisfied based on the %MVC values of the nRMSavg.

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ACKNOWLEDGEMENTS

This study was conducted with the approval of our University IRB and with financial assistance from the Ministry of Education, Culture, Sports, Science and Technology, the Grant-in-Aid for Scientific Research (Study No.18500523).

Table 1: EMG data obtained during measurements of balancing ability

	nRMSavg	maximum	minimum
with occlusion	61.3 SD 29.4% MVC	95.2% MVC	29.4% MVC
without occlusion	4.3 SD 1.7%MVC	6.9% MVC	2.1% MVC

ELECTROMYOGRAPHIC ANALYSIS OF THE ANTERIOR, MEDIAL AND POSTERIOR PARTS OF THE TEMPORAL MUSCLE DURING MASTICATION

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INTRODUCTION

According to American Academy of Orofacial Pain the prevalence of temporomandibular disfunction (TMD) in the American population is very high (Siqueira & Teixeira, 2001), and proposes a classification based on its etiology, that is, myogenic, arthrogenic and mixed, with the last being considered the most prevalent. According to Bérzin (2004), an alteration in the kinetics of the three parts of the temporal muscle is one of main etiologies, because the temporal muscle controls the position of the mandibular chondyle in temporomandibular articulation during mastication.

Due to the presence of hair in the region, studies using surface electrodes are rare. Electromyographic studies are more often carried out with needle or wire electrodes, which involve a small area of signal capture, pain, discomfort, difficult with the fixation of electrodes and other methodologic problems, which could produce unsatisfactory results (Basmajian & De Luca, 1985).

The aim of this work was to study the kinesiology of the three parts of the temporal muscle during mandibular movements using surface electrodes.

METHODS

The study involved 7 young subjects without any signs of orofacial pathology, with complete dentition and with the temporal region shaven in conformity with social convention. Electromyographic measurements were made during mastication of parafilm, with left, right and bilateral chewing, and at rest. The subjects were examined in the seated position, with eyes open and the Frankfurt plane parallel to the floor. The surface electrodes used were from Noraxon USA Inc., model 27-2, connected to an electromyograph (Myosystem, Datahommis Tec. Co., 12 bytes resolution, CMRR of 112 db at 60Hz, sampling frequency of 2000 Hz, passband filter 20-500 Hz, and signal value calculated in RMS.

Three measurements were taken for each individual, and after a critical evaluation of the RMS values observed a single outlier was excluded. The means of these measurements were then calculated, which based on the central limit theorem makes the data normally distributed, satisfying a basic condition for the application of analysis of variance. As the data were not normalized, inter-individual differences were suppressed by adopting a mathematical model that considered each individual as a block, and consequently analysis of variance followed a model adapted for a randomized block design experiment, with two factors (side of mastication of the individual and part of the temporal muscles analyzed) and interaction. A significance level of 5% as well as Tukey’s test were determined a priori as appropriate for this study.

RESULTS AND DISCUSSION

First, the results of the analysis of variance individualized in each of the movements studied, are presented in Table 1.

The four movements studied showed level ($p<0.05$) of the existence of differences in means of true RMS values among the different parts of the muscle examined, but there were no observed indictions of a significant effect for the side of the muscle nor indications of interaction between the two principal effects.

Since indications of significant effect were demonstrated, Tukey's test was applied for comparison of the RMS means of the different parts of the temporal muscle.

It was observed initially that at rest, the RMS values are much lower than those observed in the different activities examined. In the resting condition, a significantly greater activation was observed for the medial part of the temporal muscle in relation to the anterior and posterior parts, with a mean RMS of 186 versus 97 and 99 for the anterior and posterior parts, respectively.
In the three types of mastication examined, it is always seen that the posterior part of the temporal muscle displays a lower mean RMS, indicating that this part displays less electrical activity compared to the anterior and medial parts.

SUMMARY/CONCLUSIONS

The three parts of the temporal muscle act as an agonist in mastication, where the posterior part is the least active electromyographically. There is no significant difference in muscular electrical potential in any of the three muscle regions whether mastication occurs on the left or right side or bilaterally. However, at rest, the medial part of the muscle is significantly more active than the anterior and posterior parts

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We thank Dr. A. Leyva for his help in the translation and editing of this work.

Table 1. p values obtained in the analysis of variance of the RMS data for movements studied.

Cause of Variation	Mastication Right	Mastication left	Mastication bilateral	Rest
Side of mastication	0.7138	0.0585	0.3309	0.3842
Part of muscle	0.0013	0.0164	0.0045	0.0006
Side * Part	0.4923	0.4162	0.8319	0.2760

ELECTROMYOGRAPHIC AND CEPHALOMETRIC CORRELATION OF MANDIBULAR BIOMECHAMICAL WITH THE PREDOMINANT MASTICATORY MOVEMENT

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INTRODUCTION

The study of the craniomandibular biomechanics is related to the system of predominant lever and of the relation between function and mechanically induced deformations (Hylander, 1975, 1977, 1985). The aim of this study was to evaluate the chewing muscular dynamics and correlating the chewing movement side that is more vertical and/or more horizontal set up by the photomeasurement of Planas' Masticatory Functional Angle (MFA) to the muscular activity behavior, shown in the surface electromyography and in the radiographic images.

METHODS

Seventeen people of both sex, medium aged about 25, were selected, white skin and presenting Class I of Angle without apparent sign and symptom of muscles disorders masticatory. The panoramic radiography and telerradiography in lateral position and surface electromyography were made. The acquisition of radiographic images followed the rules established by the Piracicaba Dental School (UNICAMP). The electromyographic data were obtained bilaterally from the masseter (muscles), anterior temporal portion and supra-hyoids muscles at the postural position and isometric position. Medtrace® passive bipolar surface electrodes were used coppled to a pre-amplifier, forming a circuit corresponding to a differential circuit. The registrations of the electric signals were caught by EMG- 800C equipment of Brazil EMG System Ltda with eigth channels, sample frequency of 2 KH and 16 bits resolution, digital filter with a band pass of 20-500 Hz. A pressurized transductor was used, which consists of a pressurized rubber tube connected to a sensor (MPX 5700)* to obtain maximum biting force. A mandibular goniometer (EMG System of Brazil) was used to measure the opening size. A comparison and correlation between the groups with MFA < 5° and MFA> 5° through "*t*" Student or a Man-Whitney test according to the normality or not of the distribution, respectively (Bérzin, 1995, 2004).

RESULTS AND DISCUSSION

The results have shown important differences between the groups with MFA < 5° and MFA > 5° but without sexual disformity to the biting force measurements (38,70 ±, 10,88 and 27,28 ±11,40), for the maximum opening. (40,04 ± 11,82 and 26.86± 11,70) and masseter muscle isometric (174, 16± 49,67 and 116, 41± 51,11).

* Motorola SPS, Austin, TX, USA

A strong correlation between the masseter muscles to the biting force (r = 0,63 p =0,0001) occurred for the groups with MFA > 5°. On the Hyoid triangle (Bibby & Preston, 1981), for both groups, the vertical and angular behavior of the hyoid bone represented by H–H' and PHA showed important correlations to the mandibular dynamics recorded on the panoramic radiography.

SUMMARY/CONCLUSIONS

These results indicate that to a strong relationship between the aspects anatomic physiological with asymmetric mandibular function.

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THE CORRELATION OF MOBILE TRACING OF SOMATIC GRAVITY CENTER AT THE TIME OF STANDING UP WITH ACTING SPEED

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INTRODUCTION

The action of standing up is the most basic daily action for humans. There have also been many studies on standing up. Some reports have shown that particularly subjects with motor hypofunction are characterized by high angle of somatic anteversion during actions. It is thus widely recognized that subjects with motor hypofunction have kinetic characteristics in mobile tracing of somatic gravity center in the action of standing up. There have been many previous reports on determinants for such kinetic characteristics, which include muscle activities of the lower extremities and articular torque as parameters. It has also been known that subjects with motor hypofunction show low acting speed in the action of standing up. The present study was designed to elucidate the influence of acting speed on determinants for mobile tracing of somatic gravity center at the time of standing up. **Specifically**, characteristics of mobile tracing of somatic gravity center were regarded as curvature, and the correlation of curvature with acting speed was analyzed.

METHODS

The subjects of the analysis were 10 healthy adults [5 men and 5 women aged 21-23,with a mean age of 21.7 years], and their mean height was 163.2cm [±7.5]. The contents of the experiment were explained to the subjects, and their consent was obtained. This study was conducted with the approval of the Ethics Committee of Tokyo Metropolitan University.

A. The method for calculation of the position of somatic gravity center

Photographs of actions were taken with a three-dimensional action analyzer (VICON mx). The position of somatic gravity center was calculated from the coordinate data on the four extremities, which were obtained from the photographs.

B. The method for calculation of acting speed and curvature

Mobile tracing of somatic gravity center was recorded as two-dimensional motion on a sagittal plane. With regard to the recorded two-dimensional tracing of somatic gravity center, the data on gravity center for the time from the start of motion to stasis was extracted, and the time required for actions was regarded as acting speed. The subjects of tracing of somatic gravity center included 80% of 100% the whole time for actions (the remaining 10% and 10% corresponded to the time for the start and the time for the end, respectively, of actions). Curvature was calculated in 0.1-second unit according to the methods shown in Figure 1, on the basis of the collected mobile tracing of gravity center. Curvature indicates the quantity, which expresses sharpness of a curve; e.g., curvature of the circumference of a circle with a radius (r) is 1/r. When the curve is sharp, curvature increased.

C. Experimental procedures

Each subject was instructed to voluntarily stand up 9times at low to highest speed. The height of the chair was set at which the subject's knee joints show flexion of 90°.

D. Methods for statistical analysis

The maximum curvature calculated on each occasion was used as dependent variable, and the subject's height and acting speed were used as independent variables for multiple regression analysis.

RESULTS AND DISCUSSION

As a result of the analysis, the mean maximum curvature was 15.41rad/mm [±5.96],and the mean acting speed was 1.68sec[±0.80]. The result of the multiple regression analysis showed the multiple correlation coefficient was 0.72. The prediction formula is shown below.
These data indicated that acting speed is highly correlated with mobile tracing of somatic gravity center in the action of standing up. Healthy adults can perform these actions at various acting speeds. However, the results of the present analysis showed fixed correlation between acting speed and mobile tracing of somatic gravity center concerning the standing up action, suggesting that people with motor hypofunction can stand up only by characteristic movement of somatic gravity center because there is a limit to acting speed.

SUMMARY/CONCLUSIONS

A relationship between mobile tracing of somatic gravity center at the time of standing up and acting speed was analyzed in healthy adults. The results of the analysis showed high correlation between these parameters. This indicates that action patterns are determined by a limit to acting speed in moto r hypofunction.

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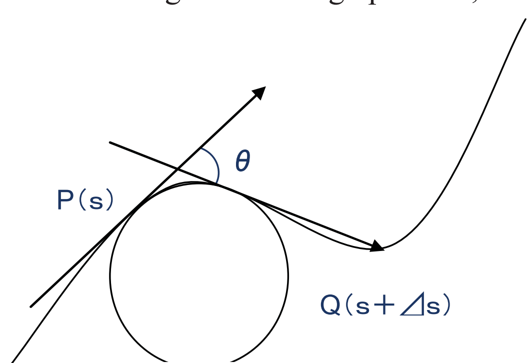
2)Osamu Nitta, et al.: Power assistance type handrail which synchronizes with user's standing up operation, The World Congress of The International Society of Physical and Rehabilitation Medicine,ABSTRACTS,S268, 2005.

[y=5.16*(speed:rad/sec)+-0.22*(hight:cm)+43.23]
prediction formula



$$\chi = \lim_{\Delta s \to 0} \left| \frac{\Delta \theta}{\Delta s} \right| = \left| \frac{dt}{ds} \right| = \frac{1}{R}$$

Figure1: The method for calculation of acting speed and curvatur

DOES KNEE ANGLE ALIGNMENT AFFECTS POSTURAL CONTROL?

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INTRODUCTION

Knee hyperextension is a condition in which knee extends beyond the neutral position (180˚). More frequently found in females, subjects may experience pain and may be predisposed to a greater risk of anterior cruciate ligament and other knee injuries (Loudon et al. 1998; Devan et al. 2004).

Postural alignment, although poorly documented, is thought to interfere in postural control. The purpose of this study was to verify if knee angles change during quiet standing position under visual and proprioceptive perturbation, and if postural control differs between hyperextended and aligned knee groups.

METHODS

19 healthy young female adults performed 3 trials of 30 seconds during quiet standing position under four test conditions: (i) stable surface/eyes open, (ii) stable surface/eyes closed, (iii) unstable surface/eyes open and (iv) unstable surface/eyes closed.

Kinematic data were acquired by means of a Panasonic digital camcorder at 60 Hz and Ariel Posture Analysis System (APAS) software. Range of motion (ROM) and mean angle (Ak) of the knee joint were calculated. Kinetic data were acquired using an AMTI OR6-7 1000 force plate at 100 Hz. RMS value and Mean Velocity (MV) of the Center of Pressure (COP) displacement in the anterior-posterior direction and the area (AREA) of the ellipse that contains 85% of the total COP displacement were calculated.

Initially, subjects were classified in aligned (n=5) or hyperextended group (n=14) using a criteria in which the knee joint angle was shorter or higher than 180˚, respectively, during the whole trial in test condition (i). Analysis of variance (ANOVA) with Tukey post-hoc test was applied to perform between-groups comparison of kinematic variables (α=5%).

Afterwards, as subjects initially classified in hyperextended group, in subsequent trials with higher postural demands, exhibited a flexion of their knees they were not hyperextended anymore. Therefore, all trials in all postural conditions tested were reclassified in hyperextended or aligned using the same criteria as initially. ANOVA with Tukey post-hoc test was applied to perform between-groups comparison of kinetic variables (α=5%).

RESULTS AND DISCUSSION

Within-groups comparison of postural conditions tested showed no statistical differences of ROM and Ak in the aligned group. However, in the hyperextended group Ak and ROM were respectively shorter and higher in condition (iv) than in the condition (i) and (ii) ($p<0.01$). See Table 1.

Between-groups comparison showed no significant differences in conditions (i), (ii) and (iii), for any COP variable. However, in condition (iv) the hyperextended group showed lower values than the aligned group for all variables. Figure 1 shows the between-groups comparison for MV.

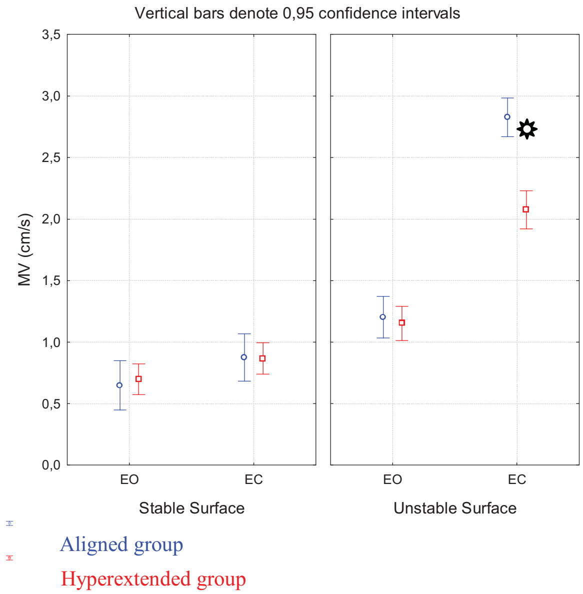

Figure 1: Mean and 95% confidence interval for the MV in the aligned and hyperextended group in all conditions. EO- eyes open; EC- eyes closed.✡ $p<0.01$

There is a lack of reports regarding the relationship of postural alignment and postural control. Ferreira et al (2007) found no correlation between 11 variables of postural alignment in sagittal plane and COP variables in stable surface, eyes open condition. Our results suggest that in normal subjects this correlation might occur as demands in postural control increase.

Subjects who under conditions with full sensory input keep the knees hyperextended, when under more unstable condition tend to flex their knees. 27% of the subjects that had hyperextended knees in the most stable condition had knee joint angle below 180˚ during the unstable surface, eyes closed condition. 63% of the subjects had Ak shorter in the most unstable condition than in the most stable and only 10% had higher Ak.

This result suggests that although knee hyperextension is thought to provide stability by means of capsule and ligaments locking, this might occurs only in stable conditions. When balance is more demanding one tends to change strategy and recruits muscular activity upon the passive knee postural maintenance. This new strategy, leading to an increment in knee motion might contribute to the higher values of COP parameters found in the aligned group.

SUMMARY/CONCLUSIONS

These results suggest that knee angles change under visual and proprioceptive perturbations. Also, knee angle alignment affects postural steadiness.

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Table 1: Mean angle of knee (^{o}) in the 4 test conditions in aligned and hyperextended groups.

Group/Condition	(i)	(ii)	(iii)	(iv)
Aligned	177.13± 0.7	176.46± 1.1	175.92± 0.9	174.68± 1.9
Hyperextended	185.24± 3,7	184.93± 3.8	184.03± 4.7	181.84± 5.8*

(i) stable surface/ eyes open, (ii) stable surface/ eyes closed (iii) unstable surface/ eyes open, (iv) unstable surface/ eyes closed. *Within-group comparison (i) > (iv) e (ii) > (iv) $p<0.05$

COORDINATION TRAINING OF POSTURAL MUSCLES – DOES IT IMPROVE POSTURAL CONTROL?

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INTRODUCTION

Physically hard work such as cleaning work is known to have a tearing rather than a training effect on the body. Cleaning work includes a lot of standing and walking and working with arms above shoulder height, which result in a prolonged exposure on the trunk muscles. Danish cleaners seldom do physical training in their spare time. Still, these workers may need a well-planned training program in order to enhance physical work ability and thereby reduce relative exposure. Training of trunk muscles must focus on both strength and coordination in order to enhance workability in standing work. We tested the hypothesis that 4 weeks of coordination training reduces the sway area in a postural sway test and the displacement in a perturbation test.

METHODS

18 healthy female workers participated in the study. The subjects were divided in two groups based on their work place; an intervention group (n=10, 46 yrs, 167 cm, 68 kg) and a control group (n=8, 51 yrs, 163 cm, 75 kg). For four weeks the intervention group was offered 3 weekly training sessions at the work place. The training included 7 coordination exercises involving postural muscles at an intensity corresponding to mean emg range for the muscles erector spinae 14-57%EMGmax, rectus abdominus 5-61%EMGmax, obliquus externus 12-70 %EMGmax, latissimus dorsi 8-44 %EMGmax, and trapezius 5-57 %EMGmax. Before and after the four week intervention period the subjects' postural control were tested in five postural sway tests and one perturbation test on a force platform. The postural sway tests included standing in Romberg position and tandem position on a force platform with both open and closed eyes and standing on one leg with open eyes only. The subjects were instructed to stand as still as possible in the position looking at a fix point for one minute (30 seconds in the one-legged position). Subjects had 6 trials for each test. The resulting sway area (95% confidence ellipse) was calculated, a smaller area represents better postural control and the four best trials for each subject were used in the statistical tests. In the perturbation test subjects stood in a Romberg position with their arms elevated to horizontal and holding a bar. A load (2.2 kg) was magnetically anchored to the centre of the bar. The magnet would demagnetize and the load drop from the bar at a randomly selected time point in an interval between 5 and 15 seconds after the trial began. The maximal anterior-posterior displacement during the first second after the load drop was used as the result of the perturbation, a smaller displacement representing a better postural control.

An ANOVA was used to calculate whether there was a difference in postural control in the two groups before and after the intervention period. The statistical significance level was set to 0.05. Tests of the one-sided hypothesis was deemed significant if a two-sided P-value was less than 0.1.

RESULTS AND DISCUSSION

There was no difference between the intervention and control group in age, height, weight or sway area and displacement before the intervention period. This was also the case after the intervention period. Both the intervention and the control group had a significant reduction in sway area. None of the groups had a change in their perturbation test result. A typical force curve for the sway test and for the perturbation test is shown in figure 1 and 2, respectively.

Both the intervention group and the controls seemed to have improved their postural control. The exercise program performed by the intervention group trained specifically the coordination of trunk muscles in a static position and in a static position while using the arms. We therefore hypothesized that it would improve postural control, but the complexity of the tests may be too low to reveal an improvement.

The fact that both groups improved sway indicates that there was a learning effect of the test, that didn't disappear during the four weeks between the tests. The lack of a better result for the intervention group might be the result of a type two statistical error coming about by the small number of subjects further strengthened by general improvement among the subjects because of the possible learning effect.

SUMMARY/CONCLUSIONS

Both the intervention and the control group reduced their sway area, but there was no difference in perturbation before and after the intervention period for either of the groups. The lack of intervention effect may be due to insufficient sensitivity of the test methods together with the rather small sample size. Futher, a significant general learning effect may have masked a possible quite small training effect. Improved methods and larger sample size may be requested to reveal if the coordination exercises can result in an increased postural control.

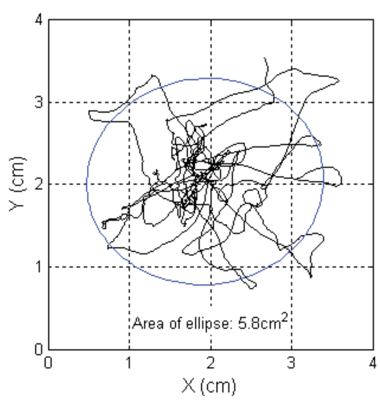

Figure 1: The graph illustrates a typical curve for the result of a postural sway test**.** The circle illustrates the 95% confidence ellipse.

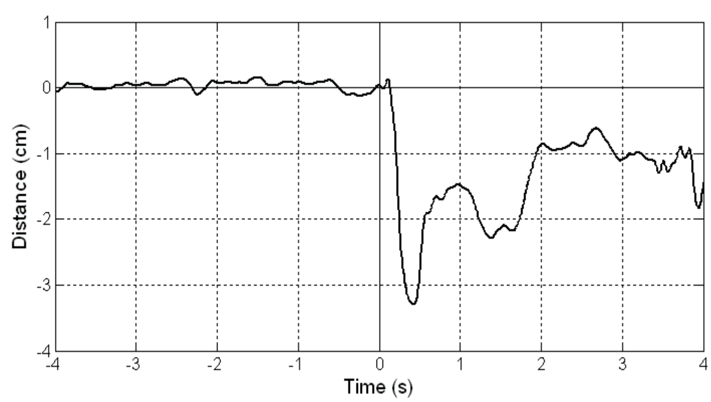

Figure 2: The graph illustrates a typical curve for the result of a perturbation test. Time 0 is the time for demagnetization.

CENTER OF PRESSURE EXCURSION DURING VOLUNTARY TRUNK MOVEMENT IN SITTING

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INTRODUCTION

Voluntary reaching, in sitting, is often used to assess trunk control in patient populations (e.g. Reisman and Scholz, 2007; Messier et al., 2004), as it is both a functional and a relatively easily reproducible task. During this task, the trunk contributes to both transport (e.g. hand position when reaching) and to the maintenance of a dynamically stable seated posture.

The purpose of this abstract is to describe the initial development of a test of voluntary trunk control and mobility, in sitting, using the excursion of the centre of pressure (**CoP**) to assess the challenge to stability posed by target-directed trunk movements.

METHODS

Three young, healthy subjects (1 male, 2 female) provided informed consent for this preliminary protocol, which was approved by the local ethics committee. Subjects sat on an elevated force plate, with 75% of the length of their thigh (greater trochanter to lateral condyle) on the plate, and their feet unsupported. Subjects performed three tasks: 1) quiet sitting (**QS**) for 5 minutes; 2) three maximum voluntary leaning (**MVL**) trials, in each of 8 directions, at 45^{o} intervals about the full circle (random order); 3) three targeted leaning (**TL**) trials at 3 speeds (10, 20 and 30^{o}/s – paced by a metronome), 3 distances, and 5 directions. Target direction and speed were randomized, and target distance was progressively increased (Figure 1). Subjects were instructed to: 1) look at the target; 2) lean towards the target and touch it with their head; 3) return to an upright sitting position. Subjects were also instructed to keep their legs hanging vertically downward and relaxed.

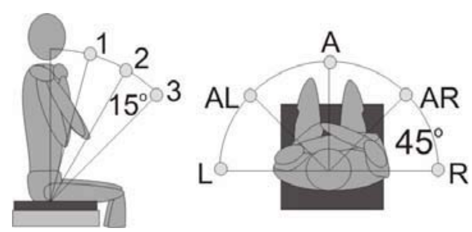

Figure 1: **Left**: target distances. **Right**: target directions.

A best-fit ellipse of CoP motion during QS was determined using a method described by Fitzgibbon et al.(1999). A similar ellipse was fit to the points of maximum CoP excursion from the 3 MVL trials, delineating the limits of stability (**LoS**). The distance from the points of maximum CoP excursion, during the TL trials, to these LoS were then determined, and averaged across the 3 trials. Values were normalized to the length of the anterior-posterior base of support (**BoS**) to compare between subjects.

RESULTS AND DISCUSSION

During QS, all 3 subjects maintained their CoP near the centre of their BoS, with minimal postural sway. The length of the major axis of the best-fit ellipse was ~0.01 of the A-P length of the BoS for all subjects, and was rotated between 14 and 21^{o} from the A-P axis, indicating some directional preference in postural sway. The LoS ellipse for two of the 3 subjects, was virtually symmetrical (0.53^o and -0.97^o rotated), while the third subject's LoS was slightly rotated (-9.75^o). These best-fit ellipses are illustrated in Figure 2.

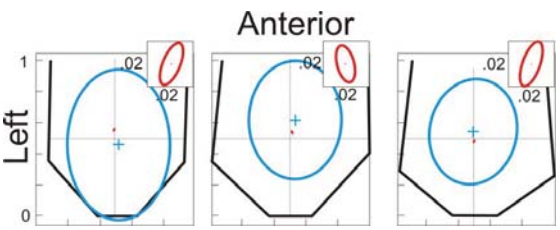

Figure 2: Best-fit ellipses for LoS (blue) and QS (red – inset) for the 3 subjects. BoS (black) has points at the PSIS, greater trochanters and the edge of force plate (dotted lines = BoS center).

For the targeted leaning trials, the CoP excursion increased with an increase in the target distance for all subjects, in all directions. None of the 3 subjects was able to reach the target at the 3rd distance for the left (L) or right (R) directions (Figure 1) without moving their lower limbs to counterbalance the weight of the torso (these trials were considered failed). CoP excursion for subject 1, at 10^o/s, for the 3 target distances, is illustrated in Figure 3.

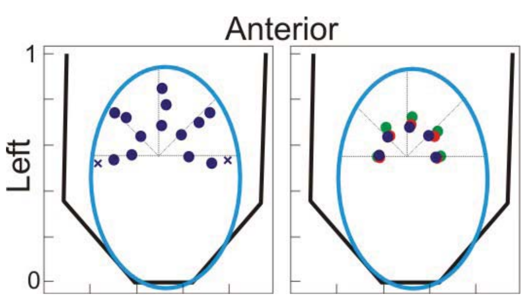

Figure 3: Mean CoP excursion for subject 1. **Left**: 3 target distances at speed 1. The "X" indicate failed trails. **Right**: 1st target distance at speeds 1 (blue), 2 (red) and 3 (green). The LoS ellipse (cyan) and the 45^o axes from the QS position (dotted lines) are also shown.

The mean CoP excursion at the different speeds of trunk movement was similar for each of the target distances, reflecting the instructions to move only as far as the target. There was some tendency for greater CoP excursion at the faster speeds, particularly at the first target distance (Figures 1), although this was not consistent across subjects. An example of the CoP excursion at the three movement speeds, for the first target distance is illustrated in Figure 3.

SUMMARY/CONCLUSIONS

These findings describe preliminary results in the development of a test of voluntary trunk control and mobility, in sitting, based on the excursion of the CoP during targeted trunk movements, relative to pre-determined LoS and QS parameters. The similar excursion of the CoP at the different movement speeds, the increase in CoP excursion with target distance, and the inability of the subjects to reach the 3rd target distance in the lateral directions, provide validity for this test. Further testing is required to establish the repeatability / reliability of these measures, and to support the validity of these tests for a greater number of subjects, and for different subject populations.

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THE RELATIONSHIP BETWEEN MUSCLE POWER AND BALANCE IN POSTURAL CONTROL

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INTRODUCTION

Falls in the elderly lead to fractures and a reduced ability to perform activities of daily living. The cause of these falls is multifactorial. Muscle power and balance are both important factors, and are known to decline with age. However, the relationship between muscle power and balance is not clear. The purpose of this study was to determine the relationship between balance and muscle power in postural control.

METHODS

Subjects included 21 healthy individuals (age range 21-81yrs). First, static balance was assessed using a Romberg balance test (standing with eyes opened and eyes closed) combined with a GRAVICORDER® (Anima, Japan). Active balance was then assessed with the EQUITEST SYSTEM® (NEUROCOM, Clackamas, USA). Subjects were then assessed during an Adaptation test in the toes up (ATT-up) and toes down (ATT-down) conditions. Additionally, the amount of activity in the first and fifth trials were compared to evaluate any learning effects. Subjects' muscle strength was determined by knee extensor power and knee extensor time to peak torque using a Biodex® (Biodex, USA), and by ankle plantar flexion and ankle dorsiflexion power using a μ-Tas® (MT-1; Anima, Japan). Statistical analyses were performed using Spearman's rank-correlation coefficient (SPSS for windows 15, Chicago, IL).

RESULTS AND DISCUSSION

There were no significant correlations observed between muscle power and static balance. Correlations existed between active balance as measured by the ATT-up test and knee extensor power (r=0.486), knee extensor time to peak torque (r=0.582), ankle plantar flexion power (r=0.639), and ankle dorsiflexion power (r=0.468).

The lack of a significant relationship between muscle power and static balance may be due to decreased ROM and loss of spinal flexibility common in the elderly. Additionally, the elderly use a hip strategy for postural control, which may be affected by arthritis, deformed joints, muscular weakness and other conditions. This may explain why length/envelop area in elderly subjects were sometimes better than in young subjects. As a result, static balance may not predict falls, which occur more often in active situations such as walking, turning around, and ascending stairs.

Significant correlations were observed with the ATT-up test between latent muscle time to peak torque and active balance. Researchers have previously demonstrated that onset latency times for the tibialis anterior and quadriceps postural muscles become progressively longer in older adults as they become more unstable. Thus, the relationship between muscle time and peak torque can be used to evaluate a delay of postural control.

The amount of activity observed during the fifth trial was less than that of the first trial in subjects with high muscle power. This indicates that there may have been a learning effect present in subjects with high muscle power.

SUMMARY/CONCLUSIONS

There is a significant relationship between muscle power and active balance. Individuals who have high muscle power and short times to peak torque possess a high capacity for balance and for learning improved postural control. Static balance may not be a predictor of falls due to low levels of body sway secondary to generally poor ROM seen in elderly individuals. Future studies are needed to evaluate the relationship between muscle power and balance in dynamic situations.

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ELECTROMYOGRAPHIC ANALYSIS OF THE STERNOCLEIDOMASTOIDEUS MUSCLE DURING HEAD DYNAMIC MOVEMENTS

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INTRODUCTION

The sternocleidomastoideus muscle (SCM) is located in the neck region and presents two clearly distinct parts in its thoracic origin: an internal part, inserted on the sternum, and another external part, inserted on the clavicula (Testut and Latarjet, 1988; Williams et al., 1995).

Considering the sternum and the clavicula as a fixed point, the SCM muscle acts on the head performing anterior flexion, inclination to the same side and rotation to the opposite side (Testut and Latarjet, 1988).

It has been reported that each part of the SCM muscle may present a single contraction (Duchenne, 1949; Sousa et al., 1973).

The aim of this work was to determine the action of the belly, the sternal and clavicular parts of the SCM muscle during voluntary dynamic movements of the head and respiratory movements using a digital system and active surface electrodes.

METHODS

Ten volunteers (5 male and 5 female) aged 18-40 years were selected for this study. Three pairs of electrodes were positioned on the muscle belly, the sternal and clavicular parts of both SCM muscles (right and left sides). Each volunteer performed head dynamic movements (flexion, extension, rotation, inclination, rotation combined with inclination) and respiratory movements (normal and forced inspiration).

EMG signs were captured using active differential surface electrodes (Lynx Electronics Ltda, São Paulo, SP, Brazil) consisting of two silver parallel rectangular bars distanced of 10 mm (input impedance higher than 10 Gohm, CMRR higher than 90 dB and gain of 20 times). Recordings were made on 12-channel equipment of simultaneous EMG signal acquisition (Myosystem I, Datahominis Tec. Co.). The analog EMG signal was digitized using a 12 bit A/D converter at a sampling rate of 4 kHz. After digitalization, the signal was filtered by a digital pass-band of 10-500 Hz. Myosystem I software (version 2.22) was used to visualize and to process the EMG signal.

Statistical analysis was performed using the Kruskal-Wallis and the Dunn post-hoc tests to compare among the three parts of the SCM muscle in the same side and the Mann-Whitney test to compare between the two sides in the same part of the muscle. A value of $p < 0.05$ was considered significant.

RESULTS AND DISCUSSION

Table 1 shows that the SCM sternal and clavicular parts presented significantly higher electric activity than its muscle belly in the flexion movement for both sides while the clavicular part was the most active in the extension movement. These findings support that the SCM muscle is active in head flexion and extension (Testut and Latarjet, 1988; Williams et al., 1995).

In single rotation movement, the muscle belly and the sternal part showed the greatest activity, confirming that the SCM sternal part has a rotation component higher than its clavicular part (Duchenne, 1949). In inclination movement, the three SCM parts act equally because this muscle is likely responsible to this movement.

In combined movement (rotation with inclination) toward the right side, the SCM muscle belly and the sternal part of the left side showed significantly higher EMG activity than its clavicular part. The same was observed for the movement toward the left side. Previous reports also demonstrated that the SCM muscle acting unilaterally, performs head inclination toward the same side and rotation toward the opposite side (Testut and Latarjet, 1988; Williams et al., 1995).

In normal inspiration movement, the SCM sternal and clavicular parts had significantly higher electric activity than the muscle belly while in forced inspiration the clavicular part was the most active and the muscle belly was the least active. These findings disagreed from previous reports showing that the SCM, especially its sternal part, acts as a potential inspiration muscle (Testut and Latarjet, 1988).

SUMMARY/CONCLUSIONS

This study showed that there is individual contraction of the SCM muscle belly, the sternal and clavicular parts. Also, the SCM sternal part presents a rotation component higher than the clavicular part, and this latter is more active than the muscle belly and the sternal part in inspiration movements.

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Table 1: RMS (μV) values (mean ± SD) of the sternocleidomastoideus muscle of 10 volunteers during head dynamic movements.

Movement	RMS (μV) of Sternocleidomastoideus					
	Right			Left		
	Belly	Sternal	Clavicular	Belly	Sternal	Clavicular
Flexion	3.1 ± 1.3	$6.8 \pm 4.5^{\#}$	$7.1 \pm 4.9^{\#}$	3.4 ± 1.7	$7.8 \pm 5.3^{\#}$	$8.6 \pm 5.6^{\#}$
Extension	12.5 ± 9.4	$17.9 \pm 12.3^{\#}$	$22.9 \pm 14.3^{\# \psi}$	12.2 ± 8.7	14.9 ± 10.5	$22.7 \pm 12.1^{\# \psi}$
Rotation/right	$7.2 \pm 3.1^{*}$	$6.4 \pm 2.3^{*}$	$8.9 \pm 6.5^{* \psi}$	76.8 ± 48.2	67.4 ± 34.5	$32.1 \pm 24.2^{\# \psi}$
Rotation/left	$78.1 \pm 6.1^{*}$	$36.2 \pm 20.6^{*}$	$21.36 \pm 20.6^{*\# \psi}$	10.01 ± 8.9	6.4 ± 2.9	9.6 ± 6.8
Inclination/right	$31.8 \pm 21.6^{*}$	$31.2 \pm 24.7^{*}$	$24.9 \pm 13.5^{*}$	3.7 ± 2.1	4.1 ± 1.4	$5.9 \pm 2.8^{\# \psi}$
Inclination/left	$3.3 \pm 1.3^{*}$	$4.9 \pm 1.8^{*\#}$	$6.3 \pm 3.3^{*\#}$	21.5 ± 17.8	26.5 ± 25.4	$32.4 \pm 31.0^{\# \psi}$
Combined /right	$11.6 \pm 4.5^{*}$	$11.1 \pm 4.7^{*}$	$14.1 \pm 7.3^{*}$	127.9 ± 78.1	128.2 ± 81.5	$79.1 \pm 46.8^{\# \psi}$
Combined /left	$129.0 \pm 88.7^{*}$	$126.0 \pm 119.7^{*}$	$74.7 \pm 68.8^{*\# \psi}$	11.8 ± 8.7	17.4 ± 37.7	16.5 ± 19.4
Normal inspiration	2.1 ± 0.9	$3.8 \pm 1.3^{\#}$	$3.5 \pm 1.5^{\#}$	1.9 ± 0.5	$3.3 \pm 1.1^{\#}$	$4.0 \pm 1.2^{\# \psi}$
Forced inspiration	13.7 ± 9.8	$18.6 \pm 10.2^{\#}$	$44.7 \pm 27.5^{\# \psi}$	13.5 ± 12.5	20.4 ± 13.4	$37.1 \pm 20.8^{\# \psi}$

* $p < 0.05$ in relation to respective contralateral part; # $p < 0.05$ in relation to homolateral belly; ψ $p < 0.05$ in relation to homolateral sternal part.

FUNCTIONAL ANALYSIS OF THE ANTERIOR, MEDIAL AND POSTERIOR PARTS OF TEMPORAL MUSCLE DURING MOVEMENTS OF THE MANDIBLE – ELECTROMYOGRAPHIC STUDY

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INTRODUCTION

An electromyographic examination of three parts of the temporal muscle, when studied by means of surface electrodes, is hampered by the presence of hair in the region. Authors such as Petrovic & Horvat-Banic (2007) studied parts of this muscle using needle and wire electrodes. However, as commented by Basmajian & De Luca (1985), this type of examination involves a small area of signal capture and causes pain and discomfort, in addition to presenting difficulties in the fixation of the electrodes and other methodologic problems, which can lead to unsatisfactory results.

The correct understanding of the kinesiology of the three parts of the temporal muscle in mandibular movements which occur more than 2000 times a day, is fundamental for the diagnosis of temporomandibular dysfunction (TMD) which according to the American Academy of Orofacial Pain the prevalence of temporomandibular disfunction (TMD) in the American population is very high (Siqueira & Teixeira, 2001), in whom the alteration of the kinetics of the temporal muscle plays an important role (Berzin, 2004).

The objective of this work was to investigate the kinetics of three parts of the temporal muscle using surface electrodes during movements of the mandible.

METHODS

Twenty-one male subjects with normocclusion (Angle's class I) were studied; they were shaved before electromyographic examination of mandibular movements of lowering, elevation, propulsion, retropulsion, protrusion, and ipsilateral and contralateral retrusion. The subjects were examined in sitting position, with eyes open and the Frankfurt plane parallel to the floor. The surface electrodes utilized were from Noraxon USA Inc., model 27-2, connected to a Myosystem electromyograph from Datahommis Tec. Co., with 12 bytes resolution, CMRR of 112 dB at 60 Hz, sampling frequency of 2000 Hz, passband filter of 20-500 Hz. The EMG signal was calculated in RMS and was not normalized because it was not necessary for data comparison.

Statistical analysis was performed using the SAS system and was adopted a mixed model with repeated measures to compare means of the ranks of each part of temporal muscle in several movements. The use of the ranks were decided after the Shapiro-Wilk test showed a very weak adherence to a normal distribution. All tests were conducted using a significance level of 5%.

RESULTS AND DISCUSSION

The results show that the three parts of the temporal muscle act in all the mandibular movements studied. Each one of these parts displays different electrical potentials, confirming literature findings that do not consider the temporal muscle as the principal agonist in the elevation of the jaw but as stabilizer of its movements, acting not only as an agonist but also as an antagonist and stabilizer of the mandible. Depending on the movement, one or the other part of the muscle is more recruited, as can be observed in Table 1.

Table 1: Cluster of movement with same pattern of temporal part activation.

Cluster	Pattern	Movement
1	A<M and A<P	Slight drop Maximal drop Retropulsion Homo. laterality Contra. laterality
2	M>A and M>P	Protrusion
3	P>M and P>A	Retropulsion
4	A>P	MVC
5	M>P	Propulsion

SUMMARY/CONCLUSIONS

The electrical activity of the anterior, medial and posterior parts of the temporal muscle during different mandibular movements was studied by means of surface electrodes in 21 volunteers who were shaved. The temporal muscle, through its three parts, acts like puppet strings controlling and harmonizing the movements of the mandibular chondyle inside the mandibular cavity of the ATM.

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EFFECT OF THE APPLICATION OF AN EMBEDDED ENVIRONMENT ON THE MIGRATION OF THE COP DURING THE MAINTENANCE OF THE BIPED POSITION IN YOUNG WOMEN

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INTRODUCTION

The different motor strategies for the maintenance of the balance are represented by the path of the center of pressure (CoP) on the base of sustentation, the resultant oscillations are quantified in frequency and Amplitude (through posturo-graphy). Some authors have employed the different frequencies in which different subsystems deposit information from the exterior world. Values correspond to >0.1 Hz somatosensorial system, <0.5 Hz otoliths and between 0.5 - 1.0 Hz semicircular channels, whereas the human vision show discharge frequencies lower than 0.1 Hz. Embedded environment is understood to mean a non-inertial space related to certain subject, which allows to create and manipulate a virtual 3D environment, which in turn might effects the amplitudes of the different frequency bands studies. The aim of this study was to quantify the effects of the application, in young women, of an absorbed visually modified environment, registering in this manner the oscillations of the COP in different frequencies and extents with static posturography.

METHODS

Average Age	Average height	Average Body Mass	BMI
21.5 years (± 1.65)	1.6 m (± 0.05)	56.6 Kg (± 7.32)	21.7kg/m² (±4.16)

TableI: Random sample of 31 young healthy women.

A cubic environment was constructed with 15 m² of base area, with lateral walls (black color), in front of a white screen (projection zone) and a posterior wall in which there was fixed a projector of digital video. The posturographic machine was located inside the room with a view of the projection area, (A=2arctg (S/2D)), see Fig.I. A software of three-dimensional reconstruction 3D V. 11 demo was used to construct a digital 3D room, observed from the center, in first person and with an object in each of four walls (door, picture, window and photograp). The 3D video possessed 3 phases which were synchronized with the posturographic machine.

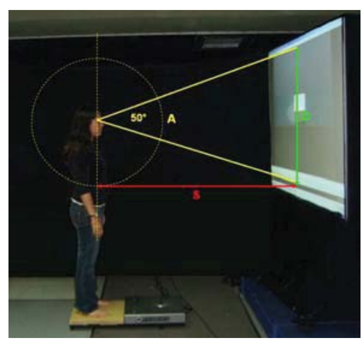
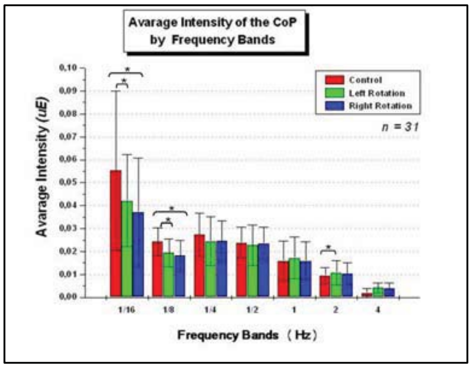


Figure I: Shows the basic conditions of the assessment.

The first phase was a fixed scene of the room (Static Phase and Control), whereas the phases of right and left rotation modified of the temporal frequency, by increasing the angular velocity of the rotation of the scene. For the comparison of the amplitude for bandwidth (1/16Hz, 1/8Hz, 1/4 Hz, 1/2 Hz, 1Hz, 2Hz, 4Hz) between the three conditions a T-Student test for paired data was used. The significant level was 95 %.

RESULTS AND DISCUSSION



Graphic I: Shows the frequency band of the assessment.

Statistically significant differences were observed in the group control - left rotation and control - right rotation for the bandwidths of the frequencie 1/16Hz and 1/8Hz simultaneously. (P <0,05). Probably, the inertial environment (Static Phase) generates a decrease of the visual feedback, which was translated in a major oscillation of the CoP and major amplitude of large part of the observed bands. It would point to a decrease in the of the neuronal potential, due to the fact that the neurons of V1, they triggered in relation certain stimuli of spatial orientation since they present different receptive fields, so in this phase there would be place a smaller number and frequency of triggered neurons. In a rotating environment the oscillations of the CoP diminished due to the increase of the visual feedback, this probably associates to the increase of the number and magnitude of the action potentials activated, in presence of greater available information. This increase in the feedback might be provoked by the major presence of saccadic movements, Fixation and Follow-up movements. With this the vestibular participation increases and the reflex tonic - ocular activity. Improving the postural responses. The bandwidth of the frequency 2Hz also presented significant differences, but only between the control group and left rotation (P<0,05). The propioceptive and somatosensorial subsystems would increase their participation before the new oscillatory conditions of the position. The brain would be visuospacially cheated, therefore this deficit should be compensated with sensory systems of high frequency. On the other hand, the vestibular system is kept relatively constant in three conditions, principally in the range of bands corresponding to the semicircular channels (1/2Hz-1Hz). Another factor to consider is the decrease of the bands of lower frequency in the rotations, with regard to the control group. We believe that it would be related with the motor dominance between both hemibodies, since 80,6 % of the studied population was right-handed and 19,35 % was left-handed and the observed intensities relate in their distribution. Though this was not analyzed quantitatively it might be associated with a major adjustment of certain movements of the eye in the predominant sense. This predominance would not be reflected in the bands of frequencies 1/4H, 1/2Hz, 1Hz, where significant differences were not observed. The motor dominance would not be relevant to maintain the biped position, of symmetrical characteristics.

CONCLUSION

We concluded that the facing of embedded virtual environment produces changes in postural control.

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THE VELOCITY OF THE CENTER OF PRESSURE FLUCTUATION DURING QUIET STANDING REFLECTS THE NEURAL MODULATION OF THE ANKLE TORQUE

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INTRODUCTION

Postural stability is reduced with age or neurological disorders. Such change can be evaluated by various statistical center of pressure (COP) measures during quiet standing. It is well known that the velocity of the COP fluctuation is more sensitive to the change in postural stability compared to the COP fluctuation itself (Prieto et al., 1996). However, the cause for this has never been investigated, let alone understood.

The COP fluctuation is proportional to the ankle torque fluctuation, which controls the body equilibrium during quiet standing. The ankle torque is decomposed into a passive/mechanical torque and an active/neural torque. Since most of total ankle torque is provided by the passive torque (Loram and Lakie, 2002; Casadio et al., 2003), this component alone should also be proportional to the COP fluctuation. We hypothesized that the fluctuation of the COP *velocity* reflects the fluctuation of the active torque, and therefore that the COP velocity is more sensitive to changes in postural stability induced by aging and neurological disorders. Thus, the purpose of this study was to test this hypothesis by decomposing the ankle torque into active and passive torque using a linear control model.

METHODS

Ten healthy male subjects (30.7 ± 4.1 yr) participated in this study, and three of them have been analyzed. Each subject was asked to maintain a quiet stance posture standing barefoot with eyes open. A force platform was used to measure the subject's ankle torque fluctuation. A laser sensor was used to measure the subject's anterior-posterior ankle angle fluctuation. The surface electromyogram (EMG) was measured from the right soleus to investigate the fitting of the model for the neuro-muscular system, but respective results are not presented here due to the limited space. All data were sampled at a sampling frequency of 1 kHz and low-pass filtered using a 4th order, zero phase-lag Butterworth filter with a cut-off frequency of 2 Hz.

For the torque decomposition, we adopted a feedback control model with linear controllers. The ankle torque was controlled by a mechanical and a neural controller, both of which were proportional and derivative controllers. While the output of the mechanical controller directly translated into the passive ankle torque, the output of the neural controller, represented by EMG, was delayed by a constant time delay and fed into the neuro-muscular system yielding the active torque. Using body angle data from quiet standing experiments, the feedback model was used to generate the controlled ankle torque data. The data were then compared with experimental ankle torque data and optimized by tuning the gains of the controllers (Kp and Kd for the neural proportional and derivative gain, K for the mechanical proportional gain) as well as the natural frequency (ω) of the 2nd order dynamics of the plantar flexors. The mechanical derivative gain was set to B = 5 Nms/rad.

RESULTS AND DISCUSSION

Fig. 1A shows an example of the torque decomposition. It reveals that the optimized torque (red line) matches the experimental torque (black line) well and that the larger portion of the optimized torque is generated by the passive (blue line), not the active torque (green line). Fig. 1B shows the derivatives of the optimized torque (red line), the passive torque (blue line), and the active torque (green line). It can be seen that the derivative of the active torque is about equivalent to the derivative of the passive torque and that both equally contribute to the derivative of the optimized torque.

In Table 1, the optimized variables are listed for each subject. In addition, the variance ratios of the active torque to the passive torque (var Tact/ var Tpass) and of the active torque derivative to the passive torque derivative (var dTact/ var dTpass) are shown. The results imply that the contribution of the active torque is much higher in terms of the variance of the torque variation (0.455-0.801) than in terms of the variance of the torque magnitude (0.184-0.269).

In summary, we demonstrated that the contribution of the active torque is higher to the *variation* in total ankle torque than to the *magnitude* of the total ankle torque. Therefore, the COP velocity rather than its position can be used to characterize the neural control mechanism of balance and to detect changes in postural stability induced by aging and neurological disorders.

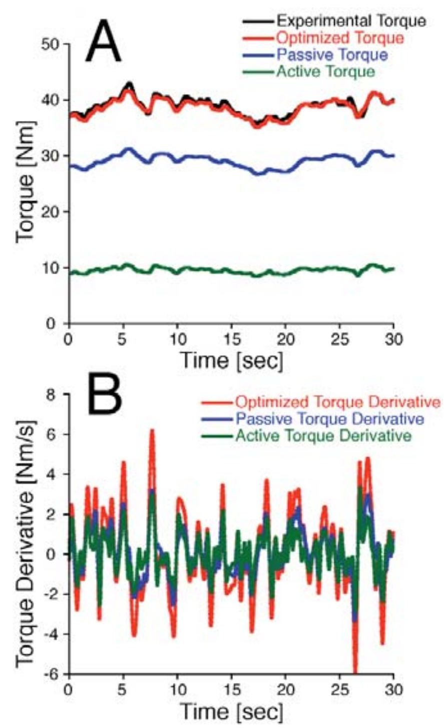

Figure 1: Results of the torque decomposition. See details in the text.

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Table 1: Optimization results and variance ratio. See details in the text.

Subject	Height [m]	Weight [kg]	Kp [Nm/rad]	Kd [Nms/rad]	K [Nm/rad]	Omega [1/s]	var Tact / var Tpass	var dTact / var dTpass
A	1.77	76.3	242	104	510	10.00	0.239	0.455
B	1.81	73.1	229	127	531	8.44	0.269	0.561
C	1.82	72.6	173	159	533	9.68	0.184	0.801

MODELING THE NEUROMUSCULAR CONTROL FOR HUMAN ERECT POSTURE

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INTRODUCTION

Recently, several strategies have been used seeking to elucidate the mechanisms responsible by the control of human movements and posture. Many of them with the sole objective of facilitating early diagnosis of diseases that affect the normal behaviour of the postural control system (e.g., Parkinson's disease, peripheral neuropathy, strokes, etc.). Generally, the main lines of action in the past few years have been: experiments with animals, experiments with humans and theoretical modeling and computational simulation. Each one of those strategies has its advantages and deficiencies. In studies with animals, for instance, the results should be analyzed in a way that assures us of their validity for humans as well, which sometimes is impossible due to the individual characteristics of each species. On the other hand, results obtained by researches involving humans can be limited by ethics and by the use of non-invasive methods. Theoretical modeling allied to computational simulation, emerges as an alternative to overcome those difficulties.

In this context, this work presents a new physical-mathematical model dedicated to the study of the control of human erect posture. The proposed model, in contrast with more traditional ones, includes the contribution of the extrafusal (EF) and intrafusal (IF) muscles, and the intrinsic reflex responses coming from three sensorial sources: muscular fuses, muscular spindles, Golgi tendon organs and Renshaw cells. In the same way as other models (Morasso and Sanguineti, 2002; Verdaasdonk et al., 2004), ours considers the body as a single link inverted pendulum, where the ankle is operated by a pair of antagonistic muscles, that is, by a single dorsiflexor (tibialis anterior) and a single plantarflexor (solius and gastrocnemius). The model considers that an angular deviation from the erect posture leads to a variation in the muscular length which is detected by the sensorial systems (muscular fuses and Golgi tendon organs). This sensorial information is then used to modulate the neural excitation of the muscles allowing us to generate corrective compensatory responses to disturbance torques.

METHODS

A simplified diagram of the proposed model is shown in Fig. 1.

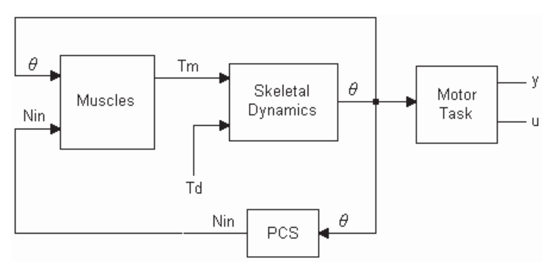

Figure 1: Block diagram of the proposed model.

The model has been developed based on the assumption that the postural control system (PCS) perceives any deviation (θ) of the body, that is different from the one associated to an erect posture of reference (θ = 0), and sends neural stimuli (Nin) to the muscles in order to generate the necessary muscular torque (Tm) to resist the deviation.

A linearized inverted pendulum model has been used to represent the dynamics of the body. The pendulum oscillations around the erect posture of reference are simulated by the insertion of an appropriate disturbance torque (Td). The positions of the center of gravity (y) and of the center of pressure (u), are then calculated as a function of θ.

The muscular forces were modeled based on muscle-reflex actuator suggested by Winters (1995a, b). To the model we added both the extrafusal and intrafusal muscular fibers. The extrafusal model is constituted by a contractile element (CE), a parallel element (PE), that represents the passive elasticity of the conjunctive tissue and a series element (SE), which considers the instantaneous response of the muscles and tendons to sudden changes of load. The intrafusal fibers are modeled in the same way as the extrafusal ones, except for the fact that they do not contribute for the muscular strength. We also consider that the afferences of the muscular spindle and Golgi tendon organs have a Gaussian response (Loeb, 1984) whose the maximum value (center) occurs when the muscle is resting (relaxed).

All the simulations were carried out using the Simulink software version 5.0 of Matlab® 6.5. The anthropometric parameters of the body were adopted for a typical adult male (Maurer and Peterka, 2005).

RESULTS AND DISCUSSION

The simulations show the consistency of the model in several aspects, such as, for example: (a) the generation of torque control is always aimed to canceling the disturbance torque; (b) the levels of muscular activation obtained in response to disturbance are very low, as expected for the maintenance of the erect posture; (c) the center of pressure and center of gravity oscillations are compatible with the ones observed experimentally in erect postures. In addition, the results allow us to conclude that the model is consistent with fundamental properties of the somatosensorial system and with the results obtained experimentally by other researchers that studied the erect posture (Fukuoka et al., 2001). The simulations also showed that the somatosensorial reflexes are extremely important for the maintenance of the erect posture and for supporting the body in the upright position. Note that such information cannot be obtained in purely EF models.

CONCLUSIONS

This paper presented a new physical-mathematical model of the neuromuscular control of human erect posture which includes the contribution of the extrafusal and intrafusal (IF) muscles and the intrinsic reflex responses coming from three sensorial sources: muscular fuses, muscular spindles, Golgi tendon organs and Renshaw cells. The results obtained by the simulations show that the behavior of the model is compatible with fundamental properties of the somatosensorial system.

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INFLUENCE OF VISUAL INPUT IN MOTOR STRATEGIES IN YOUNGS ADULTS

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INTRODUCTION

To maintain posture in relationship to space it's required that the CNS consider several factors, like visual, vestibular and somatosensory inputs. These are combined with internal and external influences such as expectation, attention, experience, environmental context and intention, thus the CNS determine the response in front of an environment change. (Forth KE, 2007).
Posture perturbations in bipedal stance are useful to obtain information about motor control exerted by the organism for maintain and recover balance. (Horak, F., et al., 1997)
The response of CNS to a posture perturbation is translated in muscle activation patterns and its associated trajectories called motor strategies. There are two principal types of strategies that have been described as a response for a perturbation. This are the ankle and hip strategies, they also could be combine and generate the called mixed strategy (Horak F.B., Nashner L.M. 1986). Previous studies pointed that in young males the more common used strategy is the ankle strategy (Runge, CF, et al. 1999).

METHODS

The data is acquired from male subjects ages between 18-30 years.
Are exclude subjects with central alterations or no controlled systemic damage.
All subjects gave their informed consent to the study which was approved by the local ethics committee of the Universidad Mayor.
Procedure: The subjects hold static bipedal position, isolated from auditive inputs, standing in a force platform with nine active markers creating the different body segments of foot, leg, thigh, pelvis and trunk.
The subjects are submitted to a forward perturbation by a pendulum under two conditions, opened eyes and closed eyes.
The perturbation was in relationship to a percentage of the subjects mass, thus the perturbation transmit a proportional energy to the mass of every subject. Each test consists in two perturbations for every condition.
Previous to the perturbation the subjects were asked to stay quietly over the platform to acquire the basal data for 60 seconds.
The registered data it's the surface EMG, (Delsys Bagnoli 8) from the right side of body, from muscles Tibialis anterior, Gastrocnemious medialis, Rectus Femoris, Biceps Femoris, Rectus Abdominal, Longissimus Thoracis. Other data are acquired from AMTI force platform and kinematic data acquired from two video cameras with APAS system. We used Igor Pro 5.01 software, (Wavemetrics) to data processing. All data was synchronized in relationship with the time of the perturbation.

RESULTS AND DISCUSSION

In processed data can be observed (figure 1) an ascending component in the X axis torque in relationship with a silence in the EMG activity especially in muscles related with knee and ankle.
The behavior in the descent part of X axis torque is different in the opened, closed eyes

conditions, where in opened eyes it can be observed a sinusoidal and extended response to return to basal magnitude, in closed eyes condition the response have less inflexions and is quicker in return to basal magnitudes of the X axis torque.
The EMG electric silence related with the ascending component of the X axis torque could be by a forward inertial component of de body mass. Also could be explained by the activation of the discharge reflex described by Feldman (Latash, M. 1998).
The different observed control strategies could be because in the first condition the subjects have environmental information about the location of they body segments in relationship to space to control their dynamic balance. However when the subjects are without visual input, they have to quickly search for their new dynamic balance after the perturbation. In other hand the morphology of the recovery phase of the x axis torque could be explained by the PDI (proportional, derivative and integrative) control mechanism in both conditions.

SUMMARY/CONCLUSIONS

1. - It was observed that in the stage of privation of visual input the time to recover the basal line of the torque in X axis was reached before than in the open eyes stage.
2. - The EMG response was higher in Gastrocnemius Medial in the same condition (Closed Eyes), in relation to Open Eyes.
3. - It was also observed that after the first burst of the EMG follow an "electromyographic silence" associated to an increase of the torque in X axis.

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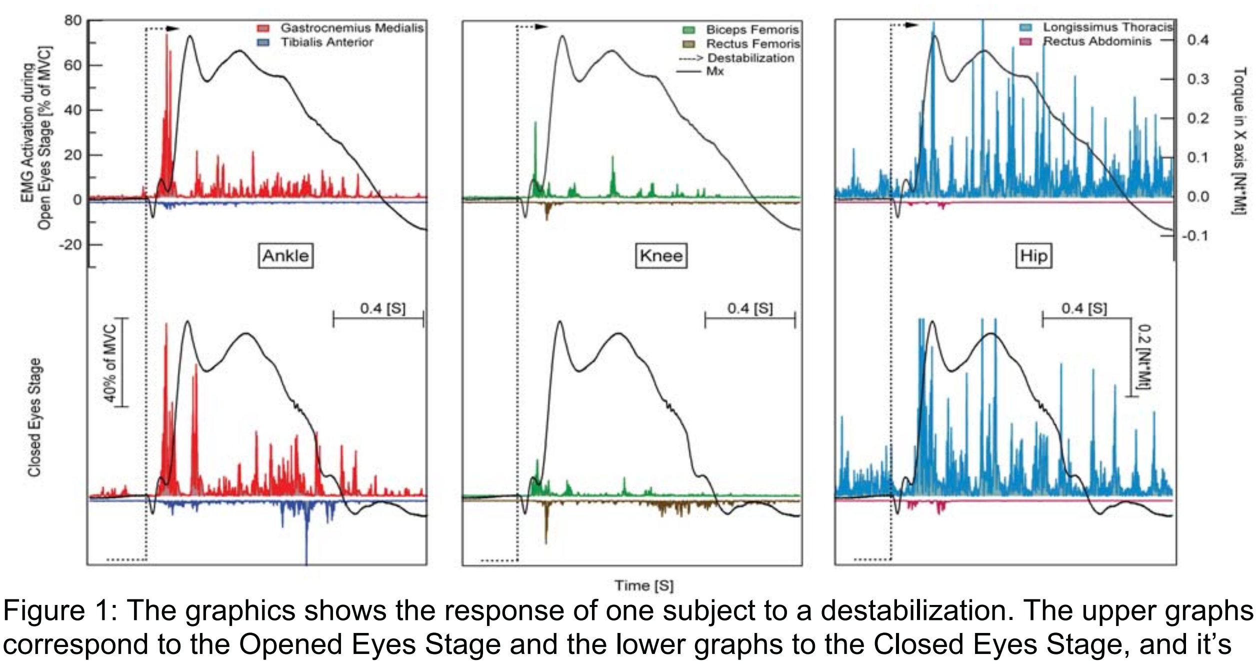

Figure 1: The graphics shows the response of one subject to a destabilization. The upper graphs correspond to the Opened Eyes Stage and the lower graphs to the Closed Eyes Stage, and it's divided according to the muscles involved to the strategies of every articulation (Ankle, Knee and Hip). The Left axis correspond to the SEMG response, normalized to the Percentage of MVC. The right axis represent the Torque response of the subject in the X axis in Nt*Mt.

INFLUENCE OF THE COMBINED USE OF IMPROVING STRENGTH OF THE MAXIMAL GLUTEAL MUSCLE AND STRETCHING OF THE GREATER PSOAS MUSCLE ON AN ANGLE OF INCLINATION OF PELVIS AND SWAY OF GRAVITY CENTER

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INTRODUCTION

We encounter clinical cases in which changes in an angle of inclination of pelvis have influence on the site of gravity center in a standing position and keeping balance in the position. Various investigators have reported variation of a gravity center line against an angle of inclination of pelvis and changes in sway of gravity center (postural sway) after improving muscular strength or stretching, which are involved with angles of inclination of pelvis. We investigated changes in an angle of inclination of pelvis and in postural sway by the combined use of exercise for improving strength of the maximal gluteal muscle, which has a retroverted action on pelvis, and stretching of the greater psoas muscle, which has an anteverted action on pelvis, in this study.

METHODS

The subjects were pivot feet of 19 healthy persons (7 men and 12 women, with a mean age of 30.4 years) whose consent about the experiment was obtained. The items to be measured were: An angle of inclination of pelvis (an angle made by a line made between ASIS and PSIS with a horizontal line in a standing position with feet closed) measured by means of a level (Shinwa Rules Co., Ltd.; Multilevel A150); Postural sway [measurement of the length of total locus, length of unit locus, the outer periphery area, and the rectangular area for 1 minute in a standing position with feet closed by means of a gravicorder GS-10 (Anima Corp.)]; and Strength of the maximal gluteal muscle measured [in a prone position with the knees in a flexed position at 90°by means of a hand-held dynamometer (Anima Corp.) attached to the distal end of the femoral bone, which was retained for 5 seconds in 3 occasions (the maximum value was adopted)]. With regard to intervention methods, the subjects were randomly assigned to two groups, A and B. In group A exercise for improving muscular strength was conducted (for 4 weeks: only exercise for improving strength of the maximal gluteal muscle), and in group B stretching was also conducted with the exercise [exercise for improving strength of the maximal gluteal muscle for the former 2 weeks and stretching of the greater psoas muscle was also conducted (in addition to the exercise) for the latter 2 weeks]. In both groups voluntary practice was performed once a day. With the method for exercise for improving the maximal gluteal muscle, Thera-Band was wound around the bilateral femurs in a prone position. The extended position was retained for 6 seconds in 20 occasions, and two sets of the maneuvers were conducted on each of the right and left femurs. With regard to stretching of the greater psoas muscle, 30-second self-stretching was performed in 3 occasions on each of the right and left femurs. Two and 4 weeks after the start of the measurements, the measurements similar to the initial ones were conducted, and two-way layout analysis of variance, multiple comparison (Bonferroni method), and Wilcoxon's rank sum test were conducted according to SPSS (statistical package for social sciences). The P value, at which differences were regarded as significant, was designated less than 5%.

RESULTS AND DISCUSSION

There was no significant difference in the mean angle [°] of inclination of pelvis among the occasions in group A, while retroversion was significantly recognized 4 weeks (15.1±5.0) after the start, as compared to the initial occasion (18.6±5.9) in group B. The mean strength of the maximal gluteal muscle (kg・m) was significantly increased 4 weeks (6.6±1.7) after the start, as compared to the initial occasion (5.3±2.4) and 2 weeks after the start (5.4±2.3) in group A. In group B as well, the mean muscular strength was significantly increased 4 weeks after the start (6.3±2.7), as compared to the initial occasion (5.4±2.7). There were no significant differences among the occasions in both groups regarding the items to be measured for postural sway.

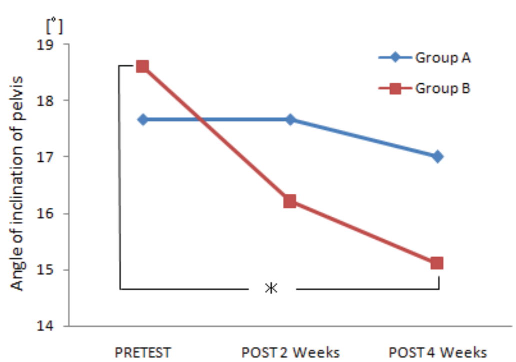

Figure 1: Mean angle of inclination of pelvis. *: P<0.05

The reason for the absence of change in the angle of inclination of pelvis in group A seems to be that exercise for improving strength of the maximal gluteal muscle with Thera-Band wound around both feet requires an immobilizing action on the foot on the opposite side to the hip joint during extension of the hip joint, leading to action of flexors of the hip joint and resulting not only in improving strength of the maximal gluteal muscle but in improving strength of the greater psoas muscle, which has an anteversed action on pelvis. It was considered in group B that an anteversed action on pelvis was attenuated by the combined use of stretching of the greater psoas muscle, resulting in significant retroversion of an angle of inclination of pelvis. The reason for the absence of change in postural sway in both groups seems to be that the gravity center line usually passed through the region in the vicinity of the hip joint in a static standing position to ensure lack of necessity of high muscular activity and no influence of the increased strength of the maximal gluteal muscle. It is also considered that center of gravity is maintained by strategy for each of hip joint, knee joint, and ankle and postural sway is not altered by hip joint strategy alone, such as change in the direction of retroversion of pelvis.

SUMMARY/CONCLUSIONS

The results of the present study suggest that the combined use of exercise for improving strength of the maximal gluteal muscle and stretching of the greater psoas muscle is useful for change in an angle of inclination of pelvis in the direction of retroversion and that the combined use is applicable to treatment for improvement of pelvic alignment.

QUANTIFICATION OF FORWARD HEAD POSTURE AND ANALYSIS OF STERNOCLEIDOMASTOID AND UPPER TRAPEZIUS PROPERTIES IN ASYMPTOMATIC YOUNG ADULTS

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INTRODUCTION

Discomfort and pain around head, neck and shoulder (HNS) regions have long been considered as consequences of a habituated faulty posture (Kisner & Colby, 2002; Szeto et al., 2002; Walker-Bone & Cooper, 2005). An anterior positioning of the cervical spine and the head, i.e. forward head posture (FHP), is one of the most common posture faults (Griegel-Morris et al., 1992). Patients with upper crossed syndrome have been described with characteristic posture of forward head and their HNS muscles such as sternocleido-mastoid (SCM) and upper trapezius (UT) tended to be tight and shortening (William & Michael, 2001). It's interesting to know whether the SCM and UT properties changed in asymptomatic subjects with FHP faults. In this study, we utilized forward head quantification, force measurement and surface EMG technique to study the connection between forward head and muscle strength, performance as well as muscle endurance during static sitting.

METHODS

Thirty-six young and asymptomatic adults (15♂/21♀) were included to evaluate their posture and muscle properties. For evaluation and analysis of FHP, lateral views of their HNS and trunk regions were captured with a digital camera during quiet and habitual sitting. As showed in Fig.1, the pasted marks including the external auditory meatus (EAM), the corner of the eye (CE) and the middle line of trunk (MLT) were identified and lined. The severity of forward head was quantified as the distance that EAM shifting from MLT and was normalized by dividing the distance of EAM-to-CE (% of his/her EAM-to-CE distance).

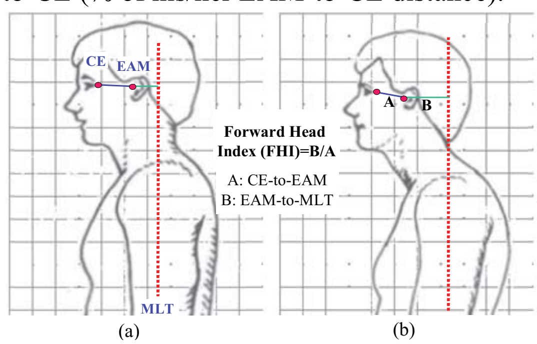

Figure 1: Quantification of FHP with forward head index (FHI) in a normal subject (a) and a subject with an obvious FHP (b).

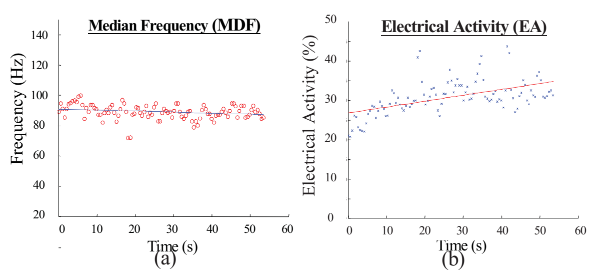

Figure 2: Muscle fatigue of L't SCM in one subject during one-minute $MVC_{30\%}$ contrac-tion. The downward trend of MF (a) and the upward trend of EA (b) indicated this muscle fatigued.

A force-recording device was developed to measure the maximal voluntary contraction (MVC) force of cervical flexion and shoulder elevations. Surface EMG of SCMs and UTs were then measured during three one-minute contractions of $MVC_{30\%}$ force (cervical flexion, right and left shoulder elevations, respectively). As showed in Fig.2, short-time (0.5s) median frequency (MF) and electrical activity (EA) during one-minute $MVC_{30\%}$ contraction were derived from EMG signal to observe if the muscle was fatigue during contraction. A temporal decrease in MF accompanied an increase in EA can be regarded as the result of muscle fatigue (Luttmann et al., 2000). Furthermore, MVC force and force performance error (error percentage during $MVC_{30\%}$ contraction) were used to represent the muscle performance properties.

RESULTS AND DISCUSSION

Thirty-six subjects were split into two groups according to their FHI (G1: FHI≤0.5, 11♀; G2: FHI>0.5, 15♂&10♀). As showed in Table 1, significant higher MVC force was noted in G2 (all $P<0.05$) and can be attributed to the lack of male subjects in G1. Force performance were generally better in shoulder elevation than in cervical flexion (all $P<0.05$) for both G1 and G2 subjects. Only cervical flexion performance of G2 was found significantly better than G1 ($P<0.05$). Further-more, muscle fatigue were less seen in SCM (1/11~8/25) but frequently found in UT (5/11~ 14/25) for all subjects. Fatigue frequency of SCM was significant higher in subjects with larger FHI (20%~32%) than subjects with smaller FHI (9.09%). However, there was no significant difference between groups in UT muscles. These results indicated a better muscle endurance in SCM of subjects with less forward head posture.

CONCLUSIONS

Unbalanced loading from FHP seems not alter the MVC force and performance accuracy of cervical flexion and shoulder elevation for asymptomatic young adults. However, SCM endurance was likely easy to be decreased in adults with more head protraction.

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Table 1: Summary of parameters: forward head index (FHI), maximal voluntary contraction (MVC) force and performance error of cervical flexion (F), right and left shoulder elevations (R and L). Number of fatigue for right SCM, left SCM, right UT and left UT were also included.

Parameters		FHI	MVC force (N)			Performance error (%)			Fatigue frequency (%)			
			F	R	L	F	R	L	RSCM	LSCM	RUT	LUT
G1 (N=11)	Mean	0.31	4.39	15.05	14.77	9.49	5.87	5.16	9.09% (1/11)	9.09% (1/11)	45.45% (5/11)	54.54% (6/11)
	SD	0.19	1.05	5.88	6.19	6.00	2.11	1.36				
G2 (N=25)	Mean	0.81	6.07	22.98	21.92	5.93	4.54	4.82	20.00% (5/25)	32.00% (8/25)	44.00% (11/25)	56.00% (14/25)
	SD	0.23	1.79	6.96	7.85	2.55	1.73	2.67				

INTENSIVE RETURN TO WORK REHABILITATION IMPROVES STABILIZATION PATTERNS IN WHIPLASH-ASSOCIATED DISORDER INDIVIDUALS.

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INTRODUCTION

A whiplash can be described as a fast movement of the head which occurs during the 100 to 200 ms following an impact in any direction (Sjostrom et al., 2003). About 10% of the victims of car accidents involving a rear impact develop a pathology, and 18 to 40% will develop chronic symptoms. Post-traumatic pathologies are generally grouped in a common class called whiplash-associated disorders (WAD). The mechanisms of whiplash and how they influence the systems involved in postural control are not well understood. The altered postural equilibrium observed in WAD individuals could result from disorders in one or more of the sensory-motor elements involved in maintaining balance.

Although postural reactions in seated healthy individuals have previously been described (Blouin et al., 2003), much less is known about how they are affected by neck pathology or if symptoms of postural impairment can be reversed following rehabilitation. The first objective of this study was to characterize kinematic and electromyographic postural stabilization patterns of individuals with chronic WAD and to compare these patterns with those of subjects in an able-bodied control group. The second objective was to evaluate the effects of a seven-week intensive return to work rehabilitation approach on postural stabilization patterns of WAD individuals.

METHODS

Ten individuals with chronic WAD aged 23-53 years and ten healthy controls aged 20-46 years participated in this study. Six WAD individuals repeated the experiment after an intensive return to work rehabilitation protocol, while five healthy individuals repeated the experiment after a seven-week period. The rehabilitation protocol lasted about seven weeks (5 days/week, 5.5 hours/day) and included treatment by a multidisciplinary clinical team (medical doctors, physical and occupational therapists, psychologists and kinesiologists).

Subjects sat on an adapted ergonomic chair firmly bolted to a moveable support surface. They were submitted to forward and backward translations (15 cm in 500 ms). A randomized sequence of 15 perturbation trials was administered, with 5 forward translations, 5 backward translations and 5 unperturbed trials. Motion of the head, arm and trunk segments was analyzed from 3D position coordinates of the markers placed on anatomical body landmarks. Reflective markers were also placed on each corner of the moveable platform. The activity of 16 muscles of the neck and trunk was acquired using bipolar surface electrodes: scalenus (SCA), sternocleidomastoid (SCM), cervical paraspinal (CP), upper trapezius (UT), erector spinae (thoracic level; TES), erector spinae (lumbar level; LES), rectus abdominis (RA), external obliques (EO).

Kinematic data was used to compute head and trunk angular displacements as well as abdomen, lower thorax, upper thorax, and head centers of mass (COMs). The COM of the head, arms and trunk (HAT) was computed from a combination of individual segments COM. Onsets were identified for angular and COM displacements as well as for EMG bursts. The amplitude and the time-to-peak of the first angular peak position were also computed. Data was then averaged over trials for each subject, and group averages were computed. T-tests and analysis of variance with Tukey post-hoc were performed ($p < 0.05$).

RESULTS AND DISCUSSION

In response to perturbations, in both groups, the trunk segment reached its first peak angular position earlier than the head. This caudo-cranial sequence was also reflected in the orderly displacement onsets of COMs. The HAT COM began moving before in the healthy group, in both directions of perturbation. Conversely, the angular displacement onsets and time-to-peak extension occurred earlier in the WAD group (forward perturbations only). This suggests two distinct initial response patterns characterized by a more linear, *en bloc* spine displacement in the direction of support surface travel in the healthy group, and an initial rotation of the spine in the WAD group.

Although there was a significant angle effect in peak angular displacement amplitudes, there was no significant group effect. Still, there was a significant angle x group interaction effect in forward perturbations, with the head-trunk peak extension difference being significant in the WAD group only. This suggests an impaired ability of WAD subjects to spatially distribute their response across the entire postural chain of the spine.

In the forward direction, WAD individuals recruited their SCM muscle later in comparison with healthy controls, with other flexors (SCA, EO) displaying the same trend. These findings are consistent with our observed kinematic pattern of the head moving earlier after the platform onset in WAD subjects. As neck extension is passively provoked by forward perturbations, the stretched neck flexor is activated later in the WAD group, delaying its action as a neck extension decelerator.

Following the intensive return to work rehabilitation, the time-to-peak head extension of WAD individuals occurred significantly later, so that there was no significant difference between the healthy and WAD groups at the second visit.

SUMMARY/CONCLUSIONS

Our results suggest that individuals with WAD present with altered timing of their postural response across the spine segments that is likely due to delayed muscle responses. They also suggest that intensive return to work rehabilitation is successful at improving postural reactions in WAD individuals. The results of our study highlight the importance of considering WAD as a pathology affecting many spine levels and of accounting for temporal characteristics of postural response patterns in studying and treating WAD individuals.

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EFFECT OF FATIGUE AND PROTECTIVE CLOTHING ON FUNCTIONAL BALANCE OF FIREFIGHTERS

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INTRODUCTION

This study investigated the effect of fatigue and load carriage on balance. Occupations such as structural firefighters (FFs) must perform hard work wearing personal protective equipment (PPE) consisting of specialized clothing and a self-contained breathing apparatus (SCBA) to protect themselves. However, the use of PPE can potentially hinder FFs due to increased heat stress and fatigue rate and impaired mobility. This study was designed to assess the effect of firefighting activity (FFA) and PPE on balance. Specifically, we developed a new functional balance test protocol, and investigated changes in balance due to PPE ensembles and 18 min of strenuous simulated FFA.

METHODS

Forty-nine male FFs (ages 18-50) participated in this study. Subjects were divided into control (n=29) and intervention (n=20) groups. The control group (standard PPE) wore PPE typically used by US firefighters, i.e., helmet, heavily insulated hood, bunker gear, and rubber boots. The intervention group (enhanced PPE) wore PPE designed with an industrial partner that included: 1) a lighter helmet, 2) more breathable hood, 3) bunker gear with reduced insulation, maximum breathability, and a passive cooling system to assist with heat transfer; and 4) lightweight leather/Kevlar boots. Both PPE ensembles meet current guidelines for thermal protection and breathability. Both groups wore identical SCBA packs and masks.

To assess the effect of PPE ensemble and FFA, subjects were evaluated at three testing periods: initially in normal clothing (baseline), before FFA in PPE (pre-activity), and after FFA in PPE (post-activity). FFA consisted of 18 min of alternating work-rest cycles that included four simulated firefighting activities in a burn facility that contained live fire. Post-activity balance testing occurred within 1-2 min after completing the simulated FFA.

To assess functional balance, the subject began on a raised platform (15 cm (H)), stepped down and walked on a narrow plank (3m (L), 15 cm (W), 4 cm (H)), stepped up and turned around within a defined space (61×61 cm^2) on a second raised platform (10 cm (H)), and walked back to stop within a defined space (61×61 cm^2) on the original platform. Subjects were instructed to perform the task as quickly as possible, but safely. The task was made more challenging by placing an overhead obstacle (a lightweight rod) across the center of the pathway and set at 75% subject height. The rod was designed to fall away if contacted directly. The subject performed 2 trials with no obstacle, 4 with the obstacle, and finally 2 no obstacle trials. Each trial was timed by two investigators. Average time per trial was penalized by 5s for a major error (rod fall) and 2s per minor error (touch rod or floor, failure to turn or stop within defined space).

Performance times were calculated by summing the four adjusted trial times per obstacle condition.

Three-way repeated-measures ANOVA tests were used to assess the effect of different PPE group, testing period, and obstacle presence on the performance time. Statistical analyses were run on SPSS (SPSS Inc., Chicago, IL; v15).

RESULTS AND DISCUSSION

Significant increases in performance times were found due to testing period ($p<0.001$), obstacle presence ($p<0.001$) and the interaction between the two ($p<0.001$), Figure 1. No significant differences in time were found due to PPE (standard versus enhanced), or any interactions with PPE gear ($p>0.05$).

Performance times increased significantly between baseline and pre-activity ($p<0.001$), but not from pre- to post-activity. These results suggest that the use of PPE (baseline to pre-) significantly impacted balance by increasing the performance time; however, 18 min of simulated FFA did not affect balance performance. Previous studies also found that PPE impairs FF balance performance (Kincl, 2002; Punakallio, 2003). We are not aware of other studies that have investigated the effect of FFA on balance.

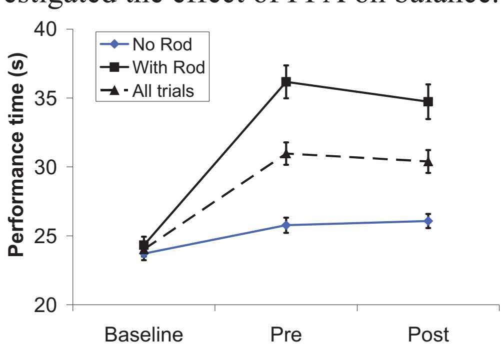

Fig 1. Mean time with standard error bars

Adding the challenge of walking under an obstacle reduced speed and increased errors. Performance time increased substantially with the addition of both the obstacle and PPE.

Although not statistically significant, performance scores for enhanced PPE declined slightly between pre- and post-activity, whereas standard PPE scores increased slightly (Figure 2). These results suggest that the enhanced gear may have some positive effect on balance performance. It is also possible that subjects were learning to move in enhanced gear during the study. The intervention gear is quite different from the traditional PPE, which is closer to the standard PPE ensemble.

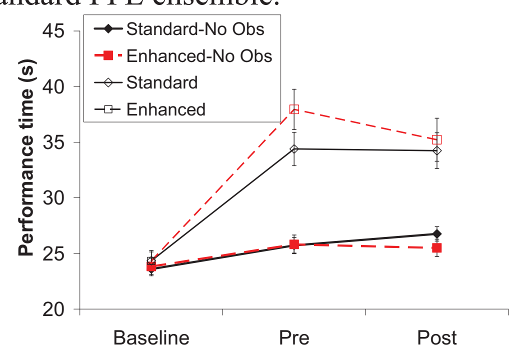

Figure 2. Gear effect with and without obstacle

SUMMARY/CONCLUSIONS

The effects of fatigue and personal protective equipment on balance performance were investigated. To assess balance performance, a new functional balance protocol was developed. Wearing any PPE was found to significantly impair balance performance. Balance performance times, however, were not significantly affected by a short bout of strenuous FFA or modified PPE.

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ELECTROMYOGRAPHY OF ORBICULARIS ORIS MUSCLE IN CHILDREN

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INTRODUCTION

Previous studies on the relationship between tongue-lip pressures and tooth position has shown that the lips and cheeks, rather than the tongue, are the most important environmental determinants of tooth position; a second finding was that pressure at rest and not functional pressure is the dominant factor as its effect is continuous.

The aim of this study was to evaluate the functional characteristics of the lower and upper fascicles of *orbicularis oris* muscle, by EMG activity, in children with anterior open bite at the mixed dentition stage, comparing to children with normal occlusion, at rest, during contraction, and while suction of liquids with different consistencies.

METHODS

The local Research Ethical Committee approved the project. Ninety-eight Brazilian children from the patient files of the Piracicaba Dental School were recruited and examined, thirty were selected and distributed in two groups: Group Open bite (OB) – 15 children with anterior open bite (11 dental, 4 skeletal), composed of 9 boys and 6 girls (mean age 9.0±1.0 years) and Group normal occlusion (Noccl) NO - 15 children with normal occlusion, composed of 9 boys and 5 girls (mean age 9.4±1.1 years). The anterior open bite was the consequence of non-nutritive sucking habits for a long period (until the end of primary dentition), atypical swallowing, tongue trusting, mild mouth breathing, and/or disharmony in the skeletal morphological pattern. The groups were submitted to EMG examination of the orbicularis oris muscle at rest and contracted state, and during suction of water and yogurt (conditions involving lip participation).

EMG evaluation: It was used a 16-channel signal conditioner (MCS - V2 from Lynx Eletrônica Ltda São Paulo, Brazil*)* with 12-byte dynamic band resolution, Butterworth–type band pass filter set at 509 – 10.6Hz, gain of 600 times, 1 KHz sampling frequency, A/D converter board (model CAD 12/36, *Lynx Eletr nica Ltda,* SP, Brazil). *Beckman* Ag-AgCl, bipolar disc surface electrodes, with 3 mm of detection surface diameter, were used. The interelectrode distance was 10 mm. The software *Aqdados*, version 4.18 (*Lynx Eletr nica Ltda*, SP, Brazil) displaying several raw and conditioned signals simultaneouslywas used.

Procedures: The child sat with a straight back and head oriented with the Frankfort's plane parallel to the floor. Both the skin and the electrodes were cleaned with 70 percent GL ethyl alcohol. The ground electrode was fixed to the right wrist. One pair of the electrodes was fixed at 2 mm above the free edge of the upper lip, and another pair at 2 mm below the free edge of the lower lip.

The EMG signal acquisition was performed with the muscle relaxed (at rest) and in maximum isometric contraction of closing the mouth and contracting the lips (contracted state). The acquisition time for each evaluation, in both positions, was 5 s.

The muscle activity while sucking water and yogurt using a straw was executed using 200 ml plastic glasses and 6.5 mm diameter straws, during 5 s. The sample order was water-yogurt and the signals collection was repeated three times for each liquid, and the averages were considered.

Data analysis: Signals were processed by Matlab software routines (Version 5.0 The MathWorks, Natick, Mass) The absolute EMG signal amplitude values (μV) were normalized with respect to the values obtained in the isometric contraction.

Statistical analysis: Paired, unpaired *t* tests, Mann-Whitney *U* test, ANOVA and Tukey test as *post-hoc* were used when appropriate. The integrated EMG activity of upper and lower fascicles was also computed and analyzed. SPSS 13.0 (SPSS, Chicago, IL) with significant level at $p<0.05$ was used.

RESULTS/DISCUSSION

The orbicularis oris generated slight electrical activity at rest and the Group OB showed significantly more activity in the upper fascicle, probably due to the malocclusion characteristics, supporting that the resting pressure is the dominant factor in the tooth position and arch form. There was no difference between groups in the contracted state. There was greater EMG activity for the lower fascicle than the upper in Group Noccl on contraction, and this could be explained by the fact that the lower lip depends on the lower orbicularis oris and the mentalis muscles, the latter placed below the orbicularis oris, providing mechanical advantage to the lower fascicle during the movement. Moreover, the lower fascicle assumes greater kinetic demand on closing lips. The lack of difference in potential action between the fascicles in Group OB, in the contracted state, was probably due to the postural position of the upper lip, which must have recruited more motor units to perform the contraction, generating the same magnitude as the lower lip.

Besides, there was no difference among contracted state and sucking liquids between groups, comparing the two fascicles or their integrated activity.

The orbicularis oris activity was not so influenced by the difference in consistency of the two liquids, but it was found a tendency of higher levels of activity for yogurt than water in the integrated activity in Group OB, but these were not significantly higher. Moreover, the subjects were free to suck any amount of liquid and because of the higher consistency of yogurt; the Group OB could have sucked a smaller amount of yogurt than water. Larger volumes of liquid require active engagement of the labial muscles to prevent oral spillage of liquids prior to initiation of the oral swallow, and the bigger volume of water could have increased the activity reaching about the same values of yogurt.

CONCLUSIONS

The electrical activity of the *orbicularis oris* could be altered in presence of open bite and the sucking ability seems not to be decreased by this malocclusion in the studied sample.

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RATE OF FORCE DEVELOPMENT AND CO-ACTIVATION AROUND THE KNEE IN ELDERLY FALLERS

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INTRODUCTION

The falls in people over 85 years are associated with the raise of mortality and morbidity and the reduction of functionality, being the main cause of death (Lee and Chou, 2006). After the 60th year of life, either the incidence as the severity of falls are related to age, and 35-40% of these subjects fall yearly (American geriatrics society, British geriatrics society and American academy of orthopedic surgeons panel on falls prevention, 2001). Some aspects related to the fall risk increase are the reduction of quadriceps force and increase of hamstrings activity, generating an excessive co-activation around the knee, the lower capacity of a rapid force generation, and lower motor units recruitment, that had being investigated using the electromyography (EMG). Therefore, the main objective of this work was to correlate the fall risk and the EMG ratio with the rate of force development during knee isometric flexion and extension in elderly fallers.

METHODS

Seven elderly women (74.86$\pm$7.88 years, 151.43$\pm$5.50cm height and 58.46$\pm$9.55kg weight) considered as fallers (score less or equal than 46 in Berg Balance Scale [BBS] – Lajoie and Gallagher, 2004) agreed in participate in this study and signed an inform consent previously approved by the Local Ethics Committee. They were excluded if presented any cardiac, orthopedic, neurological, vestibular or other conditions that incapacitated them to do the tests proposed. The volunteers executed 3 maximal knee flexion and extension isometric contractions (3 second length and 5-minutes rest), in random order at 90 degree flexion position. The best trial was analyzed. Bipolar Ag/AgCl electrodes were positioned over the vastus-lateralis (VL) and biceps femoris (BF) after shaving, abrasion and skin clean according to the SENIAM project. Muscle activity was recorded at 1,000 Hz with a four-channel telemetric system (Telemyo 900 - Noraxon®) with a band-width from 10 to 500 Hz. The RMS values (obtained from the second that presented less variation in the load cell data) from the BF and VL during the flexion and extension contraction respectively, were normalized to the peak value and were used to calculate the BF/VL EMG ratio. Also, the rate of force development (RFD) was determined as the slope between the torque data and the time, after the torque raised above 10N/m for the first 50ms, 100ms and 200ms of contraction (RFD50, RFD100 and RFD200 respectively). The BBS score, and the BF/VL ratio were correlated with RFD50, RFD100 and RFD200 by the Spearman's Correlation test and statistical significance were considerate when $p<0.05$

RESULTS AND DISCUSSION

The mean BBS score was 44.14$\pm$1.57 (arbitrary units – a.u.). The mean BF/VL ratio was 1.10$\pm$0.79 a.u. and the RFD were 9.83$\pm$6.96 Nm/seg, 9.32$\pm$7.21 Nm/seg and

8.55±7.26 Nm/seg during the first 50, 100 and 200ms of knee flexion contraction respectively. Therefore, the mean RFD was 3.13±1.81Nm/seg, 2.72±1.46Nm/seg and 2.32±1.10Nm/seg during the first 50, 100 and 200ms of knee extension contraction respectively. Through the observation of the RFD results it could be observed that the capacity to quickly generate knee extension force is less than the capacity to generate knee flexion force rapidly, probably due to the reduction of type I fibers during aging observed in the quadriceps muscles (Sadeghi et al., 2004). This is also evident when the BF/VL ratio is observed, since the activity of BF was greater than the VL (ratio greater than 1). According to Sadeghi et al. (2004) the co- activation around the knee in elderly could be the result of a greater hamstring activity, supplying the lost of force of the knee extensors muscles, and can result in an abnormal movement pattern, or an elevated energy expenditure (Mian et al., 2006). Also, according to Aagaard et al. (2002), the lower ability to generate force in a small period of time (represented as lower RFD) could represent a risk of fall, since it represents the incapacity to recovery quickly from a drop or a slip. These aspects can be confirmed according to table 1, since a significant negative correlation was observed with the BF/VL ratio and the RFD100 during extension and RFD50 during flexion and extension, what represents that the higher BF/VL ratio the greater activity of BF) capacity of generate force rapidly became reduced

Beside the greater correlation found between the BBS score with others variables (RFD and BF/VL ratio), the absence of significant correlation could be explained by the notion that this scale involves others aspects than the neuromuscular ones, and the neuromuscular and kinesiology variables analyzed in this study are very specific.

Table 1: Spearman´s correlation coefficients between the BBS (a.u.) and BF/VL ratio (a.u.) and the RFD50, RFD100 and RFD200 (Nm/seg) during flexion and extension contractions. *p<0.05; **p<0.01

	Flexion			Extension		
	RFD50	RFD100	RFD200	RFD50	RFD100	RFD200
BBS	0.764	-0.400	-0.600	-0.709	-0.655	-0.654
BF/VL ratio	-0.643	-0.714*	-0.571	-0.929**	-0.857*	-0.714

CONCLUSIONS

It can be conclude that the evident activity of the hamstrings in elderly fallers is highly correlated to lower values (RFD), representing part of the causes of falls in this population.

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Track 08

Muscle Fatigue (MF)

TIME COURSE OF NEUROMUSCULAR CHANGES DURING HUMAN TRICEPS SURAE LOW-FREQUENCY ELECTRICAL STIMULATION

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INTRODUCTION

In recent years, considerable research has focused on the benefits of Neuromuscular Electrical Stimulation (NMES) in muscle strengthening and functional rehabilitation programs. The muscle fiber recruitment patterns of NMES differ from those of voluntary muscle activation, causing greater metabolic stress and thus a more rapid onset of fatigue.

Fatigue is a complex and multifactorial phenomenon that appears at several sites from brain to muscle. It involves diminished central drive to the motoneurons, decreased motoneuronal excitability, and altered muscle excitability at a peripheral level. Recent studies that investigated fatigue-induced neuromuscular reorganizations (Boerio et al, 2005; Zory et al, 2005) used NMES sessions with high stimulation frequencies (80 Hz) devoted to muscle strengthening. These frequencies provoked high fatigue, however, and no information was given on the time courses of the observed changes.

The purpose of the present work was thus to examine the time course of neuromuscular modifications induced by a NMES protocol with stimulation parameters minimizing fatigue (frequency 30 Hz).

METHODS

Nineteen healthy subjects participated in an electrostimulation protocol for the triceps surae. The protocol was composed of 3 series of 17 stimulation trains (4 s on - 6 s off, pulse duration 450 μs, frequency 30 Hz, at a maximal tolerated intensity). Neuromuscular tests were performed before, during (at the end of every 17-train bout) and immediately after the protocol. Torque and EMG activity of the gastrocnemius medialis muscle were continuously recorded (sample frequency 4096 Hz).

Alterations in the muscle's characteristics (excitability and contractile properties) were evaluated by neurostimulation at a supramaximal intensity and analysis of the subsequent muscle compound action potential (M-wave) and twitch torque. Motoneuronal excitability was assessed by the H reflex (H_{max}), obtained with neurostimulation at submaximal intensity, whose amplitude was then normalized to the M-wave maximal amplitude (H/M). In order to identify changes in the central command, the twitch interpolation technique was used and the root mean square (RMS and RMS/M) was calculated during maximal voluntary contraction (MVC). A one-way ANOVA test with repeated measures was used for statistical analysis.

RESULTS AND DISCUSSION

Neuromuscular fatigue was attested by a significant decrease in maximal voluntary torque production from the first 17-train bout (- 6.6%, $P<0.001$). The decrease persisted throughout the stimulation protocol

(-10.32% and -11.53% at post34 and post51, respectively).

This fatigue was not due to peripheral factors since muscle excitability was preserved, as indicated by the absence of any significant changes in M-wave amplitude or duration, while muscle contractile characteristics were significantly improved from the early phase of the protocol (Table 1). In contrast, significant decreases in RMS and RMS/M values were observed (-11.19%, -18.44% and -17.95% for RMS/M at post17, post34 and post51, respectively, $P<0.001$). These decreases indicated a reduced descending influx to the motoneurons and/or an inhibition of motoneurons excitability. The non-significant changes in H_{max} amplitude and H/M ratio throughout the protocol were evidence of preserved motoneuronal excitability during the NMES protocol. A reduced central command to the motoneuronal pool was thus assumed to be the origin of the decreased RMS and RMS/M values. Activation of III and IV afferents could have led to an impairment in the descending cortical signal (Taylor et al, 2006), while nociceptive afferents from the skin would have acted directly on the brain and thus reduced the cortical command.

SUMMARY/CONCLUSIONS

This study provided evidence of reduced corticospinal influx responsible for a decrease in voluntary torque production induced by low-frequency NMES. The results indicate the implication of the central nervous system in electrically-induced fatigue, which is highly consistent with the not purely peripheral character of this technique.

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	pre	**post17**	**post34**	**post51**	***F***	***P***
Pt (Nm)	12.32±0.58	17.1±0.97***	16.15±0.91***	16.33±0.97***	31.87	0.001
CT (ms)	118±7.2	95±4.3***	95±3.8***	100±4.3**	7.23	0.001
MRFD (Nm/s)	0.34±0.02	0.49±0.03***	0.47±0.02***	0.47±0.02***	43.53	0.001
HRT (ms)	96±4.7	86±4.6*	84±4.5**	82±4.6***	5.06	0.01
MRFR (Nm/s)	0.18±0.01	0.22±0.01***	0.21±0.01***	0.22±0.02***	9.58	0.001

Table 2: Contractile properties before (pre), during (post17 and post34) and after (post51) the exercise protocol. Pt peak torque, CT contraction time, MRFD maximum rate of force development, HRT half-relaxation time, MRFR maximum rate of force relaxation; *$P < 0.05$, ** $P < 0.01$, ***$P < 0.001$ significantly different from pre values.

COMPARISON OF MUSCLE FATIGUE INDICES DURING DYNAMIC MUSCLE CONTRACTIONS

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INTRODUCTION

This paper presents a comparison of indices of muscle fatigue derived from the surface myoelectric signal (MES) of human participants under static (i.e. isometric) and dynamic (i.e. where muscle force and joint angle are time varying) conditions.

The decrease in action potential conduction velocity (CV) due to fatigue causes a spreading of the MES in time which corresponds to a spectral compression towards lower frequencies. (DeLuca, 1984) Thus, power spectral parameters such as mean frequency (MF) or median frequency have been heralded as the gold standard for fatigue estimation.

A multi-variable approach known as the fatigue mapping index (MI) trains an artificial neural-network (ANN) to assess fatigue based on four time-domain features (MacIsaac, 2006). While this approach requires subject-specific training data for each estimate, it was shown that an ANN trained from the data of other subjects performs as well (Rogers, 2007). This approach is known as the generalized mapping index (GMI).

A general power spectrum method (GPSM) was introduced based on the fractal properties of the MES (Talebinejad, 2007). GPSM models the MES power spectrum according to equation 1, where c is a scaling constant, f_0 is the characteristic frequency (CF), g is the low frequency parameter and q is the high frequency parameter.

$$\hat{P}(f) = \frac{c\left(f/f_0\right)^{2g}}{\left[\left(f/f_0\right)^2 + 1\right]^{q+g}} \tag{1}$$

(Talebinejad, 2007) showed that CF is sensitive to CV and is therefore a potential index of fatigue. This work compared CF to MF and GMI under static and dynamic conditions.

METHODS

Two sets of data (training and test) were collected from the right biceps of 10 human participants under static, cyclic and random conditions according to the protocol described in (Rogers, 2007). An adhesive eight-channel Ag-AgCl electrode strip with 5mm spacing was placed approximately parallel to the muscle fibers. Seven channels of single-differential MES were collected using the acquisition system described in (Martin, 2006).

The data was bandpass filtered between 10 and 300 Hz and segmented into five second windows. Each segment was characterized by the time-domain features described in (Hudgins, 1993). The entire training set was used to train a single multi-layer perceptron (MLP) according to (Rogers, 2007). GMI fatigue estimates were generated by simulating the trained network using the features extracted from the test data set.

The power spectrum of each segment was estimated using Welch's averaged periodogram method using 0.5 second 50%

overlapped Hamming windows. MF was calculated directly and GPSM parameters were calculated using an iterative least-squares algorithm.

GMI, MF and CF fatigue estimates were normalized such that the line of best fit, $\hat{s}(t)$, spanned the range [0, 1]. Estimates were evaluated using the signal to noise ratio (SNR) given by equation 2.

$$SNR = \frac{R}{E_{RMS}} = \frac{\max(\hat{s}(t)) - \min(\hat{s}(t))}{\sqrt{\frac{1}{N}\sum_{i=1}^{N}\left(\hat{s}(i) - s(i)\right)^2}} \qquad (2)$$

RESULTS

The SNR for each index was averaged across all subjects as shown in Figure 1. A Bland & Altman mean bias 95% confidence interval was evaluated and tabulated in Table 1 (statistical differences were shaded for clarity).

According to Table 1, GMI was better than MF and CF under all conditions and MF was better than CF only for random contractions.

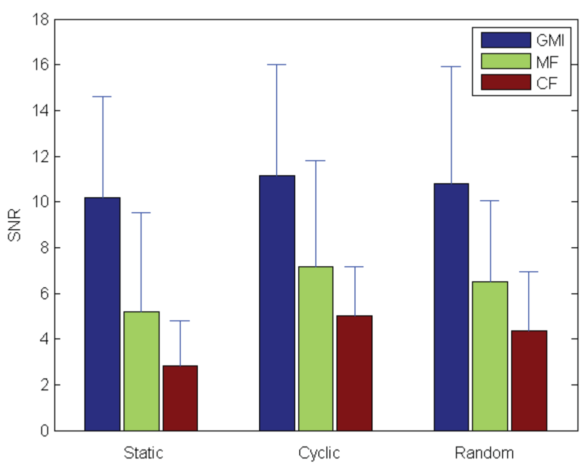

Figure 1: Mean SNR across all subjects under static, cyclic and random contraction conditions

Table 1: Upper and lower bounds for Bland & Altman mean bias 95% confidence interval.

Contraction Conditions	Mean bias 95% confidence interval			
	Indices compared		*Lower bound*	*Upper bound*
Static	GMI	MF	0.643	6.25
	GMI	CF	1.42	8.76
	MF	CF	-0.357	3.64
Cyclic	GMI	MF	0.923	5.22
	GMI	CF	1.73	7.73
	MF	CF	-0.415	3.73
Random	GMI	MF	0.830	5.79
	GMI	CF	1.75	8.13
	MF	CF	0.544	2.71

CONCLUSIONS

It was shown that GMI outperforms both MF and CF under static, cyclic and random contraction conditions. While CF is not a robust index of fatigue on its own, future work will examine the use of GPSM features in conjunction with MI and GMI.

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EFFECT OF MOTOR UNIT FIRING RATE ON THE ACCUMULATION OF EXTRACELLULAR POTASSIUM AND MUSCLE FIBER CONDUCTION VELOCITY

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INTRODUCTION

During sustained fatiguing muscle contractions, a progressive decrease in muscle fiber conduction velocity (MFCV) is observed. This reduction in MFCV forms the basis of commonly used indices of muscle fatigue.

Recent studies have indicated that the reduction in MFCV may be primarily due to changes in ionic concentrations, in particular the accumulation of extracellular potassium (K^+) ions. K^+ ions are released during each action potential, and accumulate extracellularly in both the interstitial space and the transverse tubular system during sustained muscle contraction. The accumulation of extracellular K^+ ions causes a depolarization of the transmembrane potential and a reduction in membrane excitability. To stabilize the transmembrane potential, the extracellular K^+ ions are transported inwards by both the Na^+- K^+ pump and the inward rectifier K^+ current (Sejersted and Sjøgaard, 2000).

At higher force levels, the rate of fatigue and MFCV decay is increased. This may be due to the faster accumulation of extracellular K^+ due to the greater number of action potentials elicited at higher firing rates. The aim of this study was to examine the effect of motor unit (MU) firing rate on the accumulation of extracellular K^+ and on the transmembrane action potential and MFCV, using model simulations.

METHODS

A model of the propagating muscle fiber action potential was developed, based on previous muscle models and using values for mouse skeletal muscle at 20 ºC (Wallinga et al., 1999). The model consists of both the surface membrane and the tubular system and accounts for the accumulation of K^+ ions in extracellular space and the corresponding loss of intracellular K^+ ions.

The model of the surface membrane incorporates the membrane capacitive current, I_C, the Na^+ current, I_{Na}, delayed and inward rectifier K^+ currents, I_{DR} and I_{IR}, the Cl^- current, I_{Cl}, and the Na^+- K^+ pump current, I_{NaK}. The sodium and delayed rectifier potassium currents are described as in the classic Hodgkin-Huxley model, with the addition of their slow inactivation processes. At each node, the tubular membrane is represented by 16 concentric compartments and includes all of the channels described in the surface membrane model, with the conductances scaled according to the ratio of channel density between the tubular and surface membranes.

The model was simulated during repetitive firing of the MU at a range of different firing rates, to examine the effect of firing rate on MFCV and K^+ accumulation.

RESULTS AND DISCUSSION

MFCV decreased as the extracellular K^+ ions accumulated in both the interstitial space and the tubular-system. As the firing rate was increased, the rate of extracellular K^+ accumulation and of MFCV decrease was observed to increase, Figure 1.

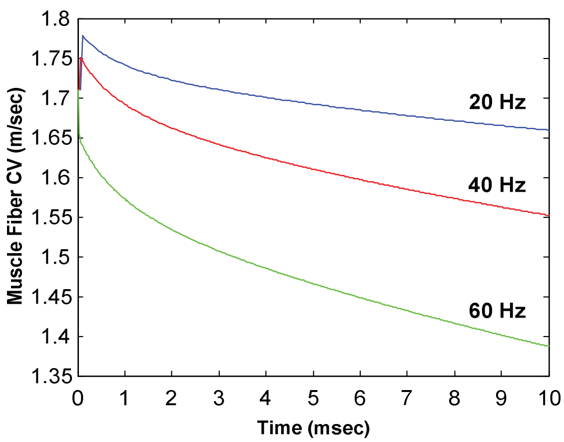

Figure 1: The effect of extracellular K^+ ion accumulation on MFCV for motor unit firing rates of 20, 40 and 60 Hz at 20 ºC.

In addition to the effect of the MU firing rate on K^+ accumulation and hence MFCV, variations in the instantaneous firing rate also influenced the CV of the second action potential in the MU action potential train. The ratio of the action potential CV of the second action potential (CV_2) of the train to the first (CV_1) was found to be dependent on the inter-pulse interval, Figure 2. A similar variation in MFCV with inter-pulse interval, at higher frequencies, has been shown experimentally (Mihelin et al., 1991). However, the underlying mechanisms for this dependency are unknown.

Blocking the tubular current was found to considerably reduce the change in membrane potential and the resulting variation in CV, suggesting that the dependency of CV on instantaneous firing rate may be due to the contribution of the tubular current to the membrane potential.

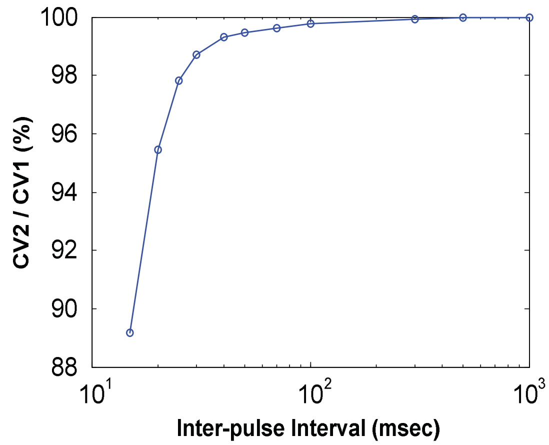

Figure 2: The effect of inter-pulse interval on the ratio of CV_2 to CV_1.

CONCLUSIONS

The results of this study illustrate an increasing rate of extracellular K^+ accumulation and MFCV decrease as MU firing rate increases. The results also indicate that the ratio of the CV of the second action potential to the first action potential is dependent on the firing rate. The simulation results suggest that this dependency of CV on the instantaneous firing rate is due the contribution of the tubular current to the membrane potential.

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NOVEL FRACTAL INDICATORS WITH DISTINCT SENSITIVITIES TO LOCALIZED MUSCULAR FATIGUE DURING STATIC CONTRACTIONS

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INTRODUCTION

This paper presents a set of novel fractal indicators (FIs) with unique properties in sensing localized muscular fatigue. These FIs are obtained by myoelectric signal (MES) modeling using a fractional-order differential equation with characteristics, allowing it to provide an accurate approximation for MESs. We have shown that the FIs computed using this methodology, namely the general power spectrum method (GPSM), are capable of separately sensing the force and joint angle effects during isotonic, isometric contractions of *biceps* (Talebinejad, 2007a). This unique property is based on the asymptotic behavior of the proposed fractional power-law in GPSM. Moreover, homomorphism of the model results into power spectral separation of FIs and makes them insensitive to spectral compression which results into the insensitivity of FIs to the effects of decreased muscle fiber conduction velocity (CV) in the form of compression of frequency content towards lower frequencies during localized muscle fatigue (De Luca, 1984; Talebinejad, 2007b). The asymptotic behavior of the fractional power-law is governed by a critical exponent, termed the characteristic frequency (CF), in which the power-law asymptotes are changed. It was also shown using simulated MESs that the CF is sensitive to changes of muscle fiber CV, meanwhile insensitive to changes of motor unit (MU) recruitment strategies and shape. In this paper we evaluate previous hypothesis based on simulated data with human experimentation.

METHODS

The data used in this experiment were collected from the right *biceps* of 10 human participants under static conditions according to the protocol described in (Rogers, 2007). Preprocessing involved segmenting MESs to 4 portions each of which, containing 25 % of the total physical time of exhaustion (80 to150 sec). The power spectrum estimation was done for each individual portion using the Welch's method with a sliding window with 1s temporal width and 50 % overlap. The GPSM was then applied to the estimated power spectrum to compute the FIs. For GPSM the modeling frequency interval was between 10 to 300 Hz. For comparison the median frequency (MDN) was also computed in this frequency range from the Welch's estimated power spectrum. Numerical evaluation was performed by defining a signal to noise ratio (SNR) assuming the ideal noise-free parameters are linearly decreasing from 1 to 0 during the total exhaustion time in this form,

$$SNR = N(\hat{\theta}) / \sigma\left(\theta, \hat{\theta}\right) \qquad (1)$$

In Eq.(1), $N(\hat{\theta})$ is a normalized estimated parameter set between 1 to 0 and $\sigma\left(\theta, \hat{\theta}\right)$ is the deviation between the estimated parameters and noise-free ideal parameters. For this preliminary study of the 10 subjects, 5 subjects were removed as they did not exhibit a consistent decrease in MDN trend associated with muscular fatigue or the GPSM modeling error was high. The high error could be explained by the fact that the estimated CF was small denoting the low frequency asymptote is estimated in a small frequency range and might not be highly accurate because the least square solution of GPSM is not exposed to sufficient amount of information. The objective of the experiment is to evaluate sensitivity of FIs to muscle fatigue.

RESULTS

The results are shown in Fig.(2). The fractional-orders *q* and *g* are not exhibiting consistent trends and have small variations with time (SNR = 0.12 and 0.07). The CF is showing a strong correlation with time (SNR = 2.34) similar to MDN. The scaling factor, *c* (a measure of energy) also shows a trend similar to the root mean square (RMS) of the signal.

CONCLUSIONS

A set of novel FIs was presented with capability of providing both muscle fatigue-sensitive and -insensitive parameters. The results are consistent with the previous studies and promising complementary geometrical measures for fatigue studies during dynamic contractions. The temporal resolution of GPSM is poor and we are developing alternative estimation approaches to resolve this limitation.

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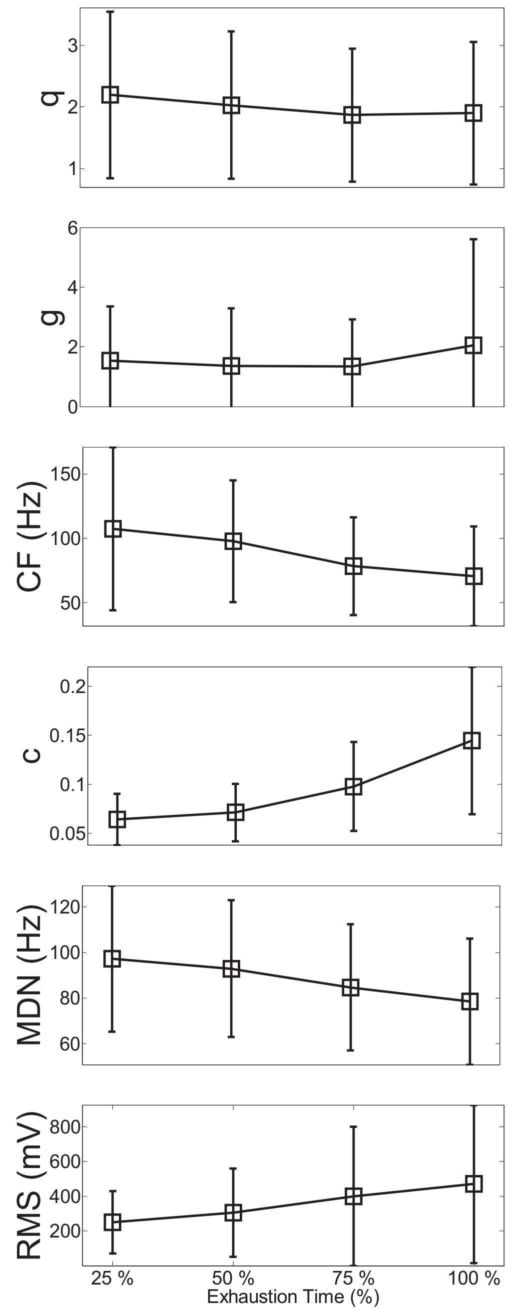

Figure 1: The plots are showing the averaged parameters and their standard deviations (i.e. inter and intra subject variations). The fractional-orders, *q* and *g* are insensitive to muscle fatigue, the CF is mimicking the MDN and the scaling factor, *c* is showing a trend similar to RMS.

A NEW METHOD TO ESTIMATE MYOELECTRIC MANIFESTATIONS OF MUSCLE FATIGUE

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INTRODUCTION

Fatigue may be described as a feeling or sensation of weakness or muscle pain or a decrement of performance, not easily suitable for quantification or measurement. During a muscle contraction, albeit in absence of mechanical manifestations of fatigue, strong modifications in surface EMG signal can occur: the so-called myoelectric manifestations of fatigue. Two main physiological factors are at the origin of the myoelectric manifestations of fatigue: 1) the decrease of the conduction velocity (CV) of motor unit action potentials (MUAP) ("peripheral fatigue" in the following), and 2) the increase of motor unit (MU) synchronization by the central nervous system ("central fatigue" in the following). We suggest quantifying the myoelectric manifestations of muscle fatigue through a two dimensional vector based on two distinct measures of central and peripheral fatigue, instead of using a scalar estimator.

METHODS

The following fatigue indexes were investigated: 1) mean spectral frequency – MNF, 2) median spectral frequency – MDF, 3) root mean square – RMS, 4) average rectified value – ARV, 5) muscle fibre conduction velocity - CV, 6) percentage of determinism (obtained from recurrence quantification analysis) – %DET , 7) spectral indexes defined as the ratio between the signal spectral moment of order -1 and moments of order k=2÷5 (FI_k), 8) mean frequency of the power spectrum density estimated by autoregressive analysis – MNF_{AR}, 9) mean frequency of the power spectrum density estimated by Choi-Williams time-frequency representation – MNF_{CWD}, 10) mean frequency of the power spectrum density estimated by continuous wavelet transform – MNF_{CWT}, and 11) fractal dimension – FD.
The indexes were tested on epochs (0.5 s of duration) of simulated signals using a model with a planar description of the volume conductor.
Simulation 1. Stationary signals (20 signals 20 s long) to assess the reliability of the indexes expressed on the basis of the coefficient of variation (COV).
Simulation 2. Global estimation of fatigue tested measuring the rate of change of the indexes when applied to non stationary signals (20 signals 20 s long, with increasing level of synchronization from 0 to 20% and decreasing CV with -0.1 m/s^2).
Simulation 3. Selective information on either peripheral or central myoelectric manifestation of fatigue was assessed. For 40 random distributions of the MUs within the muscle, signals of duration 0.5 s with mean of CV distribution in the range 3-5 m/s (with 11 steps), and synchronization level in the range 0-20 % (11 steps) were simulated. Mean CV and level of synchronization were mapped linearly to the range (0, 1). Each fatigue index was obtained as a function of the two variables. Regression planes were

obtained by minimizing the mean square error. The sensitivity of a fatigue estimator to a variation of CV or synchronization was defined as the first and second component of the vector normal to the regression plane.

RESULTS AND DISCUSSION

Simulation 1: MNF, FD, S, CV had lower COV (about 2%), compared to amplitude estimations (3%). The spectral index FI_k and %DET showed high COV (about 6% and 25%, respectively).
Simulation 2: CV, amplitude and frequency estimators had comparable rate of changes (about 1%/s), which were found higher than that of FD (0.2%/s). The spectral index FI_k was very sensitive (about 1.5%/s) to what?.
Simulation 3: CV estimation is uncorrelated to the level of simulated synchronism, being the most promising index for the estimation of peripheral fatigue. FD was the estimator least affected by CV changes and most related to the level of synchronism, being the most promising estimator of this manifestation of central fatigue (see Figure).

SUMMARY/CONCLUSIONS

The availability of two independent estimators of central and peripheral fatigue provides new insights into myoelectric manifestation of fatigue describing the phenomenon as a whole and assessing, by means of a two dimensional vector (CV, FD), the relative role of both central and peripheral fatigue.

ACKNOWLEDGEMENTS

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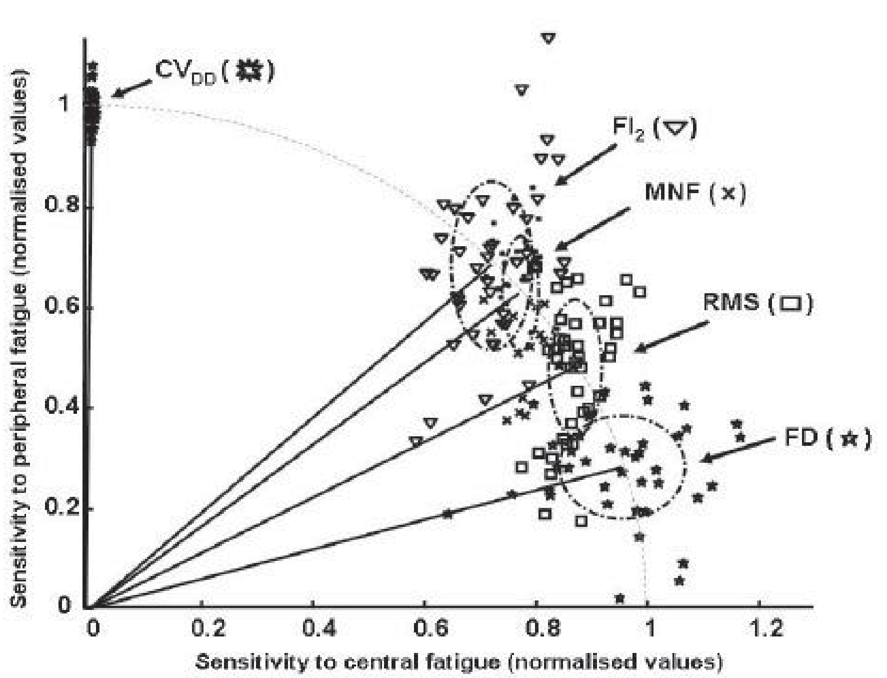

Sensitivity of CV, RMS, MNF, FI_2 and FD to peripheral and central fatigue. The sensitivity is defined as the two components of the vector of the best fit plane of the indexes as a function of simulated CV and synchronism. Mean and standard deviation of the two components of such a vector obtained from 40 simulated distributions of the MUs within the muscle are shown (after normalisation of the vectors of each index with respect to the mean modulus).

ASSESSMENT OF MUSCULAR STATUS FROM SURFACE ELECTROMYOGRAPHY DURING DYNAMIC PROTOCOLS

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INTRODUCTION

Surface ElectroMyoGraphy (sEMG) provides objective information about the functional status of muscles by giving insights on the physiological processes related to force exertion, movement production and motor control. Electrical parameters (i.e. amplitude and mean spectral frequency) extracted from sEMG signals during muscular activations have been extensively used for the assessment of both fatigue and force. However, while spectral modifications and the relationship between signal envelope and muscular force have been clearly understood for static protocols, an unique interpretation is still missing for dynamic conditions.

Many applications related to rehabilitation, ergonomics and athletes training, could greatly profit from a reliable indication of the functional status of muscles involved in the motor task. To deal with this problem it should be considered that in dynamic protocols the muscular functional status evolves following complex physiological phenomena (fibers recruitment, muscular synergies; etc.). During movement, the muscles exert different levels of force, experience fatigue, and recover from fatigue conditions. Electrical parameters extracted from sEMG can help providing an answer on this. In dynamic protocols a single electrical parameter is not sufficient to monitor the functional status of muscles (generally, this does not depend on a single phenomenon, such as in the status of fatigue where conduction velocity and time-varying muscular force concur). Luttmann and coworkers (Luttmann et al., 1996) proposed to use both the electrical parameters and their variations with respect to a reference level to assess the muscular status. However, the use of a single reference level is not correct in every experimental condition, since it varies on the basis of both the motor task and the dynamic and cinematic constraints. In this study, we propose a novel implementation of a muscular status coder based on the variations of amplitude and mean spectral frequency of sEMG signal. The coder will detect five different muscles' statuses (steady state, force increase, force decrease, recovery, fatigue) by using adaptive algorithms developed to deal with non stationary signals. The computational complexity of these algorithms will allow a real-time implementation of the coder.

METHODS

The proposed method is based on the estimation of electrical parameters, that is mean frequency (Conforto and D'Alessio, 1999) and signal amplitude (D'Alessio and Conforto, 2001), carried on by algorithms already proposed by the authors. The modifications of the parameters through time can help monitoring the muscular status during the exercise. By calculating the percent instantaneous relative variation of amplitude (Δa%) and mean frequency (Δf%), the muscular status can be coded as: steady state (no notable variation of either parameter), force increase (Δa%, Δf% both positive), force decrease (Δa%, Δf% both negative), recovery (Δa% negative, Δf% positive), fatigue (Δa% positive, Δf% negative). Ten able body subjects volunteered for an experimental protocol based on a laboratory cycling session. Subjects were asked to cycle for 30 minutes, at about 90 RPM, with the resistance of the flying wheel (RFW) at about the 10% of the maximum load of the cycle-simulator. A movement analysis system (StepPC©, DEM-Italy), has been used to record cardiac activity, sEMG signal from rectus femoris, and angular displacement at knee joint. 180 seconds of data recorded in the central phase of the session (ex-1) were compared against those recorded at the end of the exercise, where subjects underwent a recovery exercise (ex-2) by decreasing the RFW. In particular, sEMG signals have been analyzed and coded by using the five status approach. Every pedaling cycle, obtained by the knee angular displacement signal, has been associated to a muscular status. The relative number of muscular statuses in the five categories, in ex-1 and ex-2, was thus extracted and underwent a statistical analysis.

RESULTS AND DISCUSSION

Table 1 reports the preliminary results of the coder. Descriptive statistics for the number of muscular status has been calculated for both the cycling exercises. One way ANOVA has been calculated to assess the statistical difference between the number of muscular statuses in the two exercises.
It is interesting to notice that the coder manages to differentiate between fatigue and recovery status in the two different exercises. No differences have been found in force conditions, as requested by the protocol.

Table 1: Descriptive statistics and statistical significance of the test (two exercises). Values are expressed in percentage of number of muscular status.

Status	**Mean± STD (%)**		**p<0.05**
	Ex-1	**Ex-2**	
Force Increase	2.7±1.7	1.3±1.5	-
Fatigue	3.6±1.2	2.4±1.9	*
Force Decrease	1.4±1.3	1.3±1.2	-
Recovery	1.3±1.3	3.2±2.0	*
Steady State	91.0±2.1	91.8±3.1	-

SUMMARY/CONCLUSIONS

The proposed coder opens interesting scenarios on the assessment of muscular status. The preliminary results are consistent with the experimental conditions and the algorithms used for the estimation of the electrical indicators show a good performance when dealing with non stationary time series. Even if the biological plausibility of the method has to be further investigated, the first insights seem to be encouraging. Since the computational complexity of these algorithms allows for a real-time implementation of the coder, future development will move toward the implementation of a portable device to be used in ergonomics and rehabilitation.

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ELECTROMYOGRAPHIC MANIFESTATIONS OF FATIGUE AND TISSUE OXYGENATION IN LOW-INTENSITY TRUNK EXTENSOR CONTRACTIONS

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INTRODUCTION

Muscles of the extremities, can sustain contractions at low intensities (below 10% of maximum) over a substantial period. However, during such contractions their force producing capacity does substantially decrease, and electromyographic fatigue manifestations are evident (Krogh-Lund 1993). The trunk extensor muscles are mostly active below 10% of maximum (Mork and Westgaard, 2005). However, for example during office work, these muscles are active for 93-98% of the time (Dieën et al. 2001). The trunk extensor muscles might be adapted to such sustained activity associated with their postural role, as indeed the predominance of type I fibers, the well-developed capillary network and high glycogen concentration and aerobic metabolic enzyme activity suggest (Jørgensen et al. 1993). In addition, the mechanical redundancy of this muscle group appears to be exploited to prevent or delay fatigue development by alternating activity between muscle parts (Dieën et al. 1993).

This study was designed to determine whether trunk extensor fatigue develops during low-intensity contractions and whether this is associated with a decrease in muscle tissue oxygenation. EMG feedback was used to impose constant activity in a part of the trunk extensor muscles.

METHODS

Twelve volunteers performed 30-min contractions at 2% and 5% of the maximum EMG amplitude (EMGmax) at the feedback site. Contractions at the two levels were performed on separate days.

EMG was recorded from 6 sites over the lumbar extensor muscles. Feedback was provided of the linear envelope of the signal recorded at the left L3 level, 3 cm paravertebral. The means and coefficients of variation over time of the EMG linear envelopes were calculated, to express mean level and variability of activity. The mean power frequency (MPF) was estimated over 1-second windows and results were regressed against time to use the slope as a fatigue indicator.

Near infrared spectroscopy (NIRS) was used to measure changes in oxy-hemoglobine and deoxy-hemoglobine contents at the feedback site. Before and after the sustained contraction a short 50% MVC contraction was performed to check stability of the NIRS recordings.

RESULTS

The mean EMG amplitude was not significantly different between electrode sites, whereas the coefficient of variation was lower at the feedback site, compared to all other recording sites.

Significant decreases of the EMG MPF occurred in all subjects (Table 1). At the feedback site, the MPF decreased consistently, i.e. the group-averaged slope was significantly lower than zero, without a significant difference between conditions. At 5% EMGmax, all ipsilateral sites showed a consistent MPF decrease. At other sites, MPF decreases were significant in some subjects, while group averaged slopes were not significantly lower than zero.

High correlations between the changes in hemoglobine contents before and after the sustained contractions confirmed that stable NIRS recordings were obtained. However, no significant changes in oxy-hemoglobine content during the sustained contractions were detected.

DISCUSSION

Our results suggest that even at very low contraction intensities muscle fatigue did develop. This occurred in absence of a decrease in tissue oxygenation. Combined with previous NIRS data collected during short-lasting low-intensity contractions (Jensen et al. 1999), and during high-intensity contractions (Yoshitake 2001; Kell et al. 2004), this suggests that oxygenation is a limiting factor in this muscle group only at contraction intensities higher than approximately 50% MVC.

The limited variability of muscle activity at the EMG feedback site and at ipsilateral locations appeared to enhance fatigue development. In line with previous findings (Dieën et al. 1993), this suggests that temporal variation in load sharing between the extensor muscles may attenuate fatigue development.

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Table 1. MPF slopes (%/min), with Student t-values (T) and corresponding p-values for a difference between average and zero and the number of subjects in whom the slope was significantly smaller than zero (n < 0) out of the total number of subjects (n).

	2% EMGmax					5% EMGmax				
	mean	SD	T	p	n<0 / n	mean	SD	T	p	n<0 / n
lL3m	-0.136	0.218	-2.16	0.027	10/12	-0.165	0.114	-5.00	<0.001	11/12
rL3m	-0.045	0.220	-0.72	0.244	7/12	0.038	0.277	0.48	0.320	3/12
lL3l	-0.072	0.363	-0.69	0.253	6/12	-0.193	0.244	-2.74	0.010	8/12
rL3l	-0.005	0.196	-0.08	0.467	6/12	-0.045	0.189	-0.82	0.215	7/12
lL1	-0.121	0.416	-1.00	0.169	6/12	-0.278	0.300	-3.21	0.004	9/12
rL1	-0.099	0.250	-1.37	0.100	7/12	0.052	0.256	0.70	0.249	6/12

lL3m = electrode site to left of L3 medial (3 cm) feedback site, rL3m = right of L3 medial, lL3l = left of L3 lateral (5 cm), rL3l = right of L3 lateral; lL1 = left of L1 (3cm), rL1 = right of L1

LONG TERM DIFFERENTIAL ACTIVTION OF THE BICEPS BRACHII MUSCLE PREVENTS FATIGUE AT A LOW CONTRACTION LEVEL

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INTRODUCTION

Several motor control strategies are proposed to reduce the progressing fatigue during prolonged contractions (Enoka and Stuart 1992).

In particular, intra-muscular redistribution of activity (Holtermann et al 2006) is assumed to prevent fatigue during prolonged contractions. Farina and colleagues (2006) confirmed this notion by revealing a positive association between changes in spatial distribution of trapezius muscle activity and time to exhaustion with the shoulders at 90º abduction. On the contrary, Holtermann and colleagues (2007) observed a negative relation between the frequency of differential activations between the short and long heads of the biceps brachii and time to exhaustion in a prolonged elbow flexion contraction at 25 % of maximal voluntary contraction (MVC). This finding could be due to the relatively moderate contraction level, not permitting parallel changes in local circulation with the differential activation.

Therefore, the main aim of this study was to investigate the influence of differential activation on fatigue during prolonged low level contractions.

METHODS

15 males participated in the study. The subjects performed isometric elbow flexion using a dynamometer. First, the subjects carried out three MVCs. Then, the subjects performed a prolonged contraction at 5% MVC for 30 min receiving target and visual feedback of the force. Immediately after the 30 min prolonged contraction, the subjects performed a MVC.

Surface electromyographical (sEMG) signals were recorded using an electrode-grid device covering 6 x 4.5 cm of the skin surface. The middle of the grid was located on the partition of the two heads of the biceps brachii muscle. The amplitude of the sEMG signals was calculated with RMS in 0.5 s time-windows. The average RMS from the channels located on the long and short head of the biceps brachii was calculated. The differential activation was defined as 33 % difference in average RMS between the muscle heads. Long and short term differential activations were quantified by low and high-pass filtering at a cut-off frequency corresponding to a 10 s time-period. The frequency and duration of differential activations throughout the prolonged contraction were quantified.

RESULTS AND DISCUSSION

Long and short term differential activations between the heads of the biceps were observed in all but one subject. The average frequency of the differential activations varied from 0.3 to 9.74 min^{-1}. The average duration of the long term differential activations was 15 ± 7.2 s. The average duration of the short term differential activations was 0.82 ± 0.17 s. The relation between the average duration of long term differential activations and fatigue-induced reduction in maximal force was highly significant, $R^2 = 0.53$, $p<0.005$. No significant association was observed between the average frequency of long term differential activation or frequency and duration of the short term differential activations and fatigue-induced reduction in maximal force.

The observed differential activation in almost all subjects indicates that differential activation is an innate motor control strategy of the biceps brachii muscle during prolonged monotonous contractions.

Long term differential activations were observed during the prolonged contraction at 5% MVC. In contrast, long term differential activations were not observed during prolonged contractions at 25 % MVC (Holtermann et al 2007). The lack of differential activations of longer duration at the contraction level of 25 % MVC may be due to high cost of sustaining an activity level of 33% larger than the other head at an already relative high activation level.

Reciprocal differential activations between the long and short head were often observed during the prolonged low level contraction. Reciprocal differential activations did not occur during the prolonged contraction at 25 % MVC, in which the differential activation was generated by increased activity in one head while the activity remained constant at the other head of the biceps brachii muscle (Holtermann et al 2007).

The high association between average duration of long term differential activations and fatigue-induced reduction in MVC designates that long term differential activation has a positive influence on prevention of the progressing fatigue during prolonged low level contractions.

SUMMARY/CONCLUSIONS

The average duration of the long term differential activations was shown to explain about half of the variation in fatigue-induced reduction in maximal force. This finding supports that the duration of differential activations influences the ability to prevent the progressing muscle fatigue during prolonged low level contractions.

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BACK AND HIP EXTENSOR MUSCLE FATIGUE IN HEALTHY SUBJECTS: COMPARISON OF TWO SORENSEN TEST VARIANTS

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INTRODUCTION

Various positions have been tested to study back muscle fatigue (Da Silva et al., 2005; Koumantakis et al., 2001), and the most common procedure reported in the literature is the Sorensen test.

Electromyography (EMG) is instrumental in examining neuromuscular mechanisms associated with muscle fatigue. A decline in the median (or mean) frequency of the EMG power spectrum (MPF) during sustained isometric muscle contraction is a valid indicator of muscle fatigue (De Luca, 1997).

Careful examination of previous studies results comparing paraspinal fatigue testing positions indicates a task-dependency effect. Task-dependency refers to the different neuromechanical or neurophysiological mechanisms involved in each specific fatiguing contraction. Elfving and Dedering (2007) assessed paraspinal muscle fatigue and obtained a greater rate of fatigue during a seated test compared to a modified Sorensen test (on a 40^o roman chair). Da Silva et al. (2005), testing three different fatigue protocols, showed that back muscle fatigue was lower during a lift position in comparison to the Sorensen test and an upright test. These previous studies, however, investigated only the task-dependency of the paraspinal muscles. Since hip extensor muscles contribute to trunk extension movements (Leinonen et al., 2000), it is important to investigate their relative involvement in the task-dependency effect during back muscle fatigue assessment. Therefore, the aim of the present study is to evaluate and compare the rate of back and hip extensor muscle fatigue during two variants of the Sorensen test. It is hypothesized that the rate of muscular fatigue of the hip extensor muscles will vary according to test position.

METHODS

In this cross-sectional study, 20 healthy subjects (age: 24.7±3.0 years; height: 177±10 cm; mass: 76.4±10.5 kg; body mass index: 24.3±3.4 kg/m^2) performed body weight-dependent isometric back extension (Sorensen test) in two positions: on a horizontal table (S1) and on a 45^o roman chair (S2). EMG of the paraspinal muscles at T10 and L5 levels, gluteus maximus (GM) and biceps femoris (BF) were recorded. Muscle fatigue was estimated by the rate of MPF decline.

The MPF was calculated from consecutive 3-s intervals by Fast-Fourier transformation from the entire recorded signal. Least square linear regression analysis was applied to the MPF time series to estimate the rate of decline (MPF/T slope) and the coefficient of determination (R^2).

RESULTS AND DISCUSSION

The subjects achieved significantly higher endurance time during S2 compared to S1 (262 ± 81s vs 163 ± 70 s). Figure 1 shows that there was no effect of the test variants on MPF/time slope values (mean±SEM: -0.8±0.13 vs -0.42±0.7). A significant between-muscle effect was found, and post-hoc comparisons revealed that the L5 level presented significantly higher

MPF/time slope values than all other muscle groups. Interaction test variants by muscles disclosed a significant *p* value where the L5 paraspinal muscles gave a greater value for the S1 test than S2.

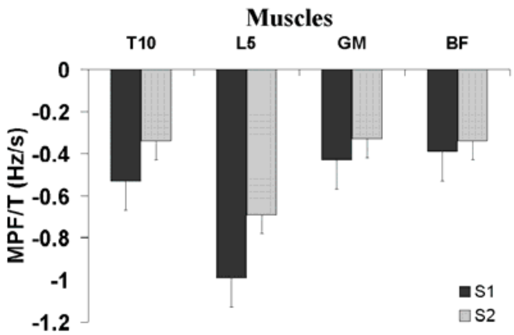

Figure 1: MPF/time slope during two Sorensen variants. Data are mean ± SEM.

For R^2, a significantly lower value was apparent during S2 compared to S1 (mean±SEM: .31±.02 vs .40±.02). A significant between-muscle difference was also observed, and post-hoc comparisons determined that L5 had a higher R^2 value than all other muscles (mean±SEM: .58±.05 vs .33±.03, .25±.03 and .25±.04 for T10, GM and BF, respectively). None of the interaction effects had a significant *p*-value.

The aim of this study was to assess the task-dependency effect on back and hip extensor muscles during two Sorensen variants. It was expected that hip extensor fatigue indices would differ between the horizontal and 45º roman chair positions. Our results did not support this hypothesis because the GM and BF did not present significant differences between the Sorensen variants.

One important finding from this study was that hip extensor muscle fatigue simultaneously with paraspinal muscle during both Sorensen variants. However, only L5 level EMG fatigue indices showed a task-dependency effect between S1 and S2. Hip extensor muscles appear to contribute to load-sharing of the upper body mass during both Sorensen variants, but to a different extent because L5 level fatigue differ between the Sorensen variants. Tveit et al. (1994) noted a significant effect of lumbar lordosis curvature on lever arm lengths of the back extensor muscles. Further studies with kinematics analysis should be conducted to assess the relationship between lordosis curvature and lumbo-pelvic extensor muscles fatigue during the Sorensen protocol.

Interestingly, compared to the hip muscles, a higher R^2 value was obtained for the L5 paraspinal muscles, disclosing that the MPF data best fitted a simple regression analysis technique. Another study has demonstrated that more complex statistical models are not valuable on fatigue-related EMG MPF characteristics (Coorevits et al., 2005).

SUMMARY/CONCLUSION

The present study found that task-dependency has to be considered during the assessment of back extensor muscle fatigue. More experiments are required to investigate task-dependency of the lumbo-pelvic muscles in low back pain subjects.

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ELECTRO- AND MECHANOMYOGRAM DURING LOW-FORCE STATIC CONTRACTION IN SUBJECTS WITH EPICONDYLITIS LATERALIS COMPARED TO HEALTHY CONTROL

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INTRODUCTION

Prolonged static contractions and ongoing repetitive low-level activity in the forearm muscles is well-known risk factors for development of epicondylitis lateralis (EL). Further, pain may affect the force generating capacity. For prevention purposes the electromyogram (EMG) has been used to identify muscle fatigue development, but the sensitivity during low-force contractions is limited. However, mechanomyogram (MMG) has been shown to be promising to identify muscle fatigue during low-force contractions (Blangsted et al. 2005).

The purpose of this study was to test the hypothesis that in subjects with clinically diagnosed EL the maximal voluntary contraction (MVC) force in wrist extension is lower and the fatigue development more pronounced compared to healthy controls (CON). Further, that MMG may detect fatigue development not reflected in EMG.

METHODS

Seven females and 3 males (41 (SD: 14) years, 171.4 (SD: 8.9) cm, 69.9 (SD: 12.2) kg) with unilateral EL, and 10 CON matched with regard to age and sex (40 (SD: 13) years, 172.7 (SD: 7.5) cm, 69.5 (SD: 12.0) kg). The subjects were sitting with the elbow flexed 90 degrees, the forearm pronated and resting on a horizontal platform. In this position 3 MVC was performed against a force transducer. Moment arm was measured and the wrist extension torque calculated. Next, static wrist extension was performed at 10% MVC for 10 min.

Bipolar surface EMG was recorded from m. extensor carpi radialis (ECR), m. extensor carpi ulnaris (ECU), and a forearm flexor (FLEX). MMG was recorded by an accelerometer placed on ECR. EMG and MMG were analyzed in 30 s-periods for root mean square amplitude (rms), and mean power frequency (mpf).

Furthermore, perceived exertion (RPE) was rated on a Borg scale from 0 (no perceived exertion) to 10 (maximal perceived exertion), and pain sensation (VAS) was rated by a VAS scale from 0 (no pain) to 10 (worst imaginable) every minute during the 10% MVC contraction.

Muscular tenderness, measured as pressure pain threshold (PPT) was determined with an electronic pressure algometer. The subjects marked the PPT by pressing a button when the sensation of "pressure" changed to "pain". All PPT measurements were conducted 3 times and the mean value was calculated. All results are presented as mean (SD).

RESULTS AND DISCUSSION

There was no significant difference in torque between the EL (10.0 (4.2) Nm) and CON (11.6 (3.8) Nm). The mean PPT tended to be lower (t-test: p=0.056) for the EL (344 (155) kPa/s) compared to the CON (527 (310) kPa/s). EMG and MMG results are presented in table 1. The increase in RPE was significantly higher (p=0.01) for the EL (0.6 (1.1) and 6.7 (2.9) in the first and last min, respectively) compared to the CON (0.2 (0.3) in the first and 4.0 (1.5) in the last min) of the 10% MVC contraction. Also pain ratings were significant higher (p<0.01) for EL, with a increase in VAS from 0.6 (SD: 1.4) to 4.2 (SD: 3.8) compared to CON, 0 (SD: 0) in the first min and 0.8 (SD: 1.1) in the last min.

The inflammation of the epicondylitis lateralis, probably originate from excessive activity of the wrist extensor muscle. Nevertheless, this was not reflected in a reduced maximal torque of the muscle. For both EL and CON the EMG and MMG responses were significantly different from first to last min of the contraction, i.e. the rms-values were increased whereas the mpf values decreased, indicating fatigue. However, although there were significant differences between EL and CON regarding RPE and VAS, this difference was not seen in the EMG responses, but only in the MMG, indicating a change in the mechanical properties of the EL muscle.

SUMMARY/CONCLUSIONS

This study shows that MMG in contrast to EMG is modulated in EL. MMG could possible be useful as a non-invasive method to investigate changes in the neuromuscular system and may be developed to serve as a prediagnostic tool for identification of musculoskeletal disorders.

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ACKNOWLEDGEMENTS

The Danish National Research Foundation supported this work (Grant no. 22-04-0488).

Table 1: EMGrms, EMGmpf, MMGrms, and MMGmpf during 10% MVC wrist extension (ext) for 10 min (mean (SD)). * 10th min sign different from 1st min (p<0.05). # EL sign different from CON (p<0.05).

10% MVC wrist ext	**EMGrms**			**EMGmpf**			**MMGrms**	**MMGmpf**
	ECR (µV)	**ECU (µV)**	**FLEX (µV)**	**ECR (Hz)**	**ECU (Hz)**	**FLEX (Hz)**	**ECR ($m*s^{-2}$)**	**ECR (Hz)**
CON 1st min	43.82 (26.42)	57.06 (17.63)	7.64 (6.45)	107.02 (10.30)	116.91 (13.14)	122.09 (30.56)	0.02 (0.01)	29.17 (10.68)
CON 10th min	53.90 (28.22) *	70.10 (19.77) *	8.63 (3.23)	96.38 (11.54) *	109.86 (11.42) *	110.30 (24.55)	0.06 (0.05) *	25.06 (5.15) *
EL 1st min	44.83 (14.12)	48.46 (18.46)	4.97 (1.75)	104.94 (8.49)	116.76 (9.16)	122.35 (13.67)	0.04 (0.03)	37.75 (6.05)
EL 10th min	59.84 (27.43) *	65.25 (19.53) *	7.86 (3.72) *	91.58 (12.45) *	107.56 (8.82) *	109.15 (16.96) *	0.14 (0.13) * #	28.54 (3.37) *

ISOMETRIC LEG EXTENSION ENDURANCE TIME PREDICTION USING ELECTROMYOGRAPHIC SIGNAL

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INTRODUCTION

Measurement of endurance time (tlim) is an indicator of the muscle resistance to fatigue (Merletti and Roy, 1996). However, the assessment of the endurance capacity requires exhaustive exercises that may be deleterious. Moreover, these tests are problematic because they might be influenced by some psychological factors, as motivation and tolerance to pain. Thus, it could be useful, in the field of sport as for patients suffering of weakness and chronic fatigue, to estimate the tlim without requiring the subject to perform an exhaustive test.
Surface electromyography (sEMG) represents an objective technique to assess fatigue. It has been shown that changes in sEMG parameters, during isometric contractions sustained until exhaustion, were related to the endurance time. These relationships have been reported during mono-articular tasks soliciting back extensors (e.g. van Dieen et al., 1998) and upper and lower limbs muscles (e.g. Maisetti et al., 2002; Merletti and Roy, 1996). In most of these studies, changes in sEMG signal computed over the first half of the test duration appeared to allow the prediction of the tlim. Recently, we tested relationships between sEMG changes and tlim during a submaximal multi-joint task, equivalent to a 20% of maximal voluntary isometric torque (MVIT) contractions of leg extensors (Boyas et al., in press). Early changes in the sEMG did not allow us to predict endurance time for trained and untrained subjects in isometric contractions. The important number of muscles and joints involved in this task and the relative low intensity applied at each joint probably induced muscular coordination and central fatigue that may have impaired our ability to predict endurance time. Hence mono-articular tasks seem to be more adapted to tlim prediction. However, in these conditions, studies focused mainly on exercise intensities equal or greater than 50% MVIT. Therefore, this study aimed to determine the ability to predict endurance time using early changes in sEMG signal during a submaximal (20% MVIT) isometric leg extension sustained until exhaustion. This was tested for trained and untrained subjects in isometric sustained submaximal contractions.

METHODS

Twenty-seven healthy men volunteered to this study, seven high level sailors trained in isometric leg extensions, and twenty students in sports sciences (untrained). Subjects performed test seated on a Biodex dynamometer with hip and knee angles fixed respectively to 70 and 40°. After a familiarization session and a specified warm-up, subjects realized three maximal isometric leg extensions to determine their MVIT and to record associated maximal muscular activity. Then, subjects sustained a torque equivalent to 20% MVIT until exhaustion. The electrical activity of the superficial leg extensors (i.e. rectus femoris, vastus medialis and vastus lateralis) was recorded throughout the entire duration of the test. SEMG signals were filtered (bandwidth: 6-500 Hz) and changes in the root mean square (RMS), mean power frequency (MPF) and relative power in the 6-30 Hz frequency band (FB1) were calculated every 2% tlim. SEMG parameters were normalized according to values recorded during the maximal extension (RMSmax) for RMS and according values recorded at the start of the fatigue test (MPFi, FB1i) for MPF and FB1. Changes in sEMG parameters during the fatigue test were expressed via the slope of the linear regression or using the area ratio (Merletti et al., 1991), and their relations with tlim were tested using Bravais Pearson correlations.

RESULTS AND DISCUSSION

Trained subjects (535 ± 159 s) maintained the task longer than untrained ones (276 ± 68 s, $p<0.01$) because of neuromuscular adaptations probably induced by training. However, sEMG parameters evolved similarly for both groups and we observed a raise in RMS (from 14 to 41% RMSmax) and in FB1 (+ 48% FB1i) and a drop for MPF (- 5% MPFi) suggesting fatigue. Changes in sEMG were more important for the rectus femoris indicating that this muscle was the most fatigued at the end of the test. Significant relationships between tlim and changes in FB1 of the most fatigued muscle recorded during the 25 first % tlim were found for trained subjects ($r^2=0.74$, figure 1).

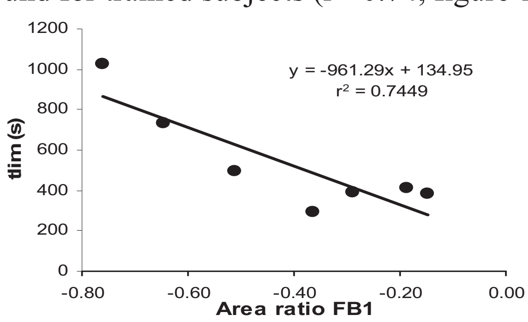

Figure 1: Relationship between the area ratio of the early changes in FB1 (120 s) of the most fatigued muscle of trained subjects.

The use of the changes in low frequency band in tlim prediction is in accordance with the literature (Maisetti et al., 2002) and suggests that this sEMG parameter, influenced, among others, by changes in the velocity conduction of action potentials and motor unit frequency may be used to tlim prediction. Though, no relationships were observed for untrained subjects contrary to previous studies performed at higher intensity. For these subjects, we suppose that lower exercise intensity may extend the tlim and influence the respective parts of central and peripheral fatigue that may influence our ability to predict tlim.

SUMMARY/CONCLUSIONS

The present study suggests that early changes in sEMG signal (particularly the relative power in the 6-30 Hz frequency band) may be used to predict the endurance time of a sustained submaximal (20% MVIT) isometric leg extension performed by a group of trained sailors. However, tlim prediction was not possible for untrained subjects which is not in accordance with the literature. We could suppose that the intensity of the exercise plays a role in the etiology of fatigue and influences our ability to predict tlim using the sEMG signal.

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ALTERNATING MOTOR UNIT RECRUITMENT DURING BICYCLE EXERCISE, INDICATIONS FROM HDsEMG AND ^{31}P NMR SPECTROSCOPY.

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INTRODUCTION

The majority of the knowledge about motor unit recruitment patterns has been obtained during isometric contractions. As a consequence, it remains largely unknown if this knowledge is also valid for dynamic contractions.

High-Density surface EMG and ^{31}P NMR spectroscopy were used to study muscle energetics and electrophysiological behavior during bicycle exercise at fatiguing intensities. Here, we report about largely unexpected muscle behavior considering traditional notions.

METHODS

We used a bicycle ergometer constructed from non-ferrous components. The mechanical load was controlled by adjusting the brake force exerted on a wooden flywheel. A metronome was used to set the pedaling frequency to 80 rpm.
Six physically fit, male subjects performed exercise at three different fatiguing workloads (brake force: 30N – 50N – 65N). Typically, subjects could continue these exercises for respectively: 3, 2 and 1 minutes.

^{31}P spectra were acquired from the vastus medialis and were recorded at rest and during exercise (time resolution 12s). PCr, inorganic phosphate (Pi), and ATP resonances were fitted in the time domain. We calculated absolute concentrations and pH according to the method described by Jeneson and Bruggeman (2004).

The exercise protocols were repeated outside the magnet to record the HD-sEMG data. An adhesive electrode grid containing 130 electrodes (10x13 with inter electrode distance of 4mm) was placed over the vastus medialis so that propagation of motor unit action potentials was clearly visible. Data were band-pass filtered and a double differential montage was used for subsequent analysis. Muscle fiber conduction velocity (MFCV), and RMS amplitude were calculated for each cycle. The method described by Farina et al. (2004) was used to calculate the MFCV.

RESULTS AND DISCUSSION

Figure 1 shows typical ^{31}P NMR spectra at rest and at the end of the 30N and 65N exercise protocols of the same subject.

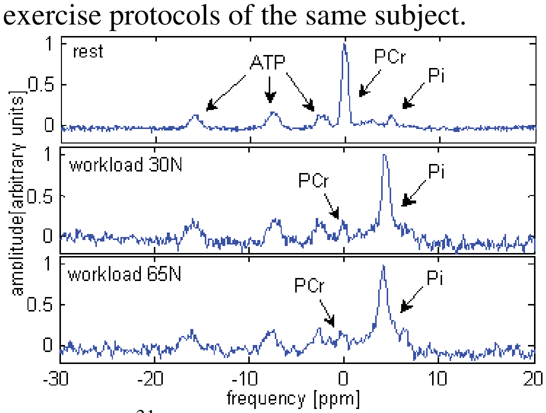

Figure 1: ^{31}P NMR spectra at rest and at the end of the 30N and 65N exercise protocols. PCr and Pi concentrations at the end of exercise at both workloads are equal.

At all workloads the PCr concentration dropped from 32.2±3.0mM (mean±std) to 5.1±2.6mM at exhaustion. This large drop can only be explained if all motor units are recruited during the exercise. This means that also at the end of the sub-maximal exercise level all motor units still are or recently have been recruited. The intramuscular pH dropped only little during exercise from 7.10±0.03 at rest to 6.82±0.05 at exhaustion.

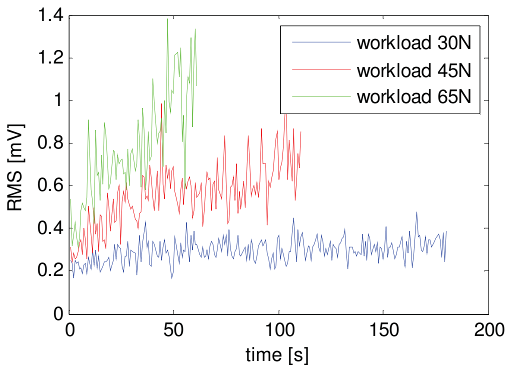

Figure 2: RMS amplitude during bicycle exercise at three different workloads.

The corresponding EMG results are shown in figure 2 and 3. The RMS increases for the workloads 45N and 60N but remains almost constant for the 30N workload. The increase in RMS amplitude could be explained by more MUs becoming active and/or by an increase in the firing rates. However, the almost constant RMS amplitude at 30N contradicts with the NMR results, which predict that all MUPs must have been activated during the contraction. We expected that RMS amplitude would increase also at this lower level if more MUs become active.

The increase in initial MFCV with increasing workloads (Figure 3) is significant ($p<0.001$, t-test) and could be interpreted as the result of an increased contribution of larger motor units, but could also be attributed to higher firing rates. In all three exercise protocols, the MFCV does not decrease as a result of fatigue; this is probably the result of the small drop in pH. This explanation is supported by a study of muscle fatigue in McArdle's disease (Linssen et al., 1990). These patients also showed no decrease in MFCV in absence of a decrease in pH during intermittent fatiguing isometric contractions.

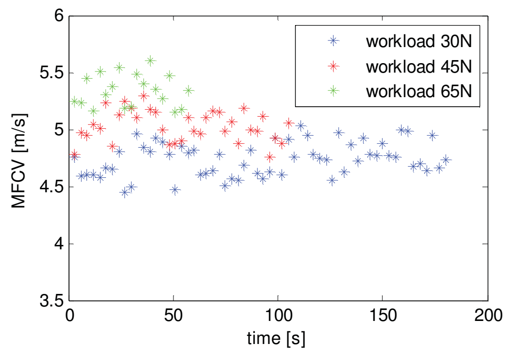

Figure 3: MFCV during bicycle exercise at three different workloads.

SUMMARY/CONCLUSIONS

We showed during bicycle exercise at sub maximal exercise levels, that all motor units must have been recruited. The MFCV does not decrease during bicycle exercise at fatiguing intensities as a result of only a small decrease in pH. However, MFCV also does not increase during the 30N exercise suggesting that firing rates remain stable. Therefore, we hypothesize that alternating recruitment must play a role in muscle activation during bicycle exercise.

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EFFECTS OF FATIGUE ON SIGNAL-DEPENDENT NOISE DURING ISOMETRIC FORCE PRODUCTION

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INTRODUCTION

An inherent feature of human's motor system is that force production exhibit unavoidable variability or noise. This noise is signal-dependent (signal-dependent noise, SDN), that is, the force variability scales linearly with the mean force (Jones et al. 2002). This feature is responsible, at least in part, for the stereotypical structure of human movements. Indeed, recent modelling studies on optimal control have demonstrated that the central nervous system (CNS) plans movements in order to minimize the effects of SDN in the force output of muscles (e.g. Harris and Wolpert 1998). Based on this assumption, these studies were able to reproduce many observed kinematics invariant of motor behaviour, for instance Fitts' law (Fitts 1954).

It is assumed that the variability of force output increases with fatigue (e.g. Garland et al. 1994). However, the precise way fatigue affects the relationship between force variability and mean force is currently unknown. Also, the mechanisms responsible for the increase in force variability with fatigue are poorly understood.

In a series of experiments, we studied the effects of muscular fatigue on the relationship between the force and its variability. Our aim was (1) to characterize this relationship over a broad range of forces, and (2) to investigate the possible origins of the increase in force variability with fatigue at various force levels.

METHODS

Experiment 1

This experiment was designed (1) to characterize the effect of fatigue on SDN, and (2) to test the hypothesis that the increase in force variability is due to the increased effort needed to achieve a given force during fatigue. Participants had to match steadily 4 levels of isometric elbow flexion force from 7 to 53 % of their maximal capabilities (MVC) before and after a fatigue protocol.

Experiment 2

This experiment was designed to demonstrate that there are other factors than the increase of effort that are responsible for the increase in force variability during fatigue, especially at low force levels. Participants had to match steadily 4 levels of electromyographic activity (EMG) corresponding to isometric elbow flexion force from 7 to 53 % of MVC before and after a fatigue protocol. By this way, the levels of effort were identical pre- and post-fatigue.

Experiment 3

This experiment was designed to investigate the role of the variability in the contractile mechanisms of the muscle itself in the increase of force variability with fatigue. Here, isometric forces were evoked by neuromuscular electrostimulation (NMES) applied on the muscle belly of the triceps surae before and after a fatigue protocol, to control the variability in the muscle excitatory drive. Stimulations parameters were chosen to evoke various forces of peripheral origin (i.e. force was generated by direct muscle stimulation).

In the 3 experiments, the fatigue protocol consisted of the repetition of 20-s isometric contractions. The workload was fixed at 60% of the MVC. Contractions were elbow flexions and extensions performed alternately, separated by periods of 15 s of passive rest. Typically it induced a loss of maximal force of about 30 %.

RESULTS AND DISCUSSION

Experiment 1 showed that SDN increased with fatigue at each force level. In particular, the relationship between the mean force and its variability could be equivalently described by the same model pre- and post-fatigue, but the magnitude of the variability was significantly more important during fatigue. This increase in force variability was not concomitant with an increase in muscle activation (EMG) for the 2 lower force levels.

Experiment 2 showed that when participants have to match the same level of EMG (i.e. of effort), the coefficient of variation of force increased with fatigue for the 2 lower force levels, but was unchanged for the 2 higher forces. This indicated that the increase of effort was the sole responsible for the increase in force variability with fatigue for moderate but not low forces.

In experiment 3, the variability of the force evoked by muscle stimulation was unaffected neither by force level, nor by fatigue.

SUMMARY/CONCLUSIONS

All together, these results provide a precise a precise characterization of the effect of fatigue on the relationship between the muscular force and its variability over a broad range of forces. Also, we show that the increase in the central drive to muscles is partly responsible for the increase in force variability during fatigue. However, our results suggest the existence of alternative mechanisms of neural but not muscular origin, especially at low force levels.

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SPATIAL DEPENDENCY IN SURFACE ELECTROMYOGRAPHY AFTER ECCENTRIC ARM EXERCISE

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INTRODUCTION

Eccentric exercise is known to result in a prolonged loss of muscle force and has been suggested to be accompanied by muscle cell membrane damage (Koskinen et al. 2006). Such damage may be expected to cause failure in action potential (AP) propagation. This phenomenon has been studied with surface electromyography (sEMG), however, introducing some controversial findings. In this regard, location of surface electrode can play an important role.

Purpose of the present experiment was to examine if any spatial dependency exists in changes of sEMG variables after intensive eccentric elbow flexor exercise in humans.

METHODS

Nine subjects performed 50 maximal eccentric contractions (20 s intervals, angular velocity of 1 rad·s^{-1}) with elbow flexors of the right arm on a motorized isokinetic dynamometer. Maximal isometric voluntary contraction (MVC) was determined for the exercised arm with elbow angle of 120° before (BEF), two hours after (2H), 2 days (2D) after and 4 days (4D) after the exercise. During MVC test sEMG signals were recorded (bandwidth: 10–750 Hz, gain: 200 and sampling rate: 2048 Hz) with grid of 64 electrodes (8 mm inter-electrode distance) from short head of biceps brachii muscle (BB). The electrode grid formed 2-D-multichannel system with 59 bipolar channels (12 rows and 5 columns) aligned in longitudinal direction of BB.

Root mean square (RMS) and mean power frequency (MNF) were calculated for each bipolar electrode separately from 1 second epoch during MVC test. Similarly, mean fiber conduction velocity (CV) was estimated based on three adjacent bipolar channels (McGill, K. & Dorfman, 1984). In addition, global values for RMS, MNF and CV were calculated as mean of all acceptable channels in the multichannel system. Prior to this, channels representing innervation zones (IZ), tendon regions or high noise were excluded.

Repeated measures analysis of variance was conducted for 1) MVC 2) global sEMG variables and 3) sEMG variables derived from each individual channel.

RESULTS AND DISCUSSION

MVC was 21.3 ± 5.6 % lower ($P < 0.001$) on 2H and 12.6 ± 11.1 % lower ($P < 0.01$) on 2D as compared to BEF values, and was recovered on 4D (Fig. 1.). Global RMS, MNF and CV decreased from their initial levels (RMS: from 1.31 ± 0.48 mV to 1.08 ± 0.38 mV, $P < 0.01$, MNF: from 92.6 ± 10 Hz to 85.2 ± 11 Hz, $P < 0.01$ and CV: from 4.03 ± 0.29 m/s to 3.86 ± 0.34 m/s, $P < 0.05$) on 2H, and in this case were recovered already on 2D (Fig. 1.). Individual channels showed statistical differences only in MNF and in six channels (see Table. 1.).

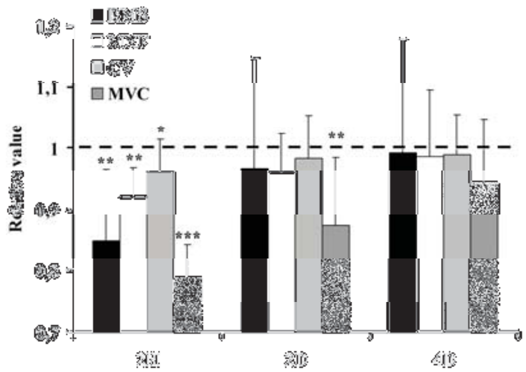

Figure 1: Relative changes in global sEMG variables and isometric MVC after the exercise. * = p<0.05, ** = p<0.01, *** = p<0.001 with respect to the initial level (dashed line).

Although, sEMG variables in most of the single channels followed the global changes, there were some channels with opposite behavior (Fig. 2). This could possibly be explained with exercise induced geometrical changes in the muscle, including a potential shift in IZs with respect to sEMG electrodes.

Table 1: Spatial map of MNF in each 59 bipolar channel. Channels with significant changes (P < 0.05) in green, opposite behavior in red, continuous decrease yellow and IZ locations encircled.

Col. 1	Col. 2	Col. 3	Col. 4	Col. 5
	12	24	36	48
1	13	25	37	49
2	14	26	38	50
3	15	27	39	51
4	16	28	40	52
5	17	29	41	53
6	18	30	42	54
7	19	31	43	55
8	20	32	44	56
9	21	33	45	57
10	22	34	46	58
11	23	35	47	59

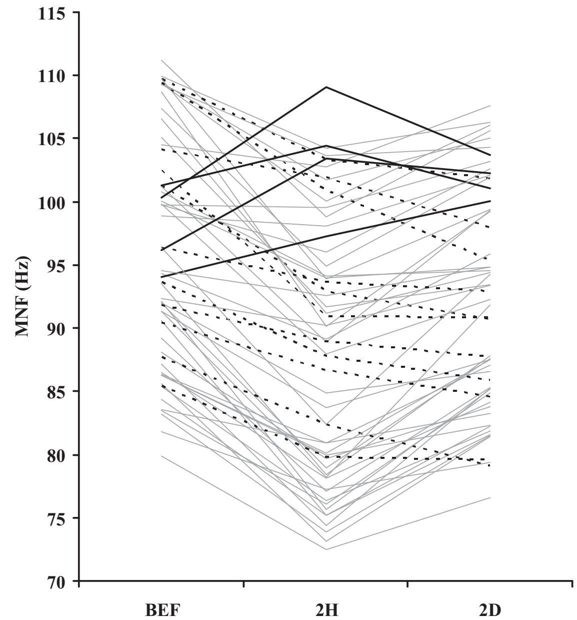

Figure 2: Average changes in MNF in each 59 bipolar channel up to 2D. Most of the channels followed the global MNF changes (grey lines). There was an opposite behavior in four channels (black lines). 11 channels showed a continuous reduction until 2D (dashed lines).

CONCLUSIONS

Spatial dependency can exist in sEMG variables after eccentric exercise. This can be due to some geometrical changes in the muscle, such as IZ shift. Therefore, location of single sEMG electrode over the muscle belly may, in worst case, reverse the physiological interpretations. However, with the use of multichannel sEMG systems, spatially dependent factors can be taken into account and thus improve specificity and validity of the physiological sEMG based interpretations.

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COMPARATIVE STUDY BETWEEN EMG FATIGUE ESTIMATORS

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INTRODUCTION

During the last few years, several studies of muscle fatigue using surface EMG (sEMG) signals have been conducted for different types of exercises, and different parameters have been proposed to estimate it. For example, the mean sEMG signal amplitude, the median and mean frequency (Potvin, 1997), new spectral parameters calculated in the frequency domain (Dimitrov, 2006), and the instantaneous frequency calculated in time-frequency domain (Molinari, 2006). The purpose of this work is to compare and study the adequacy of several classical and new proposed parameters as indicators of muscle fatigue in a dynamical contraction exercise. As a direct fatigue indicator we used the exercise mechanical power.

METHODS

15 physically active male subjects took part in the experiment. The protocol consisted on 5 sets of 10 leg press contractions with the maximum load that the subject could move during 10 exercise repetitions. The rest period between sets were of 2 minutes. The contractions started with 90º and 45º of knee and hip angles respectively and finished with 180º and 90º angles. During the extension actions, sEMG was recorded from the vastus medialis using bipolar surface electrodes. Peak power output was also measured as the maximum of the product between the force exerted against the platform and the velocity movement of the load. The movement was divided in 4 intervals of 22.5º of the knee movement.

The following well-recognized parameters were calculated for the first interval (from 90º to 112.5º of knee movement): mean EMG amplitude (MEA) (Aagaard, 2002), mean (F_{mean}), median frequency (F_{med}) and Dimitrov's spectral index (DSI). We also tested three new parameters, calculated from the Choi-Williams time-frequency distribution: negentropy (Neg), variance of instantaneous mean frequency (F_{mean_var}) and mean of frequency variance (F_{var_mean}) (Boashash, 1992).

RESULTS AND DISCUSSION

During the 50 contractions the peak power decreased significantly as shown in the example of Figure1. In Table 1, we present the correlations between the tested fatigue parameters and the peak power output. The high correlation between MEA and negentropy showed that the negentropy was very much influenced by the amplitude of the sEMG signal, not providing new valuable information.
The mean and median frequencies provided very similar information, as the correlation

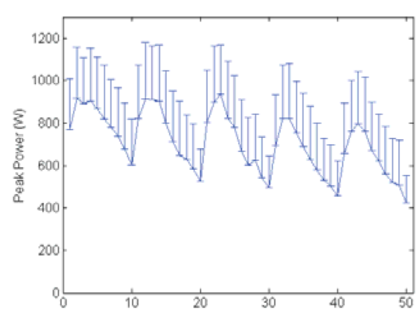

Figure 1: Peak Power (mean and standard deviation) during the 50 contractions

between them was very high. The rest of parameters did not show very high correlation one another, which means that they contained different information about the sEMG signal characteristics.

The parameters with higher correlation with the peak power measurement were F_{mean}, F_{med} and DSI.

SUMMARY/CONCLUSIONS

We studied several sEMG-based classical fatigue index parameters and three new proposed parameters. We compared their behavior in free dynamic contractions using the peak power output as a direct indicator of muscle fatigue.

Dimitrov's spectral index, the median frequency and the mean frequency proved as the best fatigue indicators in our experiments. Moreover, Dimitrov's index was only partially correlated with the mean and median frequency, which means that they carry different and complementary information for muscle detection which could be combined to yield more robust and precise fatigue detection.

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This work has been done with the support of the Spanish Ministry of Education and Science (Plan Nacional I+D+I 2004-2007 Strategic Action: "Sport and Physical Activity". Ref: DEP2006-56076-C06).

Table 1: Normalized correlations between sEMG parameters and peak power output.

	MEA	**F_{med}**	**DSI**	**Neg**	**F_{mean}**	**F_{mean_var}**	**F_{var_mean}**	**Power**
MEA	1	-0.329	0.633	0.966	-0.244	-0.257	-0.464	-0.298
F_{med}		1	-0.501	-0.283	0.879	0.389	0.711	0.407
DSI			1	0.605	-0.421	-0.241	-0.522	-0.428
Neg				1	-0.197	-0.225	-0.396	-0.270
F_{mean}					1	0.397	0.619	0.402
F_{mean_var}						1	0.443	0.213
F_{var_mean}							1	0.271
Power								1

ANALYSIS OF MAXIMAL GRIP FORCE USING EMPIRICAL MODE DECOMPOSITION

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INTRODUCTION

Grip strength is often used in the assessment of elderly and handicapped individuals as it is an indicator of autonomy and overall strength, particularly for the elderly (Ishizaki et al. 2000). High grip strength is associated with independence (Ishizaki et al. 2000), while low midlife grip strength can predict old age disability and mortality (Rantanen 2003). Hand-grip strength is also an indication of risk factors for functional decline in the elderly, such as nutritional level (Wang et al. 2005).

The standard grip-strength measurement is to take the maximal value over a number of tests, which, although reproducible (Li et al. 2008), does not provide much information about muscle function. The recent development of digital devices capable of providing continuous grip-force data combined with innovative signal processing techniques could provide parameters related to underlying muscle function.

The aim of the study was to develop a methodology to identify new parameters obtained from continuous measures of grip force. The physiological significance of the new parameters will be verified using surface electromyography.

METHODS

A pilot group of six male subjects was used for the study, with informed consent obtained from each subject prior to testing. Subjects were tested using the Myogrip dynamometer, which provides a continuous digital signal of grip force (Li et al. 2008).

Subjects performed a 30-s maximal grip strength test using their dominant hand when seated, with shoulders adducted, the testing arm close to the body, elbow extended to 180°, and wrist slightly extended. Subjects were continuously given verbal encouragements throughout the trial.

Empirical Mode Decomposition (EMD) was used to decompose the force signals into Intrinsic Mode Functions (IMF), with further details available in (Huang et al. 1988).

Surface EMG signals from the dominant arm were collected from 10 forearm muscles including flexor carpi ulnaris, extensor digitorum communis, and brachioradialis muscles using In Vivo Metric E263A surface electrodes in a bipolar configuration in the direction of the muscle fibers. The inter-electrode distance was 10 mm. Data was acquired at 1000 Hz with a band pass of 10-500 Hz. Electrode preparations were in accordance with SENIAM guidelines, with locations were based on Perotto's recommendations (Perotto 2005).

RESULTS AND DISCUSSION

An example of the EMD decomposition for a 30-s maximal grip-force test is presented in Figure 1. The decrease in force over the 30-s contraction averaged 39 ± 12 % for the six subjects. Mean power frequency of EMG signals decreased by 24 ± 10 % on average across all muscles for all subjects. Fatigue, as identified by the last IMF of the EMD, decreased by an average of 28 ± 11 %. EMD analysis of maximal grip strength divided force signals into high- and low-frequency components, as well as providing the slope of the fatigue response in the last IMF of the decomposition.

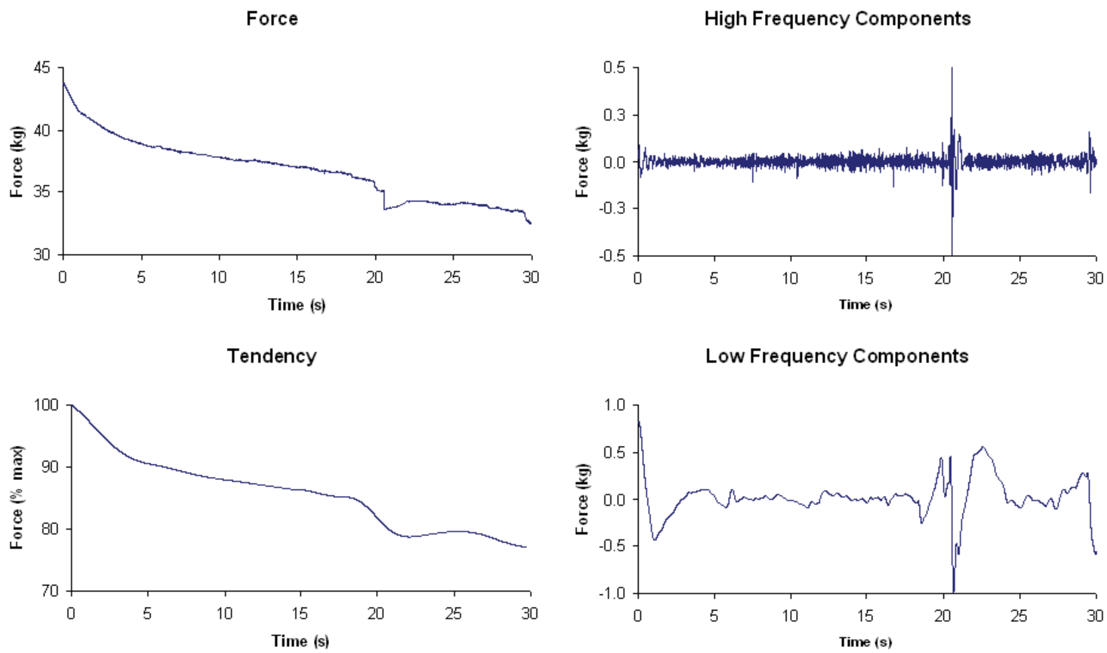

Figure 1: Empirical Mode Decomposition of a 30-s maximal grip-force contraction into constituent low-frequency, high-frequency, and tendency components.

SUMMARY/CONCLUSIONS

The major changes observed in the low-frequency components appear to occur when force varies markedly. Additional analysis is needed to correctly interpret the different components (IMFs) resulting from EMD. It would be of interest to examine this relationship in the context of the EMG signals, particularly with regards to inter-muscular and intra-muscular coordination. Future work will examine sub-maximal force recordings, with one perspective being to detect whether or not subjects are exerting a maximal contraction.

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ELECTROMYOGRAPHY FATIGUE THRESHOLD OF DELTOID AND UPPER TRAPEZIUS DURING SCAPULAR PLANE ARM ABDUCTION

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INTRODUCTION

The prevalence of shoulder disorders can lead to pain and disability (Green et al. 2005).

Shoulder muscle exercises must be properly applied since shoulder muscle fatigue can be associated with alterations in scapulothoracic motion (McQuade et al. 1998).

Furthermore, local muscle fatigue has received attention because it can contribute to the development of chronic muscle pain. This fact has motivated new investigations (Nussbaum, 2001). In this light, determining objective fatigue indexes of shoulder muscles would be relevant in the clinical setting.

The aim of this study was to determine the electromyography fatigue threshold (EMGFT) of the deltoid muscle and upper trapezius during isometric shoulder abduction in the scapular plane on different days and with various loads.

METHODS

Seven healthy men (aged: $\overline{x}$ = 22.3 ± 1.5 years), without musculoskeletal disorders, who did not perform upper trapezius and deltoid training in previous months, took part in this study. A 16-channel surface EMG was used, having a frequency range of 20 to 450 Hz, and total amplification gain of 2000. Each channel was coupled to two active electrodes and one reference. The electrodes were connected to a high impedance preamplifier (1.0 x 10^{12} Ohm), with a common mode rejection ratio of 120 dB. The signals were collected with a sample frequency of 2000 samples per second per channel. Electrodes were placed in the anterior, posterior and medium deltoideus and upper trapezius, in accordance with SENIAM recommendations. The reference electrode was placed in the nondominant wrist, and during abduction a load cell was pulled (traction – 1,961 N).

Three maximal voluntary isometric contraction (MVIC) trials were performed and the peak contraction value was used to determine 50%, 40%, 30% and 20% of MVIC loads at 90° isometric shoulder abduction in the scapular plane. In the first protocol, the loads were carried out in a random order until exhaustion, over four days (i.e., one load per day). The second protocol followed the same sequence of loads as the first protocol, but with all loads performed on the same day. The electromyographic signals were analyzed by root mean square (RMS). The RMS was normalized by the MVIC value of the respective day.

The analysis of variance with repeated measures in general linear model was used to determine the difference in % of MVIC. The significance was set at 5% ($P < 0.05$).

RESULTS AND DISCUSSION

The three portions of the deltoid remained stable during the tests and only the upper trapezius presented a fatigue slope (i.e., significantly different than zero). Therefore, even with the deltoid being an important arm abductor, it was not possible to detect its EMGFT fatigue slope, since subject interruption of the test seemed to be related to the upper trapezius fatigue (Fig. 1).

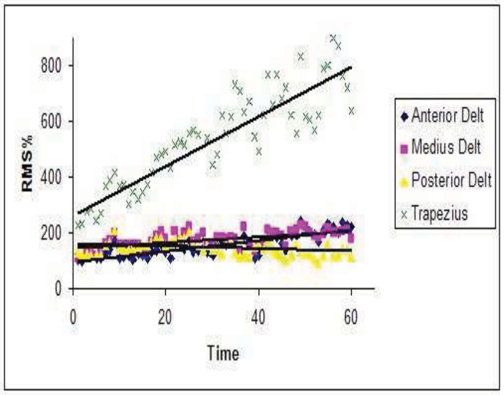

Figure 1: Most common activity of each muscle by time.

The results of EMGFT in different protocols (one single day versus four days) for the upper trapezius did not show any significant difference (Table 1).

	Protocols	
	1 Day	4 Days
EMGFT (% MVIC)	$\overline{X}$ (SE)	$\overline{X}$ (SE)
	16.6 (3.7)	7.7 (4.4)

$\overline{X}$ = mean; SE = Standard Error

Table 1: % MVIC of EMGFT to upper trapezius in each protocol.

Minning et al. (2007) have reported fatigue in the deltoid, but with subjects in a seated position. These different forms of assessments may have caused the differences in the results.

Carpenter et al. (1998) found a decrease in shoulder joint position sense after a fatigue protocol. Therefore, it is of great clinical value to determine a resistance limit to the onset of fatigue, which would serve as an ideal load for shoulder abductor exercises.

Ebaugh et al. (2006) found an increased scapulothoracic motion in relation to external humerus rotation after a fatigue protocol. The upper trapezius fatigue in the present study could account for this alteration in scapulothoracic motion.

CONCLUSION

This protocol presents identifiable fatigue only in the upper trapezius. Tests to evaluate EMGFT on a single day or distributed over four days for the upper trapezius did not show any significant difference.

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EFFECT OF ECCENTRIC STRENGTH TRAINING ON MEDIAN FREQUENCY AND TIME OF FATIGUE IN DIFFERENT LEVELS OF ISOMETRIC CONTRACTION.

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INTRODUCTION

The eccentric contraction has been recommended for strength training (ST) programs because, comparing it to any other contraction modes, it induces more hypertrophy (Kramer et al. 2002) and less metabolic demands (Hugget et al. 2004). The aim of this study was to evaluate the effect of eccentric strength training of knee muscles (extensors and flexors) on median frequency and time of fatigue in different levels of isometric submaximal contractions,

METHODS

The training group, TG (9 men, 62 ± 2 years old) was submitted to 12 weeks of ST for both knee muscles, extensors and flexors (2x/week, 2-4 series of 8-12 repetitions, 70-80% of eccentric peak torque). The control group, CG (8 men, 63 ± 3) did not carry out ST. Both groups were evaluated before and after the period of ST. The maximal isometric peak of torque (PT) of the dominant leg knee extension was tested at 60° of knee flexion (full extension = 0^o) by using an isokinetic dynamometer (Biodex Multi Joint System III, Biodex Medical System Inc., Shirley, NY, USA). The submaximal isometric contractions (SIC) of knee extension (15, 30 and 40% of isometric PT) were performed during 240s or until exhaustion, with a 15-20 min resting period between each level. The EMG signal during the SIC were recorded with bipolar surface electrodes (Ag/AgCl, 1-cm diameter: 20mm between electrodes) that were placed over the muscle vastus lateralis (SENIAM, 1999). Reference electrode was placed on wrist. The EMG signals were amplified (x1000), bandpass filtered (20-500Hz) and digitized at 1000 samples/s. Data analysis: the eccentric PT, the isometric PT and the duration of SIC were quantified before and after the ST period. To evaluated the muscle fatigue and EMG activity were calculated a) the median frequency over the first 30s and the last 30s of SIC. The ANOVA for repeated measures and post hoc Unequal N HSD were used for statistical analysis ($p<0.05$). For the eccentric PT, the isometric PT and the duration of SIC were evaluated the effect of ST (pre *vs* post), group (TG *vs* CG) and interaction between both effects. For the median frequency was evaluated the effect of ST (pre *vs* post training), group (TG *vs* CG), levels of contraction (15% *vs* 30% *vs* 40%), the beginning to the ending of contraction (first 30s *vs* last 30s) and interaction between these effects.

RESULTS AND DISCUSSION

As shown in table 1, no significant changes in the isometric PT and in the execution time of each sub-maximal isometric contraction were observed after the ST in both groups. However, only for TG it was possible to observe the interaction between ST and group effects, i.e. a significant increase on eccentric force, showing the specificity of the training used (Paddon-Jones et al. 2001). The table 2 shows the median frequency. For this variable it was possible to observe the effect of the beginning to the ending of contraction, i.e. for all level of effort, pre and post training, and for both groups the median frequency had a significant decrease from the first 30s to the last 30s of contraction. This result suggests that all SIC causes muscle fatigue (Merletti et al.2004). Also, we observed the interaction between ST and group effect, i.e. the TG had an increase of MF while the GC had a decrease of MF in post ST. This result suggests that ST can reduce the myoeletrical fatigue.

SUMMARY/CONCLUSIONS

Eccentric ST improves eccentric force and reduces the myoeletrical fatigue in SIC but unchanged isometric force and the time of fatigue.

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Table 1: Eccentric and isometric peak torques (PT) and time of fatigue in isometric submaximal contractions (SIC).

	Training group		Control group		p values		
	pre	**post**	**pre**	**post**	**ST**	**G**	**I**
Eccentric PT (N.m)							
Extensor	210 ± 38	253 ± 61	203 ± 33	215 ± 40	ns	0.004	0.008
Flexor	118 ± 25	133 ± 27	126 ± 20	135 ± 26	ns	0.001	0.003
Isometric PT (N.m)	178 ± 25	195 ± 32	172 ± 27	176 ± 26	ns	ns	ns
Time of fatigue in SIC (s)							
15%	240 ± 0	240 ± 0	240 ± 0	240 ± 0	ns	ns	ns
30%	203 ± 55	218 ± 50	189 ± 66	205 ± 57	ns	ns	ns
40%	136 ± 57	145 ± 56	132 ± 67	152 ± 56	ns	ns	ns

Data are reported mean ± SD. PT= peak of torque, SIC= submaximal isometric contraction, G=group effect (training *vs* control), ST=strength training effect (pre *vs* post training), I=interaction between G and T effects, ns=not significant.

Table 2: Median frequency (MF) on the first and the last 30s of contraction.

	Training group				Control group			
	pre		post		pre		post	
MF (Hz	**first 30s**	**last 30s**	**first 30s**	**last 30s**	**first 30s**	**last 30s**	**first 30s**	**last 30s**
15%	73 ± 9	71 ± 7	80 ± 10	77 ± 10	85 ± 11	80 ± 10	80 ± 14	74 ± 9
30%	73 ± 8	70 ± 7	79 ± 10	78 ± 12	84 ± 6	79 ± 14	78 ± 10	75 ± 12
40%	74 ± 9	69 ± 9	83 ± 10	76 ± 12	84 ± 10	75 ± 12	80 ± 10	73 ± 11

Data are reported mean ± SD.

EMG SIGNAL ANALYSIS DURING TREADMILL RUNNING AT ELECTROMYOGRAPHIC FATIGUE THRESHOLD INTENSITY

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INTRODUCTION

For the muscular fatigue analysis during incremental protocol, the EMG signal amplitude (RMS or iEMG) breakpoint has been analyzed in function of exercise intensities (HUG et al., 2006) or in function of the significant increase on the EMG signal amplitude between the beginning and the end of each incremental protocol stage (HANON et al., 1998). Another methodology using surface electromyography had been used by Matsumoto et al. (1991), called electromyographic fatigue threshold (EMG_{FT}) and used in recent studies (OLIVEIRA and GONÇALVES, 2007; SILVA and GONÇALVES, 2006).

In this context, the purpose of this study was to test the hypotheses that the EMG amplitude analysis in incremental treadmill running test allows identify the EMG_{FT}, and validated this index during treadmill running test performed at EMG_{FT} intensity.

METHODS

Ten healthy males (age: 21.8 ± 4.1 years, height: 177.0 ± 3.8 cm, body mass: 72.5 ± 10.2 kg) volunteered for the study and gave their written informed consent to participate in the study, wich was approved by the local Ethics Committee.

In order to determine the EMG_{FT} the subjects first performed an incremental test, and after (minimum 48 hours) were required to run for 1 hour or until to exhaustion at EMG_{FT} velocity. Both tests were carried out on a treadmill with inclination fixed at 1%.

The EMG signal from the right VL muscle was recorded continuously throughout the both tests using a bipolar configuration of pre-amplified Ag/AgCl surface electrodes (MediTrace). The electrodes were applied to the skin and located according to the SENIAM (HERMES et al. 1999).

The EMG signals were obtained using a four-channel biological sign acquisition modulus with input range from -10 to +10 Volts, calibrated with sampling frequency rate of 1000Hz, total gain of 2000 times (20x in the sensor and 100x in the amplifier), high-pass filter of 20 Hz and low-pass filter of 500 Hz. The WINDAQ software was used for the acquisition and analysis of the EMG signals.

During both tests blood samples and HR were taken. The lactate threshold (LT) and the anaerobic threshold (AT) were determined. The RMS values from VL muscle were obtained by specific routine (MatLab), and the EMG_{FT} was obtained according Matsumoto et al. (1991).

Shapiro Wilk, Levene test, and analysis of variance (one-way ANOVA) with Tukey test post hoc were used. For all statistical analyses, the level of significance was set at $p < 0.05$.

RESULTS AND DISCUSSION

Table 1: Velocity values at VL EMG_{FT}, LT and AT thresholds, and maximum values of velocity (Vmax), heart rate (HRmax) and blood lactate concentration ([Lac]max) obtained during incremental treadmill running test.

VL EMG_{FT} ($km.h^{-1}$)	**LT ($km.h^{-1}$)**	**AT ($km.h^{-1}$)**	**Vmax ($km.h^{-1}$)**	**HRmax (bpm)**	**[Lac]max (mmol/l)**
9.3* ± 1.9	9.3* ± 1.3	11.9 ± 1.4	15.3 ± 1.1	198 ± 8.0	9.5 ± 2.1

* $p < 0.05$ significant difference to AT.

Table 1 shows the VL EMG_{FT}, lactate threshold (LT) and anaerobic threshold (AT) values, as well as maximum velocity ($km.h^{-1}$), HR (bpm) and lactate concentration (mmol/l) for each subject. There was no significant difference between VL EMG_{FT} and LT values, therefore, both were significant lower than AT.

During the EMG_{FT} treadmill running test no significant difference was found to the HR (bpm) and blood lactate concentration (mmol/l) values between 10th minute, 30th minute and the end of the test (exhaustion or 1 h). There were no significant changes over time to the VL RMS normalized values (% RMS initial) obtained during EMG_{FT} treadmill running test.

SUMMARY/CONCLUSIONS

In conclusion, the electromyography may be used as non-invasive methodology to analyze muscular fatigue and presents the EMG_{FT} as an aerobic index that may be determined by means of incremental protocol; however, further studies should be conducted in order to analyze whether it could be used for evaluate performance, prescribe exercise intensity and evaluate training effects, as achieved by the LT and AT.

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NEUROMUSCULAR FATIGUE OF THE LUMBAR EXTENSORS INFLUENCES WALKING GAIT PARAMETERS IN HEALTHY WOMEN

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INTRODUCTION

Neuromuscular fatigue influences modifications in spatial-temporal parameters of gross motor tasks. Changes in walking gait parameters may become manifest when introducing external perturbations to the low back system (Arendt-Nielsen et al., 1996; Graven-Nielsen et al., 1997). Since temporal and spatial parameters of the neuromuscular patterns can be modified (Chow et al., 2004; Clark et al., 2007) this would present different loading patterns at the lumbar and lower extremity joints. Thus it is possible that neuromuscular fatigue may be a precursor to joint pain and injury, particularly at the low back.

The purpose of this study was to observe muscle activity patterns and kinematics variables in determining significant differences in walking gait. It was hypothesized that fatigue of the paraspinal muscles would facilitate modified gait parameters in healthy individuals.

METHODS

Fourteen healthy female volunteers (age: 27.5 ± 12 yrs, body mass: 62.2 ± 7 kg, height: 1.67 ± 0.07 m) participated in this study. None of the participants reported a history of low back or lower extremity pain/disorder within the last year. Each participant was required to perform the protocol in two sessions, separated by 7 days. Participants walked across a 10 m walkway at their preferred walking velocity four times before and after the fatigue protocol. Infrared sensors positioned at waist height in the middle 3 m of the walkway were used to collect walking stride and velocity data.

The fatigue protocol included testing the maximal trunk extension effort (MVE) of each individual while positioned at 30 degrees from erect stance. Participants then performed static trunk exertions at 50% or 80% of MVE for either five minutes or two minutes, respectively.

Surface electromyogram (EMG) signals were collected from the lumbar paraspinals (LP) (L3), rectus abdominis (RA), external oblique (EO), rectus femoris (RF), biceps femoris (BF), and lateral gastrocnemius (LG) on the individuals' right side. EMG signals were bandpass filtered 20-500 Hz, with a CMRR of > 90 dB at 60 Hz and an impedance of > 10 MΩ and sampled at 1000 Hz. EMG signals were subsequently centered, full-wave rectified, and digitally smoothed with a fourth order Butterworth filter at 4 Hz.

Foot switches were adhered to the plantar surface of the heel and first metatarsophalangeal joint of the forefoot to determine heel contact and toe-off instances during the gait cycle. These signals were time synchronized with the EMG.

Walking gait cycles were normalized to 100% (common number of 100 data points). EMG amplitudes were normalized to the initial peak amplitude from the first walking trials.

Intrasubject analyses of the dependant variables walking velocity and stride time were performed using a repeated measures one-way ANOVA. Timing of EMG peaks per muscle group during walking (> 20% of peak amplitude), and the EMG peak amplitudes were also analyzed within subjects. Alpha was set at 0.05.

RESULTS AND DISCUSSION

Stride time was not significant in the 50% session, but decreased significantly in the 80% session from 0.98 (0.05) to 0.969 (0.05) s ($p < 0.05$). Mean velocity did not change within either 50% session (1.75 ± 0.27 vs. 1.66 ± 0.5 m/s, $p > 0.1$) or 80% session (1.69 ± 0.15 vs. 1.60 ± 0.45 m/s, $p > 0.1$).

EMG amplitude significantly increased in the first peak (heel contact) of the LG muscle in the 50% session (0.375 ± 0.24 vs. 0.39 ± 0.3 mV, $p < 0.001$). Amplitude of the EMG signal decreased significantly in the RF muscle in the 80% session (0.34 ± 0.19 vs. 0.25 ± 0.12, $p < 0.02$). Significant changes in other muscle groups were not observed in either session.

A significant timing of the peak LP amplitude was observed during the corresponding contralateral heel contact/ipsilateral pre-wing phase (52.6 ± 12.9 vs. 55.5 ± 12.2 % of the gait cycle, $p < 0.001$). Significant timing modifications of corresponding peaks during heel contact, toe-off, or limb transitions during the gait cycle were not observed.

SUMMARY/CONCLUSIONS

It was hypothesized that timing and amplitude changes in the muscles assisting with walking gait would be modified after the performance of sub-maximal fatiguing of the trunk extensor muscles. Stride time was reduced statistically, however this may not be physiologically significant to the system's overall task performance. Amplitude changes in the LG and RF muscles indicate a possible modification in neuromuscular recruitment strategies of the system. Timing of the peak amplitudes during the gait cycle may provide more conclusive evidence to the influence of neuromuscular fatigue on gait patterns. Future research should investigate the influence of neuromuscular fatigue on the spatial-temporal kinematics of the trunk and lower limbs in determining gait modifications along with EMG parameters.

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CLINICAL TEMPORAL ANALYSIS OF HAND GRIP STRENGTH IN ATHLETES

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INTRODUCTION

Many sports use the hand as a tool to achieve victory. The hand is a tool of the human body used in many situations and sport activities; its movements with a high degree of skill, strength and endurance, together, enable achieve a good result in sports performance.
The occurrence of fatigue in muscles responsible for the hand grip strength in decisive moments during sport activities is a problem that can lead to the athlete defeat. Knowledge of the hand grip strength behavior on athletes is not really wide (NICOLAY and WALKER, 2005).
There are no standardized and validated hand grip strength evaluation protocols that discriminate the type and duration of muscular contraction. Also, the protocols do not differentiate these characteristics (type and duration of muscular contraction) between sports and there are just a few specific sports modalities recorded (HENNIG and SCHNABEL, 1996).
The aim of this study was to evaluate the hand grip strength performance of maximum continuous contraction during two minutes using the maximum strength and the strength decrease variables (difference between maximum and final strength) between the dominant and non-dominant hand in different sports and non-athletes.

METHODS

The subjects of this study were 29 adult male athletes who use the hand grip movement in the sports practice (six individuals from aikidô, six from jiu-jitsu, eight from judo and six individuals from rowing) and 16 non-athletes subjects.
The inclusion criteria in this study were: the subject should not present problems with the radiocarpal, carpometacarpal, metacarpophalangeal and interphalangeal joints.
The handgrip dynamometer (figure 1) was used for assessment of hand performance; it (mechanic structure, hardware and software) has been developed in the Laboratório de Instrumentação - LABIN/CEFID/UDESC. The instrument has continuous adjustment of grip aperture, load limit of 800 N, excellent coefficient of linearity ($R^2 = 0.9999$), 8 bits resolution and 1000 Hz as maximum sample rate.
The subject position during the evaluation was based in American Society of Hand Therapy Protocol (BOHANNON, 1991 in HAIDAR et. al, 2004). The subject was seated with the elbow maintaining an angle of 90 ° of flexion, the grip aperture of the dynamometer with forearm on half pronation and wrist neutral, and it could be moved up to 30 ° degrees of extension. The subject's arm was suspended in the air with his hand placed on the dynamometer that is sustained by the evaluator (figure 1).
The grip aperture used was 0.055 m (RUIZ-RUIZ et al, 2002), adjusted in case of discomfort of the assessed subject. The subjects began and ended the test after verbal command. The sample rate used was 100 Hz.
For purposes of this investigation, the maximum strength and strength decrease differences between dominant and non-

dominant hand was defined by paired t test and differences between groups was defined by analysis of variance (ANOVA) with bonferroni post hoc. Statistical analyses were performed using SPSS® 11.0 software (SPSS Inc. Chicago, Illinois).

RESULTS AND DISCUSSION

The table 1 shows de descriptive results of dominant maximum strength (Dmax), non-dominant maximum strength (NDmax), dominant strength decrease (DDecrease), non-dominant strength decrease (NDDecrease).

Figure 1: subject position during the evaluation

The result of paired t test shows significant differences of the maximum strength between dominant and non-dominant hand in the groups: aikidô (106 N, p=0.05), non-athletes (45 N, p=0.004) and jiu-jitsu (64 N, p=0.03).
We have found significant differences of the strength decrease between dominant and non-dominant hand in the groups: jiu-jitsu (-4%, p=0.001) and non-athletes (4.6%, p=0.04).
The result of Anova shows significant differences of the dominant strength decrease between rowing x non-athletes (-7.86%; p=0.033).
For maximum strength of the dominant hand, significant differences were found between groups: jiu-jitsu x rowing (132 N, p=0.005), jiu-jitsu x aikidô (142 N, p=0.002) and jiu-jitsu x non-athletes (121N, 0.002).
There are also significant differences of the non-dominant maximum strength between groups: jiu-jitsu x aikidô (184N, p=0.002) and rowing x aikidô (138N, p=0.03).

CONCLUSIONS

The results show that the values of the strength decrease are virtually the same, between dominant and non-dominant hand and between groups. The maximum strength also presented differences between dominant and non-dominant hand and between groups. These data can be used as parameters for evaluation of performance and asymmetry contributing for the treatment of injuries and methods of specific training for each sport.

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Table 1. Descriptive results (mean± SD).

	Aikidô	Jiu-jitsu	Judo	Rowing	Non-athletes
Dmax(N)	433.4±68.6	575.8±43.3	494.4±48.9	443.7±42.5	454.7±+71.0
NDMax(N)	327.4±77.6	511.8±66.6	442.6±95.1	466.2±93.1	409.5±60.7
Ddecrease(%)	77.6±5.3	76.1±3.7	75.7±6.4	71.5±6.4	79.4±4.6
NDDecrease(%)	76.0±3.6	80.1±18.8	76.8±7.5	71.8±4.3	74.8±9.4

MYOELECTRIC FATIGUE PROFILES DURING ELECTRICALLY-ELICITED CONTRACTIONS OF VASTUS LATERALIS, VASTUS MEDIALIS OBLIQUUS, AND VASTUS MEDIALIS LONGUS MUSCLES

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INTRODUCTION

The analysis in the time and spectral domains of the surface EMG signal detected during selective electrical stimulation of the motor point of a muscle allows the assessment of the myoelectric manifestations of fatigue, which are useful to differentiate between muscles with different characteristics.

METHODS

In eighteen healthy male subjects (nine sedentary subjects and nine rowers) surface EMG signals were detected during isometric electrically elicited contractions of three muscles of the dominant thigh, in the following order: vastus medialis obliquus (VMO), vastus lateralis (VL), and vastus medialis longus (VML). A linear array of eight electrodes (interelectrode distance 5 mm), placed over the VMO muscle, and two bidimensional arrays of 30 electrodes (5 columns x 6 rows, interelectrode distance 8 mm), located over the VL and VML muscles, were used for the detection of the compound muscle action potentials or M-waves. M-waves were recorded as the muscles were stimulated at 25 Hz for 60 s with one cycle of a biphasic sinusoidal waveform of 152-μs duration. A recovery period of five minutes was applied between each contraction. Examples of M-waves detected from the three muscles in one subject during the first and last second of stimulatio n are reported in Figure 1. Rate of change and normalized rate of change (calculated as the percentage ratio between rate of change and initial value) of muscle fiber conduction velocity (CV) were adopted to assess myoelectric manifestations of fatigue.

RESULTS

In the comparison of the three muscles, CV rate of change and normalized rate of change were significantly (Kruskal-Wallis test, $p<0.01$) greater for the VL muscle with respect to the VM muscles, whereas no significant difference was found between VMO and VML muscles [median (interquartile range), *CV rate of change*: VL, -0.031 (0.006) m/s^2; VMO, -0.025 (0.008) m/s^2; VML -0.026 (0.007) m/s^2; *CV normalized rate of change*: VL, -0.61 (0.11) %/s; VMO, -0.50 (0.19) %/s; VML, -0.47 (0.13) %/s].
CV rate of change and normalized rate of change for the VML muscle were significantly (Kruskal-Wallis test, $p<0.01$) greater for the sedentary subjects with respect to athletes [median (interquartile range), *CV rate of change*: sedentary subjects, -0.028 (0.005) m/s^2; athletes, -0.021 (0.007) m/s^2; *CV normalized rate of change*: sedentary subjects, -0.50 (0.16) %/s; athletes, -0.38 (0.12) %/s].

SUMMARY/CONCLUSIONS

In summary, myoelectric manifestations of fatigue were greater for the VL with respect to VM muscles, in agreement with the known histochemical difference between the vasti muscles. Furthermore, the VML muscle of rowers resulted less fatigable with respect to sedentary subjects, whereas no differences were found between rowers and sedentary subjects in the myoeletric fatigue profile of VMO and VL muscles.

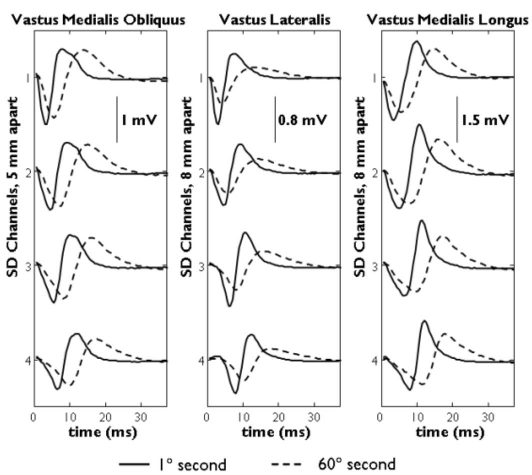

Figure 1: Four channels of M-waves detected (from one subject) on one side of the innervation zone of vastus medialis obliquus, vastus lateralis, and vastus medialis longus, stimulated for 60 s at 25 Hz. For each muscle, the continuous line shows the array of (averaged) waves detected during the first second, whereas the dotted line shows the array of (averaged) waves detected during the last second of electrical stimulation. A widening of the single differential (SD) M-wave from the beginning to the end of the contraction is evident for each muscle as a function of CV reductions (VMO: 1° s: 5.1 m/s, 60° s: 3.8 m/s; VL: 1° s: 4.7 m/s, 60° s: 3.6 m/s; VML: 1° s: 6.1 m/s, 60° s: 4.8 m/s) that are possibly related to a progressive separation of the values of conduction velocity between different types of motor units.

ACKNOWLEDGEMENTS

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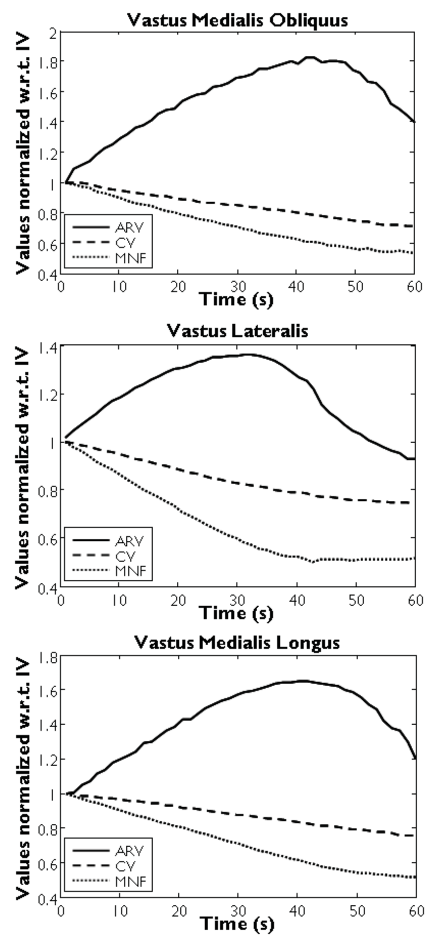

Figure 2: Fatigue plots showing myoelectric manifestations of fatigue for the three muscles (from the same subject of Figure 1). ARV: average rectified value; MNF: mean frequency of the power spectrum; CV: muscle fibre conduction velocity. Rate of change of CV is estimated as slope of the regression line (fit over 30 s) divided by the Initial Value (IV).

ELECTROMYOGRAPHIC ASSESSMENT PRE TO POST SURGERY OF THE ERECTOR SPINAE IN PATIENTS WITH HERNIATION OF NUCLEUS PULPOSUS

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INTRODUCTION

The surgical treatment of a herniated nucleus pulposus (HNP) results in the rapid relief of neurological symptoms and pain, however controversy exists about when complete muscular recovery occurs due to post surgical evidence of persisting low back pain syndromes and deficits in motor control.

The aim is evaluate the electromyographic activity of the erector spinal muscles in an isometric test of sub maximal resistance in patients with HNP pre and post hemilaminectomy decompression surgery (HMLD).

METHODS

The fatigue capacities of the muscles iliocostalis lumborum (ILI) and multifidus (MUL) were assessed using electromyography (EMG) in 10 patients diagnosed with posterolateral HNP at L4-L5 or L5-S1 admitted in the spine section of the Hospital del Trabajador Santiago to be submitted to a HMLD.

A spinal isokinetic device was used to maintain isometric extension at an individually predetermined torque. Recordings were taken at submaximal isometric resistance at 20 per cent of total body weight. Of the obtained EMG signals, the indices of local muscle fatigue were determined and compared for possible bilateral myoelectric changes. Pre and post surgical pain scores on the visual analog scale (VAS) were also recorded.

The non parametric test, the Sign Test, with a significance level of $p<0.05$ was performed using specialized software (Minitab Release 14). Average values and standard deviations of the normalized fatigue indices were calculated in both muscles.

RESULTS

No statistically significant differences were seen in the index of muscular fatigue of the Iliocostalis Lumborum muscle pre and post surgery ($p=0.0625$) on the side affected and non affected by HNP. The same was seen for the Multifidus muscle (AS $p=0.0625$, NS $p=0.5$). Pain ratings on the VAS had a tendency to decrease post surgery.

The graph 1 and 2 shows the tendency of the MFslope for each muscle group. Graphs 1 show a lesser tendency of the ILI to fatigue after surgery in both the affected and non affected sides. This is similar to graph 2 where a tendency towards greater resistance to fatigue is shown for the MUL on the non affected side, and a tendency towards greater resistance to fatigue post surgery.

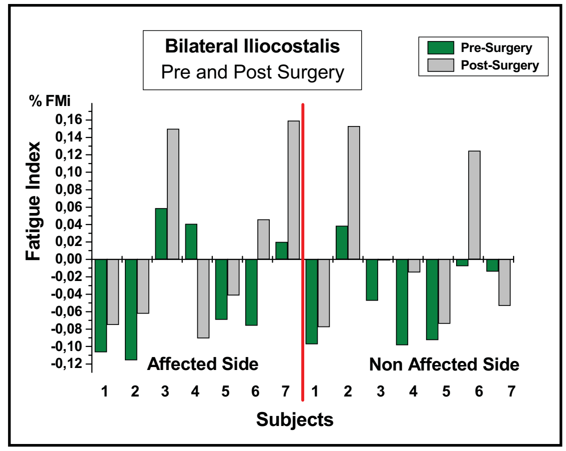
However for the MUL on the affected side (graph 2), heterogeneous behavior can be seen.

DISCUSSION

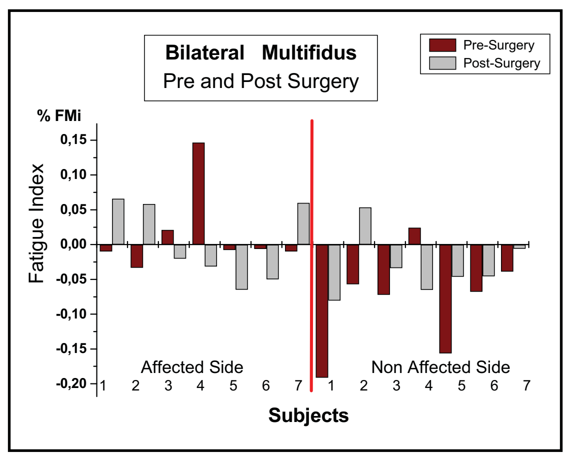
The laterality of the HNP has relationship with the decrease of the bilateral spinal erector resistance, the surgery not. Both they are related with adaptative strategies of the CNS for to keep the indemnity of the spine and an alteration mechanisms of lumbar motor control (M. Solomonow. 2002).

The aim of this study was to investigate the influence of surgery on the fatigue resistance of the paraspinal muscles. Although no statistically significant differences were found in the behavior pre and post surgery, these results show a tendency towards a general improvement in the fatigue resistance post surgery. This may indicate that the damage caused by the HNP not only affects the ipsilateral musculature, but also the contralateral in a similar way, and although the muscle damage generated by the HNP and the surgery are important they are not necessarily linked to the resistance capacity of the muscle but instead with an adaptive response to damage designed by the CNS to protect the stability function of the paraspinal muscles (M. Panjabi, 2003).

As our study is of small sample size, it is not possible to give a tangible conclusion on this topic. However, this study can serve as a guide, describing the events that occur at this level in the presence of acute lumbar spine pathology and based on that which has been established in the literature, may indicate a new factor to consider in the rehabilitation of this type of patient. The focus on improving the stability, strength and or muscular resistance should be directed together towards the restoration of normal motor control, both on the side affected and non affected by HNP in order to avoid the advance of changes in neuromotor control strategies following an acute episode of LBP that have the potential to lead to chronicity.



Graph 1: Shows the indexes of muscular fatigue for the Ilicostalis Lumborum (p=0.0625) affected side and non affected side pre to post surgery.



Graph 2: Shows the indexes of muscular fatigue for the Multifidus affected side (p=0.5) and non affected side (0.0625) pre to post surgery.

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A FATIGUE STUDY IN AUXILIARY NURSES

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INTRODUCTION

Auxiliary nurse (AN) work is associated to musculoskeletal disorders (MSD) affecting the low back and the scapular hip. In a previous study we could not found that the prevalence of high risk of MSD using OWAS procedure was less that 0.05. However, among this group of workers we observe a large prevalence of MSD complains and work related injuries are frequent. For this reason we decided to set up a new study examining the musculoskeletal load in the back muscles.

Muscular fatigue can be explored physiological by means of the change in the spectral components of the surface electromyography (SEMG). It is widely accepted that a muscle unit as it fatigues the power spectral density mean or median moves towards the low frequency ones. As a consequence, a fatigued muscle has a lower mean or median and the speed of the median decay will be faster.

METHODS

We presented the hypothesis and the study proposal to the Occupational Committee for acceptance and assistance. Subject were 22 voluntary AN, 10 belonging to a residential nurse (all female mean age 42.3 (24-59) and 12 from a Stroke Unit in a General Hospital (9 females age mean 40 [37-61], and 3 male mean age 39 [37-41]) that had been out of work at least for 3 days. We excluded pregnant women and those bearing a MSD that prevented form doing the tests. One worker was excluded at the beginning of the study because of pain and trochanteritis and a second one left the study because of back pain. Work load was similar in all workers in each work environment. After the purpose of the study was explained they were asked to lay supine for 10 seconds resting on the birth, to obtain a basic SEMG and then to sustain, for 90 sec. the upper body suspended in horizontal with the hands crossed in their neck for a constant isometric contraction. Previously 4 surface electrodes were placed bilaterally over the erector spinae muscles dorsalis (3 cm away from the apophysis spinosa at T9) and lumbaris (3 cm away of the apohysis spinosa L3) and the neutral electrode in the epycondile. Subjects were recorded at the beginning and end of the working shift and in the nursing home in day one and three of their workin cycle. We used DelSys SEMG hardware for the signal adquisition, and Delsys software for the analysis of the signal and SPSS 12.0 for the statistical evaluation. We discarded those signals that did not achieve the standards of quality. At the end we used 16 workers (80%) for the analysis of the erector dorsalis (ED) and only 11workers (55%) for the lumbaris (EL). We calculated the initial median frequency (MDF) as the mean of 20 seconds at the beginning of the study, once discarded the first 10 seconds. The association between MDF and time was examined by contrasting the regression lines in the different test before and after the work day and at the beginning and end of the work cycle by means of contrasting the regression coefficient with an alpha level of $p<0{,}05$.

SUMMARY/CONCLUSIONS

The different muscles behaved differently across subjects and studies Tables 1 and 2. Not all fatiguing test could produce a negative slope in the MDF. Only between the 18% and 33% of the muscles demonstrated a significant decrease in the starting level of the MDF before and after the day. The increase in the negative slope was more frequent than not in most of the muscles, but it was present in more that 50% of the tests only in the RED when it was tested for significance. The question here is if this frequency of fatigue found is of any clinical interest. In a preliminary study we could not find any increase in the negative slope in 4 control subjects after a day work. It seems, from these data that most likely, fatigue is a problem between the NA. In the other side, we were surprised to find that in 6 muscles test out of 54 we found that the slope after a day work was less negative, as if they fatigued less after a day of work. In summary, from our data, and those of other that did a similar study, we are prone to conclude that the there is an accumulation of efforts done with the muscles of the back that could result in MSD

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Table 1. SEMG results of the fatigue study on the Stroke Unit

Muscle group	**Dorsal muscles**		**Lumbar muscles**	
Muscle side	**LED**	**RED**	**LEL**	**REL**
Significant negative slope of MDF in the fatiguing test	9/11(82%)	9/11(82%)	9/9(100%)	9/9(100%)
Increase on the slope after day work	6/11(54%)	9/11(82%)	6/9(67%)	4/9(44%)
Significant increase on the slope after day work	4/11(36%)	7/11(63%)	4/9(44%)	2/9 (22%)
Decrease in the starting MDF after a day work	2/11(18%)	3/11(27%)	3/9(33%)	2/9(22%)

Table 2. SEMG results of the fatigue study on nursing home

Muscle group	**Dorsal muscles**		**Lumbar muscles**	
Muscle side	**LED**	**RED**	**LEL**	**REL**
Significant negative slope of MDF in the fatiguing test	10/12(83%)	11/12(91%)	6/6(100%)	6/6(100%)
Increase on the slope after day work	9/18(50%)	13/18(72%)	5/15(33%)	4/12(25%)
Significant increase on the slope after day work	8/18 (44%)	10/18(55%)	3/15(20%)	4/15(27%)

LED left erector dorsalis; RED right erector dorsalis; LEL left erector lumbaris; REL right erector lumbaris

THE EFFECT OF HYPERTHERMIA AND PROLONGED SUBMAXIMAL CYCLING ON NEUROMUSCULAR CONTROL OF MAXIMAL VOLUNTARY CONTRACTION OF THE KNEE EXTENSORS

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INTRODUCTION

It is likely that a major cause of premature endurance exercise fatigue in a hot environment has been governed by the CNS (Nybo 2007). From this it has been suggested that this fatigue acts as a protective mechanism to protect the body from excessive damage (Noakes 2000). It has been demonstrated (Nybo and Nielson 2001) that voluntary muscular activation is reduced during hyperthermic conditions as is neuromuscular recruitment (Tucker et al 2004). However, an important neuromuscular factor is muscle fibre conduction velocity (MFCV) which is a good indicator of motor control strategies (Andreassen and Ardent-Nielson 1987). It has been previously demonstrated that passively heated muscle results in elevated MFCV(Farina et al 2005; Gray et al 2005) as well as an increase in intramuscular lactate during a maximal 6 second sprint (Gray et al 2005). Conversely, it has also been shown that a reduction in intramuscular pH results in a concomitant decline in MFCV (Brody et al 1991). Therefore, it is likely that accumulation of lactate from high intensity submaximal exercise in the heat will indirectly inhibit the increased muscle temperature effect on MFCV.
Accordingly, the aim of this study was to determine if inducing hyperthermia would alter MFCV after 50 minutes of high-intensity cycling.

METHODS

8 healthy, well trained recreational cyclists (mean $\pm$ SD age, VO_{2max} and body mass 35 $\pm$ 9.9, 57.4 $\pm$ 6.6 $ml.kg^{-1}.min^{-1}$ and 76.4 $\pm$ 10.8 kg) on 3 separate occasions in random order completed; 1) 50 minutes of cycling at 60% of their peak power output (HOT); 2) rested for 50 minutes in 40°C, 35% humidity (PASS). and 3) repeated the same cycle protocol in 19°C, 20% humidity (NEUTRO). Before and after this took place all subjects had their muscle temperature of the Vastus Lateralis recorded after which they were seated on the isokinetic dynamometer to perform 3 x 6 second maximal isometric contractions of the knee extensors (MVC). Following the second set of contractions the subjects were also instructed to perform a 100 second MVC (SMVC). During the contractions we measured EMG amplitude (RMS), MFCV and force. During the 50 minutes for HOT, PASS and NEUTRO we measured rectal temperature, venous lactate and heart rate.

RESULTS AND DISCUSSION

The HOT condition successfully induced hyperthermia by demonstrating significantly ($p<0.05$) higher core and muscle temperatures (Figure 2), lactate values (Figure 3) and heart rate (mean cycle 15 $\pm$ 3 %) than the PASS and NEUTRO. This resulted in significant decline in force during SMVC in the HOT condition (Figure 1C). This decline is likely to be partly from a reduction in motor unit recruitment as shown by the RMS data which revealed a slight interaction ($p=0.083$) in the decline between conditions and a significant ($p<0.05$) reduction for the HOT condition at 75 minutes (Figure 1B). There was highly significant ($p<0.01$) increase in MFCV (pre to post 3.8 ± 0.3 vs 4.7 ± 0.3 $m.s^{-1}$) following PASS. However, during SMVC the HOT condition showed no difference to NEUTRO for MFCV (Figure 1A). This would suggest that any increase in MFCV bought about by increased muscle temperature is likely to be offset by a decline in pH from an increase in metabolites as evidenced by the significantly ($p<0.05$) higher lactate values shown in the HOT condition (Figure 3).

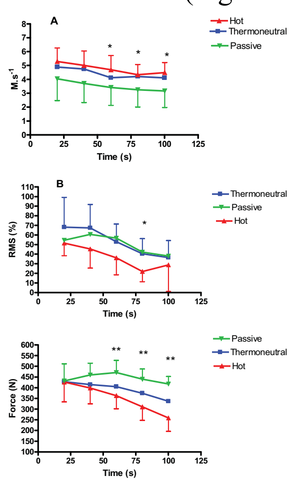

Figure 1 100s MVC post 50min cycle for; A. Muscle Fibre Conduction Velocity (MFCV) *$p<0.05$ passive slower than hot; (B), EMG amplitude (RMS) $p = 0.083$ interaction between conditions. *-$p<0.05$ hot less than thermoneutral and C force were highly significant ($p<0.01$) between the conditions, over time and interaction with each other.

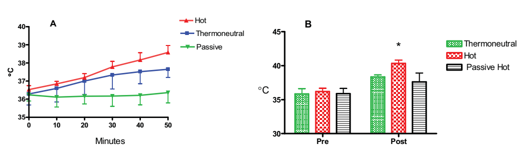

Figure 2 Temperature taken from the A. Core (rectal) taken before and during the 50 min cycle Significant ($p<0.05$) differences are shown between all 3 conditions and B Muscle of the Vastus Lateralis taken before and after the 50 min cycle. Muscle for All muscle temperatures highly significantly ($p<0.01$) increased after all 3 conditions while the post Hot temperature was significantly (*$p<0.05$) greater than the other 2 conditions.

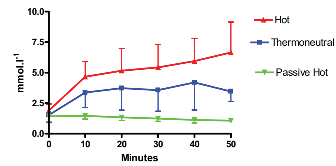

Figure 3. Blood lactate collected before and during the 50 min cycle for Hot (38˚C) and Thermoneutral (19˚C) conditions and 50mins passive rest (38˚C conditions). Significant ($p<0.05$) differences are shown between all 3 conditions

SUMMARY/CONCLUSIONS

This study has shown that despite elevated muscle temperature MFCV is reduced due to a likely increase in metabolites following high intensity submaximal exercise. The decline in force production as a result of hyperthermia may be partly due to a reduction in motor unit recruitment.

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Track 09

Neurophysiology (NP)

THE EFFECT OF A PRE-CONDITIONING ACTIVATION ON MUSCLE FIBER CONDUCTION VELOCITY AND TWITCH FORCE DURING DOUBLET SUPRAMAXIMAL STIMULATION

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INTRODUCTION

The conduction velocity (CV) of muscle fiber action potentials and the muscle twitch force depend on the history of activation of the fiber membrane. The velocity recovery function (VRF) and the twitch recovery function (TRF) of muscle fibers have been separately investigated during doublet and triplet activation (Griffin L. et al. 2002, Kamavuako E.N. et al. 2007, Karu Z.Z. et al. 1995). However the relationship between changes in CV and twitch force when varying inter stimulus interval (ISI) has been less explored.

Aim: The aim of the study was to investigate the effect of a conditioning activation on the VRF and TRF.

METHODS

The experiments were conducted on 8 healthy men (20 - 32 yr, mean 24.7 yr). Compound muscle action potentials (CMAPs) were recorded from the tibialis anterior muscle and were elicited with electrical stimulation of the peroneal nerve below the fibular head. The EMG signals were detected in bipolar derivation with a linear adhesive array of 8 electrodes (5-mm inter-electrode distance) located between the innervation zone and distal tendon. Signals were amplified 1000 times, filtered (20-500 Hz) and sampled at 10 kHz. The right leg was fixed to a pedal for twitch force measurement. The force signal was acquired in the bandwidth 0.01 Hz - 100 Hz). Stimulation intensity was set to 150% of the intensity that produced the maximal CMAP. In the first part of the experiment, the VRF and TRF were assessed from doublet stimulation: a stimulus (S1) was first delivered alone (single activation) and then delivered with a second delayed test pulse (S2) that followed S1 at varying ISI (doublet stimulation) (Fig. 1*A*). Twenty two ISI values were randomly investigated (4 - 1000 ms). CMAPs were analyzed off-line immediately after the stimulation sequence and the RFs were obtained. The VRF was used to determine two inter pulse periods (IPPs) to be used in the second part of the experiment (Fig. 1*B*,C). The first IPP corresponded to the peak of the VRF (IPP1) and the second to 50% of the peak value, on the right side of the peak of the VRF (IPP2) (Fig. 1*D*).

In the second part, two triplet stimulation sequences were investigated, fixing the first and second interval using IPPs. For the same ISI values as in the first part, a custom written LabVIEW program controlled the following sequence of stimulations: 1) a single stimulation, 2) doublet stimulation, 3) doublet stimulation with fixed IPP1, 4) triplet stimulation using IPP1 as first interval, 4) doublet stimulation with fixed IPP2. Digital subtraction was applied to separate the CMAPs and TF. The VRF/TRF was the ratio between CV/twitch peak (TP) of S2/S1 and CV/TP of S1/S0 at each ISI.

RESULTS

The RFs of the triplet were significantly different from the RFs of the doublet for some of the ISI (F = 11.65, *P* = 0.006; Fig. 2*A*). The effect of a preconditioning activation was significant for ISI below 125 ms and 40 ms for CV and TP respectively (Fig. 2*B*).

CONCLUSIONS

Both CV and twitch force depends on the distance with respect to previous activation. On both variables, a pre-conditioning stimulus has the effect of prolonging the intervals of ISI where the maximal value is reached (Fig. 2A).

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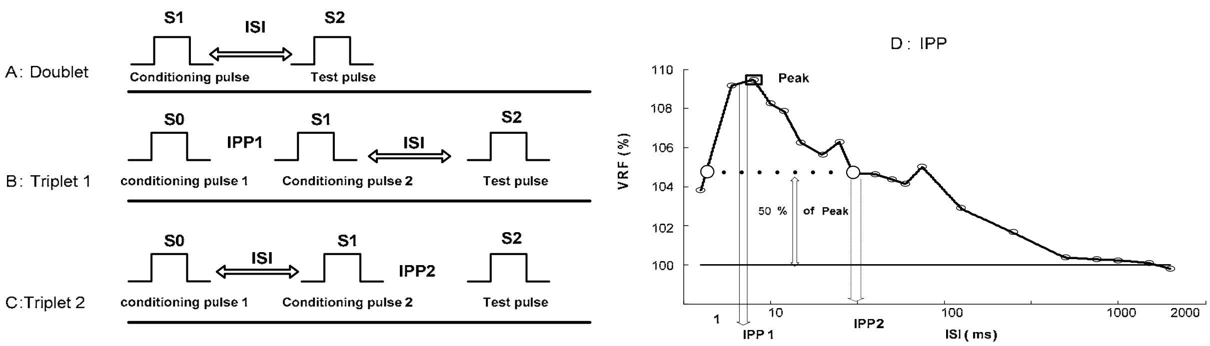

Figure 1: Experimental conditions. A: Doublet activation with varying the time interval (ISI) between stimuli S1 and S2. B: Triplet activation with fixed IPP1. C: Triplet activation with fixed IPP2. C: Selection of inter pulse periods (IPPs).

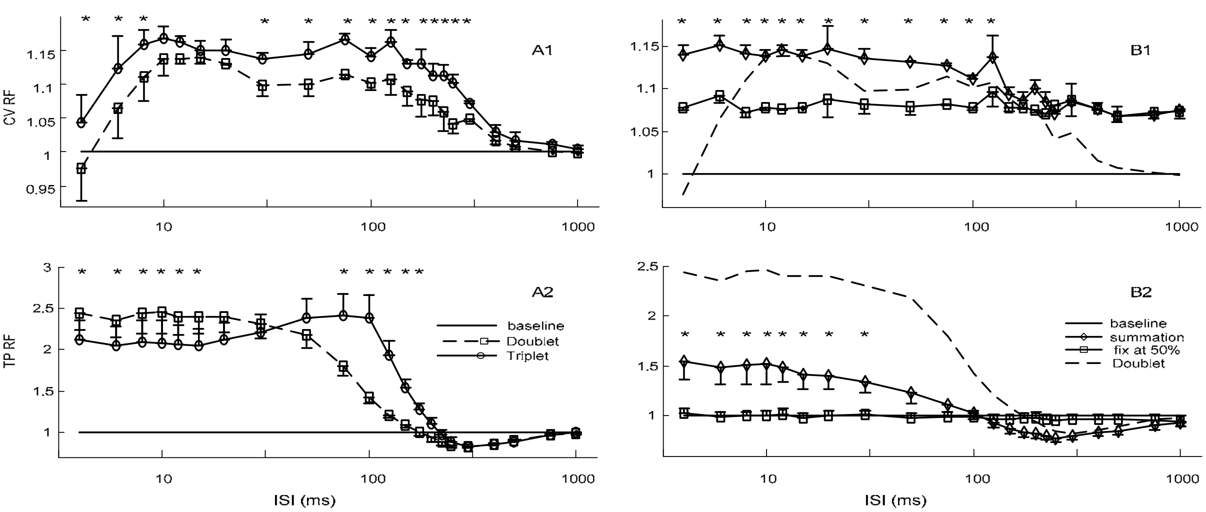

Figure 2: Mean ± SE for A1) VRF and A2) TRF for doublet (dash lines with square) and triplet with fix IPP1 (open circles); and B1) VRF and B2) TRF for doublet (squares) and triplet (diamonds) at fix IPP2. The dashed line is the mean RF for doublet. Asterisks indicate ISI values for which the triplets were significantly different from the doublet. The abscissa is in logarithmic scale (base 10).

SELECTIVE INCREASE IN CORTICOSPINAL EXCITABILITY WITH TACTILE EXPLORATION

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INTRODUCTION

Active contraction is known to lead to facilitation of motor responses evoked from transcranial magnetic stimulation (TMS) of the motor cortex. This facilitation of motor evoked potentials (MEPs) is thought to reflect an increase in excitability both at the spinal and cortical level (ie, number and size of descending volleys). While MEP facilitation could be elicited with simple finger movements, several studies have shown that the magnitude of this facilitation could be modulated in a task-dependant manner. For instance, Bonnard et al showed that when the precision demands increased for the thumb-index grip force, corticospinal (CS) excitability also increased selectively in either the 1st dorsal interossei (FDI) or the abductor pollicis brevis (APB), but not in both. Sale et al (2005) reported similar findings showing a selective increase in MEP facilitation in the FDI during a scissor action as compared to isolated contraction or a power grip.

In the present study, we sought to determine whether engaging the fingers in tactile exploration would enhance CS excitability in intrinsic hand muscles when compared to simple motor tasks.

METHODS

Eighteen healthy young participants volunteered to participate in this study (age, 24 ± 2 yrs, 9 males, 9 females).

Corticomotor excitability of the hand motor representation was assessed under two task conditions: 1) Tactile Exploration (TE) wherein subjects were asked to explore raised letters (6 upper case characters, 6-mm high) with the tip of the index finger for tactile recognition; and 2) Button Pressure (BP), wherein subjects depressed a button with the index finger. Five subjects were tested under an additional condition, ie, scanning a smooth surface identical in size to the surface used for tactile recognition using a similar exploratory movement. In all conditions, TMS pulses (Magstim 200, figure-of-8 coil) were delivered @ 2500 ms in the course of the 5000 ms trials (intensity, 110% MT) over the contralateral motor cortex (hot spot for FDI muscle). Twelve trials were recorded in each condition. Variations in MEP amplitude and latency recorded in the FDI were subject to a repeated measure analysis of variance (ANOVA) to determine the effects of task conditions and hand dominance.

RESULTS AND DISCUSSION

The ANOVA revealed a large effect of "TASK" ($F= 31.4$, $p<0.001$) on MEP amplitude, but no laterality effect of "Hand (R vs.L)" ($F=0.3$, $p=0.62$). As shown in Figure 1, the large effect of "TASK" was attributable to the increased CS excitability noticed under the TE condition. In fact, MEPs were, on average, 25% larger for TE, than in the BP condition. The influence of background EMG level produced in the two tasks (TE and BP) on the corresponding MEP amplitude was further examined with ANOVA and showed no significant effect (F<2.8, p>0.1). Thus, MEP facilitation was independent of the level of EMG produced in the two tasks.

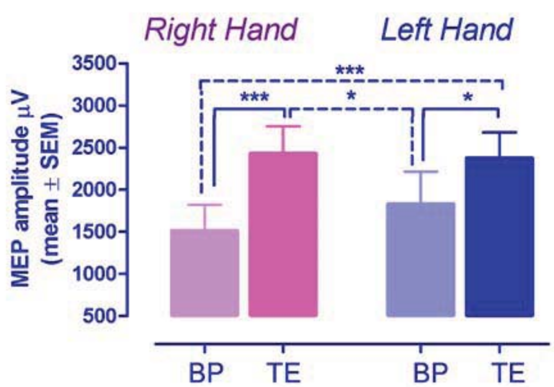

Figure 1. Mean variations in MEP amplitude under the task conditions. *significance levels from the Tukey's test.

Active finger scanning without tactile letter recognition led to MEP facilitation comparable to that seen with the BP task (n=5, Scan: Right, 1235 µV; Left, 1290 µV).

The present results confirm and extend previous observations on the influence of task conditions on CS excitability of intrinsic hand muscles (Muir & Lemon 1983; Bonnard & al 2007). The novel finding was that engaging the index finger in TE for sensing raised letters produced a selectively larger facilitation in the FDI than the other tasks (ie, either BP or finger scanning).This selective increase in CS excitability associated with tactile sensing likely reflects influences exerted at the cortical level. Tactile letter recognition requires fine precise control over the finger, as it scans the contour to allow for recognition. Such dextrous tasks are known to lead to greater activation in the motor cortex as opposed to tasks requiring no dexterity or precision (Bonnard et al 2007). Thus, when compared to BP, the level of precision required for the TE might have led to a greater engagement of the motor cortex in controlling the index finger to optimize sensory acquisition. The large attentional demand associated with the TE task may have also exerted top-down influences to further enhance CS excitability under the TE condition. This conclusion is supported by the fact that finger scanning without tactile recognition led to facilitation levels comparable to the BP task. In this regard, our results are entirely consistent with recent TMS studies showing that attention greatly enhanced sensory responsiveness to tactile inputs in the CS system (Rosenkranz & Rothwell 2004, Stefan et al 2004).

SUMMARY/CONCLUSIONS

Corticomotor excitability is greatly enhanced when the index finger is used to explore and recognize tactile forms as compared to tasks involving finger movements to produce minimal force.

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H-REFLEX VARIABILITY UNDER PRESYNAPTIC INHIBITION

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INTRODUCTION

The H-reflex is obtained by stimulating the Ia afferent axons of a mixed nerve and recording the resulting reflex in the homonymous muscle.

The excitability of the reflex pathway may be controlled by means of presynaptic inhibition (PSI) of the Ia afferents that excite the motoneurons. For instance, it has been suggested that the amplitude of the H-reflex decreases due to PSI when subjects change position from lying to seated, and from seated to standing (Zehr 2002). There is an inherent variability in the H-reflex amplitude. Even with a rigorous control of the peripheric factors, the soleus H response varies in a random manner and also shows dependencies on the motor task, posture, state of alert, etc (Mezzarane & Kohn 2002; Zehr 2002; Bonnet et al. 1997).

It has been reported in cat that the variability of the monosynaptic response (MSR) decreases under PSI conditioning (Rudomin and Dutton 1967). In the present work we sought to verify if PSI in humans causes a decrease in the variability of the H-reflex. A positive finding would imply possibly similar mechanisms in cat and humans and provide a technique to further investigate the human spinal cord by electrophysiological means.

METHODS

Four healthy subjects (two males and two females) gave a written informed consent to participate in this preliminary study (27 ± 5 years). Subjects were seated in a specially designed armchair with the angles of the hip, knee and ankles at 110, 135, and 110 degrees, respectively.

The right foot was strapped to avoid movement of the ankle. The H-reflex was evoked in the soleus muscle by a percutaneous electrical stimulation at the popliteal fossa. The surface EMG electrodes were located 4cm below the inferior margin of the two heads of the gastrocnemius muscle.

The conditioning stimulus was applied to the common peroneal nerve (on the head of the fibula) 100ms before the test stimulus was applied. This is known to evoke a PSI of the Ia afferents arising from the soleus muscle spindles and usually causes an attenuation of the H reflex amplitude.

Five hundred stimuli were applied with a frequency of 1Hz to evoke either kind of responses, control (without conditioning stimulus; **C**) or test (with presynaptic inhibition conditioning stimulus; **P**). One to three sequences for each kind of response were obtained for each subject.

The standard deviation (STD) of the last 450 H-reflex amplitudes was evaluated (the first 50 were discarded to avoid transients). The mean H reflex amplitudes were maintained similar in the two conditions by adjusting the test stimulus intensity.

RESULTS AND DISCUSSION

A decrease was observed in the variability (STD) of the H-reflex under PSI (**P**), compared to that without any conditioning (**C**) (figure 1). This result is in agreement with Rudomin's observations in cats (Rudomin and Dutton 1967). The degree of variability decrease for each subject was 39, 19, 38 and 20%.

One possible explanation for the variability reduction of **P** could be the refractoriness caused by the presynaptic conditioning volley in some point along the path leading to the last order interneuron (e.g., PAD interneuron) (Rudomin and Dutton 1967). As a consequence, the fluctuation in the membrane potential of the Ia afferent terminals is reduced.

From an application viewpoint, the variability of the H-reflex in humans could be used as an aid to detect the level of the PSI in different circumstances (under the condition of no significant changes in the ongoing EMG level). The congruent results obtained in both species (cat and humans) may help to improve the understanding of the spinal cord mechanisms of humans, either in health or disease.

SUMMARY/CONCLUSIONS

The conditioning of the H-reflex pathway in humans by an electrical stimulation that increases the level of presynaptic inhibition of the Ia afferents caused a decrease in the H reflex amplitude variability. This approach may open new possibilities for studying the human spinal cord in health and disease.

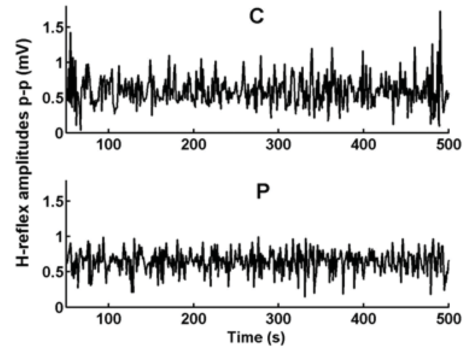

Figure 1: The sequence of 450 H-reflex amplitudes control (**C**) and conditioned by PSI (**P**) obtained along the time. Data from a single subject.

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CHANGES IN REFLEX EXCITABILITY CAN SUGGEST PREPARATORY ACTIVITY IN THE SPINAL CORD FOR MUSCLE CONTRACTION

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INTRODUCTION

In the last decade there have been reports that the spinal cord is also involved in preparatory activity. Armand et al. (1997) showed that there are anatomical possibilities for a spinal cord contribution because large densities of corticospinal projections reach interneurons in intermediate lamina.

Bonnet et al. (1997) verified that imagined motor tasks are able to increase spinal reflex excitability and Prut and Fetz (1999) characterized involvement of interneurons in the monkey spinal cord related to preparatory activity.

In this study we aimed to verify the activity of spinal cord neuronal circuits preceding a voluntary movement, under an instructed delay period paradigm.

METHODS

Eleven healthy subjects participated in the experiments, approved by the local Ethics Committee. The mean H reflex amplitude in control conditions was kept between 10% and 40% Mmax.

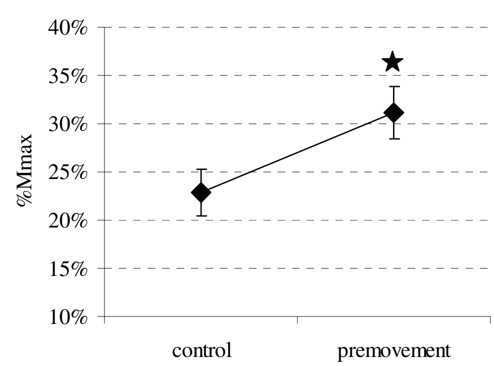
Electrophysiological recordings of the H reflex were employed. The epochs of H reflex recording were associated either with a resting period (control condition) or with premovement periods (instructed delay period) preceding contraction of the right leg soleus muscle (SO).

The subject received a cue at an appropriate time about the type of contraction: plantarflexion or dorsiflexion. Peak to peak H reflex values were computed at rest and at 1000 ms before action within the instructed delay period.

Percent values of H amplitude with respect to maximum M values (%Mmax) were computed when the SO was an agonist of contraction during plantarflexion.

RESULTS AND DISCUSSION

The results showed that the H amplitude (%Mmax) values increased significantly at premovement when compared to control (graph 1).



Graph 1: This graph illustrates reflex excitability of the homonymous Ia afferents-

motoneuron pool at rest (control) and at 1000 ms (premovement) before plantarflexion. Vertical bars represent standard error and dark star represents a significant difference between control and premovement detected by student t-test.

Differences in reflex excitability preceding action observed in instructed delay period can represent preparatory activity, occurring 1000 ms before movement as it was observed by Prut and Fetz (1999) in monkeys.

CONCLUSIONS

The results showed activity in spinal cord neuronal circuits previous to muscle activation. The pattern of activation suggests preparatory mechanisms for movement.

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INTERNEURON MEDIATION OF MUSCLE CO-CONTRACTION: FCU LATE RESPONSES AFTER DISTAL ULNAR STIMULATION.

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INTRODUCTION

In the clinical subspecialty of Electrodiagnostic Medicine, F-waves are routinely used to evaluate the proximal segments of peripheral nerves for demyelination. In addition, when F-waves are overly stable, spinal cord or brain pathology must be suspected. An F-wave is defined as an action potential (AP) that is evoked intermittently by a supra-maximal electric stimulus to the nerve due to antidromic stimulation of motor neurons. Amplitudes are typically 1-5% of Compound Muscle Action Potential (CMAP) amplitudes, with variable morphology and latency. The neurophysiologic mechanism of F-wave generation has traditionally been attributed the retrograde propagation of an incident antidromic AP from the motor neuron [Eccles, 1955]. This theory does not explain how a retrograde AP might propagate past the refractory axon segment adjacent to the hillock, or how an AP might remain latent during this refractory period, nor does it account for the noted 1ms delay between the incident and retrograde APs [Eccles, 1955]. We postulate the existence of a pathway in which branches from the motor axon project to interneurons. Such a pathway may explain a 1ms delay that allows the proximal segment to overcome its refractory period, and may also offer insight into neurophysiologic mechanisms of the activation of co-contracting muscles.

To explore this concept, we designed an experiment to look for F-waves in the flexor carpi ulnaris (FCU) muscle after stimulating the ulnar nerve at the wrist.

METHODS

A two-channel ulnar CMAP study was set up on a VIASYS™ clinical EMG machine for a human subject. An E1 sensing electrode was placed over the hypothenar muscle mass, E2 over the distal little-finger dorsum, and electrical wrist stimulation applied to the skin overlying the ulnar nerve 7cm from E1. A second sensing (bar) electrode was placed over the FCU, parallel to the ulna, with E1 placed 8cm distal to the medial epicondyle, and 2cm medial to the ulna, with E2 oriented distally. 22 trials of supra-maximal shocks were run; in 5 trials, the FCU was facilitated with slight voluntary contraction.

RESULTS AND DISCUSSION

F-waves from the hypothenar muscles and late responses from the FCU were observed. Figure 1 shows a typical pair of simultaneous CMAP studies with the hypothenar responses in black, and the FCU in red. Both responses were evoked by a single electrical stimulus of the ulnar nerve at the wrist. The initial off-scale waveform is the hypothenar CMAP. The later 250µV response at arrow *A* is the F-wave. The FCU curve shows an initial small-amplitude response that we interpret as a far-field potential emanating from the ulnar-innervated intrinsic hand muscles. Arrow *B* indicates the FCU late response. Table 1 summarizes the results. Amplitudes of the hypothenar F-waves and FCU late responses were comparable from 40-400μV, and FCU latencies averaged 1.4ms less than hypothenar (3ms if facilitated). In addition, slight voluntary contraction increases FCU late response amplitude 2-3 fold.

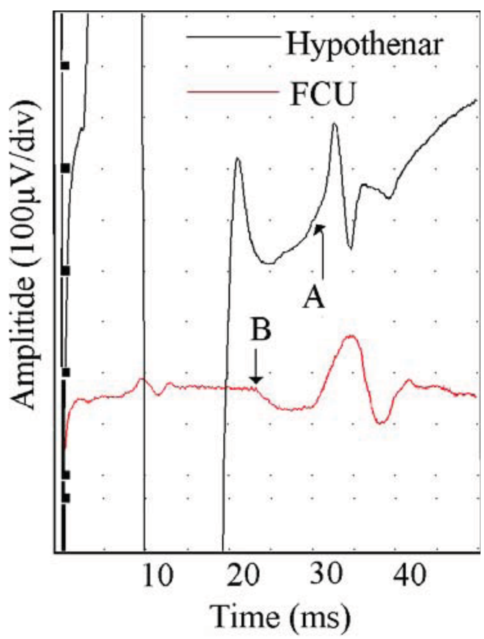

Figure 1: F-waves in the hypothenar muscle group and FCU after ulnar wrist electrical stimulation (distal to the FCU).

SUMMARY/CONCLUSIONS

Late responses in the FCU after ulnar nerve stimulation at the wrist are not explained by retrograde firing of motor neurons. The point where the ulnar nerve is stimulated is far distal to the motor nerve branches to the FCU as shown by the lack of a CMAP. Volume conduction of F-waves from ulnar intrinsic hand muscles is not tenable as the FCU responses are earlier, and far too large. In addition, the FCU late response amplitudes are facilitated by voluntary contraction. These results are consistent with our assertion that the FCU late responses are in fact F-waves, and with our postulate that branches from motor axons project to interneurons as part of the F-wave pathway. Such a pathway must have at least one synapse, which would account for a 1ms delay between the incident and retrograde action potentials. This result is also consistent with our postulate that such a pathway may have a role in mediating the co-contraction of muscles.

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	Hypothenar	FCU	Facilitated FCU
Ave. Latency	27.9 +/- 1.2 ms	26.5 +/- 3.2 ms	24.8 +/- 2.1 ms
Ave. Base-Peak Amplitude	269 +/- 87μV	91 +/- 86 μV	212 +/- 102 μV
Ave. Peak-Peak Amplitude	284 +/- 128 μV	159 +/-178 μV	410 +/- 227 μV

Table 1: Statistics of Hypothenar F-waves and FCU late responses from 22 trials of ulnar electrical stimulation at the wrist. For 5 trials, the FCU was slightly voluntarily contracted (Facilitated FCU), showing increased amplitudes and decreased latencies.

MUSCULAR RESPONSE TO QUASI-STATIC TENSILE LOADING OF THE CERVICAL FACET JOINT CAPSULE

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INTRODUCTION

The cervical muscles may play a role in the development of chronic whiplash pain secondary to facet joint capsule (FJC) damage. The FJC is richly innervated with mechanoreceptors, most of which are thought to be primarily involved in proprioception, as well as free nerve endings which function as nociceptors (McLain, 1994). The FJC has been shown to undergo significant deformation during a whiplash-like perturbation, with maximum capsular ligament strains reaching sub-failure injury thresholds (Pearson et al., 2004). Damage to the mechanoreceptors embedded in the injured capsule is hypothesized to result in corrupted muscular activation patterns; which over time could lead to greater stresses in the muscles and muscular fatigue (Panjabi, 2006).

In comparison to healthy controls, whiplash patients exhibit a reduced ability to recruit the deep cervical flexor muscles, an increased reliance on the sternocleidomastoid muscles to execute cervical spine flexion, and a reduced ability to relax the superficial extensor muscles (Falla et al., 2004a, 2004b). It remains unclear whether these changes are a response to disruption of the mechanoreceptors embedded in the FJC and thus, occur immediately in response to non-physiologic FJC loading; or whether the changes develop over time (i.e. days, weeks, or months) as a learned behavior (e.g. pain avoidance, fear of re-injury, etc.). Therefore, the purpose of the present study was to investigate cervical muscle response to physiologic and non-physiologic facet joint capsule loading.

METHODS

Five adult female Lamancha goats (39-63 kg) were utilized in this study. Anesthesia was induced and maintained throughout surgical preparations by inhalation of isoflurane (1-3.0%) and oxygen (2 L/min). Once surgical preparations were complete, α-chloralose was administered (60 mg/kg) and maintained (10-15 mg/kg) throughout the test procedures. The goats were positioned in a test fixture which accommodated a spine fixator, a load cell-actuator system and stereoimaging system to measure capsular strain (Lu et al., 2005). The left C5-C6 FJC was surgically isolated from its bony attachments and connected to the load cell-actuator system. The FJC underwent a series of controlled uniaxial stretch tests in 4 mm increments at a rate of 0.5 mm/s until the capsule ruptured. Each test had a trapezoidal loading pattern consisting of a load ramp to a specified displacement, a 10 second hold, and a release ramp back to the original position. Tensile "re-tests" were repeated between each increment of the tensile loading paradigm. Each re-test test had the same trapezoidal loading pattern and displacement rate as the other tensile tests, but the FJC was only displaced to 4 mm.

EMG activity of the trapezius (TR), multifidus (MF), sternomastoid (SM) and longus colli (LC) muscles was recorded via 8 pairs of bipolar fine-wire electrodes made from 125 µm Teflon®-insulated platinum wires (2mm exposed tip), during and after incremental tensile loading of the FJC. Oxidized brass spheres applied to the FJC were tracked via stereoimaging to obtain 3D capsular strain. EMG, load, actuator motion, and video data were collected synchronously.

RESULTS AND DISCUSSION

All muscles exhibited discharges above baseline activity in response to FJC stretch. TR, MF, and LC were responsive to both physiologic and non-physiologic stretch, while the SM responded only to non-physiologic stretch (test increments ≥ 12 mm). In the physiologic range, strains eliciting MF (18.0 ± 15.5 %) and LC (17.4 ± 10.9 %) activity were not significantly different (p = 0.99), but significantly preceded those eliciting TR activity (32.3 ± 15.8 %) (p = 0.003). Normalized rectified EMG (NREMG) was more significantly correlated with peak capsular load than with strain. In particular, quadratic regression revealed significant proportions of shared variance between peak capsular load and re-test NREMG in 4 of 8 muscles tested: C6 TR (R^2 = 0.560, p = 0.00003), C5 MF (R^2 = 0.480, p = 0.0003), C6 MF (R^2 = 0.326, p = 0.01), and left SM (R^2 = 0.431, p = 0.001). The finding that NREMG was better correlated with load than strain agrees with the findings of others investigating the neural response of the feline knee joint capsule to stretch (Khalsa et al., 1996).

SUMMARY/CONCLUSIONS

Preliminary results revealed that MF and LC activate at approximately the same strain and precede TR activity by approximately 12% strain in the physiologic range. Further data analysis in this study will address the link between non-physiologic tensile FJC loading and changes in neck muscle recruitment. Sustained over time, decreased activation of the deep cervical muscles may lead to increased activation and fatiguing of the superficial cervical muscles, eventually leading to muscular degeneration and chronic pain.

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COMPARISON OF TWO METHODS OF DETERMINING THE LATENCY OF MOTOR EVOKED POTENTIALS TO THE INFRASPINATUS MUSCLE

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INTRODUCTION

Motor evoked potentials (MEP) allow researchers to investigate the excitability of the central nervous system and the conduction of stimuli along the central and peripheral motor nerves. MEPs are commonly induced by transcranial magnetic stimulation (TMS). MEP latency is used to determine the velocity of nerve conduction along the motor neuron pathway from the motor cortex to a particular muscle. MEPs may be induced while the target muscle is at rest or while it sustains a voluntary contraction, typically between 10% and 30% of maximum (MVC) (Rossini et al., 1994).

There are two possible methods of measuring MEP latency at the target muscle. MEP latency may be determined by taking the first crossing of the MEP over the zero voltage line; this technique is recommended for MEPs recorded while the target muscle is at rest. The MEP latency may also be calculated using the first crossing of a threshold that is two standard deviations (2SD) above the baseline level of the electromyographic (EMG) data recorded prior to the stimulus. This method is recommended for MEPs recorded while using facilitating contractions. It is unclear which technique provides the most reliable MEP latency measurements, therefore the purpose of this study was to determine which method is more appropriate for MEPs obtained from the infraspinatus muscle during a facilitating contraction held at 20% of the subject's MVC.

METHODS

The study was approved by the Queen's University Health Sciences Research Ethics Board and informed consent was obtained from all participants. Healthy adults over 18 years old were recruited if they had no pre-existing orthopaedic or neurological conditions, and no contraindications to TMS. Both infraspinati were tested, one at a time, in random order.

Adhesive Ag/AgCl surface electrodes were applied in a monopolar configuration over the motor point of the infraspinati and the posterior aspect of the acromia. A carbon rubber (5 by 10 cm) reference electrode was centred across the C7 spinous process.

An AMT-8 Bortec™ Octopus EMG amplifier was used to amplify (x1000) and filter (10Hz to 1 kHz) the MEPs. The signals were sampled at 2 kHz using a16-bit A/D board and custom Labview™ software. For each MEP, the computer recorded 50 ms of baseline activity, triggered the stimulation and continued to record the response for a further 300 ms.

Subjects sat upright with the test shoulder abducted to 90° and in neutral rotation and with the elbow in 90° of flexion. The subject pulled on a strap to generate a measurable shoulder external rotation torque. EMG and torque values were recorded for three, five-second maximal voluntary contraction (MVC) efforts and the maximal response was used to determine the

20% MVC contraction required for facilitation. (Inoue et al., 2003).

A Magstim 200^{2} ™ monophasic stimulator was used with a single 90 mm coil. The TMS coil was placed over the vertex of the skull. The threshold stimulus intensity was deemed the intensity that produced a distinct MEP in three of six consecutive stimulations (Wassermann, 2002). The intensity was then increased to 120% of threshold and 10 consecutive MEPs were recorded.

Each of the 10 trials for each side was analysed separately. The first zero voltage-crossing of the MEP after the stimulus artefact was designated as MEP latency$_{ZC}$ (Figure 1). To determine the MEP latency using the point where the MEP voltage exceeded 2 SD of the baseline level (MEP latency$_{2SD}$), 25 ms of pre-stimulus baseline data were averaged and the standard deviation (SD) was determined. The first time the MEP signal crossed the 2 SD line was designated MEP latency$_{2SD}$ (Figure 1). If the MEP tracing did not fully reach the zero line or the 2 SD line but there was an obvious change in direction, the point at which the direction changed was considered to be the zero-crossing or 2 SD crossing.

To compare the within-subject variability of the two methods of determining MEP latency, the individually-analysed trial data were used to determine the coefficients of variation (CVs) for MEP latency. A one-way analysis of variance (ANOVA) was performed to establish which method of determining MEP latency was the most reliable (i.e. had the lower CV).

RESULTS AND DISCUSSION

Eight men and 15 women, aged 18-66 years participated in the study. One reported being left-handed.

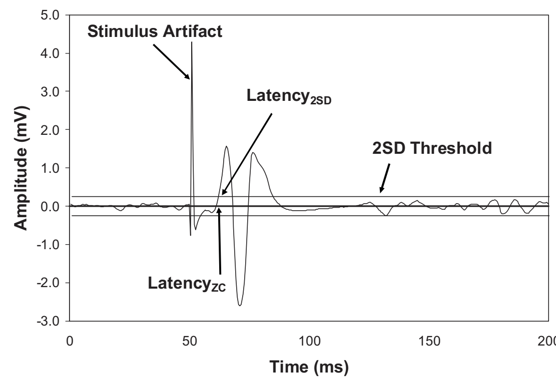

Figure 1: MEP performed with a facilitating contraction, showing latency$_{ZC}$ and latency$_{2SD}$.

The CVs computed for each subject were averaged across the sample. The CV for MEP latency$_{ZC}$ was 12.0% and for MEP latency$_{2SD}$ was 9.5%. A one-way ANOVA revealed that MEP latency$_{2SD}$ had significantly less within-subject variability ($p=0.002$) than MEP latency$_{ZC}$.

SUMMARY/CONCLUSIONS

This study shows that when analysing MEP latency in studies using facilitating contractions, the first crossing of the 2 SD line should be used as the threshold for detecting negative peak onset. This is primarily because the increased background EMG activity which occurs with voluntary contraction makes it more difficult to discern the first crossing of the zero line. This conclusion supports current practice.

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NEURAL MECHANISM IN BRAIN UPON PRE－DETERMINED RESPONSE TO PERIODIC AND APERIODIC STIMULI

- ANALYSIS OF VISUAL STIMULATION OF MOTOR MOVEMENT -

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INTRODUCTION

In Parkinson's disease, movements are usually distinctly impaired at their start and end or when they change direction, but usually not during the periods of constant speed. Many studies have implicated the nerves connecting the basal ganglia to the supplementary motor area (SMA), but there have been few reports on the involvement of other areas of the brain. Therefore, we compared neural activities in the brain, monitored by functional MRI (fMRI), during a hand grasping task performed in response to periodic and aperiodic visual stimuli in healty subjects.

SUBJECTS and METHODS

The subjects were 14 healthy adults (age: range 19 to 47 years, mean 27.6 years; 10 males) with no history of neurological disease. Eleven were right-handed and 3 left-handed, as determined by the Edinburgh Handedness Inventory. All subjects gave informed consent.

The fMRI images were obtained from subjects in a supine position in a 1.5-T MR scanner (GE Signa Horizon). The subjects performed a clench-unclench dominant hand task, adapted from Lutz, K**.** *et al***.** (2000), **i**n synchrony with appearance of a visual stimulus under two conditions, periodic (2 Hz) and aperiodic(mean frequency of 2 Hz), viewed through a mirror. The tasks were presented in alternating 30-s periods of rest and test, and each subject performed under each condition twice. Brain image data were expressed as statistical parametric maps and analyzed using SPM2. In statistically comparing the effects of the two conditions on region-specific brain activations, the level of significance was $p<0.05$(t-test) for differences within individuals and $p<0.001$(t-test) for differences between individuals.

RESULTS

1) Increased fMRI activation was seen in 9 cerebellums, 11 frontal lobes, 7 posterior lobes, and 3 supplementary motor areas with the aperiodic task, (t-test) in only 1 cerebellum and 3 frontal lobes with the periodic task. (see Figs. 1 and 2).
2) The activation occurred in the prefrontal cortex, posterior lobe, and SMA on both sides, in the cerebellum, and in the

premotor cortex on the same side as the non-dominant hand. In these areas, the signal strength (T-value) was high during the aperiodic task. Signal strengths for the periodic and aperiodic tasks in different areas of the brain including Broadman areas are shown in Table 1.

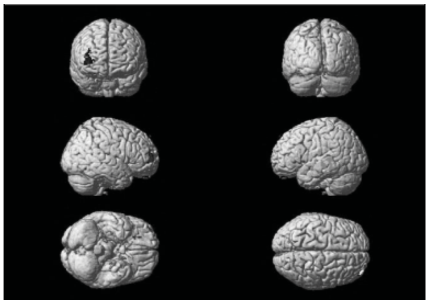

Figure 1. Activation on fMRI during periodic task

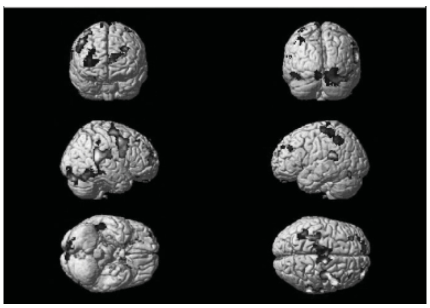

Figure 2. Activation on fMRI during aperiodic task

Table 1 Signal strength (T-value)

area	T-value		x	y	z
	Periodic	Aperiodic			
Putamen	3.05	3.95	-23	22	-1
area 37	2.39	0	42	-60	-5
area 47	3.00	3.03	27	31	2
area 24	0	6.64	-9	19	21
area 10	0	4.74	-4	55	7
PM	0	3.67	48	4	24
SMC	0	2.86	44	-4	44

X : Positive:Right hemisphere

Negative:Left hemisphere

DISCUSION AND CONCLUSION

Compared with the periodic task, activations of the brain may have been significantly more extensive during the aperiodic task for the following reasons. 1) Movement induced by outside information being unpredictable caused increased activity in the premotor cortex. 2) Increased attention demanded by the unpredictable stimulus caused increased activation in the SMA and prefrontal cortex. 3) Increased activity in the posterior lobe in response to space perception accompanying a visual operation. 4) Since the route from the cerebellum dentate nucleus to the area is extremely important for motor learning for tasks visually induced or induced by other particular receptors, motor learning induced by the aperiodic visual stimulus may have been promoted.

We will continue the study to see how using variations of rhythm and different movements affect the brain activity.

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ELECTROPHYSIOLOGICAL STUDIES IN PATIENTS WITH SUBACUTE MYELO-OPTICO-NEUROPATHY BY MAGNETIC STIMULATION

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INTRODUCTION

Subacute myelo-optico neuropathy (SMON) is the neurological intoxication of Clioquinol, and SMON has affected about 10,000 patients in Japan until when Clioquinol formulations were released .
The sensory disturbance such as dysesthesia with peripheral neuropathy has been regarded as the major symptom of SMON. However, the subclinical disturbance of pyramidal tract functions has not been investigated.
So we investigated the central motor conduction times in SMON patients with the method of the transcutaneous magnetic stimulation.

METHODS

The functions of central conduction times were studied in 31 patients with SMON (47-74 Y.O), the transcutaneous magnetic stimulation was applied to the motor cortex and spinal cord.
The motor evoked potentials (MEPs) were elicited from the abductor pollicis brevis muscle and abductor hallucis muscle.
The central motor conduction times (CMCTs) were calculated from the latency difference between the MEPs elicited from motor cortex to cervical cord ,or the MEPs elicited from motor cortex and lumbar cord(Figure1).

RESULTS AND DISCUSSION

As the results, in normal subjects (21 cases, Age :42-67Y.O. Mean:58Y.O.), CMCTs between motor cortex and cervical level were 9.13±0.92msec,and the upper limit of normal values (mean+3SD) was 11.89msec. CMCTs between motor cortex and lumbar level were 17.29±1.31msec,and the upper limit of normal values was 21.22msec.
In SMON patients (N=31), CMCTs from motor cortex to cervical root were in normal range ,but the CMCTs from motor cortex to lumbar root were abnormal(over the upper limit of normal values) in 8 cases of moderate (N=11), and 4 cases of severe cases(N=7). In 3 cases of severe cases, evoked potentials could not evoked from leg muscles by transcranial magnetic stimulation. Furthermore, in mild cases (N=13) having not the pyramidal tract signs, 3 cases also showed the abnormal conduction times (Figure 2).

SUMMARY/CONCLUSIONS

The degree of prolongation of latencies of evoked potentials elicited from abductor hallucis muscles of legs by the transcranial magnetic stimulation was correlated with the grade of severity of clinical signs with SMON. These central motor conduction tmes (CMCTs) between motor cortex and lumbar level also reflected the subclinical disorders of pyramidal tracts in mild cases. Our results suggest that the transcranial magnetic stimulation is beneficial for evaluating the subclinical disturbance of pyramidal tract signs of myelopathy such as in patients with SMON.

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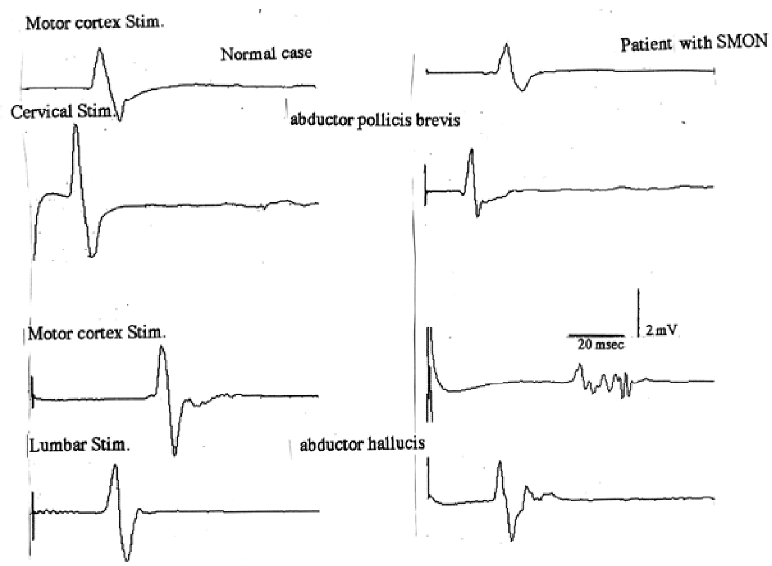

Figure 1: Motor evoked potentials elicited by magnetic stimulation

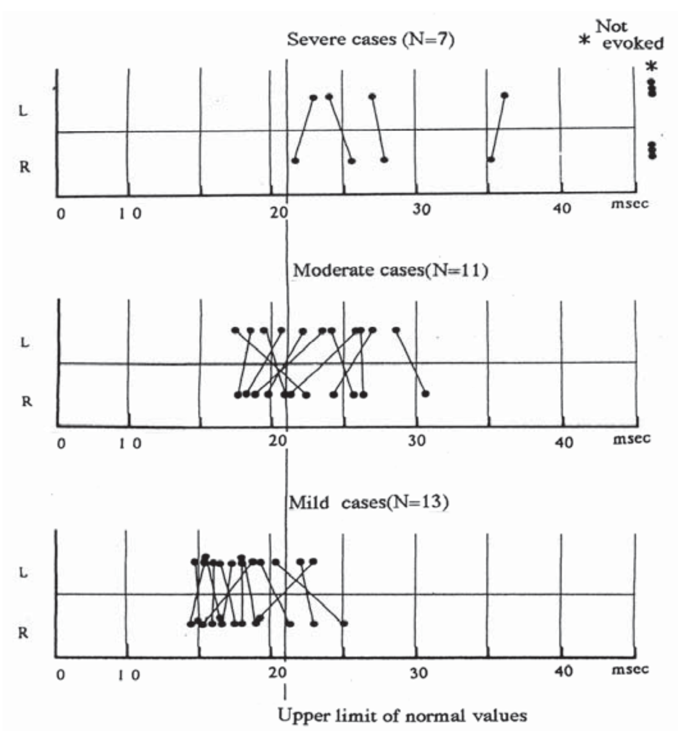

Figure 2:Central motor conduction times of motor evoked potentials from motor cortex to lumbar level in SMON patients

ELECTROMYOGRAPHY OF MASTICATORY MUSCLES IN PARTIALLY EDENTULOUS CHILDREN AFTER ORAL REHABILITATION

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INTRODUCTION

Chewing consists of rhythmic jaw-opening and -closing muscle activity, regulated by a central pattern generator. This basic rhythm is modulated by sensory. Chewing pattern may depend on the type of food and its interaction with various masticatory system components. During natural chewing, different masticatory muscles are recruited separately with variations in amplitude and occurrence of muscular contraction.

The aim of this study was to investigate the adaptation process of masticatory function to oral prosthetic appliance in children with premature loss of primary molar teeth. For this purpose, integrated activity of masseter and anterior portion of temporal muscles during chewing of silicone tablets was assessed before and six months after oral rehabilitation.

METHODS

Subjects: Two hundred and forty nine children were scanned and 23 healthy children (12 girls and 11 boys, mean age 7.10±0.74 yr) who agreed to and were cooperative with the procedures participated. Written and verbal consent was obtained from child's parents and the research was approved by the Dental School Ethics Committee.

The children were in the mixed dentition stage, with normality of their relationship, as well as normality of the oral tissues. Moreover, they should present premature loss or indication for extraction of one or more primary molars due to large caries and total coronal destruction. Both extractions and total coronal destruction should be occurred before at least one month, as well as absence of tooth pain during this time. The children received a removable partial denture made with acrylic, resin artificial teeth, clasps and passive Hawley arch (orthodontic stainless steel wires, 0.7 mm) to provide retention (Figures 1 and 2).

Electromyography: Muscle activities were measured with the EMG System MCS-V2 Electromyograph (São Paulo/Brazil) and passive silver-chloride surface electrodes (Tyco Healthcare – Kendall LTP/Canada). The children were seated upright. The muscle bulk center was located by palpation when children clenched their teeth. The electrodes were placed on the skin surface over the central portion of the masseter muscle (the area of greatest lateral distention, approximately halfway between the origin and insertion and 2 cm posteriorly to the anterior border) and for the anterior temporal, in front of the anterior border of the hair line, on the area of greatest lateral distention. The centers of the electrodes were 17 mm apart lengthwise along each muscle. In order to minimize contact impedance, the recording sites were cleaned with a piece of cotton soaked with 70% alcohol. The ground electrode was applied to the child's wrist.

The chewable test materials were provided in portions of 17 silicone cubes, with edge size of 5.6 mm (Opsotil NF®; toothpaste; vaseline; dental plaster powder; and alginate powder, in the respective percentages by weight: 58.3, 7.5, 11.5, 3, 10.2, and 4%, and 20.8 mg/g of catalyst paste). The cubes were chewed for 20 strokes, controlled by the examiner. Muscle activities were expressed as root-mean-square (RMS) values (μV). The integrated activity of the muscles was considered, i.e., the sum of the signals of the right and left masseter and anterior portion of temporal muscles.

The evaluations were performed before and 6 months after the placement of the removable partial dentures, considering the intra-individual comparisons.

Statistical analysis: The EMG data were normalized with respect to the values obtained in the isometric contraction. The comparisons were performed through paired or Wilcoxon tests, when indicating. The correlations between body variables and EMG signals were calculated by Spearman's coefficient. The level significance was $p<.05$.

RESULTS

There was no significant difference ($p>0.05$) among the variables for boys and girls therefore all subsequent calculations were performed irrespective of gender.

During chewing the EMG values were significantly higher at second evaluation (230.77±73.92 and 279.47±88.94). The children gained weight and height within the 6 months of treatment ($p<0.001$); the body mass index did not show a statistically significant difference ($p>0.05$).

The body variables showed no significant correlation with EMG, in both evaluations ($p>0.05$).

DISCUSSION

The muscle activity in the second session was significantly higher. Two explanations could be pointed out: first, children grew up during the six month**,** as seen by the significant differences in body variables between the evaluations, and as a result, muscle activity could also be increased. Nonetheless, there was no significant correlation between body variables and EMG values at the two time points, suggesting that the muscle activity was dependent on other factors besides body size, in the evaluated sample. In this way, the second explanation could be inferred upon oral rehabilitation *per se*, since it could have influenced the dynamic process, due to more number of surfaces that were in contact with the food during chewing, increasing the respective interaction and probably determining improvement in muscle function. It is commonly accepted that a new occlusal morphology is likely to vary the chewing pattern or the masticatory muscle activity.

CONCLUSION

The increase in EMG values was able to demonstrate the influence possibility of the partial removable denture on the muscles and masticatory functions in children.

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A MODEL OF THE MUSCLE-FIBER CONDUCTION-VELOCITY RECOVERY FUNCTION

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INTRODUCTION

A muscle fiber's conduction velocity (MFCV) is affected by its recent discharge history (Stålberg, 1966). MFCV increases substantially during the first several discharges after recruitment, and it can fluctuate by several percent as a result of fluctuations in firing rate. This dependence is often referred to as the velocity recovery function. MFCV fluctuation causes jitter between the components of motor-unit action potentials (MUAPs).

Lööf (1990) presented a mathematical model to describe the variation in the inter-potential interval (IPI) between the action potentials of different muscle fibers belonging to the same motor unit. IPI is inversely related to MFCV. We propose a modified model that better describes MFCV behavior after recruitment.

METHODS

The proposed model is:

$$\hat{I}_i = I_0 + x_{i-1}^* + ci$$
$$x_i = x_{i-1}^* - \frac{b}{a}(a - x_{i-1}^*)$$
$$x_{i-1}^* = x_{i-1} \exp - (t_i - t_{i-1})/T$$

where $\hat{I}_i$ is the predicted IPI for the *i*th discharge, I_0 is the base IPI, *c* models the linear trend, *x* is a state variable, x_{i-1}^* is the value of the state variable just prior to the *i*th discharge, *b* is the decrement per discharge, *a* is the limiting factor, t_i is the time of the *i*th discharge, and *T* is the recovery time constant.

To test the model, we examined EMG signals recorded during moderate isometric contractions of the brachioradialis muscles of 10 normal subjects. The signals were recorded simultaneously from different sites in the muscle, and decomposed into MUAP trains using the EMGlab program. We selected several MUAPs that had two distinct spikes, either in the same or in different channels, and that also had one or more firing gaps of at least 0.25 s. For each MUAP, we used an optimization procedure to determine the values of I_0, *a, b*, *c*, and *T* to best model the IPI between the spikes.

RESULTS AND DISCUSSION

Three examples of actual and predicted IPI, normalized to I_0, are shown in Fig. 1. IPI decreased exponentially during the first several discharges after a long gap in firing. Thereafter it varied in inverse relation to the instantaneous firing rate. IPI increased after a gap in firing, recovering to its base value after an interval of about half a second. A linear trend was often superimposed in addition.

The proposed model simulated each aspect of the MFCV behavior fairly accurately, with the rate of recovery during long inter-discharge intervals being determined by *T*, the rate of exponential decrease after recruitment by *a*, and the size of the fluctuations by *b*.

Lööf's model does not include the parameter *a* (in the current notation it equals infinity), and thus each discharge produces an equal IPI decrement. In the proposed model, the decrements are larger at recruitment and become smaller during steady firing, thus modeling the exponential decrease after recruitment.

The physiological meaning of the state variable x is not known. It may involve the accumulation of potassium in the muscle fiber's t-tubule system. Each action potential injects potassium into the t-tubule which is only removed slowly by diffusion (Adrian and Peachey, 1973). Successive discharges cause an accumulation of potassium, resulting in a slight depolarization of the muscle fiber sarcolemma and an increase in MFCV. The accumulated potassium also decreases the concentration gradient across the t-tubule membrane, reducing the amount of potassium injected per action potential.

SUMMARY/CONCLUSIONS

We have presented a model that describes the effect of discharge history on MFCV, including the initial increase in MFCV after a long gap in firing. This understanding may aid the recognition and diagnosis of abnormal MFCV behavior in pathology.

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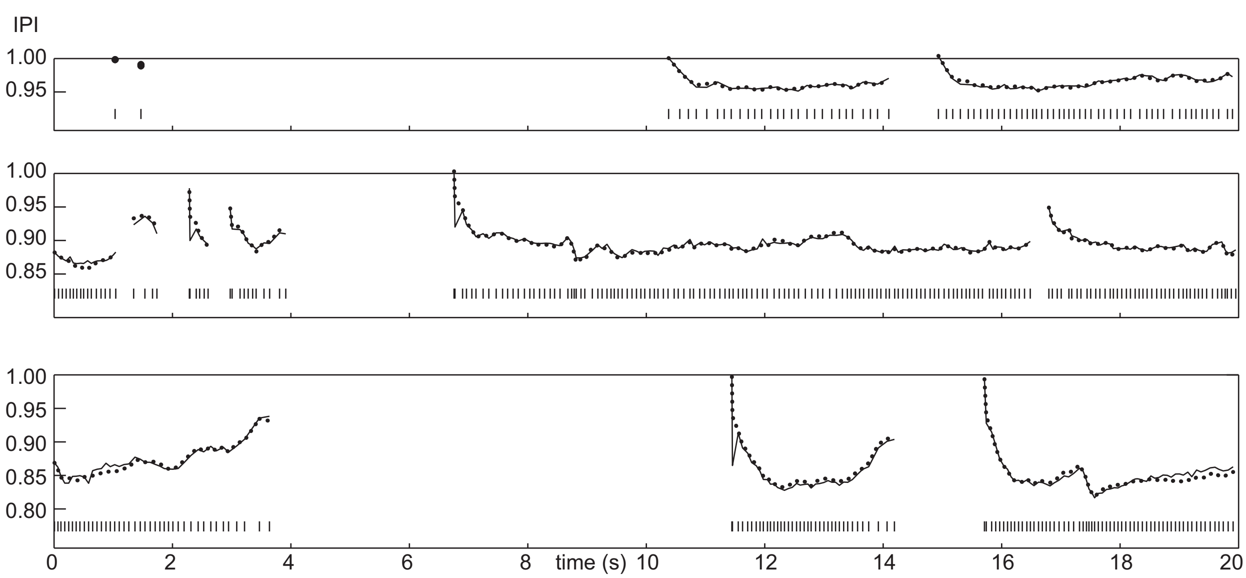

Figure 1. Actual (solid) and predicted (dotted) IPIs for three intermittently firing motor units. The discharge times of the motor units are also shown.

DO SEX, PAIN AND THE OVARIAN CYCLE INFLUENCE MASTICATORY MUSCLE ACTIVITY IN SUBJECTS WITH TEMPOROMANDIBULAR DISORDERS?

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INTRODUCTION

Masticatory muscle pain (masticatory myalgia) is one of the commonest symptoms in TMD patients (Dworkin et al., 1990) and although its pathophysiology has been studied for a long time, it is still poorly understood (Bodere et al., 2005).

Therefore, the aims of the present study were to assess variations in muscle activity at rest in relation to sex and phase of the menstrual cycle in temporomandibular disorder (TMD) patients, and to asses variations in masticatory muscle pain, in relation to phase of the menstrual cycle.

METHODS

TMD cases were 30 normally cycling women; and 23 men. Controls were 30 normally cycling women and 30 men, without TMD or other chronic pains.

Surface electromyographic (EMG) signals at rest, were recorded bilaterally from the anterior temporal and masseter muscles. The root mean square (RMS) was computed from the EMG signals and normalized to the values obtained during maximal voluntary contractions. In order to determine whether there was any effect of the two different phases of the menstrual cycle - menstrual and pre-ovulatory phases - on pain sensitivity, the VAS scores were measured during each session of EMG recording and during three menstrual cycles.

RESULTS AND DISCUSSION

There were EMG differences only in the left masticatory muscles (anterior temporal and masseter) in the men's TMD group (fig 1A and 2A). There were no statistically significant differences in the EMG activity of masticatory muscles between women with and without TMD (fig 1 and 2). The myofascial pain was significantly higher in menstrual phase compared with all of the other phases of the menstrual cycle (Fig. 3B).

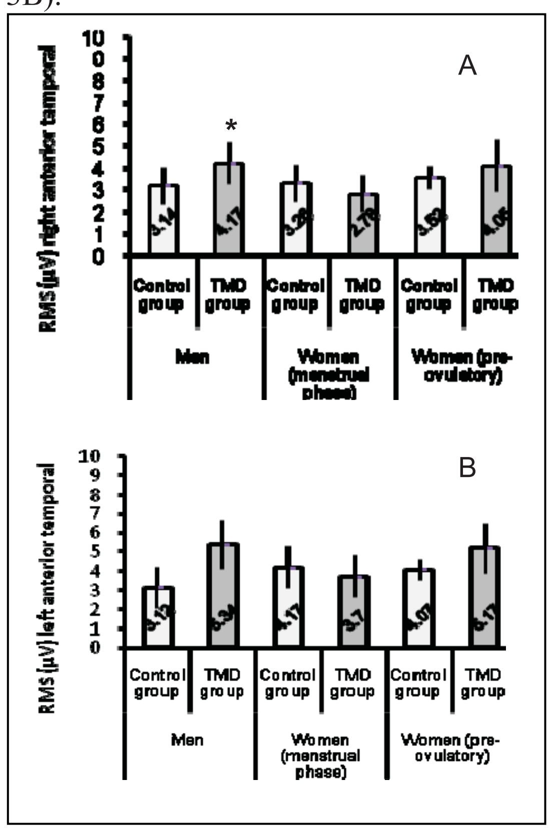

Figure1 A, B: The RMS values of mean surface EMG and the 95% confidence intervals of left and right anterior temporal

muscle. A single asterisk indicates significant differences between real mean within the same sex (men's TMD and control group) (p <0.05) in left anterior temporal muscle (fig 1A).

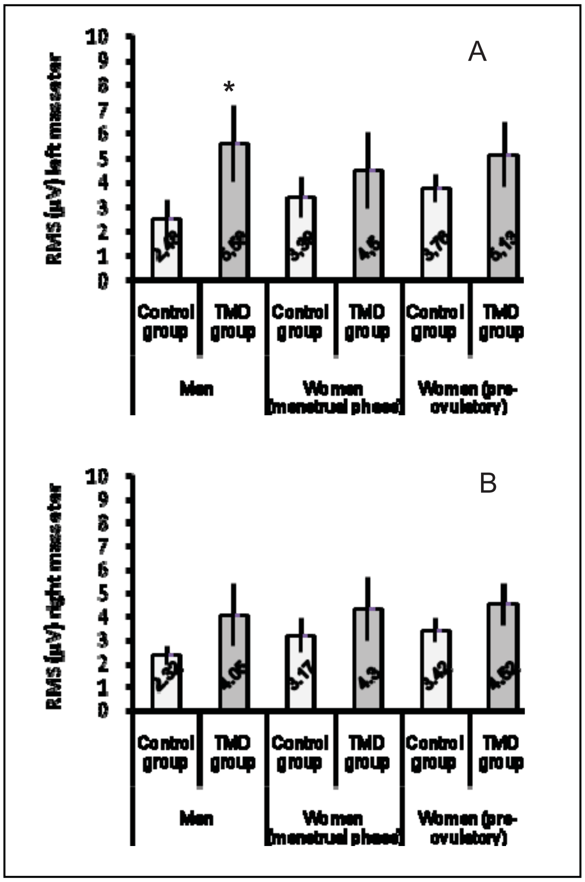

Figure 2 A, B: The RMS values of mean surface EMG and the 95% confidence intervals of left and right masseter muscles. Each column represents the mean. Error bars indicate the SEM. A single asterisk indicates significant differences between men's TMD and control group (p <0.05).

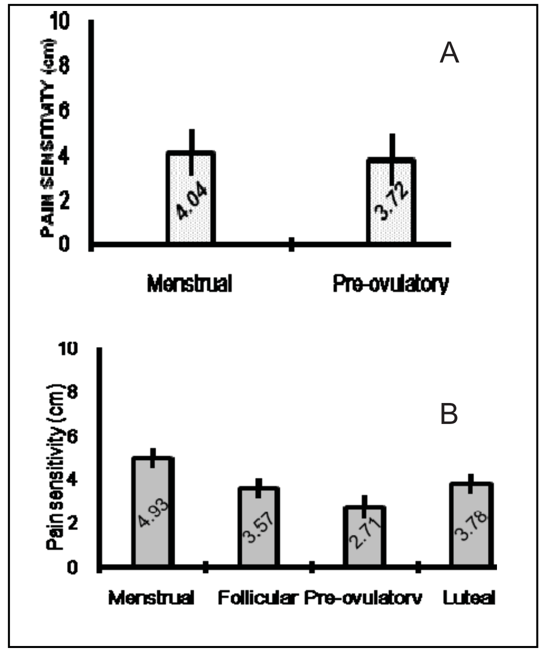

Figure 3 A, B: The VAS scores of myofascial pain, measured during each session of EMG recording of women (fig 3A). A single asterisk indicates significant differences between phases of the menstrual cycle in women's TMD group (p <0.05) (fig 3B).

SUMMARY/CONCLUSIONS

The TMD men, presented higher EMG activity on the left side of the face, where pain was more prevalent. There were no significant differences in EMG activity of women's TMD masticatory muscles, which indicates that the pain-induced changes in muscular responses could differ in men and women.

In spite of higher pain in the menstrual phase of the cycle, the EMG recording variations were unable to detect chronic-pain conditions in women with TMD.

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SCALENE MUSCLE ACTIVITY DURING PROGRESSIVE INSPIRATORY LOADING UNDER PRESSURE SUPPORT VENTILATION IN NORMAL HUMANS

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INTRODUCTION

In intensive care setting, various observations (Brochard et al., 1989; Chao et al., 1997) suggest that monitoring the activity of inspiratory neck muscle over time in mechanically ventilated patients could provide prognostic indicators, or means to detect patient-ventilator asynchrony. How to monitor this activity is a difficult question. A compromise must be found between clinical examination, simple but difficultly quantifiable, and intramuscular recordings, precise but too invasive and expertise-demanding for clinical applications, keeping in mind that differentiating the activity of the scalene muscles vs. that of the sternomastoid muscles is both difficult and critical.

Within this frame, we have previously shown that triggering the averaging of surface scalene electromyograms from the ventilatory flow signal allowed an optimised detection of the scalene activity during quiet breathing (Hug et al., 2006). On the other hand, scalene and sternomastoid are known to be recruited sequentially during voluntary tasks in humans. In dynamical situations, the activity of the sternomastoid typically starts well after the first half of the inspiratory effort, whereas that of the scalene is noticeable as early as during the first tenth of the effort duration. With these elements in mind, we hypothesized that 1) in healthy humans, the scalene would be recruited early in a situation mimicking one of the forms of patient-ventilator asynchrony (namely a progressive decrease in trigger sensitivity during non-invasive inspiratory pressure support); 2) inspiratory pressure-triggered average surface electromyograms of the scalenes would detect and quantify this early activation without interference from sternomastoid activity.

METHODS

Surface and intramuscular EMG activity of both sternomastoïd and scalene muscles were measured in ten healthy subjects. They were asked to breath quietly through a face mask for ten minutes after which they were connected to a mechanical ventilator. Recordings were then performed during three 15-min epochs where the subjects breathed against an increasingly negative pressure trigger (-5%, -10% and -15% of the maximal inspiratory pressure). The intensity of dyspnea was rated using a visual analog scale (VAS).

RESULTS AND DISCUSSION

The sternomastoïd intramuscular electrode was consistently silent. High similarity of the scalene EMG patterns was found between intramuscular and surface recordings (Figure 1).

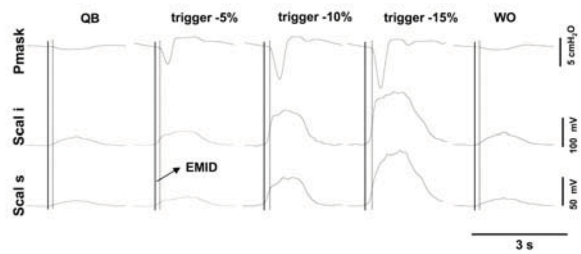

Figure 1: Individual example of the effects of decreasing inspiratory trigger sensitivity. "QB" stands for “quiet breathing”, "WO" for "washout". From top to bottom: Pmask, mask pressure; Scal i, intramuscular EMG recording of the anterior scalene (fine wire electrode) ; Scal s, surface EMG recording of the anterior scalene.

With increasing value of the inspiratory trigger, both dyspnea and the surface scalene activity increased significantly (at the 3 inspiratory triggers for dyspnea and only at triggers fixed at -10 and -15% of the maximal inspiratory pressure for scalene EMG) (Figure 2).

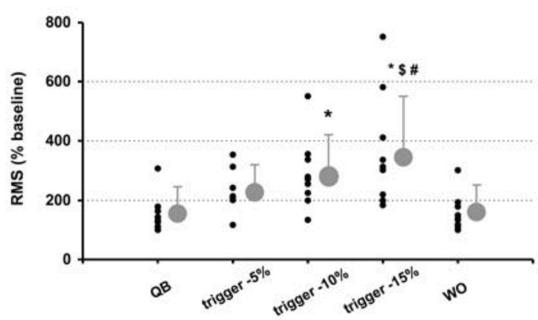

Figure : Evolution of the inspiratory activity of the scalene muscle with decreasing inspiratory trigger sensitivity. "QB" stands for “quiet breathing”, "WO" stands for "washout".
*, significant difference with QB condition; $, significant difference with Trig -5% condition; #, significant difference with Trig -10% condition

In the absence of sternomastoid activation, as in our subjects, the adequate matching of the intramuscular and surface scalene EMG recordings that we observed indicates that the averaging technique that we used can reliably identify a change in scalene activity between two conditions. This feature may open interesting perspectives regarding the use of surface recordings during mechanical ventilation.

SUMMARY/CONCLUSIONS

This study shows that in healthy subjects placed under inspiratory pressure support via a face mask, decreasing the sensitivity of the ventilator trigger in a stepwise manner (namely increasing inspiratory loading) is associated with a progressive increase in the EMG activity of the scalene muscle. At the same time, dyspnea increases, in line with the relationship between respiratory discomfort and inspiratory neck muscle activity. If our observations in healthy individuals are confirmed in patients, the inspiration-locked scalene surface EMG averaging technique could be useful to identify trigger asynchrony in a sensitive manner, namely before sternomastoid recruitment and the occurrence of ineffective inspiratory efforts

** the two senior authors contributed equally to this work*

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EFFECT OF PERIPHERAL SENSORY INPUTS ON SOLEUS H-REFLEX DURING ROBOT-INDUCED PASSIVE STEPPING IN HUMAN

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INTRODUCTION

It has been well recognized that the soleus H-reflex is modulated in task-dependent and phase-dependent manners in human posture and locomotion, i.e., the larger amplitude in the stance and lower in the swing phase of walking, and the reflex gain is the highest in standing and lowest in running task. However, it is still unclear to what extent both descending motor commands and peripheral sensory inputs have influences on the task- and phase-dependent modulation of soleus H-reflex.

In the present study to clarify the influence of peripheral sensory inputs we tested whether the soleus H-reflex is modulated in task- and/or phase-dependent manner without descending motor commands during passive steppings assisted by a robotic gait orthosis.

METHODS

Subject : Ten subjects (6 males, 4 females) with no neurological disorders voluntarily participated in the experiment.
Motor task and H-reflex recording: H-reflexes were elicited in the right soleus (SOL) muscle while the subjects passively standing and stepping with a robotic-driven gait orthosis (Lokomat, Hocoma, Switzerland). The two motor tasks were performed with 40% body weight unloading on a treadmill, and with 100% unloading without ground contact (air standing and air stepping). For the two passive stepping tasks both the maximal M-wave (Mmax) and the H-reflex with M-wave amplitude of 10% Mmax were recorded at six different phases of step cycle. EMG activities were recorded from the right tibialis anterior (TA) and SOL muscles to ensure if there is/are any activity in these muscles in the passive motor tasks.

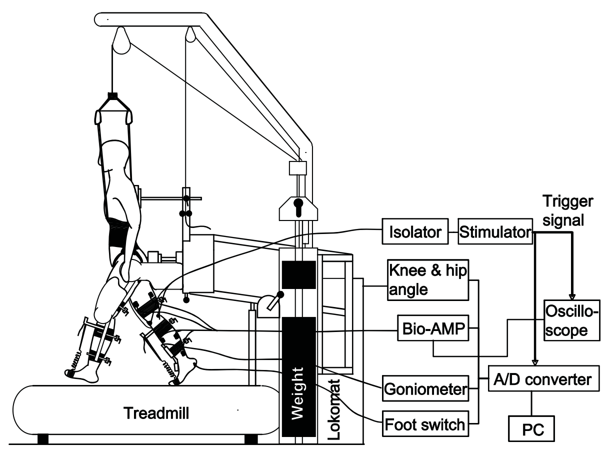

Figure 1: The experimental set up of this study.

RESULTS AND DISCUSSION

Fig 2 shows a typical example of H-reflex recordings during the passive stepping task on a treadmill. The H-reflex amplitude showed a clear phase-dependent modulation in the both stepping tasks. Fig 3 summarizes mean values of H-reflex, M-wave, Mmax and background SOL and TA activities for all subjects (n=10). The overall sizes of H-reflex during both stepping tasks were reduced significantly than those in the standing tasks. In both stepping tasks statistically significant reduction of H-reflex amplitude at the early swing phase (80% of step cycle in the figure) was observed, indicating there were phase-dependent modulations in the H-reflex amplitude in both passive stepping tasks. Both M-wave and Mmax were kept at almost constant level throughout the step cycle.

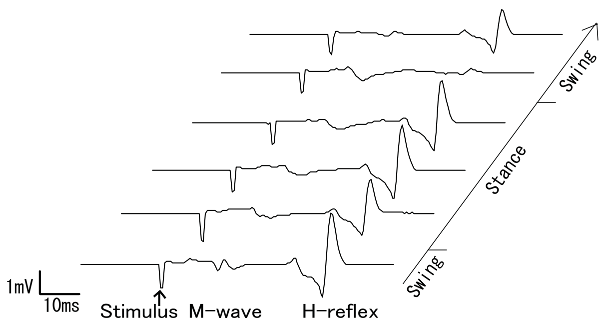

Figure 2: A typical example of the H-reflex recording during the passive stepping task.

The main findings of this study were 1) the H-reflex was significantly reduced in the stepping tasks compared to those in the passive standing tasks; 2) the H-reflex was modulated in a phase-dependent manner in the both passive stepping tasks.
These results are similar to those observed H-reflex modulation in normal standing and walking conditions in human subjects. Since there were negligible background EMG activities in both TA and SOL, it was conceivable that there were little influence from the higher nervous centers in the passive motor tasks in this study. It is strongly suggested, therefore, that the locomotor task dependent and phase dependent H-reflex modulation in normal conditions are in large part influenced by peripheral sensory inputs accompanied with stepping movements. Further, load related sensory inputs might have little influence on the suppression and phase-dependent modulation of H-reflex during stepping.

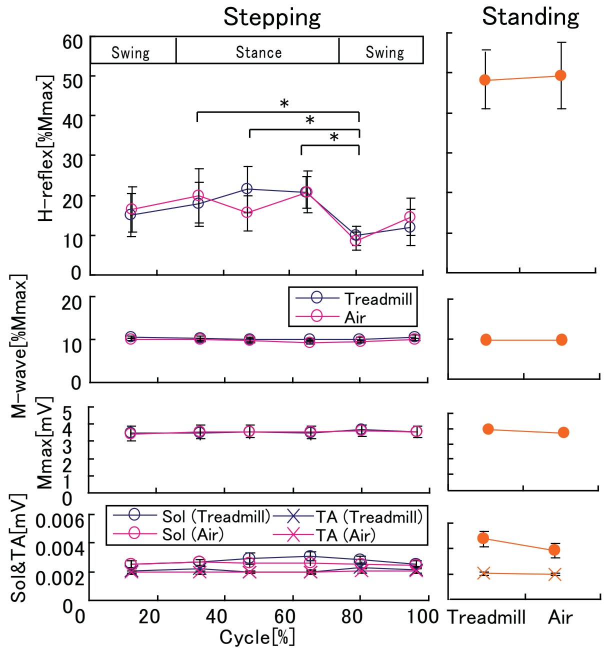

Figure 3: Means and standard errors (n=10) of H-reflex, M-wave, Mmax and background EMG activities in the passive standing and passive stepping tasks. * $p<0.05$

SUMMARY/CONCLUSIONS

Modulation of SOL H-reflex was tested in passive stepping and standing tasks provided by a robot gait orthosis. The results showed that the H-reflex was significantly reduced in the stepping tasks compared to those in the passive standing tasks; 2) the H-reflex was modulated in a phase-dependent manner in the both passive stepping tasks. These results suggest that somatosensory afferent inputs have a significant role in task- and phase-dependent modulation of H-reflex in human bipedal posture and gait.

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NOVEL INSIGHTS INTO CRAMP PATHOPHYSIOLOGY FROM M-WAVE ANALYSIS DURING CRAMP DISCHARGE

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INTRODUCTION

We recently developed a neurostimulation method for inducing cramps in an intrinsic foot muscle, the abductor hallucis, by means of electrical stimulation of the main muscle motor point: it was reliable, tolerable, and suitable to be used concomitantly with multichannel surface electromyography (EMG) (Minetto et al. 2008).

The present study aimed to examine if different frequencies of electrical stimulation, applied to the muscle motor point, trigger different muscle cramps and to analyze their EMG behaviour in time and frequency domains.

METHODS

One session of electrical stimulation of the dominant abductor hallucis muscle was performed in 15 volunteers: stimulation trains of 150 monophasic square pulses with duration of 152 μs were applied. Frequency was increased starting from 4 pps and with steps of 2 pps until a cramp was induced. The current intensity was 30% higher than that eliciting maximal M-wave and recovery between trials was 1 min. Once a cramp was induced, 30-min rest was provided to the subject before a second cramp was elicited with a stimulation frequency increased by 50% with respect to that eliciting the first cramp: the two cramps will be indicated as threshold cramp and above-threshold cramp.

EMG signals were detected with a surface electrode array (8 contacts, 5 mm apart) which was placed between the motor point and the distal tendon.

RESULTS

Figure 1 shows two typical examples of SD channel EMG from the post-stimulation recording (first second after the last M-wave) and relative power spectrum. For every subjects, we found a greater amplitude of the signal and a compression of the power spectrum for the above-threshold cramps with respect to the threshold cramps.

M-wave analysis showed that:
a) M-wave changes (ranging between small modifications of M-wave amplitude to complete M-wave disappearance) occurred in both threshold and above-threshold trials,
b) an important interindividual variability was observed in the time of cramp onset (that is, in the time of occurrence of M-wave changes) during the stimulation train, and
c) M-wave changes tended to increase throughout the stimulation train. Significant positive correlations were found between estimates of EMG amplitude during cramp

and estimated reductions of M-wave amplitude.

SUMMARY/CONCLUSIONS

In summary, stimulation trains of the same amplitude but different frequencies, applied to the muscle motor point, triggered different sized cramps of the abductor hallucis and changes in M-wave amplitude occurred during the course of threshold and above-threshold stimulations.

Our interpretations of these experimental findings fit well with the concept that action potentials during cramp are generated at the level of the terminal branches of motor axons.

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ACKNOWLEDGEMENTS

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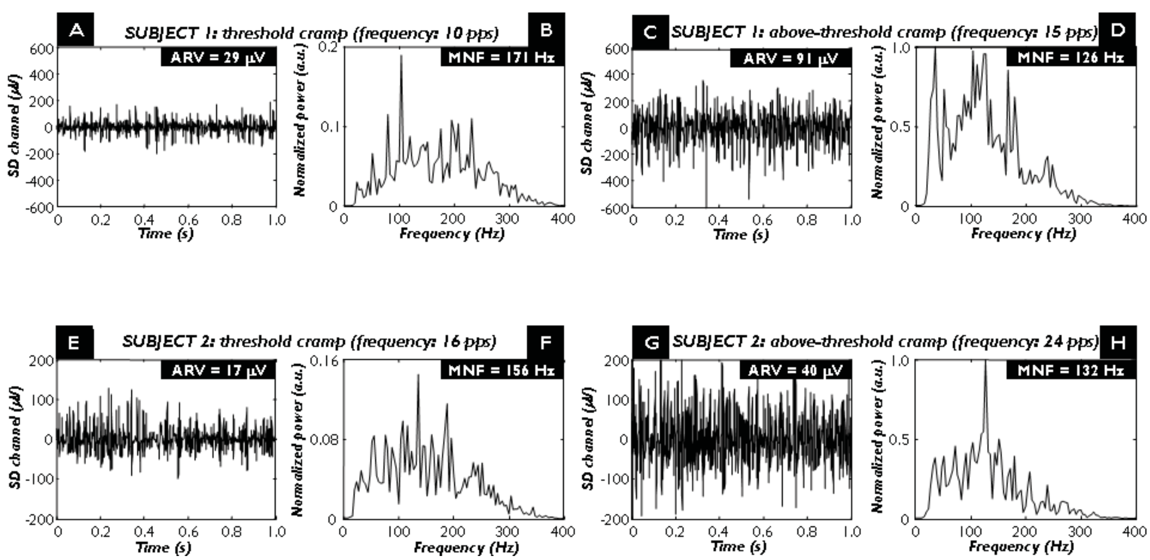

Figure 1: *Examples of threshold (panels A and E) and above-threshold (panels C and G) cramps of the abductor hallucis muscle in the post-stimulation EMG recording (1-s long signal epoch, immediately after the last M-wave) from two subjects, with the relative smoothed power spectra (obtained by means of a moving average performed on 4 points and normalized with respect to the peak of the power spectrum of the above-threshold cramp: panels B, D, F, and H). Average rectified value (ARV) and mean frequency (MNF) estimates are reported for each signal and relative power spectrum. Greater amplitude of the signal and compression of the power spectrum were evident for the above-threshold cramp with respect to the threshold cramp.*

MOTOR RESPONSE EVOKED DURING PROLONGED STIMULATION TACTILE: ELECTROMIOGRAPHIC AND KINEMATIC EVIDENCE.

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INTRODUCTION

Prior work has put forward the hypothesis that complex movement is constructed by the combination of stereotypic motor patterns 'stored' in cortical and spinal networks. Combinations of such "building blocks" could be selected by the motor cortices providing and advantageous computational procedure to optimally plan and rapidly execute complex movements. Force field correlates of such "primitives" have been recorded in the cord of spinalized reptiles and amphibians, but its existence and physiological role in intact mammals - particularly humans- remains very controversial. In the current study we used tactile stimulation of specific body regions in healthy humans to elicit a set of stereotypical but highly complex involuntary motor patterns, providing EMG and kinematics evidence of their existence

METHODS

A group neurologically sound subjects (n=4) were positioned in *supine decubitus* with a 30 degrees lateral head rotation. Then they were constantly stimulated for 5 minutes with an anterior-to-posterior, lateral-to-medial and cephalic force vector applied digitally on the pectoral region between 6th and 7th intercostals spaces in the left hemiside (following VOJTA therapy principles). In a calibrated 3D space, kinematics coordinates of the contralateral ankle, knee, hip, shoulder, elbow and wrist and EMG correlates from the Medial Deltoids (DELT), Biceps Brachii (BB), External Abdominal Oblique (OBL), Rectus Femoris (RECT), Tibialis Anterior (TA) and Semitendinous (ST) muscles were synchronously recorded.

RESULTS

1.- The results show a sequential pattern of involuntary muscle activations and movements starting at with the OBL (45.1 sec, post stimulus onset) and the ST (43.6 sec.), then the TA (53.1 sec.) and RECT (56.5 sec), followed much later by the DELT (88.5 sec) and BB (94.3 sec). Kinematically, we observed a sequential and progressive pattern towards unilateral and/or bilateral tight flexion, leg extension followed by shoulder abduction, forearm flexion, and eventually wrist supination. Both EMG and kinematic activation chains were highly consistent within and across subjects.

2.- Repetition of each pattern at 5 minutes intervals for up to 3 times provided evidence of a *progressive facilitation* of the EMG response components, appearing earlier and reaching higher values. Subjects maintained consciousness and normal cognitive abilities during the phenomena, but reported *"feeling their extremities being manipulated like if they were puppets"*. All subjects were able to interrupt such induced "involuntary" patterns at will when requested.

3.- Differences in the time of appearance of the global motor pattern stereotyped in all the subjects were observed, diminishing the time interval between the beginning of the activation and the obtaining of the complete pattern between 1 test, 2 test and test 3.

4.- In each experiments and all the subjects, the muscular contraction present display a similar temporary sequence: oblique abdominal external, semitendinous, tibial previous, rectus femoris, median deltoid and biceps brachial.

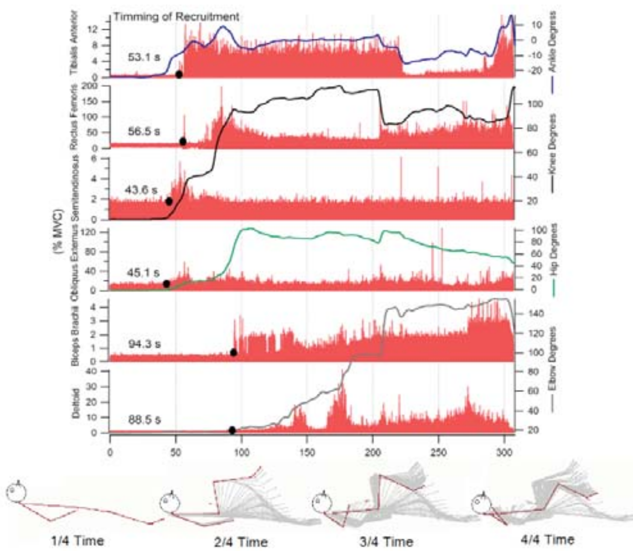

Figure 1. Observed response times muscle associated with the kinematics, caused by the prolonged tactile stimulation.

DISCUSSION AND CONCLUSIONS

1.- The preliminary results show the appearance of expressed motor answers in global and stereotyped patterns of movement with a similar kinesiology configuration.

2.- In registry SEMG, the sequence of appearance of the muscular answers and the progressive muscle recruitment different, until forming a global motor pattern. This addition of the registered muscular answers observed as much in the greater intensity of the muscular contraction like in a smaller time interval between 1ª, 2ª and to 3er experiment could be explained like a phenomenon of facilitation of the muscular contraction.

3.- We conclude that propioceptive stimulation of body regions is able to elicit complex patterns of cortico-spinal "sinergies" that might be used as modules by the motor cortices to optimally plan and execute movement.

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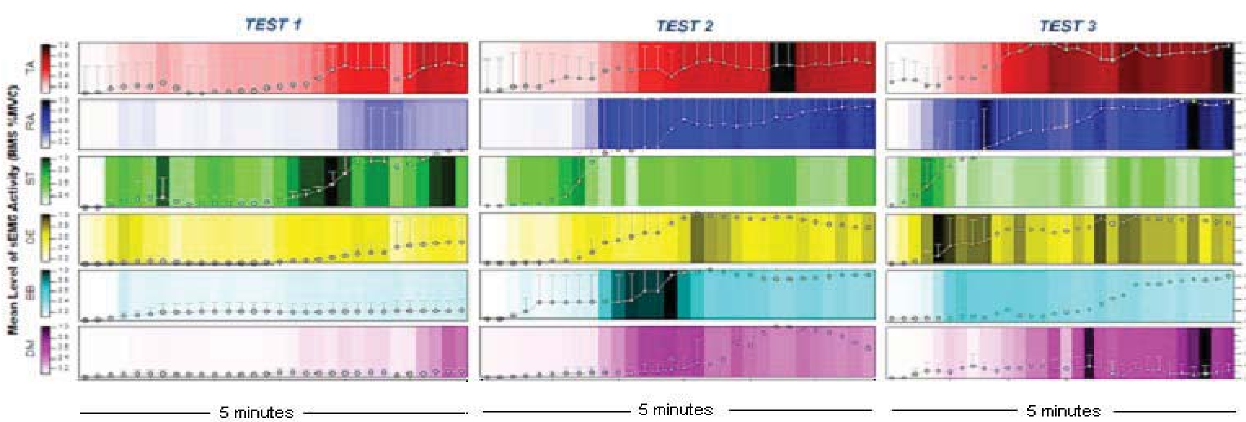

Figure 2. This graph illustrates progressive facilitation of the muscular response (normalized RMS) during 3 test. We observed the changes kinematics evoked in each test by the prolonged tactile stimulation.

Track 10

Physical Medicine and Rehabilitation (PM)

MEASURING THE EFFICACY OF A NOVEL BIOFEEDBACK DEVICE FOR POST-STROKE REHABILITATION OF THE UPPER EXTREMITY

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INTRODUCTION

Rehabilitating motor function after brain-injury can mitigate impairment by making use of the brain's plastic properties. It has been shown that activity in the affected area of the brain can be increased during the performance of repetitive bimanual tasks [Staines 2001]. Biofeedback via electromyographic visualization has been used to improve arm function in patients with severe impairment [Crow 1989].

Presently, we introduce a novel biofeedback-based training paradigm for restoration of motor control to hemiparetic individuals following stroke. A pilot study has been performed on a single subject performing cyclic pinch-and-open tasks.

The present study approaches rehabilitation with the goal of improving functionality in chronic stroke patients using Force Myography (FMG) of the forearm to measure activity during repetitve unimanual tasks with the affected hand. The high interrater compatibility between grip force and forearm muscle activation has been demonstrated previously [Wininger, 2007].

Feedback is given during activity as a scalar visual representation of the accuracy with which an initial pinch is being reproduced, a form of biofeedback that is unique to our device.

The efficacy of this rehabilitative tool is measured by comparing subjects' activity with and without feedback. It is expected that given visual feedback, chronic stroke subjects will achieve accuracy more rapidly and sustain it for a greater duration than they do without feedback.

METHODS

Data is collected using Force Myography (FMG), collected an array of four Force Sensitive Resistors (FSRs). The FSRs are arranged on a standard therapeutic cuff, which is donned such that sensors are located on the anterior and posterior musculature of the forearm.

Muscular activity changes the radial pressure exerted by the muscles. In this way, the FSRs register signals that reflect activity of the hand [Kim 2007]. Unlike electromyography, FMG signals do not require computationally intensive processing.

First, the system is calibrated to a subject's baseline and maximum FMG values. Under supervision, the subject sets the template for "resting" and "pinching" conditions. Then, timed by a metronome set at 0.25 Hz, the subject attempts to reproduce the templates, alternating between pinching and resting.

In the With Feedback (WF) condition, visual feedback is given as a scalar representation of the distance between the template point and current activity in the 4-D sensor space. In the No Feedback (NF) condition, the subject is instructed to either Pinch or Rest.

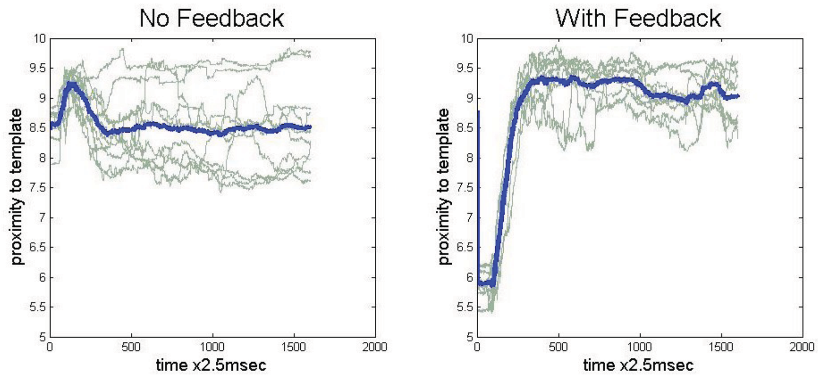

Figure 1: Individual (gray) and Ensemble-Averaged (blue) pinch distance from template, during No Feedback (left) and With Feedback (right) conditions. Higher values indicate a smaller distance.

The efficacy of our device is measured by comparing performance between the WF and NF conditions. Specifically, the rate of approach, the proximity to the template, and the duration of activity are compared. The rate of approach is given for z-normalized records, and the duration is estimated using the median of the epoch between the maximum and the end of each repetition.

RESULTS AND DISCUSSION

Results are presented for one subject, a 56 year old male in his 4th year after stroke. The distance between the subject's activity and the template are shown in Figure 1, and the results are presented in Table 1.

Notice that the subject's maximum proximity to template and the rate of approach are greater during the WF condition but may not be statistically different between conditions. However, the duration of activity is significantly greater in the WF condition, as shown in Figure 1.

SUMMARY/CONCLUSIONS

The difference between the NF and WF conditions suggests that our device significantly improves motor control in Brain Injury subjects. Without visual feedback, subjects are capable of desirable muscle activity; however, adding the scalar representation of proximity to template significantly improves their ability to sustain that activity. Future studies will assess the benefits of that sustained activity for the long-term rehabilitation of Stroke and Traumatic Brain Injury subjects.

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Table 1: Performance values during Repetitive Pinching. Unitless results are presented as mean ± std, except Duration, estimated by the median proximity after maximum.

Metric	No Feedback (NF)	With Feedback (WF)
Rate of Approach	28.5 ± 18.9	38.0 ± 7.9
Proximity to Template	9.53 ± 0.18	9.7 ± 0.15
Duration of Activity	0.146	0.871

MOTOR UNIT PROPERTIES OF THE BICEPS BRACHII OF STROKE PATIENTS ASSESSED WITH SURFACE ARRAY EMG

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INTRODUCTION

As a consequence of a stroke, both motor control and motor unit (MU) characteristics may change. For example, MU size has been reported to increase due to reinnervation (Lukacs, 2005, Cruz Martinez et al. 1982, Dattola et al. 1993). Previously, changes in MU characteristics after a stroke have been investigated using intramuscular electromyography (EMG). The development of array surface EMG enables the non-invasive investigation of MU characteristics. The aim of the present study was to investigate differences between the biceps brachii of the affected and the unaffected side of hemiparetic stroke patients using surface array EMG parameters.

METHODS

Fifteen hemiparetic stroke subjects participated. Their Fugl-Meyer score (upper extremity part) was asessed. They performed isometric elbow flexion contractions consisting of 10 force levels of 5 to 50% of the maximal voluntary contraction (MVC) with both sides. A two-dimensional 16 channel electrode array was placed on the skin above the active biceps brachii. MUAPs were extracted using the segmentation part of the decomposition software described in Gazzoni et al. 2004. MUAPs can be characterized by parameters describing their amplitude (RMS_{MUAP}) and their frequency content ($FMED_{MUAP}$, median frequency of the power spectrum of the MUAP shape). RMS_{MUAP} is related to the size of the motor unit (MU), the frequency content is related to the conduction velocity of the MU (Lindstrom and Magnusson 1977).

RESULTS AND DISCUSSION

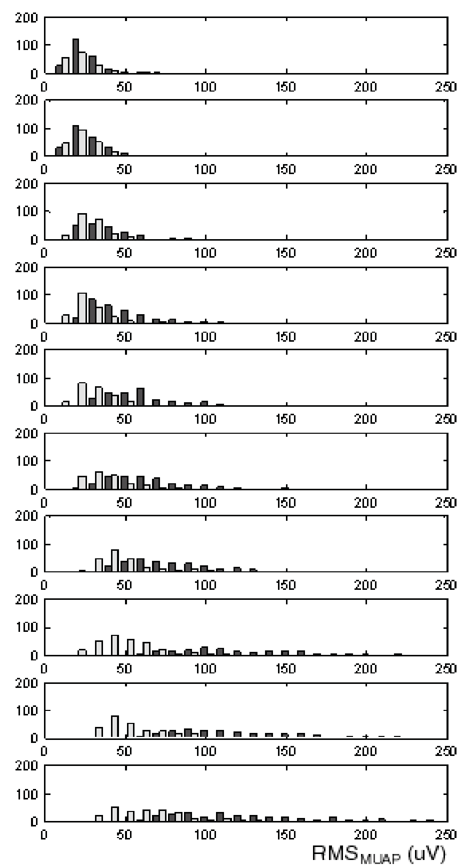

Figure 1 Example of RMS_{MUAP} distribution for the ten force levels from 5% (upper graph) to 50% MVC (lower graph). Dark bars: affected, light bars: unaffected side

An example of the distribution of RMS_{MUAP} of both sides is shown in Figure 1. Of the 15 subjects, 7 subjects showed larger RMS_{MUAP} values at the affected side, 5 subjects showed smaller values and 3 subjects did not show differences. Interestingly, the median Fugl-Meyer scores were considerably higher in the group with larger RMS_{MUAP} values at the affected side (Fugl-Meyer score 42 versus 20 out of 66).

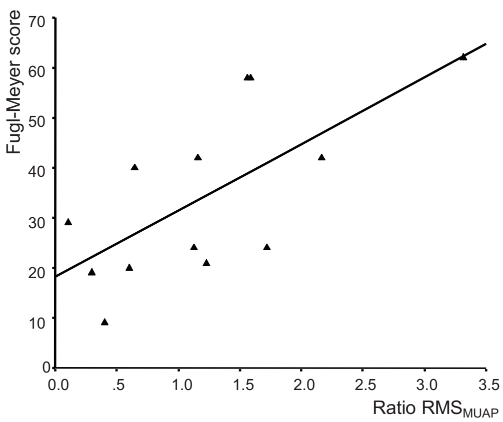

Figure 2 Relation between the ratio of RMS_{MUAP} at the affected side divided by RMS_{MUAP} at the unaffected side and Fugl-Meyer score. Data from step 6 (30 % MVC). The explained variance is 47% ($p<0.011$).

The ratio of RMS_{MUAP} of the affected side divided by that of the unaffected side correlated significantly with the Fugl-Meyer score for the force levels from 15% to 45% (Spearman's rho between 0.60 and 0.74, $p<0.039$, see Figure 2). $FMEAN_{MUAP}$ was smaller at the affected side ($p<0.001$).

A larger RMS_{MUAP} value indicates that the size of the MUAP as recorded at the skin is larger, which indicates larger MUs. This is in agreement with findings of electrophysiological studies and has been suggested to be caused by the occurrence of reinnervation of muscle fibers by collateral sprouting and branching (Lukacs 2006, Cruz Martinez et al. 1982, Dattola et al. 1993). Therefore, RMS_{MUAP} ratio might reflect the amount of reinnervation at the affected side, which in turn likely is related to the functional capacity of the muscle.

The smaller $FMEAN_{MUAP}$ values at the affected side might indicate a larger contribution of low-threshold MUs. This might be explained by selective degeneration; it has been shown that mainly type II MUs (generally high-threshold) are affected by degeneration, while type I MUs (generally low-threshold) remain, which would increase the relative contribution of low-threshold MUs. This effect might be enhanced considerably by collateral sprouting of the remaining low-threshold MUs, as has been suggested by different authors.

SUMMARY/CONCLUSIONS

High-density sEMG recordings of chronic stroke patients revealed differences in MUAP size and in frequency content of the MUAPs between the affected and the unafffected side. An interesting finding is the correlation of the RMS_{MUAP} ratio with the Fugl-Meyer score for the upper extremity, which might be a reflection of reinnervation at the affected side.

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FUNCTIONAL IMPLICATION OF GAIT AFTER LEFT OR RIGHT-SIDED STROKE

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INTRODUCTION

The cerebrovascular infarction (stroke) presents clinical manifestations that reflect location and extension of the vascular injury (Cifu, Lorish, 1994).

The scientific community recognizes that functional asymmetry exists between the cerebral hemispheres; however, the topic yet to be resolved is whether there is "a difference in functional rehabilitation of individuals with right and left-sided stroke".

Due to the mentioned facts above, the objective of this study was to quantify and compare the parameters of muscle electrical activity and ground reaction force of lower limb during gait among post-stroke volunteers with left or right side functional involvement to verify whether a difference exists regarding functional rehabilitation of these individuals.

METHODS

Data was collected on 15 post-stroke volunteers with left side functional involvement, and 15 post-stroke volunteers with right side functional involvement. Volunteers were age, gender and weight matched.

All volunteers were informed about the study protocol and signed the respective participation consent term.

Four pairs of active surface electrodes (bipolar and differential, and signal was pre-amplified in the differential electrode with a gain of 10 times and common mode rejection ratio [CMRR] of 80dB) were positioned on the motor point of the muscles rectus femoris, tibialis anterior, soleus, and medial portion of the hamstrings on the volunteer's affected side (spastic side).

The components of the signal acquisition system were connected to a signal conditioning module where analog signals were filtered with a bandpass filter of 10Hz to 500Hz, and amplified once again with a gain of 100 times. Consequently the total gain resulted in 1000 times.

After signal treatment, an average trace of the ten strides of the complete gait cycle was obtained, representing the muscle's functional activity of each volunteer. Finally, average traces representing the volunteers' functional activity were obtained and submitted to comparative statistical analysis regarding the analyzed muscles.

Regarding data on ground reaction force, it was collected with *Gaitway Instrumented - Kistler* System, consisting of two force plates based on piezoelectric transducers. A 12-bit resolution A/D converter was utilized and ground reaction force signal was sampled at 1000 Hz frequency, filtered with 4th-order digital Butterworth filter with a cutt-off frequency of 5 Hz.

The force plate not only analyzed ground vertical reaction forces, but also gait velocity, weight bearing index, cadence, and stride length. Data related to Fy1, Fy2 and

Fymin were normalized with the volunteer's body weight and expressed in N/kg.

The statistical cross-correlation analysis was applied in each point during the entire stride cycle (similarity measure between two samples), to compare electromyographic curve form variation pattern. The ground reaction force average values were submitted to the Student's t Test.

RESULTS AND DISCUSSION

Hemiparetic patients present gait with lower velocity, asymmetry, and sway more laterally towards the non-involved limb, independently of the functionally involved side. Although variability was observed on stride duration when comparing both sides, along with a longer swing phase on the hemiparetic side, no statistically significant difference was observed (p=0.6) on the gait of these patients after stroke with left or right side functional involvement.

Different electromyographic activation moments during gait, particularly on knee and ankle joints, of hemiparetic patients with stroke sequela are explained by the spasticity of the involved limb.

The average values and standard deviations obtained from vertical ground reaction force of the involved side during the entire stride cycle no statistically significant difference was observed (p=0.65). Moreover, gait velocity, cadence and stride length did not present statistically significant difference (p=0.53).

This characteristic observed for motor behavior is consistent with the results of our study and with other authors (Corrêa et al., 2005). We observed that decreased gait velocity is directly associated with the high level of muscle co-activation in post-stroke patients (Falconer, Winter, 1985).

Consequently, angular velocity could be affected by spastic hypertonia (velocity dependent) in antagonist muscles or even co-spasticity of the agonist/antagonist muscles, promoting deceleration forces that limit angular velocity magnitude.

SUMMARY/CONCLUSIONS

In conclusion, regarding the parameters of muscle electrical activity and ground reaction force of the lower limb during gait of post-stroke volunteers with left or right side functional involvement, we could suggest that there is no difference in functional rehabilitation of these individuals.

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THE EFFECTS OF CIRCUMFERENTIAL AIR-SPLINT PRESSURE ON THE SOLEUS STRETCH REFLEX IN SUBJECTS WITH AND WITHOUT CEREBRAL VASCULAR ACCIDENT

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INTRODUCTION

Circumferential pressure (CP) has been shown to decrease muscle activity in subjects without neuromuscular disorders and in individuals with spinal cord injury and cerebrovascular accidents (Robichaud 1992,1996). These conclusions have recently become under scrutiny due to the methodologies used to analyze its effects (Agostinucci 2007). Thus far, studies used to determine CPs efficacy on motoneuron reflex excitability (MNRE) have mainly used the H-reflex technique on resting muscle. Although a valued technique, the H-reflex has many shortcomings that may lead to misinterpretation of results unless strict adherence to procedural protocols is adhered to. This aspect coupled with the lack of evidence from studies when muscles are contracting have led researchers to reconsider their initial conclusions and reinvestigate CP's effect once again. Thus the purpose of this study was to investigate the effect CP has on MNRE using the soleus stretch reflex (SSR) superimposed on a ramp movement in subjects without neurological deficits (S_{nd}) and subjects with cerebrovascular accident (S_{cva}).

METHOD

Participants

Forty-eight S_{nd} and 13 S_{cva} volunteered for this study. All subjects read and signed an informed consent form approved by the University's Institutional Review Board.

Procedure

SSR's were investigated before, during and after the application of pressure to the calf. An inflated 19cm blood pressure cuff connected to a pressure transducer was used to administer and measure air pressure. Pressure was set to 40-45mmHg. For S_{nd}, SSR's were elicited by dorsiflexing the subject's ankle 10º at 180º/sec while the subject plantarflexed against a moving footplate at 20% their maximum voluntary contraction (MVC) through a 30º arc at 90 º /sec. The same parameters were used to elicit SSRs for S_{cva} except they were not required to plantarflex against the moving footplate at 20% MVC.

ANALYSIS

Twenty-five SSR's were recorded and averaged for each experimental phase. Peak to peak amplitudes were measured and normalized to pre movement mean EMG baseline activity (50ms window). Reflex latency was also measured. Two-way ANOVAs with repeated measures were used to analyze the differences between S_{nd} and S_{cva} and the change in SSR latency / amplitude from baseline values.

RESULTS AND DISCUSSION

S_{nd} demonstrated a significant mean reduction of ~ 13% while S_{cva} showed a more dramatic decrease of ~35% (Figure 1). This greater reduction in S_{cva} was not significantly different from S_{nd} and was mainly due to the large variability in S_{nd} (Figure 2) . Latencies showed no changed from baseline values.

SUMMARY AND CONCLUSIONS

Our results show CP has an over all inhibitory effect on spinal cord muscular reflexes. Results further suggest that air splints may be used to temporarily reduce lower extremity muscle activity associated with a neurological dysfunction. Clinicians, however, need to be cognizant that CP may not effect everyone in the same way.

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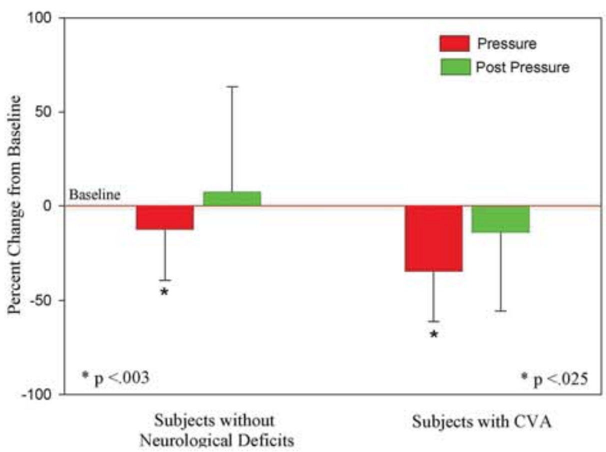

Figure 1: Comparison of SSR amplitude data from S_{nd} and S_{cva}

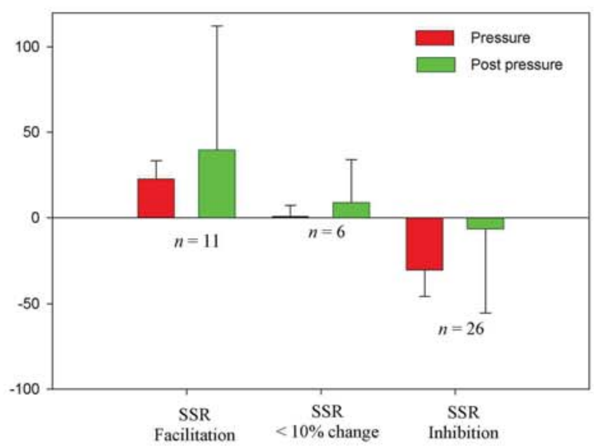

Figure 2: Soleus Stretch Reflex amplitude data for S_{nd} grouped by effect of circumferential pressure. Subjects were grouped by having a ±10% SSR amplitude change from baseline values. Means, standard deviations and number of subjects are provided. (0 = baseline level)

AN ELECTROMYOGRAPHIC ANALYSIS OF ALTERING FOOT PROGRESSION ANGLE DURING GAIT

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INTRODUCTION

Toe out gait has been shown to reduce the medial compartment loading environment during gait. This affect has been quantified through gait analysis and knee adduction moment investigation (Guo et al. 2007). The loading environment of the knee joint can also be affected by subsequent muscle forces experienced during gait. Investigating differential activation is important for understanding for neuromuscular control strategies and joint loading mechanisms (Hubley-Kozey et al. 2006, Lewek et al. 2004). While many studies support that toe out gait affects the knee adduction moment, to our knowledge no study has evaluated the neuromuscular demands of this modification in any population.

The purpose of this study was to investigate the overall magnitude of the electromyogram (EMG) of eight lower extremity muscles during neutral and toe out gait in a healthy adult population using principal component analysis (PCA).

METHODS AND PROCEDURES

Twenty-two volunteers (13 females and 9 males) between the ages of 35 and 60 ($\bar{x}$ 46.7 and SD 7.6) years participated in this study. All subjects gave written informed consent. After appropriate skin preparation procedures, bipolar skin surface electrodes (Ag/AgCl-interelectrode distance 20mm) were placed over the lateral gastrocnemius (LG), medial gastrocnemius (MG), vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), lateral hamstring (LH) medial hamstring (MH) and gluteus medius (GM). Subjects completed at least five trials of toe out walking (TO) and five trials of neutral walking (N) at their self-selected speed. Criteria for toe out gait included a foot progression angle at least 10 degrees greater than neutral and velocity within 0.1 m/s. Maximal voluntary isometric contraction (MVIC) exercises were completed for normalization purposes. Three-dimensional motion and ground reaction force data were recorded during gait for foot progression angle calculations and waveform time normalization. The myoelectric signals were collected at 1000 Hz using an AMT-8™ (Bortec, Inc., Calgary, Alberta) EMG measurement system. All of the EMG waveforms were corrected for subject bias, full wave rectified, low pass filtered (Butterworth, zero lag, 6Hz), amplitude normalized to MVIC and time normalized to represent 100% of the gait cycle.

PCA was used to extract the waveform feature from each muscle group that explained the variance in the overall magnitude of the EMG signal (Hubley-Kozey et al. 2006). *PC-Scores* were computed for each original EMG waveform. Repeated measures ANOVA models were used to test for walking condition and muscle main effects and interactions. Bonferonni post hoc adjustments were used to test significance at alpha = 0.05.

RESULTS AND DISCUSSION

The FPA was significantly different between conditions (5 ± 4.5 degrees (N) and 21.5 ± 4.5 degrees (TO)) ($p<0.05$). Self-selected velocity was not different between conditions (1.46 ± 0.14 m/s (N) and 1.47 ± 0.15 m/s (TO)) ($p>0.05$). For each muscle group, principal component (pattern) one (PC1) captured the overall magnitude of the waveforms, explaining 60%, 66% , 67% and 62% of the variance in the calf, quadriceps, hamstrings and gluteus medius waveforms respectively. The principal patterns of these muscles, except gluteus medius (shown in figure 1) have been similarly described in the literature (Hubley-Kozey et al. 2006).

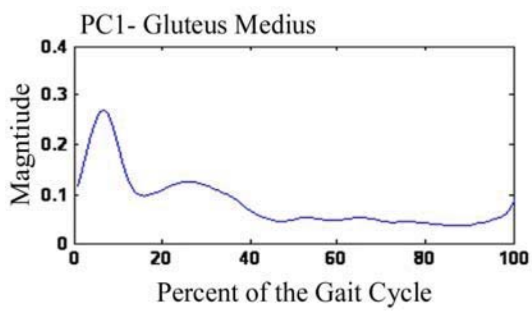

Figure 1: PC1 for Gluteus Medius

The ensemble averaged EMG waveforms, separated by condition, for each of the eight muscles are shown in figure 2.

A significant condition main effect was not found for any of the muscles ($p>0.05$) suggesting that the overall magnitude and principal shape of the myoelectric activity did not differ between subjects walking a neutral alignment at a self-selected velocity and adopting a toe out gait of approximately 15 degrees greater than neutral.

SUMMARY/CONCLUSIONS

A voluntary toe out gait modification does not affect muscle activation during gait. Where other studies have focused on the kinetics of gait, these findings suggest that this gait modification does not alter the myoelectric activity of the major muscle groups in the lower extremity that have potential to influence knee joint loading.

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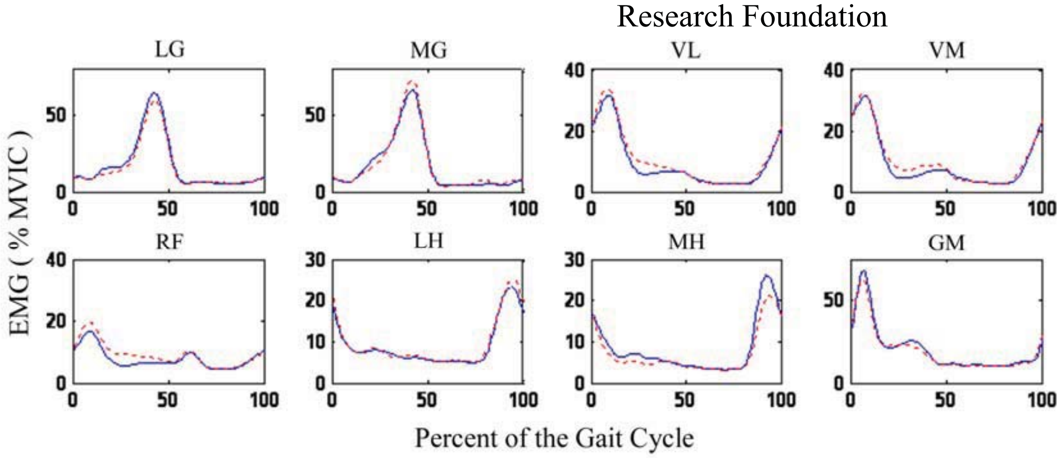

Figure 2: Ensemble averaged waveforms for the eight lower extremity muscles. Solid line (Neutral Walking (N)), Dashed line (Toe out walking (TO)).

TIME-FREQUENCY ANALYSIS OF LEG MUSCLES DURING GAIT IN PATIENTS WITH DIPLEGIC CEREBRAL PALSY

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INTRODUCTION

Cerebral palsy (CP) is the most common cause of severe physical disability in childhood and results from an injury or damage in the developing central nervous system. The characteristics are spasticity, movement disorders, muscle weakness, ataxia, and rigidity [Koman et al, 2004]. In the past, surface electromyograms (SEMGs) during gait provided valuable information with respect to on- and offset timing of muscles, the frequency content, and the amplitude. However, only one aspect could be examined in isolation. The introduction of an SEMG-specific non-linearly scaled wavelet analysis provided a simultaneous intensity, time, and frequency analysis [Von Tscharner, 2000]. The objective of this study was to investigate the time-frequency content of SEMGs over the gait cycle in patients with diplegic CP using a wavelet analysis technique.

METHODS

SEMGs were collected according to SENIAM guidelines (Ag/AgCl electrodes, 10-700 Hz bandpass-filter, sampling-rate 2520 Hz) of the gastrocnemius medialis (GM), tibialis anterior (TA), rectus femoris (RF), and semitendinosis (ST) muscles during gait at a self selected speed. For each subject, data of 12 gait cycles were analysed (foot-strike to foot-strike) and averaged to retrieve the time-frequency content using a wavelet analysis technique [Von Tscharner, 2000]. Then the mean frequencies were calculated over the gait cycle. The mean frequencies of the wavelet spectra of 12 children with diplegic CP (age: 12.6±4.1 years) and 11 healthy children (age: 11.6±3.5 years) were compared (T-test $P<0.05$). Clinical assessment included rating spasticity (modified Ashworth scale), rating manual muscle force (scale 1 to 5), and examining passive range of motion.

RESULTS

All patients showed muscle weakness. 6 patients showed spasticity at foot, knee, and hip level and 6 children only at the foot and knee. Mean frequency over the entire gait cycle was significantly higher for the children with CP compared to healthy children for all investigated muscles (GM: 149±30 Hz vs. 95±9 Hz; TA: 133±25 Hz vs. 105±18 Hz; RF: 114±16 Hz vs. 78±7 Hz; ST: 120±20 Hz vs. 86±14 Hz). Differences in frequencies were also observed for 9 individual gait phases.

DISCUSSION AND CONCLUSIONS

This study showed that a group of children with diplegic CP activate their muscles at higher frequencies during gait than a group of healthy children (Figure 1). It is well known that patients with CP have muscle weakness [Wiley and Damiano, 1998; Elder et al., 2003]. Therefore, during functional tasks such as walking, the lower force-generating capacity in children with CP creates a lower force reserve to sustain activity, so that children with CP are functionally more fatigable [Stackhouse, 2005]. Muscle fibers are classified in particular types and can be considered as slow or fast depending on its fiber-type composition. In response to the force requirements, motor units are typically recruited in a set order from slow to fast. Several main mechanisms are described in the literature contributing to muscle weakness in CP. Co-activation of antagonist muscles [Elder et al., 2003; Ikeda et al., 1998] and the inability to maximally activate the muscles [Rose and McGill, 2005] are two of them. But also altered muscle morphology may be a source of weakness. There are a variety of other factors affecting the frequency in a wavelet spectrum, the most important ones are the shape of the motor unit action potential and the conduction velocity [Wakeling et al., 2002]. These factors depend on the fibers used by the muscles and may thus indicate the new distribution or types that resulted from the disease. Treatment interventions (Botulinumtoxin-A treatment, surgery, strength training) may change this distribution and wavelet analysis may be used to monitor the effect. Wavelet analysis could potentially be a method for studying treatment intervention, help with the assessment of motor function, or give insight into the neuro-muscular mechanisms of CP.

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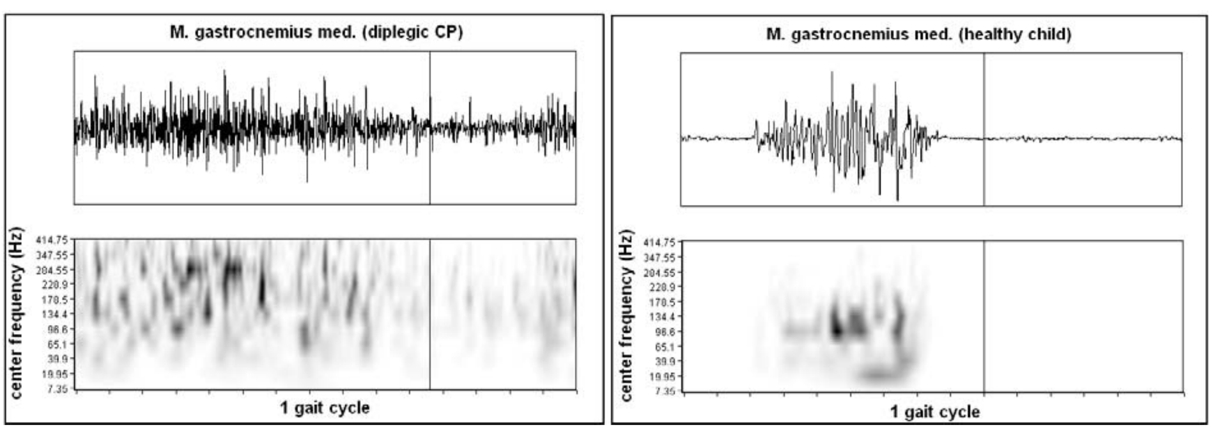

Figure 1: Raw EMG and the same data analysed with a wavelet analysis for 1 patient with diplegic cerebral palsy and 1 healthy child.

THE EFFECT OF CIRCUMFERENTIAL PRESSURE ON F-WAVE RESPONSE IN THE TIBIALIS ANTERIOR MUSCLE

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INTRODUCTION

Circumferential pressure (CP) has been shown to decrease muscle activity in subjects without neuromuscular disorders and in individuals with spinal cord injury or cerebrovascular accidents (Robichaud 1992,1996). The mechanism for this decrease is unknown although it has been hypothesized to be spinal in origin (Robichaud 1996). Spinal mechanisms of Ia reciprocal inhibition, pre-synaptic inhibition, nerve ischemia and nerve compression have all been studied and shown not to be involved (Perkins 1997, Connors 2003, King 2007). These results suggest CP's affect on the reflex arc may be on the motoneuron (MN) directly via post synaptic inhibitory mechanisms or the mechanical properties of the muscle itself. F-waves are late EMG responses to nerve stimulation produced by antidromic activation of the peripheral motor fibers resulting in recurrent discharge of MNs. Thus, investigating them indirectly studies the spinal mechanisms that affect the efferent portion of the reflex arc including the muscle.The purpose of this study was to investigate the affects CP has on the F-wave in the tibialis anterior muscle. This study's results will help determine the effect CP has on muscle activity and will also assist in assessing where in the reflex arc CP has its effect.

METHOD

Participants Thirty-two healthy volunteers (14 males, 18 females) ages 22-53 participated in this study. All subjects read and signed an informed consent form approved by the University's Institutional Review Board.

Procedure

Subjects were seated in a comfortable specially designed chair with their dominant knee flexed to 30^{o} and their ankle in neutral. Two 9 mm surface recording electrodes were positioned on the tibialis anterior muscle 3 cm apart in alignment with the tendon. A 16 cm pneumatic blood pressure cuff was placed around their calf approximately 3 cm distal to the recording electrodes. During the pressure phase of the experiment the pneumatic cuff was inflated manually to 40-45 mmHg using a sphygmomanometer. F-waves were evoked by supra maximally stimulating (20% > Mmax) the deep peroneal nerve at .2 Hz using a bipolar surface electrode located on the skin just distal to the fibular head. F-wave recordings were taken at: 1) initial/baseline, 2) with CP applied, and 3) post CP. Fifty F-waves were elicited and recorded for each recording trial.

ANALYSIS

F-waves were identified using criteria described by Mesrati and Vecchierini (2004). Three parameters were analyzed within each recording trial: mean latency, persistence, and mean F/Mmax amplitude ratio. A one way repeated measure ANOVA was performed on each of the three parameters. Alpha level was set to $p < 0.05$.

RESULTS AND DISCUSSION:

No statistically significant difference was found for any of the F-wave parameters evaluated in this study.

SUMMARY AND CONCLUSIONS

CP does not affect the F-wave in the tibialis anterior muscle. The decrease in muscle activity observed in previous CP studies were not a direct result of alpha fiber conduction velocity or in the muscle itself. Our results do suggest that CP's effect on spinal reflex excitability is more complex that initially thought.

CLINICAL IMPLICATIONS

Clinicians should be cognizant that CP may not have the same effect on limb flexor muscles that it does on extensor muscles.

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THE EFFECTS OF TIZANIDINE ON THE MEDIUM LATENCY RESPONSE IN THE M. FLEXOR CARPI RADIALIS.

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INTRODUCTION

Etiology of spasticity is not yet fully understood. It is suggested that group II muscle afferents contribute to the exaggerated stretch reflex in spastic patients (Eriksson et al. 1996). Tizanidine, an α2 adrenergic receptor agonist, has a known spasmolytic effect in man (Emre et al. 1994) and a selective group II afferent activity blocking effect in cats (Hammar & Jankowska 2003). For the lower limb it is generally accepted that ramp-and-hold stretches produce a short latency, velocity dependent EMG burst (M1 response) and a non-velocity dependent, group II mediated M2 response. For the upper limb this is still controversial. We applied ramp-and-hold stretches to the *m. flexor carpi radialis* (FCR) while systematically varying stretch amplitude and applying tizanidine in a repeated measurements design.

METHODS

Ten healthy volunteers (age 47 sd 13 years, 9♂) received 4 mg of tizanidine orally. A control group of nine (age 43 sd 13 years, 7♂) included five subjects of the experimental group who were measured a second time. Stretch of the FCR was evoked by extending the wrist using a wrist manipulator under 1 Nm voluntary flexion torque. The protocol consisted of nine series of stretches applied every 20 minutes at eight different times (T_1 - T_8) after a baseline measurement (T_0) 10 minutes before oral tizanidine intake. Total measurement time thus was 150 minutes. Stretch series consisted of three different ramp-and-hold stretches with amplitudes of 0.06, 0.10, and 0.14 rad at a fixed speed of 2 rad/s. EMG of the FCR and ECR were simultaneously recorded and digitized at 2.5 kHz. EMG was rectified, low-pass filtered (2^{nd} order recursive Butterworth, 80 Hz). The data were segmented from 400 ms before onset till 1000 ms after onset and averaged over ten repetitions. Segments in which the mean flexion torque prior to onset of the perturbation deviated more than ±0.1 Nm from the instructed 1 Nm were rejected. Three periods were recognized in the EMG: background (400 till 20 ms before stretch onset); M1 (20-50 ms after onset); M2 (55-100 ms after onset). The M1 and M2 responses were defined as the ratio between the surface of the M1 and M2 in their respective period minus the background activity, scaled with the average background activity. Statistical testing was performed using repeated measurements GLM ANOVA (SPSS 11.0).

RESULTS AND DISCUSSION

There was a strong dependency of M2 not M1 response on stretch amplitude for both tizanidine and control group (overall $p < 0.001$, interaction term *M1M2- amplitude*: $p < 0.000$, Fig 1). While M1 did not change, the M2 response in the tizanidine group lowered in time compared to the control group for all three stretch amplitudes, but with the largest effect for the highest stretch amplitude (*time* effect: p=0.02, *time-M1M2* interaction: p=0.03 and *M1M2- time-amplitude* interaction: p=0.012 for the tizanidine group vs. a *time* effect of p=0.66 for the control group). In the control group, there was a significant drop of M2 response during the first two measurement sessions (p=0.02). The selective effect of stretch amplitude on M2 response of FCR strongly suggests a II afferent origin, although it was recently shown by model simulations that the stretch amplitude and thereby input duration dependent M2 response can entirely be explained from synchronization of the motorneuron pool (Schuurmans et al. 2007). The selective effect of tizanide on M2 does suggest a partial group II afferent origin of M2 and a role of group II (over) activity in spasticity. The aforementioned synchronization, the found adaptation effect and the fact that in previous studies even with threefold higher doses of tizanidine the M2 response did not completely disappear suggests a compound origin of the M2 response, contrarily to M1.

SUMMARY/CONCLUSIONS

Significant effects on the M2 response were established for stretch amplitude (positive), repeated measurements (negative) and tizanidine (negative), whilst the M1 response was indifferent to either variable. A partial, selective group II afferent inhibitory effect of tizanide on FCR is suggested confirming a role of group II (hyper) activity in spasticity. The M2 response of FCR appears to be of more complex nature compared to the M1.

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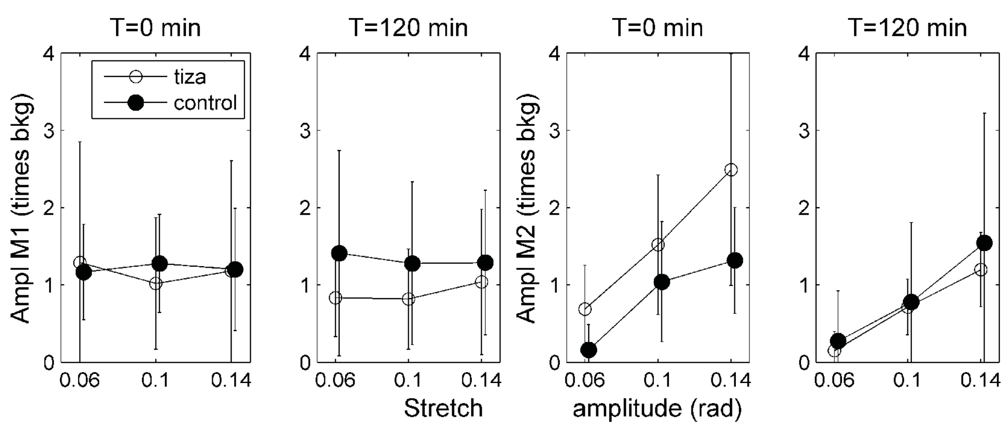

Figure 1. Stretch amplitude vs. M1 and M2 at t= 0 and 120 minutes for tizanidine (o) and control group (●)

EXPLORATION OF THE PHYSIOLOGICAL BASIS OF MUSCLE ACTIVITY WITH NERVE ELONGATION IN A CLINICAL SETTING

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INTRODUCTION

Many clinical tests to diagnose neuropathies evaluate the mechanosensitivity of the peripheral nervous system. Neurodynamic tests for example utilise elongation and compression of a peripheral nerve to assess its tolerance to mechanical loading.

Large quantities of connective tissue protect a peripheral nerve against mechanical deformation. In addition, it has been proposed that the muscular system actively protects the nervous system from excessive tensile forces. Besides symptom reproduction, this muscle activation is one of the clinical signs observed in the interpretation of neurodynamic tests.

Clinical (e.g., Goeken and Hof, 1993) and animal (Theophilidis and Kiartzis, 1996) studies have demonstrated muscle activity with nerve elongation. However, the findings are inconclusive and the neurophysiological basis of the muscle response is unknown. The muscle activity may be a nociceptive flexor withdrawal reflex or a stretch-mediated response. This study aimed to reveal the underlying neurophysiological mechanism of muscle activity in response to peripheral nerve elongation in a clinical setting.

METHODS

Muscle activity during the upper limb neurodynamic test for the ulnar ($ULNT_{ULNAR}$) and median ($ULNT_{MEDIAN}$) nerve was analysed. Eight healthy volunteers without head or neck pain or previous neuropathies were tested. Four variations of $ULNT_{MEDIAN}$ with varying degrees of mechanical loading of the median nerve were performed.

Both neurodynamic tests consist of a sequence of joint movements that elongate the length of the bedding of the median or ulnar nerve. It is important to realise that activation of the elbow flexors limits strain in the median nerve, whereas triceps activity is required to limit strain in the ulnar nerve.

Intramuscular and surface EMG was recorded from biceps and triceps brachii, brachialis and the upper trapezius muscle. Onset of pain and occurrence of submaximal pain were also recorded. We hypothesised that (1) triceps muscle activity would occur during $ULNT_{ULNAR}$ in support of a stretch-mediated response (activation of elbow flexors with $ULNT_{ULNAR}$ would suggest a flexor withdrawal reflex); (2) muscle activity would occur earlier in range during $ULNT_{MEDIAN}$ in the variations with increased nerve elongation.

Onset of muscle activity was determined from intramuscular recordings. EMG amplitude was calculated from the surface recordings and expressed as a percentage of maximal isometric voluntary contraction (MVC). EMG activity was calculated for a series of epochs, representing the mean EMG amplitude per 5 degrees elbow movement. Repeated-measures analyses of variance were used for statistical analysis. The level of significance was set at $p < 0.05$.

RESULTS

Compared to baseline levels, biceps and upper trapezius showed modest, but statistically significant increases in EMG activity during $ULNT_{ULNAR}$ ($p < 0.05$). In contrast to our hypothesis, triceps activity did not increase ($p = 0.99$). Submaximal pain was not elicited with $ULNT_{ULNAR}$.

For $ULNT_{MEDIAN}$, the onset of upper trapezius activity occurred earlier in the conditions where the mechanical loading of the nervous system was larger ($p = 0.0002$). In general, the onset occurred prior to the onset of pain, suggesting a stretch-mediated response. Although a trend for an earlier onset could be observed for brachialis, the difference was not significant ($p = 0.12$). No differences were observed for biceps ($p = 0.76$). Overall, a gradual increase in muscle activity could be observed towards the end of range for all variations of $ULNT_{MEDIAN}$ for all four muscles, although the level of activity was low, especially for triceps.

DISCUSSION & CONCLUSIONS

Although no increase in triceps EMG activity was observed during $ULNT_{ULNAR}$, the possibility of a stretch-mediated response could not be ruled out. The fact that we were unable to provoke ulnar nerve symptoms or pain in the majority of subjects may indicate that we were unable to sufficiently load the ulnar nerve to elicit a protective response in healthy subjects.

During $ULNT_{MEDIAN}$, activity was predominantly recorded in muscles that can reduce the length of the anatomical course of the median nerve. This could be interpreted as either a stretch-mediated response which can limit excessive nerve elongation or as a nociceptive flexor withdrawal reflex. The occurrence of muscle activity before pain supports the possibility of a stretch-mediated response.

Activity of the upper trapezius was present in both the $ULNT_{MEDIAN}$ and $ULNT_{ULNAR}$. Because shoulder girdle elevation limits strain in all major branches of the peripheral nervous system of the upper limb, it seems logical that the nervous system adopts this strategy as a first and generalised response to protect the nervous system from potentially harmful elongation.

Although no conclusive conclusions regarding the neurophysiological mechanism of muscle activity during neurodynamic testing could be drawn, the findings demonstrate that a non-sensitised nervous system responds to peripheral nerve elongation with a gradually increasing, low level muscle activation. This is in agreement with the fact that a healthy nervous system tolerates mechanical stresses relatively well. Because inflamed neural tissue responds to as little as 3% stretch and minimal compression (Dilley et al. 2005), future research will focus on muscle activation with nerve elongation in pathological conditions.

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ROBOTIC GAIT TRAINING IN CHILDREN WITH CEREBRAL PALSY

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INTRODUCTION

The United Cerebral Palsy association estimates that more than 500,000 Americans have cerebral palsy (CP). A decrease in walking proficiency and economy is the main physical disability in children with CP. High lower limb agonist-antagonist muscle coactivations, increased tone, tightness of hip, knee, and ankle musculature are reported as causes of abnormal gait. As a result, the gait of children with CP is characterized by slow speed and disturbed motor control.

Previous studies have demonstrated that intensive gait training using a treadmill and bodyweight support leads to beneficial effects on motor recovery and gait in children with CP (Schindl et al., 2000; Cherng et al., 2007; Dodd & Foley, 2007). Robotic devices have been developed to assist in delivering gait training. Recent evidence indicates that robotic-assisted gait training is feasible in children with CP, with improved locomotor function following such training (Meyer-Heim et al., 2007). The aim of this study was to evaluate whether gait training using a robotic driven gait orthosis (DGO) (pediatric Lokomat©, Hocoma AG, Switzerland) improves locomotor function and gait mechanics in children with CP.

METHODS

Ten children (8 males, 7-13 years) with a diagnosis of spastic diplegia due to CP capable of ambulating 50 feet without the assistance of a person participated in this study. Subjects participated in a 6-week intervention that included three 1-hour training sessions per week. Each training session included 30 minutes of robotic-assisted walking divided into 3 bouts of 10 minutes. During each bout of walking subjects were encouraged to walk continuously and as actively as possible. As training progressed, the assistance provided by the DGO, the amount of body weight support and treadmill speed was adjusted for each subject. The adjustments varied according to the subject's ability to maintain upright posture and knee extension during heel strike without compromising loading.

Pre- and post-training evaluations were performed including clinical tests of standing and walking function (Gross Motor Function Measure, GMFM, sections D and E) and walking endurance (6 minute walk test). Clinical gait analysis was also performed using a motion capture system (Vicon 512, Oxford, UK) to assess changes in gait kinematics. Spatiotemporal gait parameters and sagittal plane hip, knee and ankle joint kinematics were measured while subjects walked barefoot along a level walkway at their comfortable speed. Subjects were allowed to use an assistive device during the walking trials. For each subject mean gait parameters and joint kinematics during stance and swing for each limb were calculated from 5 walking trials. Mean data for left and right sides of the 10 subjects were pooled (i.e. 20 limbs) and pre-and post-training comparisons were made using paired t-tests ($p < 0.05$).

RESULTS AND DISCUSSION

All subjects successfully completed 18 sessions of robotic-assisted gait training. GMFM scores revealed significant improvements in standing (39% increase, p = 0.007) and walking function (36% increase, p = 0.001). Comfortable walking speed increased by 25% post-training (p = 0.001, Fig 1A), while greater walking endurance (22% increase, p = 0.061) was also an observed outcome of training (Fig 1B). These results support recent findings for the same robotic training modality (Meyer-Heim et al., 2007) and also bodyweight supported treadmill training (Cherng et al., 2007; Dodd & Foley, 2007).

The observed improvements in locomotor function were associated with significant changes in gait biomechanics, including an 18% increase in stride length (0.60m vs 0.71m, $p < 0.001$) and a 36% decrease in double support time (p = 0.003). Cherng et al. (2007) has reported similar changes post-training. Subjects showed greater hip flexion during swing (14% increase, p = 0.002) and slightly improved knee extension (10% increase, p = 0.308) during mid to late stance after training. Finally, ankle kinematics revealed a 10% decrease in excessive ankle dorsiflexion during stance (p = 0.046) and a significant increase in ankle plantarflexion during push-off (p = 0.039). Overall the changes in sagittal plane joint kinematics indicate the children were able to walk with a less crouched gait pattern following training. Ultimately, the better gait mechanics allowed for more efficient and faster forward progression.

CONCLUSIONS

The present study supports recent findings that robotic-assisted gait training can lead to enhanced locomotor function in children with CP. These preliminary results also provide evidence that the observed changes are likely related to improved motor control and quality of movement during gait.

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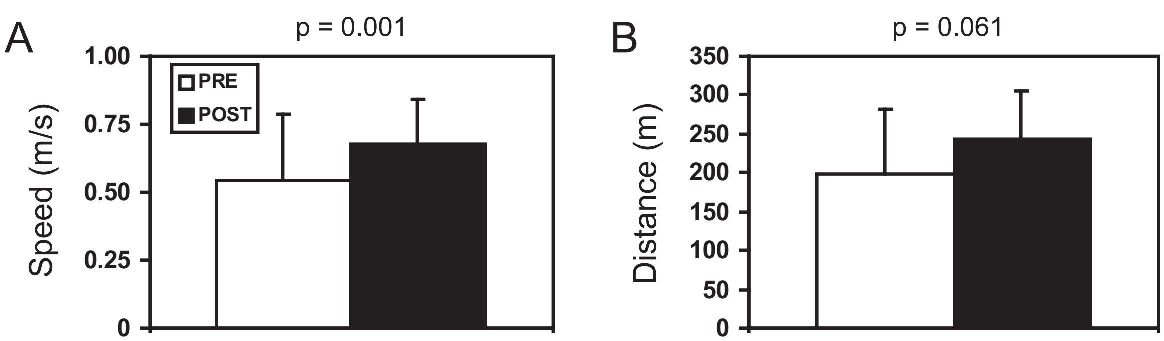

Figure 1: Mean (+SD) pre- and post-training (A) comfortable walking speed and (B) distance walked (6 minute walk test).

INFLUENCE OF PERIPHERAL AFFERENT STIMULATION ON CORTICOMOTOR EXCITABILITY: A LOWER LIMB STUDY.

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INTRODUCTION

It has recently been suggested that strategies which increase the excitability of corticospinal projections to paretic muscles in stroke patients may facilitate functional recovery (McDonnell et al., 2007). Some of these strategies concern peripheral afferent stimulation (Ridding et al., 2000). These changes in afferent stimulation input are capable of inducing organizational changes within the motor and sensory cortex (Ridding and Taylor, 2001; Kealing-Lang et al, 2002). While there is a wealth of research examining the effect of afferent input manipulation on cortical excitability in regard to upper limbs, there are few studies which examine the effect on lower limbs. In spite of their important functional role, motor pathways of proximal muscles in lower limbs are rarely investigated in neurophysiology. In 2007, Tremblay et al showed that prolonged repetitive neuromuscular electrical stimulation (NMES) associated with frequency (0.1 Hz) of transcranial magnetic stimulation (TMS) and light low voluntary knee extension (PAS paired associative stimulation) produced a significant long term higher facilitation of corticomotor spinal excitability on healthy subjects. The exact mechanism underlying this sensorymotor cortical plasticity is not well understood in healthy subjects. However, there are evidences that modulation of afferent inputs may play a central role in cortical plasticity development. The aims of this study was to investigate the effect of repeated peripheral stimulation such as NMES, transcutaneous electrical stimulation (TENS) and tendon vibratory stimulation (VIB), associated with quadriceps femoris muscle (QF) on healthy subjects. These stimulations are common modalities used in rehabilitation setting to produce motor effect (via a selective activation of a large myelinated afferent fibers) including muscle spindle activation (VIB).

METHODS

56 healthy subjects aged of 22.2 ± 1.4 yrs completed the study in five different experiments. Cortical excitability was measured using TMS with Magstim 200 stimulator connected to a double cone coil at 10% over the motor threshold. Motor evoked potentials (MEPs) induced by TMS were recorded (EMG) via surface electrodes placed over motor points of QF and expressed in % of control values (8-10 trials for each conditions). This electrophysiology testing was done before and after 30 or 60 minutes of NMES or TENS following by post-tests, 15 or 30 minutes after the end of afferent stimulation. Vibratory stimulation consisted of 5 minutes of VIB (120 Hz) over the patellar tendon. In this case, MEPs were recorded every 30 seconds during 5 min VIB and every min in the period following

VIB application until returned to baseline (Héroux et al., 2003). NMES protocol consisted in repetitive femoral nerve stimulation with large carbon silicon electrodes at I = 2.2 X motor threshold with biphasic symmetric wave of 300 µsec at 15 Hz and a duty cycle of 8on/8off sec inducing complete knee extension. TENS consisted of repetitive and continuous cutaneous stimulation under motor threshold over L2-L4 dermatomes with large carbon silicone electrodes. The wave pattern was biphasic asymmetric with a duration of 125 µsec at 100 Hz (for 20 min TENS project) or 60 µsec at 100 Hz (for 60 min TENS project). MEPs modulation (for NMES and TENS) was measured during and after the end of stimulation.

RESULTS AND DISCUSSION

The VIB stimulation (n=9) induced a significant, large and immediate facilitation of MEPs close to 600% of baseline value. This modulation declined rapidly and significantly after the end of the sensory stimulation. During TENS (n=9 and 16), a slight facilitation was seen (p<0.01) in the QF by 56±9% with a tendency to decrease 15 min and 30 min after the end of the stimulation. NMES (n=13 and 10) produced a reverse response, with a large depression during NMES close to 50% of baseline after 30 min of NMES, followed by a late facilitation rebound of 75±18%, 30 min after the end of NMES.

The figure 1 illustrates the summary of the results for TENS and NMES modalities. Vibratory stimulation (muscle spindle afferents), TENS (cutaneous afferents) and NMES (cutaneous, mechanoreceptors and reafferents of contraction) indicate that sensory stimulation can modulate corticomotor excitability in lower limb. Manipulation of peripheral afferents inputs is capable to modulate the motor cortex excitability in the short and long term complementarily in healthy subjects.

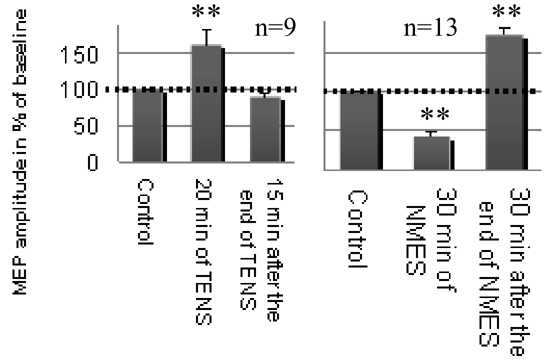

Figure 1: This figure illustrates the modulation of cortical excitability with stimulation of different types of afferents (mean±SEM).

SUMMARY/CONCLUSIONS

These findings show that peripheral afferent stimulation (VIB, TENS and NMES) can result in specific alteration in the excitability (facilitation and inhibition) of the corticospinal projections to the lower limb muscles and represent tools that may have important implications for the neuro-rehabilitation of patient such as a stroke.

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MODULATION OF NEURAL ACTIVITY DURING MENTAL SIMULATION OF KNEE EXTENSION : AGE AND TYPE OF MUSCLE CONTRACTION EFFECTS

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INTRODUCTION

Motor imagery (MI) refers to the ability of simulating motor action mentally. Mental rehearsal of movement is a common technique used by musicians and athletes to improve their performance (Hall et al., 1992; Yue, and Cole, 1992) However, there is little information demonstrating MI as means of improving motor learning and rehabilitation therapy, in spite that benefits can be present for some population (musculosquelettal immobilisation, stroke, Parkinson's disease, etc.). Previous studies have shown that actively moving and thinking about moving can produce facilitation of corticomotor response induced by transcranial magnetic stimulation (TMS) (Tremblay, et al., 2001; Fatiga, et al., 1999; Kasai, et al., 1997). In fact, imagining motor action is a congnitive task implicating parts of the brain's executive motor system (Jeannerod, 2004) including mirror neurons (Rizzolatti and Graighero, 2004). It is a question of intensity of activation in a continuum (Decety, 1996): pre-supplementary motor area (SMA), pre-frontal and parietal cortex (SII), inferior parietal area, post-cingulate cortex, insula, lateral cerebellar, basal ganglia (putamen and caudate nucleus) and primary motor cortex (MI). This cortical modulation can be selective for hands/arms imaginated movements (Abbruzzese, et al., 1996; Fatiga, et al., 1999; Hashimoto and Rothwells, 1999; Liu, et al., 2004) and legs (Tremblay, et al., 2001). In the present study, we investigated the age effect and type of mucle contraction in the dynamic and static knee extension during motor imagery of this movement with the dominant leg.

METHODS

Fourty-seven healty subjects (young: n=28, 21±2 yrs; elderly : n=19, 60.2±4 yrs; ♂ ~ 50%) participated in this study. Cortical excitability was measured using TMS with Magstim 200 stimulator connected to a double cone coil at 10% over the motor threshold. Motor evoked potentials (MEPs) induced by TMS were recorded (EMG) via surface electrodes placed over motor points of quadriceps femoris muscle (QF) and expressed in % of control values (8-10 trials for each conditions). This electrophysiologic testing was done before and during the mental simulation (eyes closed) of dynamic (about 2 seconds from 90° to 180°) and static (2-20 sec at 180°) knee extension. The Vividness of Movement Imagery Quesitonnaire (VMIQ) (cognitive testing) consisting in 24 questions (120 = maximal score), was used to measure the capacity of the subjetcs to imagine himself (kinesthetic imagery) performing different motor tasks (Isaak, et al., 1986). One-way ANOVA repeated measure were used to compare the age effect and the type of muscle contraction. Student's t- Test were used to compare VIMQ between young and elderly subjetcts.

RESULTS AND DISCUSSION

On young and elderly subjects, the MI in the dynamic knee extension condition induced a significant modulation of MEPs' amplitude in QF by 290±15% and 240±18% respectively compared to control values. In static MI, this facilitation was greater in both groups (380±20% for young and 350±25% for elderly subjects). Figure 1 illustrates the summary of the results on young and older subjects.

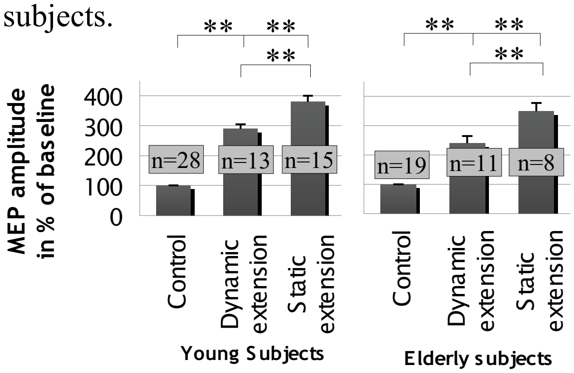

Figure 1: This figure illustrates the age and type of muscle contraction on modulation of cortical excitability during MI of knee extension (mean ± SEM).

The perception of the motor imagery's clarity (VMIQ) was lower in older than in young subjects (46.2±12.0 vs 60.3±10.0). The MI ability was preserved in older subjects (as shown by MEPs facilitation and VMIQ). On this group, the motor cortex facilitation was lower compared to younger subjects by 20% in the dynamic condition and 10% in the static condition. However, in both group, the motor imagery for the static muscle contraction condition is easier to perform, probably because of the complexity of the task. This difference of facilitation between the static and dynamic condition is greater for the elderly subjects (45% for the older subjects vs 31%). The present findings suggest that for older subjects, the neural networks sustaining motor imagery is still in operation.

SUMMARY/ CONCLUSIONS

Consistent with our previous results on young subjects (Tremblay, et al, 2001), we found that MEPs' responses were significantly facilitated in healthy non-trained young adult when the dynamic knee extension movement was imaginated. However, this facilitation was enhanced when static knee extension movement was imaginated. This study confirms that the older subjects conserved this modulation of cortical excitability in using mental motor imagery as well as in young subjects.

These results suggests possible implications in neuro-rehabilitation for clients who are cognitively stable altough suffering from motor disorders leading to poor motor control and/or mucle weakness.

MI is a simple, easy and non-costly way to activate the motor cortex.

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DIFFERENCES IN MUSCLE ACTIVATION PATTERNS BETWEEN SUBJECTS WITH CHRONIC LOW BACK PAIN AND HEALTHY CONTROLS DURING WALKING

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INTRODUCTION

In normal gait, low back muscles show a biphasic activity pattern, as assessed with surface EMG. Subjects with chronic low back pain (CLBP) may show different muscle activation patterns, i.e. higher activity in swing (Arendt-Nielsen et al.,1996), which increases at higher walking speeds (Lamoth et al., 2006) and longer activation in the periods of double support (Vogt et al., 2003). The underlying mechanisms of these changes are unknown, but one of the hypotheses is that they might be due to an inability to relax after activation.

The first aim of this study was to investigate if subjects with CLBP compared to healthy controls, show increased back muscle activity during the total gait cycle and less coordinated activity between the periods of swing compared to double support. A second aim was to investigate the differences at increasing walking speeds.

METHODS

In a cross sectional study sixty-three subjects with CLBP and thirty-three a-symptomatic controls were studied. Participants walked on a treadmill at increasing velocities (from 1.4 till 5.4 km/h). sEMG data of the Erector Spinae (L1 and L4 bilaterally) were measured using a 16 channel sEMG (Glonner) system. Smoothed rectified EMG (SRE) values of twenty strides were averaged to obtain average SRE values for the total cycle and the periods of double support and swing separately. The ratios of SRE activity in swing to double support were used as measures of coordinated activity. Statistical analysis was performed using random coefficient analysis.

RESULTS

Compared to asymptomatic controls, subjects with CLBP showed higher averaged SRE values during the total gait cycle, as well as during the periods of double support and swing separately ($\beta = 0.20$, $p = 0.001$). However, there were no differences in ratios of SRE values in swing to double support. There was also no interaction effect between group and velocity.

CONCLUSIONS

During walking, subjects with CLBP show increased back muscle activity, but do not show less coordination between the different periods of stride compared to healthy controls. The velocity induced differences in activity are also comparable between the two groups. The results do not support the hypothesis that subjects with CLBP have more difficulty relaxing back muscles after activation during walking at different velocities.

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SEGMENTAL MOTION COUPLING PATTERNS DURING MANUAL UPPER ATLANTO-AXIAL ROTATION MOBILIZATION: MORPHOLOGY OR INTERVENTION DETERMINED?

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INTRODUCTION

Motion coupling patterns in the upper cervical spine and especially in the atlanto-axial motion segment are generally explained in a mechanistic way based on the complex anatomy. At present only a few studies have reported morphometric data on the ligaments and joints of the upper cervical spine. So far no studies have been performed on the relationship between atlanto-axial kinematics and the specific spatial features of the lateral atlanto-axial joints and the alar ligaments. It is thus not known what the most determining factors are for the motion coupling patterns during manually induced motion in the atlanto-axial motion segment.

METHODS

Segmental atlanto-axial kinematics were registered in twenty un-embalmed cervical spine specimens (9 male and 11 female; mean age 80 ± 11 years) during manual regional axial rotation mobilization using a 3D ultrasound based motion tracking system (Zebris CMS20). Reference markers and anatomical landmarks were digitized using a Microscribe 3D stylus, first in the intact specimen and secondly after segmentation. Using a mathematical transformation approach the anatomical landmarks were reconstructed for the total specimen. Detailed spatial features of the lateral atlanto-axial joint surfaces and alar ligaments of each specimen were calculated separately. The relationship between the anatomical features and the spinal kinematics were analyzed using statistical regression-analysis techniques.

RESULTS AND DISCUSSION

The range of motion of the main axial rotation and the coupled lateral bending motion component could not be predicted by the 3D-anatomical features. Moreover the anatomical features could not predict the variance of the cross-correlation parameter, which has been proposed as an objective parameter for the description of motion coupling patterns (Van Roy et al. 2002; Cattrysse et al. 2006).
The range of motion of the coupled flexion-extension motion component, the ratio and the phase shift between the main axial rotation motion and the coupled lateral bending motion could be predicted for about 25% to 75% by selected sets of anatomical features (table 1).
The results indicate that the presented parameters characterising motion coupling in the atlanto-axial joint during manual

regional axial mobilization can only partially be explained and predicted by the specimen specific atlanto-axial anatomy. The results only partially confirm previously suggested relationships between specific anatomical features and joint kinematics (Cattrysse et al. 2007a; Cattrysse et al. 2007b). To fully explain the relationship between 3D-anatomy and 3D-kinematics, other kinematic parameters and upper cervical morphological features may have to be included. The Euclidian norm as a relatively stable measure expressing the overall range of motion is less dependent on the choice of the reference frame and may be a parameter to be included in the analysis. The atlanto-occipital and the atlanto-dental joints may influence kinematics in the atlanto-axial segment and their morphological features may as well have to be considered as determinants of atlanto-axial kinematics.
It has been demonstrated previously that motion coupling patterns during axial rotation mobilization can differ between therapists (Cattrysse et al. 2007b).

SUMMARY/CONCLUSIONS

The present results so far did not demonstrate a deterministic relationship from morphology on the antlanto-axial joint kinematics during axial rotation mobilization, which in turn may open positive perspectives toward the possibilities of manual therapy interventions.

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ACKNOWLEDGEMENTS

The authors wish to thank the Anatomy Department of the Université René Descartes-Paris 5, France, for the opportunity to perform this study on fresh cadaver specimens.

Table 1: Results of multiple regression analysis

Parameter	Model	Regression equation	R^2
Rom Z Flexion-extension	**1: facet orientation**	-1.316 + 0.918 axis sag angle	0.368**
	2: total	-8.175 + 0.988 axis sag angle + 0.780 alar sag angle	0.523**
Ratio	**1:facet morphometry**	5.830 – 60.633 atlas height	0.275*
	2: total	13.529 – 56.440 atlas height – 0.322 axis sag angle	0.493**
Phase Shift	**1:alar ligaments morphology.**	0.194 – 0.001 alar abs angle	0.397**
	2: facet orientation	0.192 – 0.004 axis sag angle	0.263*
	3: total	0.286 – 0.001 alar abs angle -0.004 axis sag angle	0.628**
	4: total	0.295 – 0.001 alar abs angle -0.004 axis sag angle + 0.124 axis height	0.734**

A NOVEL METHOD FOR NECK-COORDINATION EXERCISE – A PILOT STUDY ON PERSONS WITH NON-SPECIFIC CHRONIC NECK PAIN

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INTRODUCTION

Chronic neck pain is a common problem and is often associated with changes in sensorimotor functions, such as reduced proprioceptive acuity of the neck (Sjölander et al., 2006), altered coordination of the cervical muscles (Falla et al., 2004), and increased postural sway (Karlberg et al,. 1996).

In line with these findings there a studies supporting the efficacy of exercises targeting different aspects of sensorimotor function, e.g., training aimed at improving proprioception and muscle coordination. To further develop this type of exercises we have designed a novel device and method for neck coordination training.

The purpose of this study was to investigate the clinical applicability of the method and to obtain indications of the effects on sensorimotor functions, symptoms and self-rated characteristics in non-specific chronic neck pain

METHODS

Fourteen persons (10 females), 18-49 years old, with non-traumatic neck pain with duration of at least 3 months (mean duration 19 months) participated in this uncontrolled clinical study.

A newly developed device was used for the neck-coordination exercise. The exercise was an open skills task with adjustable difficulty. With visual feedback, the subject had to control the movement of a metal ball on a flat surface with a rim which was strapped on the subjects' head.

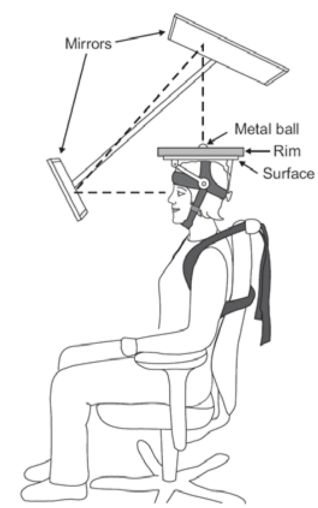

Figure 1. Schematic of the neck-coordination exercise apparatus.

The exercise was performed at eight occasions (2-3 times a week), each session lasting 10-15 minutes.

The clinical applicability was investigated by measuring the improvements in skill performance during the exercise and by interviewing the participants after the intervention period.

Before and after the intervention period pain and sensorimotor functions (including postural sway and jerkiness-, range-, position sense-, movement time- and velocity of cervical rotation) were assessed. At six-month follow up self rated pain, health and functioning was collected with questionnaires: Short Form 36 (SF-36), Neck Disability Index (NDI), Disability of Arm, Shoulder and Hand (DASH), Self Efficacy Scale (SES) and Tampa Score of Kinesiophobia (TSK).

RESULTS AND DISCUSSION

All participants improved their performance of the exercise task. The task comprehension was good and the overall opinion of the exercise was positive. This confirms that the design of the task and the progression of difficulty were well adapted.

After the intervention period postural sway and jerkiness of cervical rotation were significantly reduced. At the follow up significant improvements were seen in SF-36 (three out of eight dimensions), decreased disability measured with DASH and fear of movement measured with TSK.

Combining objective and subjective measurements is valuable in the evaluation of treatments of musculoskeletal disorders. The improvements in the sensorimotor function tests seen in this study may theoretically be due to improved function of the deep cervical muscles. Muscles known to contain a high density of muscle spindles, and therefore are important for the postural control. The questionnaire measurements indicate improvements in self experienced functioning.

The present study provides an indication of the possible effects of this neck-coordination exercise. A randomized controlled trial, involving larger groups and a more extended intervention period, is needed to ensure the effectiveness of the method.

SUMMARY/CONCLUSIONS

The results support the clinical applicability of the method. The improvements in sensorimotor functions may suggest transfer from the exercise to other, non-task specific motor functions. The results justifies a future randomized controlled trial and further investigation into the muscular mechanism involved in the exercise.

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EFFECT OF TIBIAL ROTATION IN OPEN AND CLOSED CHAIN EXERCISES ON VASTUS MEDIALIS OBLIQUUS ACTIVATION

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INTRODUCTION

Patellofemoral pain syndrome (PFPS) is a term used to describe retropatellar pain in the absence of other specific pathology. It is among the most prevalent knee problems in physically active young adults. Inappropriate tracking of the patella on the lateral aspect of the trochlear groove is thought to contribute to the etiology of PFPS. Neuromuscular weakness of the vastus medialis obliquus muscle (VMO) relative to the vastus lateralis muscle may result in lateral patellar subluxation (Tang et al., 2001), therefore most physical therapy treatments for PFPS aim to restore patellar tracking by strengthening and/or optimizing the neuromuscular control of the VMO.

It has been suggested that closed kinetic chain exercises produce greater muscle activity in the VMO than open kinetic chain exercises (Tang et al., 2001). Also, the VMO is thought to resist external rotation of the tibia and therefore be preferentially recruited with internal tibial rotation (Stensdotter et al., 2003). Adding hip adduction is thought to provide the VMO with a more stable origin and thus facilitate the preferential activation of this muscle (Hertel et al., 2003). Research investigating these theories has given conflicting results.

The purpose of this study was to determine the effect of tibial rotation and hip adduction on VMO activation in both open and closed chain knee extension exercises performed by male and female subjects.

METHODS

The study was approved by the Queen's University Health Sciences Research Ethics Board and informed consent was obtained from all participants. Healthy adults over 18 years old were recruited if they had no history of knee pain or surgery and no complicating orthopaedic or neurological conditions. The dominant leg was tested.

A Delsys™ DE2.1 electrode was placed over the muscle belly of the VMO muscle. A Delsys™ EMG amplifier was used to amplify (x1000) and filter (20 Hz to 450 Hz) the signals. Data were sampled at 1 kHz using a16-bit A/D board and custom Labview™ software. The maximum voluntary effort (MVE) of the VMO was determined for normalization purposes. The subjects then performed (in a random order) open chain isokinetic knee extension on a Biodex™ dynamometer, closed chain knee extension (squats) and closed chain knee extension with concurrent hip adduction. Three trials of each exercise were performed with the tibia positioned in neutral, in medial and in lateral tibial rotation.

EMG data were smoothed and rectified using a moving 20 ms window (10 ms overlap) and maximum activation was determined by the highest RMS amplitude of a 150 ms window across each data set. The maximum VMO EMG activity from each knee extension task was normalized to the MVE of the VMO (%MVE). A three-way repeated measures ANOVA was used to test the effects of exercise task, tibial rotation and sex on VMO activation. (α = 0.05). Post-hoc Tukey analyses were performed as required.

RESULTS AND DISCUSSION

Seven men and 7 women, aged 19-23 years participated in the study. There was no significant three way interaction between task, sex and tibial rotation. There was a significant interaction between task and rotation (p=0.018), a trend (p=0.058) suggesting a possible interaction between task and sex, and no interaction between rotation and sex (p=0.28). For both sexes, closed chain exercises (with and without adduction) created more VMO activation than open chain exercises (Figures 1 and 2).

Also for both sexes, during the closed chain exercises with and without adduction, VMO activation was significantly lower when the tibia was medially rotated (p<0.05) (Figures 1 and 2). This may be because the VMO fibres are in a shortened state and therefore at a mechanical disadvantage in this position. When the sexes were analyzed separately, for males, lateral rotation gave the greatest VMO activation while for females lateral and neutral rotation gave equally large VMO activation.

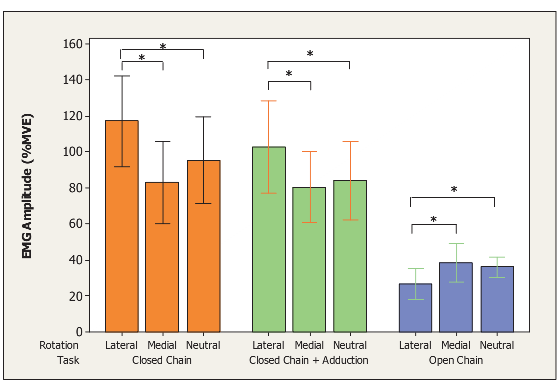

Figure 1: EMG amplitude (%MVE) for all tasks and tibial rotations for males. Error bars - 95% confidence intervals. * p<0.05.

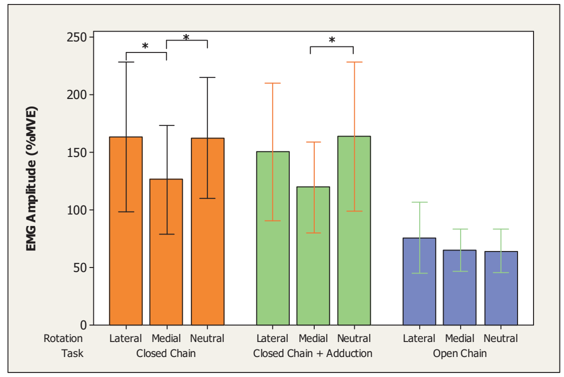

Figure 2: EMG amplitude (%MVE) for all tasks and tibial rotations for females. Error bars - 95% confidence intervals. * p<0.05.

Adding adduction to the closed chain exercise did not increase VMO activation but changed the effect of rotation such that lateral rotation produced the greatest VMO activation in males and neutral or lateral rotation produced the greatest VMO activation in females (both p<0.05). This difference between sexes may be due to the larger Q-angle seen in females, putting the VMO in a lengthened state and therefore at a mechanical advantage.

SUMMARY/CONCLUSIONS

To generate the greatest activation of the VMO, our results suggest that clinicians should prescribe closed kinetic chain knee extension exercises with lateral rotation for males and neutral or lateral rotation for females. These results pertain to healthy young adults and should not be extrapolated to older individuals or those with PFPS.

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A METHOD TO MONITOR THE STATUS OF FUNCTIONAL RECOVERY IN TENDINOPATHIES: A COMBINATION OF EMG AND STRENGTH MEASUREMENTS

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INTRODUCTION

Lateral elbow tendinopathy or tennis elbow (TE) is one of the commonest elbow problems causing pain and thus restricting performance and function. It is commonly considered a degenerative tendinopathy of the origin of extensor carpi radialis (ECR) muscle. The aetiology is likely to be multi-factorial and the optimal treatment remains undefined. Recovery time may vary between 6 months to 2 years. A reliable method is required for monitoring and assessing the status of recovery in TE. We aimed to: (a) investigate changes in muscular strength, fatigue and activity in recovered TE (RTE); (b) assess the appropriateness of EMG and strength measurements in monitoring functional recovery in TE.

METHODS

Study included three age-matched female groups of healthy controls (C) with no history of musculoskeletal problems, TE patients with local tenderness at the lateral epicondyle and pain with resisted wrist and middle finger extension, and RTE cases who were asymptomatic for at least 6 months. Measurements included metacarpophalangeal (MCP) (extension and flexion), wrist (extension and flexion), shoulder (internal rotation, external rotation, and abduction) and grip strength, total upper limb strength and EMG measures of muscle fatigue and activity for five forearm muscles including wrist extensors and flexors.

RESULTS AND DISCUSSION

Strength was greater ($p< 0.05$) for all measurements (MCP flexion 18-21%, grip 15-17%, wrist extension 16-19%, wrist flexion 18-21%, and shoulder 14-20%) in C compared to RTE and TE except for MCP extension. The total upper limb strength was significantly higher in C (122±22) than in both RTE (78±21) and TE (68±15) on the affected side ($p < 0.05$). Interestingly, there was no difference in any of strength measurements between TE and RTE groups. EMG revealed increased activity of ECR in RTE (9±5 %/min) while it was decreased in TE (–12±4 %/min).

In this study, we highlighted activation imbalance, disuse-deconditioning syndrome, and consequent global upper limb weakness in RTE. These findings suggest that regular EMG and strength measurement may provide useful information on recovery progress in the early post-injury stages, when it would be unsafe to perform some sport activities. That no difference was found for any upper limb strength measurements between RTE and TE suggests that there is sustained muscle dysfunction and weakness in RTE despite substantial pain diminutions, possibly because of inappropriate and insufficient rehabilitation. Increased activity of ECR in RTE may be attributable to relative recovery of muscle from injury and consequent reduction in the level of pain.

It is very important not only that attention is paid to pain reduction, but also to objective outcome targets in muscular strength and functional performance. More research is essential to characterize "full recovery" in both its symptomatic and functional dimensions, to assist in establishing an appropriate set of outcome criteria and related measures.

SUMMARY/CONCLUSIONS

Appropriate reconditioning of hand-wrist-forearm-shoulder musculature may be essential to achieve full recovery and prevent further tendon overload, degeneration, or relapse. Future studies should provide evidence in support of further rehabilitation after the pain has disappeared. Further research, using large sample sizes, is needed to investigate the practical reliability of our method in monitoring functional recovery in TE as well as other tendinopathies.

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MYOTEL: ADRESSING NECK SHOULDER PAIN AND ITS RELATED DISABILITIES BY ASSESSING AND FEEDBACK SEMG IN THE DAILY (WORK) ENVIRONMENT

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INTRODUCTION

Subjects with chronic pain have different muscle activation patterns compared to asymptomatic controls (eg Nederhand *et al* (2000); Veiersted *et al* (1993)). This is especially reflected in a prolonged activation of muscles after a task; i.e. a decreased ability to relax their muscles. Subjects are not aware of this muscle activation as it often concerns rather low levels of activation. Nevertheless, according to the Cinderella hypothesis these low levels of activation may contribute seriously to the development and maintenance of chronic pain, when occurring during long periods of time.

As subjects may not be aware of their inadequate muscle activation patterns, feedback on muscle relaxation during daily activities will enable subjects to change this. A myofeedback system has been developed that assesses muscle relaxation during daily activities and provides continuous feedback to the subject, when there is too little muscle relaxation. Positive effects on pain intensity and pain disability were shown in several studies in subject with neck-shoulder pain (Voerman *et al*, 2007). To improve access to the service and increase its efficiency, the system was further developed into a tele-treatment system (RSMT). With this system, subjects can view their muscle activation patterns on a PDA. Besides, the PDA automatically sends the data via GPRS to a secured sever, which is remotely accessible for the therapist, anytime and anywhere. This enables remote (e-) counseling. The objective of this study was to examine the RMST on technical efficacy for clinical use and explore changes in clinical outcome [Huis in 't Veld *et al*, 2007].

METHODS

Ten female workers suffering from work related neck-shoulder pain participated. Subjects received the RMST for four week. In addition they noted their activities and pain intensity. Weekly counseling sessions of 30 min with a therapist took place. Technical efficacy for clinical use was assessed by logging technical failures of the system and examining the hours of sEMG data available at the server during each week of wearing the system. A questionnaire, based on the Technology Acceptance Model, was used to assess satisfaction. Clinical outcomes used were pain and disability.

RESULTS AND DISCUSSION

Results show that in 78% of the weeks of wearing the system sufficient sEMG data during daily activities were available at the server to make an assessment of muscle activation patterns. Subjects reported high satisfaction with the usefulness and ease of use of RMST. However, they were less satisfied with the technical functioning, i.e. stability, of the system. Eighty percent of the subjects reported a reduction in pain intensity and disability directly after RSMT (figure 1).

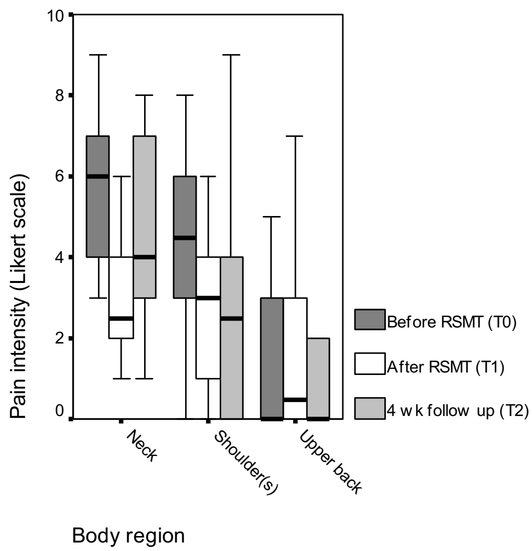

Figure 1 Box plots of pain intensity scores for neck, shoulder(s) and upper back before remotely supervised myofeedback treatment (T0), directly after RSMT (T1) and at four weeks follow up (T2) (n=10)

SUMMARY/CONCLUSIONS

The RMST was technically feasible, subjects were satisfied and the clinical changes tended to be slightly better compared to myofeedback provided *in vivo*. Nevertheless, the technical performance and the ease of use need to be optimized. Further evaluation is required in large scale clinical trials with outcomes defined on multiple endpoints like quality and costs. Such a trial will be performed in 4 different countries in the E-ten MyoTel project (www.myotel.eu).

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NEUROMUSCULAR PATTERN OF THE PELVIC FLOOR MUSCLES DURING LIMB MOVEMENTS

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INTRODUCTION

Underlying pathology of lumbopelvic pain is unclear (Vleming et al., 1997). Insufficient stabilization by the pelvic floor muscles (PFM) may be one of the causes (Pool-Goudzwaard et al., 2005). The aims of this study were to: A) develop a method in which surface EMG can be used to evaluate the neuromuscular pattern of the PFM during dynamic tests with different postural demands B) apply the method in order to investigate whether it can be used to detect an intra individually reproducible neuromuscular pattern in the PFM and C) investigate if this pattern differ inter individually in women with no lumbopelvic pain. The neuromuscular pattern in this study, is defined based on the timing (EMG onset) and the activation amplitude of a muscle.

METHODS

Four pilot studies preceded the main study in order to analyze possible sources of methodological errors and to select MVCs and dynamic tests to include in the test protocol. Ten women, with no experience of lumbopelvic pain during the last twelve months, were included in the main study. During the main study, the activity of the transversus abdominis /internal oblique (TrA/OI), rectus abdominis, erector spinae, hip adductor (HA) and deltoid muscle (DE) or rectus femoris muscle (RF), (depending on which dynamic movement) was recorded with surface electrodes (Blue sensor, M-00-S, Medicotest, Denmark diameter of active part 10 mm). An intravaginal probe (Periform Intravaginal Probe, Neen HealthCare, Dereham, England) was used to record the activity of the PFM. The two dynamic tests analyzed were a unilateral hip flexion (active straight leg raise, ASLR) performed in supine position, with and without an extra 2 kg weight and a unilateral shoulder flexion performed with the arm in a 45 degree lateral rotation and with a 5 kg weight in the hand. The trunk muscles were recorded contra laterally with respect to the movements. The reference EMG activity for EMG onset was recorded in resting positions, both in supine and standing. The EMG activity during the MVC recordings served as reference values when analyzing the activation amplitude. EMG signals were sampled at 1000 Hz by a MegaWin EMG eight-channel unit system (Mega Electronics Ltd, Finland*).

The raw EMG signals were transformed (MegaWin 2.3.4*) into the RMS averaging (frame width 0.001s) EMG signals and smoothed using a moving-window technique (frame width 0.02s) before the EMG onsets were calculated. The EMG onsets were calculated mathematically by computer using MatLab 7.1 (MathWorks Inc.). The deltoid muscle onset was excluded from further analyses due to an increase in the background activity during the upper limb movement. EMG onsets that occurred 400 ms before or after the initiation of the movement were excluded.

RESULTS AND DISCUSSION

The results of this study demonstrate that, following the designed protocol, EMG can be used to study the neuromuscular pattern of the PFM with respect to the timing

(EMG onset) and the activation amplitude during both lower and upper limb movements performed at a normal pace.

The risk for cross-talk was considered to be minimal since no correlation between the activation of the PFM and the TrA/OI or between the PFM and the HA, could be detected.

The neuromuscular pattern was considered to be stable intra individually when the EMG onset of the PFM of three repetitions did not exceed 100ms in the lower limb movements and 250ms in the upper limb movement, or if a woman lacked an EMG onset of the PFM in 3 repetitions or more. The majority of the women demonstrated a stable neuromuscular pattern during the lower limb movements (Fig. 1). The inter individual EMG onset of the PFM during the lower limb movements occurred 200 ms (SD 160) respectively 220 ms (SD140) before the initiation of the movement (Fig. 2).

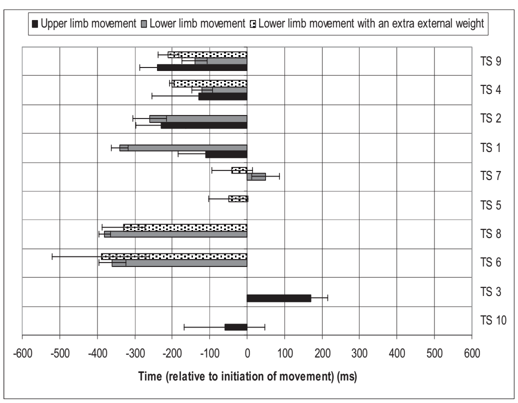

Figure 1: The mean (SD) EMG onset for the PFM during the dynamic tests for each woman. TS = test subject

A stable intra individual neuromuscular pattern of the PFM during the upper limb movement was demonstrated in 8 of 9 women (Fig. 1). The inter individual EMG onset occurred 100 ms (SD 150) before the initiation of the movement (Fig. 2). However, two of these women lacked an EMG onset of the PFM and the EMG onset could neither be detected in the MatLab nor the visual analyses.

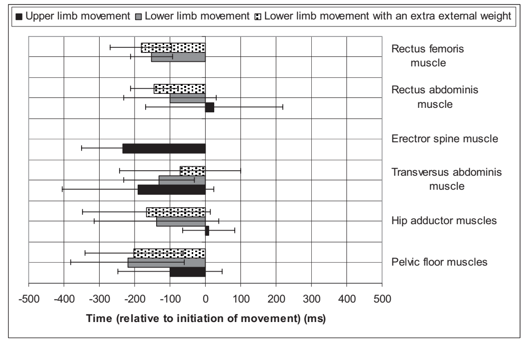

Figure 2: The mean (SD) EMG onset for the recorded muscles during the dynamic tests.

The difficulty to detect EMG onset of the PFM during the upper limb movement can be due to either measurement problems or possible lack of an EMG onset in the PFM in some women during a limb movement performed in standing position.

CONCLUSIONS

A neuromuscular pattern can be detected and reproduced during the lower limb movements performed in supine position with the developed method. It is more difficult to detect and reproduce a pattern during the upper limb movement. However, there is a tendency towards a stable pattern during the upper limb movement performed in standing position as well.

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ACKNOWLEDGMENT

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MAXIMUM PELVIC FLOOR MUSCLE ACTIVATION AND INTRAVAGINAL PRESSURE ACHIEVED DURING COUGHING IN WOMEN WITH AND WITHOUT INCONTINENCE

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INTRODUCTION

Stress urinary incontinence (SUI) is associated with pelvic floor muscle (PFM) dysfunction. Most research on the role of the PFMs in continence has focused on voluntary PFM contractions despite the fact that SUI occurs in situations that require reflex PFM contractions, i.e. coughing. The purpose of this study was to compare intra-vaginal pressure and PFM electromyogram (EMG) amplitudes generated during coughing in supine, sitting and standing in women with and without SUI.

METHODS

Three groups of women participated. Women were included if they were 35 to 60 years old, healthy, and were not pregnant within 6 months of participating. Women were excluded if, when examined by a research nurse, they displayed detrusor contraction during urodynamic testing or took medications to treat or that provoke incontinence. Incontinence severity was determined using a 3-day bladder diary.

Subjects attended one data collection session during which the researchers were blinded to their continence status. They performed three maximum voluntary PFM contractions (MVCs) and three maximum effort coughs in supine, sitting and standing. The position order was randomized. EMG data were recorded bilaterally from the PFMs using a Femiscan™ vaginal probe, the data from the side with the higher signal amplitude were used in the analysis. The EMG electrodes were interfaced with a Delsys Bagnoli™ EMG system, and a 16 bit analog to digital (AD) converter (NIDAQ PCI-MIO -16XE-10). The vaginal probe had an air filled balloon connected to a pressure transducer glued to its rear surface to record intravaginal pressure data simultaneously with EMG data. The pressure transducer was interfaced with the AD converter. Data were acquired at 1 kHz. The peak expiratory flow rate generated during each cough was measured with a peak flow meter.

All data were smoothed using a moving 200 ms sliding window (199 ms overlap) over which root mean square (RMS) values were calculated. The mean RMS was computed over 200 ms of resting data and was subtracted from all smoothed EMG and pressure values. Maximum amplitudes were determined for EMG and for intravaginal pressure during the coughs and were compared using two-way repeated-measures ANOVAs including group and position as factors (α=0.05). Originally the cough data were to be normalized by the MVC, however there were significant differences in the MVC values among the groups; therefore a nested model (subject nested in group) was used to account for inter-individual differences.

RESULTS

Twenty-four women (n=8 per group)

participated. The groups were similar in age 52.1±6.1 years (p=0.96), BMI 26.5±4.7 kg/m^2 (p=0.56) and parity 2.2±1 children (p=0.52). There was a significant group by position interaction for peak flow (p<0.01); however there was no group main effect.

The interaction between group and position was not significant for the maximum EMG amplitudes (p=0.38), nor was there a main effect for either position (p=0.66) or group (p=0.45). See Figure 1.

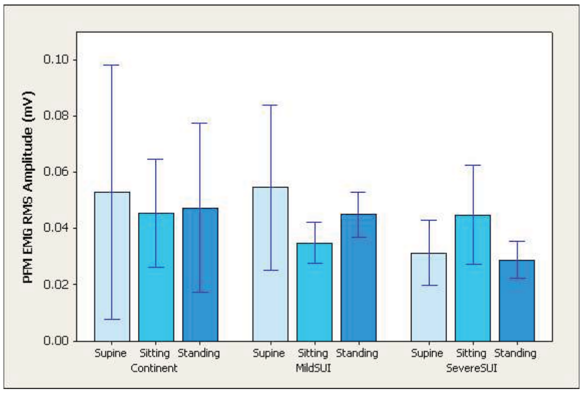

Figure 1. Maximum PFM EMG amplitude during coughing by position and group.

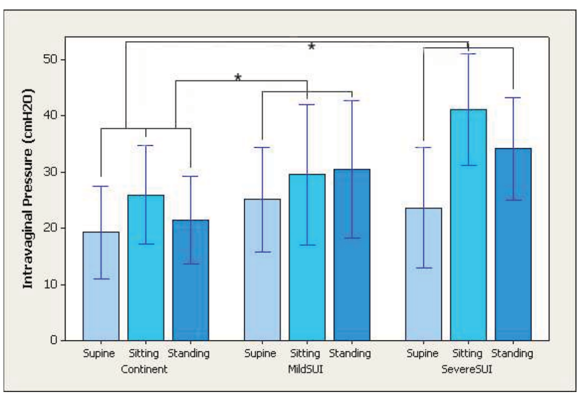

Figure 2. Maximum intravaginal pressure during coughing by position and group. * indicates significant differences between groups (p<0.05).

Maximum intravaginal pressure showed a significant group by position interaction (p=0.03). The continent women produced lower intravaginal pressures overall than did either of the groups with SUI. See Figure 2.

CONCLUSIONS

Since there were no between group differences in peak flow, we are confident that the three groups produced similar coughing efforts.

There was no difference among the groups in the maximum PFM EMG amplitudes during coughing. This suggests that the urine leakage seen during coughing in women with SUI is not related to PFM activation difficulties. Since we only included the PFM EMG data from the side that produced higher values, unilateral defects might have been missed.

The pressure data suggest that urine leakage with coughing in women with SUI may not be due to inadequate intravaginal pressure generation. Alternatively, our measure of intravaginal pressure may not be a good surrogate for intra-urethral pressure in these women. The leakage seen with coughing in women with SUI is likely due to factors other than PFM activation: i.e. unilateral PFM defects, urethral sphincter defects, fascial support defects or co-ordination defects. As we only analysed the data from the side with the higher EMG amplitudes, the other side was de facto lower, however we do not know if any of the side-to-side differences were significant.

ACKNOWLEDGEMENTS

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RELIABILITY OF EMG ACTIVATION OF THE PELVIC FLOOR MUSCLES RECORDED USING SURFACE AND FINE-WIRE ELECRODES

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INTRODUCTION

Pelvic floor muscle dysfunction is one factor thought to contribute to the development of stress urinary incontinence (SUI) in women. Electromyography (EMG) is widely used to test the neurophysiological integrity of the pelvic floor muscles (PFMs), however the reliability of the different methods of recording EMG data from the PFMs has not been established. The purpose of this study was to investigate the reliability of three different EMG detection systems commonly used to acquire EMG data from the PFMs during maximum voluntary contractions (MVCs) and during a coughing task.

METHODS

Healthy nulliparous women between the ages of 18 and 40 participated in the study. Subjects were screened by a physiotherapist to confirm the presence of pelvic floor muscle (PFM) recruitment during maximum voluntary contractions (MVCs) and during maximal effort coughs. EMG data were recorded on subjects who passed the screening while subjects performed three trials of an MVC and three coughs in supine and in standing.

Stainless steel hooked-wire electrodes were inserted into the right and left anterior pubococcygeus muscles midway between the pubic symphysis and the ischial tuberosity. Fine-wire EMG data were amplified by 10 000 and bandpass filtered from 20 – 2000 Hz. The L-shaped Femiscan™ vaginal probe (bipolar configuration) and pear-shaped Periform™ vaginal probe (monopolar configuration) were used to record surface EMG from both sides of the PFMs. Surface EMG data were amplified by 1000 and bandpass filtered from 20 - 450 Hz. All EMG data were sampled at 4000 Hz and amplified using Delsys Bagnoli-16 amplifiers.

EMG data were smoothed by calculating the root mean square (RMS) amplitude across each trial using a 100 ms sliding window with 99 ms overlap. The baseline RMS amplitude was removed from the maximum RMS value to compute the peak amplitude of the recorded signal for the cough and MVC tasks. The peak EMG amplitudes recorded during the coughs were also normalized to the peak EMG recorded during the PFM MVC for each subject and reported as a percentage of the maximum voluntary electrical activation (%MVE).

Between-trial reliability for each task was assessed using interclass correlation coefficients ($ICCs_{(3,1)}$) and coefficients of variation (CV). Between-day reliability was assessed using $ICCs_{(3,2)}$. To determine the magnitude of the difference in EMG amplitude between days, the mean absolute difference (MAD) was computed by taking the difference between mean RMS amplitudes recorded for each subject on each day. Similarly, to compare the between-day reliability across devices the MAD was normalized by dividing the MAD for each subject by the mean RMS amplitude recorded on both days for that subject and was reported as a percentage.

RESULTS

Twelve women (30.1±5.6 years old) volunteered to participate in the study. Two subjects were found through the screening examination to not perform adequate PFM contractions during MVC or cough and therefore they were excluded from the study. EMG data from the remaining ten subjects were analyzed separately for each side of the pelvic floor muscles but side data are grouped together in the tables below.

The between-trial reliability of the MVE and cough data was good to excellent for each detection device across all tasks (Table 1). The ICCs suggest that the fine-wire electrodes and Periform™ perform more consistently between trials than the Femiscan™. However, when the CVs are examined, the surface electrodes demonstrate less variability between trials compared to the fine-wire electrodes.

Table 1: ICCs and CV computed between trials for each device (across all tasks). Left and right side data combined.

	Device	*Median*	*Range*
$ICCs_{(3,1)}$	**Femiscan™**	0.85	0.61 – 0.98
	Periform™	0.91	0.81 – 0.96
	Fine-Wire	0.95	0.58 – 0.99
CV (%)	**Femiscan™**	13.8	8.5 – 20.7
	Periform™	13.0	9.6 – 19.5
	Fine-Wire	16.0	8.4 – 32.5

The ICCs indicate that between-day reliability for the MVE and cough data was poor and inconsistent across all devices and tasks, where the $ICCs_{(3,2)}$ ranged from 0.00 to 0.94 (Table 2). In particular, the normalized cough values were homogeneous across the group (79.5 – 90.2 % MVE) and therefore the ICCs computed for the normalized cough data likely underestimate between-day reliability ($ICCs_{(3,2)} =$ 0.00 – 0.55).The difference in amplitude recorded between days was more accurately reflected by the $nMAD_{day}$ (Table 2). The nMAD was considered to be a better reflection of the between-day reliability of each device, and was very good when considering the normalized cough data.

Table 2: Between day reliability

	Device	$ICC_{(3,2)}$ *Median*	$ICC_{(3,2)}$ *Range*	nMAD (%) *Median*	nMAD (%) *Range*
MVC	Femiscan™	0.65	0.36 - 0.79	31.4	20.9–42.8
	Periform™	0.69	0.54–0.89	33.3	24.9– 2.2
	Fine-Wire	0.39	0.00–0.90	78.6	55.8–97.1
Cough	Femiscan™	0.64	0.49– 0.77	33.1	27.0–47.6
	Periform™	0.76	0.42–0.94	36.5	26.9–38.7
	Fine-Wire	0.74	0.49–0.85	67.7	45.6–82.5
Cough %MVE	Femiscan™	0.08	0.00–0.30	10.4	8.6-14.4
	Periform™	0.20	0.00–0.40	9.3	7.5–12.7
	Fine-Wire	0.00	0.00–0.55	14.2	10.0–24.2

MVC (RMS amplitude during PFM maximum voluntary contractions), Cough (raw cough RMS amplitude), Cough %MVE (normalized cough amplitude), nMAD (normalized mean absolute difference). Left and right side data combined.

DISCUSSION

When EMG data are recorded within a single session, the amplitude appears to be consistent for the recording devices. Our results indicate however that EMG data recorded on separate days with these instruments should not be compared unless the data can be normalized. Although the surface electrodes outperformed the fine-wire electrodes between days, the reliability of EMG amplitude values across days was poor. A secondary finding was that ICCs present a number of limitations in assessing reliability of EMG data.

SUMMARY/CONCLUSIONS

The between trial reliability of PFM EMG data is good to excellent, but clinicians and researchers are cautioned to normalize their data if day-today comparisons in EMG amplitude are to be made using MVE or cough data recorded using these devices.

ACKNOWLEDGEMENTS

Gratefully appreciated financial support provided by NSERC.

INTRAVAGINAL PRESSURE AND PELVIC FLOOR MUSCLE ACTIVATION DURING COUGHING IN WOMEN WITH AND WITHOUT STRESS URINARY INCONTINENCE

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INTRODUCTION

The pelvic floor muscles (PFMs) play an integral role in continence; it is thought that they work in synergy with the urethral sphincters to close the urethra when intra-abdominal pressure rises. We have modeled the relationship between PFM activity and intravaginal pressure during voluntary PFM contractions in continent and incontinent women (Madill and McLean 2006; Madill, et al. 2007). Because stress urinary incontinence (SUI) occurs during increases in intra-abdominal pressure, i.e. coughing, the purpose of this study was to model the relationship between PFM activation and intravaginal pressure during coughing, in women with and without SUI.

METHODS

Three groups of healthy female volunteers participated. They were screened by a research nurse and excluded if urodynamic testing revealed detrusor contraction, if they had prolapse, had been pregnant in the six months prior to testing or took medications to treat or that provoke incontinence. Incontinence severity was determined from a 3-day bladder diary.

On testing, the subjects performed 3 trials of a maximum effort cough in supine, during which the researchers were blinded to group assignment. Electromyographic (EMG) data were recorded from the PFMs bilaterally with a Femiscan™ vaginal probe; the data from the side with the larger signal amplitude were used for analysis. Delsys Bagnoli™ amplifiers and a 16-bit analog to digital (AD) converter (NIDAQ PCI-MIO-16XE-10) were used. The vaginal probe had an air filled balloon connected to a pressure transducer glued to its posterior surface to record intravaginal pressure data which reflects intra-urethral pressure (Theofrastous et al. 1997). The pressure transducer was interfaced with the AD converter. All data were acquired at 1 kHz.

All data files were smoothed with a 3rd order Butterworth filter and normalized to the maximum smoothed amplitudes achieved during that cough. Ensemble average intravaginal pressure versus PFM EMG curves were created. Maximum cough EMG and pressure amplitudes were compared among the groups and are reported in a companion paper in these proceedings.

RESULTS AND DISCUSSION

Twenty-four women (8 per group) participated. The groups were similar: age 52.1±6.2 years (p=0.96), BMI 26.8±5.7 kg/m^2 (p=0.08) and parity 2±0.92 (p=0.30).

The curves for each group showed high variability until the data were stratified into two distinct patterns per group. In the continent women, 6 trials (from 3 different subjects) were different from the other 14 trials; in the mild SUI group, 4 trials (from 3 subjects) were different from the other 20 trials, and in the severe SUI group, 3 trials (from 3 subjects) were different from the other 20 trials. (See Figure 1) With the unusual observations removed the shapes of the curves were similar among the groups. The unusual observations were ensemble averaged to produce a "Secondary pattern" curve for each group. (See Figure 1.)

The intercepts of the curves were all significantly greater than zero suggesting that there is a large electromechanical delay. The intercept was higher in the women with severe SUI than in the women with mild SUI (p<0.05), and there was a trend (p=0.072) suggesting that the continent women had a lower intercept than the women with severe SUI.

CONCLUSIONS

The primary intravaginal pressure versus EMG curves seen in all women were not substantively different except that the women with severe SUI demonstrated a higher intercept. This difference suggests that the women with severe incontinence may contract their PFMs to a higher level before this contraction has any effect on intravaginal (and therefore intra-urethral) pressure. The need to contract at a higher level in order to generate pressure might be the result of fascial or other soft tissue damage; the early pelvic floor contraction may take up the slack in the lax support structures before it acts to generate pressure.

The importance of the secondary curves is difficult to determine. These outlier curves were generated on one or two trials for a small number of subjects. The secondary curves were substantially different between the continent women and the women with mild and severe SUI whereby a subset of women with SUI contract their PFM nearly to their maximum level before the intravaginal pressure begins to rise, then this activity does not change or drops over the course of the cough. These women may use a different PFM motor control strategy during coughing, perhaps reflecting a pre-contraction of the pelvic floor muscles.

Patterns of muscle activation need to be investigated further in this population. Perhaps by stratifying women by underlying cause of incontinence (i.e. fasical tearing versus muscle weakness) we may be able to explain the different curves. Different patterns of PFM recruitment relative to intravaginal pressure may shed light on which women would benefit most from PFM muscle training for the physical therapy management of SUI.

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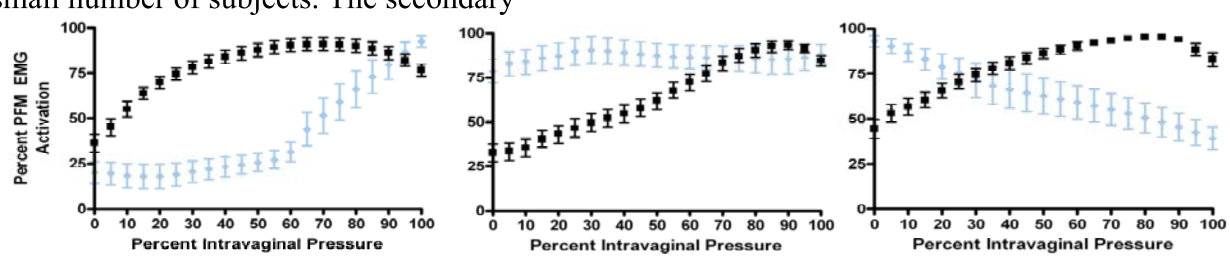

Continent n=14, n=6 Mild SUI n=20, n=4 Severe SUI n=20, n=3

Figure 1. Ensemble average PFM EMG versus intravaginal pressure curves for the three groups. Black indicates the primary curve and blue indicates the secondary curve.

PELVIC FLOOR MUSCLE RESPONSE TO A PRESSURE PAIN STIMULUS IN WOMEN WITH PROVOKED VESTIBULODYNIA

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INTRODUCTION

One common cause of painful intercourse in women is provoked vestibulodynia (PVD), a subtype of chronic vulvar pain that affects an estimated 16% (Harlow & Stewart, 2003) of pre-menopausal women. Recent reviews consider pelvic floor muscle (PFM) dysfunction as part of the pathophysiology of PVD (Farage & Galask, 2005), and recommend physical therapy as a treatment option (Weijmar et al., 2005). Physical therapy has been shown to be efficacious in decreasing pain in women with PVD (Bergeron et al., 2001) by addressing PFM dysfunction through education, biofeedback assisted exercises, manual therapy and electrotherapy. Despite the focus on deep PFMs in the physical therapy treatment of PVD, the superficial PFMs may play a larger role than the deep muscles in the pathophysiology of this condition (Reissing et al., 2005). The purpose of this study was to examine the superficial and deep PFM responses to pressure pain stimulus (PPS) applied at the vulvar vestibule (i.e., opening of the vagina) in women diagnosed with PVD. It was hypothesized that superficial PFMs would be more responsive than deep PFMs.

METHODS

Ethics approval was received from Queen's University Health Sciences Research Ethics Board. To date, eight women diagnosed with PVD by the study gynecologist have participated after providing informed consent. Surface electromyography (EMG) data were amplified using Delsys Bagnoli-8 EMG system amplifiers (gain 1000, bandpass filter 20-450 Hz, CMRR 100 dB at 60 Hz, input impedance 100 MOhms). All EMG channels were interfaced with a 16-bit National Instruments analog to digital converter (PCI-MIO-16XE-10) and sampled at 1000Hz. Differential EMG activity was recorded from the deep PFM (levetor ani) using a Femiscan™ probe (Mega Electronics Ltd, Kupio, Finland) and from the superficial PFM (bulbospongiosus) using two disposable Ag/Ag-Cl surface electrodes (Kendall-LTP 5500 Q-Trace® Gold, Chicopee, MA). Each participant began by performing a maximum voluntary contraction (MVC) of the PFM. A PPS was then applied at the posterior aspect of the vulvar vestibule using a vulvalgesiometer, a device that exerts a standard amount of pressure via a cotton swab tip (Pukall, Binik & Khalifé, 2004). Three sets of EMG data were recorded during the application of the PPS to the vulvar vestibule after the stimulus level was determined by applying sequential stimuli until the participant reported a pain rating of 6 out of 10 on a Likert scale. Raw EMG data were filtered by computing root mean square amplitudes using a 100ms moving window (90ms overlap). Baseline activity was removed from all data files and the data were normalized to the smoothed valued obtained during the MVC. Non-parametric analyses were used to test the differences between superficial and deep PFM responses.

RESULTS AND DISCUSSION

The mean age of participants was 24 (SD ± 5.0) years, mean BMI was 22.7 (SD ± 3.0) kg/m^2, and mean PVD duration was 4.0 (SD ± 4.5) years. The median PPS applied at the vulvar vestibule was 400 (IQR 100-500) g/0.4cm^2. All eight women demonstrated an increase in both superficial and deep PFM EMG activity on both sides in response to the PPS (See Table 1). This response was 18.1 (IQR 8.0-25.9) %MVC for the deep PFMs and 61.5 (IQR 43.8-75.7) %MVC for the superficial PFMs (See Figure 1).

Table 1: Combined right and left PFM EMG amplitudes shown in microvolts (median, IQR).

	Deep PFMs (μV)	Superficial PFMs (μV)
Response to PPS	6.05, (4.25)	11.20, (6.25)
MVC	35.70, (24.70)	19.40, (5.90)

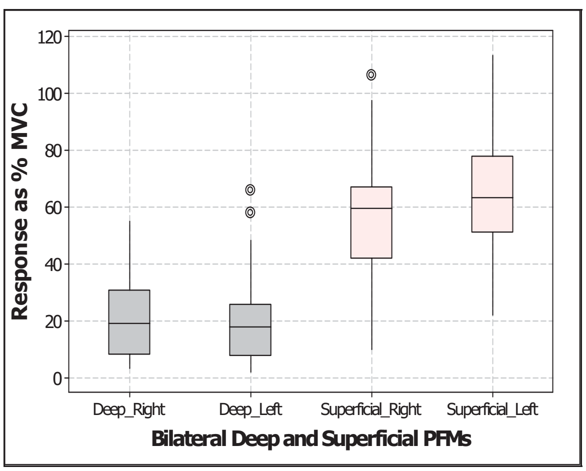

Figure 1: Box and whisker plot deep and superficial PFMs response to PPS.

A Wilcoxin signed rank test revealed that the superficial PFMs demonstrated higher EMG responses to the PPS than did the deep PFMs (Median difference = 42.8 %MVC, p=0.0001).

SUMMARY/CONCLUSIONS

These preliminary results suggest that both superficial and deep PFMs may be involved in protective reactions to vulvar pain in women with PVD, and that superficial PFMs may be more responsive. These findings may have important implications for treatment, since current therapies focus only on the deep PFM in treating PVD. As this work progresses, PFM responses will be compared to women without a history of vulvar pain, and to values collected after this sample of women have undergone physical therapy treatment for PVD.

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EMG ONSET DETECTION IN PELVIC FLOOR MUSCLES OF CONTINENT AND INCONTINENT WOMEN

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INTRODUCTION

Pelvic floor exercises are invaluable in regaining continence, but mechanisms are not fully understood. To contribute to the understanding of these mechanisms, we investigated the contraction sequence of superficial versus deep pelvic floor muscles (PFMs) in different positions in continent and incontinent women.

To our knowledge no groups to date have investigated the contraction pattern of the 2 muscle layers under various conditions. Based on nerve supply information independent functioning of the 2 muscle layers may be assumed. Because posture might influence PFM function, we compared the onset of activity of the superficial versus deep PFMs in different positions in continent versus incontinent women. This was done by comparing surface EMG using 2 pairs of intravaginal electrodes and 2 pairs of perineal surface electrodes after a command for a maximal pelvic floor contraction.

METHODS

The onset of contraction of the superficial and deep pelvic floor muscles was recorded by perineal and intravaginal surface electromyography in 32 continent and 50 incontinent women. The EMG disposable pregelled surface electrodes were placed on the perineum at each side of the urethra and anus (fig. 1).

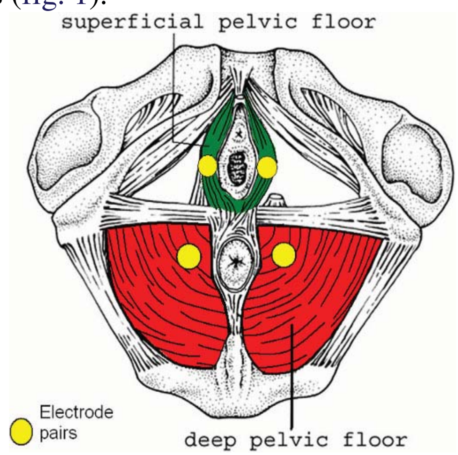

Figure 1: Perineal electrode placement

Intravaginal surface EMG was done using 2 disposable vaginal sponges each containing a pair of electrodes. The vaginal sponges were sewn together at a distance equal to the clinical digital location of superficial and deep PFM activity (fig. 2).

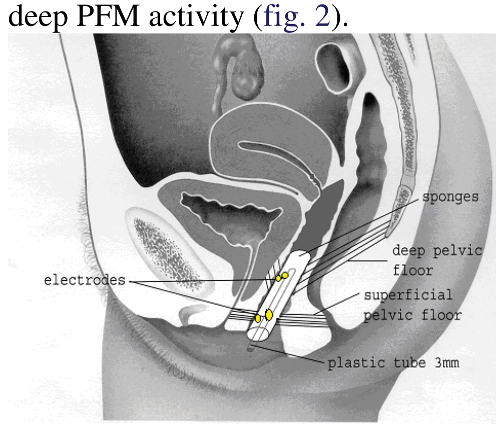

Figure 2: Intra-vaginal electrode placement

This electrode placement allowed the women to contract the PFM without feeling the materials associated with EMG recording. An EMG amplifier connected to a real time computer system processed and visualized the data. To determine the onset, the raw data was first full wave rectified and low pass filtered at 50Hz to smooth the data. Onset was detected at the time where the mean of a sliding window of 25 ms exceeding 2 standard deviations above the baseline activity. Baseline activity was detected 2 seconds prior to a maximum voluntary contraction. We subtracted the onset time of the superficial PFMs from the onset time of the deep PFMs. Therefore, a positive difference value signified that the deep PFMs contracted later than the superficial PFMs, while the reverse was observed for negative difference values. Differences were calculated for perineal and intravaginal recordings. The median and IQR (InterQuartile Range) were calculated for the continent and incontinent groups.

RESULTS AND DISCUSSION

Table 1 illustrates that perineal and intravaginal electromyography recordings used to define the onset of muscle activity showed a high level of agreement. In the continent group the superficial muscles almost always contracted before the deep muscles in all 6 positions. In the incontinent group the reverse sequence was observed in 3 of 6 positions. Higher and less consistent time differences in the onset of contraction of the 2 muscle layers were found in incontinent as compared to continent women.

SUMMARY/CONCLUSIONS

Contractions of the superficial and deep pelvic floor muscles can be recorded by intravaginal or perineal electrodes. A consistent contraction sequence can be found in continent women but it is lacking in incontinent women. This might be a possible explanation for incontinence. Including differentiated muscle contraction exercises in pelvic floor muscle exercise programs may further optimize treatment outcomes.

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Table 1: Difference in onset time of contraction of the deep minus the superficial PFM recorded by intra-vaginal and perineal surface electrodes in 32 continent and 50 incontinent women, after a request for a voluntary contraction. (All values in msec)

	Continent women				**Incontinent women**			
	Intravaginal		**Perineal**		**Intravaginal**		**Perineal**	
Position	**Med**	**IQR**	**Med**	**IQR**	**Med**	**IQR**	**Med**	**IQR**
1.supine knees-flexed	21	(20,21)	32	(30,33)	-43	(-81, -13)	-34	(-68, -32)
2.supine knees- straight	20	(17,21)	28	(9,34)	20	(-15, +55)	34	(-32, +68)
3.sit leaning forwards	34	(21,52)	32	(5.5,50)	43	(+20,+81)	45	(+22,+64)
4.sit upright	21	(20,22)	32	(23, 35)	-44	(-81, -12)	-36	(-68, -32)
5.stand leaning forwards	21	(20,23)	30	(2,32)	36	(-43, +80)	32	(-32, +64)
6.stand upright	21	(21,23)	32	(30,33)	-11	(-44, +53)	-31	(-60, +64)

MOTOR RECOVERY AFTER STROKE –OPTIMIZING RESULTS

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STROKE THERAPIES

Enhancing motor recovery after stroke has been the focus of a broad range of research efforts in the past several years, based on the improved understanding of the plasticity of the brain and our ability to modulate this response to injury. Approaches to enhancing this recovery can be divided into several categories, including: the use of therapeutic exercise/practice, provision of sensory stimulation directly to the involved limb, direct brain stimulation, medications, and the use of stem cells and growth factors. Evidence of brain plasticity in response to exercise/motor task practice treatment is well established (Liepert, 2000).

While these approaches appear to be distinct, there is, in fact, considerable overlap. In particular, the use of therapeutic exercise/practice is a component of virtually all of these therapies. The human brain develops, functions, and remodels itself based on an incompletely understood interplay between intrinsic biological programming and interaction with the external environment. With regard to the motor system, this interplay between brain and environment occurs largely through task-oriented movements. Thus treatments are combined with task-oriented motor practice. A recent example of this approach was the Everest study sponsored by the Northstar Neuroscience company, in which low-level electrical stimulation of the brain was combined with intensive exercise therapy, and compared with exercise therapy alone (Northstar, 2008). Interestingly, while this particular study found no incremental benefit of targeted brain electrical stimulation, it did show improved motor function attributed to exercise in both the study and control groups. Therefore interventions can be divided into two fundamental parallel tracks of research:
1) finding the best exercise/practice treatment program, and 2) finding the best adjunctive method(s) of facilitating activity-dependent motor recovery in the brain to enhance the effects of practice.

Critically important issues regarding exercise/practice for motor recovery remain to be resolved. These include determining the best type of exercise therapy, the optimal time window for providing specific types of exercise, the optimal duration of such training, the intensity of treatment (i.e. session length and frequency of sessions), and a method for establishing therapeutic endpoints (i.e. when to cease therapy). Even a key study establishing the efficacy of a particular form of therapeutic exercise, such as the EXCITE trial (Wolf, 2006), leaves most of these questions largely unanswered. Attempts to ascertain the optimal intensity of physical therapy have been made, but have only answered a portion of this critical question (Kwakkel, 1999).

ROBOTICS

The use of robotic and other technologies to facilitate therapeutic exercise shows considerable promise, though some caveats exist as well. Cost remains a substantial concern, as these technologies currently represent an extra cost, rather than a replacement for other expenses (i.e. physical or occupational therapist time), and the equipment itself remains quite expensive. These issues should be addressed by improvements in technology that are expected to lead to easier to use devices that do not require direct therapist supervision, and reductions in cost as technologies mature and efficiencies of scale are achieved.

Another limitation of robotic technologies is the relatively constrained nature of the range of activities that can be performed. For example, use of a robot for gait training, such as the Lokomat® (Hocoma, Inc), does not incorporate the important activities of rising from a seated position, nor practicing turns. General purpose anthropomorphic robots remain largely in the realm of science fiction at present, and existing devices are often highly specialized. The development of mobile wearable devices for the upper limb, however, such as the Myomo e100 (Stein, 2007), creates the potential for more general purpose assistive devices to facilitate motor retraining. Considerable more research and development is needed before devices truly capable of providing robotic assistance in a broad range of functional activity training become available.

Among the advantages of robotic devices is the ability to create games or virtual reality scenarios to help make the activities more engaging and/or better simulate functional tasks. Programmable devices provide a consistent and reliable form of exercise training, allowing repeated exercises to be performed within precise parameters. Many of these devices also have measurement capabilities, allowing progress to be monitored in a quantitative fashion, and improving our understanding of the underlying changes in motor performance (Rohrer, 2004). Ultimately robotic devices may be expected to be labor-saving, and allow the delivery of more extensive therapy programs without requiring increasing the personnel available. Even with the current state of development, provision of partial body weight supported treadmill training with the use of a Lokomat® requires fewer personnel than providing this therapy with a purely manual system.

FUTURE THERAPIES

The use of progenitor/stem cells to repopulate damaged areas of the brain with replacement neurons remains an appealing long-term therapeutic goal for recovery after stroke. Based on our knowledge of brain plasticity, it is clear that any replacement neurons will need to be appropriately incorporated and entrained into functioning brain networks to provide benefit. Some form of exercise therapy involving task-oriented practice will undoubtedly be a critical component of the care of these patients. Efforts to determine the best therapeutic exercise paradigms and exercise aids will create the foundation for these therapies in the future.

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EVALUATION OF THE FUNCTIONS OF THE UPPER LIMB AND THE COGNITIVE FUNCTION WITH THE IMPROVED MODEL OF A HAPTIC DEVICE SYSYTEM IN HEALTHY CHILDREN

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INTRODUCTION

We have developed the clinical application of the system for the evaluation and training of the motor functions of the upper limb and the cognitive function with a device for virtual haptic sensation (haptic device: HD), one of the virtual reality technology. HD is designed to present a subject's sensation at the time when he/she touches a virtual object on the display as virtual haptic sensation. We have inspected the quantitative and qualitative evaluation of the motor functions of the upper limb by measuring the precision of performance of the upper limb and performance speed by means of the system and by calculating the jerk as an indicator for smoothness of performance in healthy persons and stroke patients. On the basis of the achievements of our previous studies, we made attempts to add new software for the evaluation and training and to improve specification of the device. The present study was designed to obtain the basic data for rehabilitation of the disabled by measuring the functions of the upper limb and the cognitive function by means of the improved model of the haptic device system in healthy children.

METHODS

The subjects were 16 healthy children, whose consent about this study was obtained. The mean age was 8 years (3-10 years).

This system consists of HD, a personal computer, display, and the software for the evaluation and training (Figures 1). HD allows friction, elasticity, viscosity, and load being reproduced. When the subjects hold the HD's handle and move it responding to target on the display, HD produces the virtual force during operation.

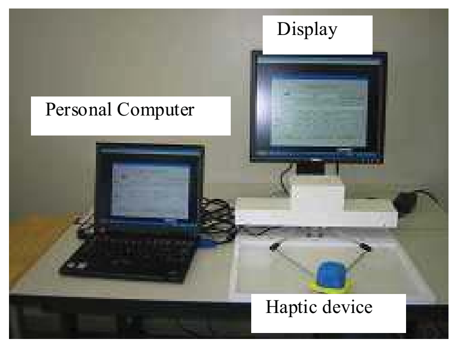

Figure 1: The improved model of a haptic device system for rehabilitation of the function of the upper limb and cognitive function (A product of MITSUBISHI PRECISION Co., LTD.)

At the site of data collection, samples of the target position, position of cursor, and force to the handle are collected and stored during the software working. At the site of data analysis, the time required for movement of cursor to the target, the error, reaction time, etc. are subjected to processing and scoring for the evaluation of smoothness, responsiveness, and precision of the performance.

The eye-hand coordination test and perception test of the evaluation program were used. The coordination test has two tasks. One is a linear task, in which an object handled is moved with the target moving on straight lines extending radially from the center of the display, and the other is a curve task, in which an object handled is moved with the target moving along the pathway of sine wave. With regard to speed of movement of the target, two types, low and high speeds, are established for each task. The perception test includes a task of discriminating friction, elasticity, viscosity, and load. In the eye-hand coordination test, tracking (error between the target and the object handled), smoothness (jerk), and the mobile distance (the range from the point of start) were calculated, and the data were expressed by scoring. In the perception test, a discrimination test for each of friction, elasticity, viscosity, and load was performed 4 times, and the accuracy of the data from each test was expressed by scoring. The total points in the perception test (the points scored for ability of perception) were calculated from individual scoring points. With regard to the points scored for tracking, smoothness, mobile distance, and contact force sensibility, we compared these points scored according to the items, and assessed relations of the points scored for each item to age and lower items of each test.

RESULTS AND DISCUSSION

A comparison of the score points for tracking, smoothness, mobile distance, and the ability of perception revealed that the score points for tracking were significantly (p <0.05) lower than those for other items. Tracking was influenced by the type of lines (p <0.001) and by speed (0.05< p <0.1). And the error in the conditions of curve and low speed was significantly (p <0.05) more than those in the conditions of straight line and low speed and those of straight line and high speed. These results indicated that slow motion on the curve was the most difficult action for healthy children. With regard to the score points for friction sensibility, elasticity sensibility, viscosity sensibility, and load sensibility in the perception test, the score points for load sensibility were significantly (p <0.05) higher than those for friction sensibility and viscosity sensibility, and the score points for elasticity sensibility were significantly (p <0.05) higher than those for viscosity sensibility. These results suggested that the contents of sensibility discriminated are related to the order of sensory development. When correlation between the score points for each item and age was investigated, there were positive correlations of the score points for tracking and of those for roughness sensibility with age (r = 0.864, p <0.001; r = 0.579, p <0.05, respectively).

SUMMARY/CONCLUSIONS

The functions of the upper limb and the cognitive function were measured in healthy children by the improved model of the haptic device system. The results suggested that characteristics of eye-hand coordination and of the ability of perception in healthy children are measured to some extent with this system.

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KINEMATIC AND ELECTROMYOGRAPHIC CHARACTERIZATION OF PHANTOM LIMB MOVEMENTS IN ABOVE-ELBOW AMPUTEES

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INTRODUCTION

In a majority of cases limb amputation is followed by the vivid sensation that the now-missing body part is still there, a phenomenon called “phantom limb”. The perceived ability of amputees to voluntarily control movement of the phantom limb is one of its most intriguing and least understood aspects, but is of clinical interest as the ability to move the phantom limb appears to be negatively correlated with phantom pain (Giraux and Sirigu, 2003). Execution of phantom movements is known to be accompanied by descending motor outputs that can be recorded both in severed nerves (Dhillon et al., 2004) and stump muscles (Reilly et al. 2006). Interestingly, distinct stump muscles electromyographic (EMG) patterns have been observed for distinct phantom movements. The aim of this study is to understand further phantom motor control and to provide quantitative measurement tools for further studies on phantom limb phenomenon.

METHODS

Five above-elbow traumatic amputees have been asked to produce different cyclic voluntary movements of their phantom and intact limbs simultaneously: 1-elbow flexion /extension; 2- wrist flexion/extension; 3- closing/opening hand; 4- finger abduction/ adduction; 5-thumb opposition/extension; 6- thumb abduction/adduction. Three trials (five cycles by trial) were performed for each movement in a randomized order. Subjects were instructed to move their intact limb through the same range and at the same speed as the phantom limb was moving, and movements of the intact limb were measured using a Cyberglove (finger and wrist) and an electrogoniometer (elbow). We recorded EMG activity from three stump muscles (biceps, triceps and deltoid) using surface Ag-AgCl electrodes, and from the three same muscles on the intact side. The kinematic data was used to separate EMG activity associated to each movement phase, to quantify duration for each movement cycle and to measure the relative amplitude of motion (in % of maximal range of motion (ROM) of the intact hand).

EMG was extracted for each half-cycle of each movement. The duration of each half-cycle was standardized. EMG was rectified, smoothed and normalized against maximal voluntary contraction (MVC) value. Several parameters were extracted from kinematic and EMG data: half-cycle duration, % of maximal ROM, normalized maximal EMG value and normalized average EMG value.

RESULTS AND DISCUSSION

Movement time varied across subjects and was generally longer in the three patients suffering from phantom pain, with an average movement time of 26 seconds for one cycle (range = 9 to 44.8 seconds), while the average movement time in the two patients without phantom pain was of 6.4 seconds (range = 5.4 to 8.2 seconds).

Average % of maximal ROM was of 55.8% for the three amputees with phantom pain, and of 62.4% in the two pain-free amputees, with large variations across movements. Interestingly, the largest amplitudes of movement were observed for movements normally associated to intrinsic hand muscles (finger abduction/ adduction and thumb opposition, average of 67.6%), followed by movements associated to extrinsic (in combination with intrinsic) hand muscles (thumb abduction/adduction, hand closing/opening, average of 56.6%). Proximal movements (elbow and wrist flexion/extension) showed the most limited amplitude, with an average of 36.3%. This proximo-distal gradient in preserved phantom movements is presumably related to the extensive representation of hand muscles within primary motor cortex.

EMG patterns in stump muscles were found to be distinct between movements (Fig. 1) and consistent across trials. On the intact side, EMG levels never exceeded 15% of MVC. In contrast the average maximal EMG level during phantom movements on the amputated side varied across subjects from 6-100% for biceps, 2-98% for triceps, and 1-15% for deltoid. Such high contraction levels in stump muscles could account for the perceived level of difficulty of phantom movement execution.

CONCLUSION

Kinematic and EMG measurements can be used to quantify the ability of amputees to perform different phantom movements. Such measurements are clinically relevant for further studies on phantom pain and prosthetic control. A larger sample size will allow us to assess the reliability of these measurements, and to assess their relationship with other parameters such as phantom pain and perceived level of effort.

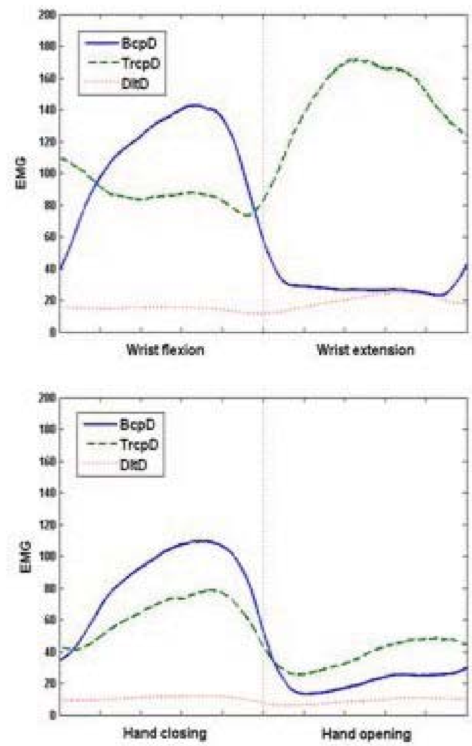

Figure 1: Example of two distinct EMG patterns recorded in stump muscles for two different phantom movements. Each trace is an average of 15 movement cycles. Movement duration is standardized and EMG amplitude is expressed in % of MVC (in this patient, EMG associated with phantom movements often exceeded EMG level associated to MVC of stump muscles).

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AUGMENTED REALITY: A TOOL FOR MYOELECTRIC PROSTHESES

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INTRODUCTION

A very popular approach for prosthesis control is based on the use of EMG signal as the control input. As these devices, known as myoelectric prostheses, use a biological signal to control their movements, it would be expected that they should be much easier to operate. In fact, as described by Soares et al. (2002), the prosthesis control is very unnatural and requires great mental effort, especially during the training stages and the first months after fitting. Those are among the main reasons why many users end up abandoning such highly advanced devices.

Augmented reality is usually defined as the overlapping of virtual 3D objects into a real environment. Generally, this can be done by combining virtual images, generated by the computer, with real images captured by cameras. The “augmented image” is then presented to the user in real-time.

In an attempt to devise better strategies for helping users to control and get used to the “behaviour” of myoelectric prostheses, especially during those critical initial months, we propose the use of a virtual prosthesis that can be fully controlled by the user in an immersive environment that combines both virtual and real scenes. The virtual prosthesis must be capable of reproducing the operation of a real prosthesis in every aspect, except by the fact that it is not really there. Therefore, it is expected that problems such as weight, heat and pain should not contribute to an already hard task.

METHODS

Figure 1 shows the basic block diagram for the proposed system.

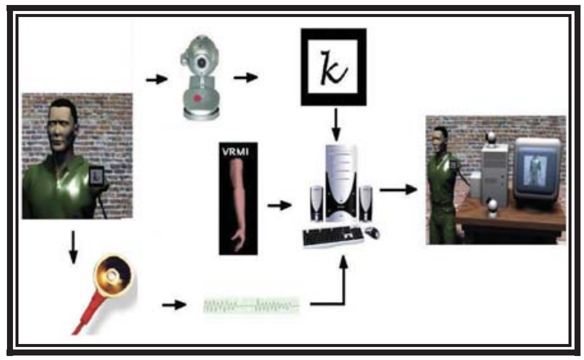

Figure 1: Augmented reality environment for upper-limb prostheses simulation.

EMG signals are collected from the muscles and processed to be used as the input signal for controlling the movements of a virtual prosthesis. A video-camera captures images of the real world that contains a marker positioned where the virtual prosthesis should be placed. A computer system combines the virtual prosthesis with the image from the camera and controls its movement according to the control signals.

EMG Processing

The EMG signals, detected by surface electrodes (Ag/AgCl, 1cm diam.), are amplified (1000x) and band-pass filtered (20Hz – 500Hz) prior to data acquisition (2kHz/12bits). The resulting signal is then processed by the combination of AutoRegressive models (Akay, 1996) and a three-layered artificial neural network (Fausett, 1994) to classify the EMG patterns into four classes of movement. In this case study, we collected signals from the biceps brachii (long and short heads) and the triceps brachii (long, lateral and medial heads) in order to discriminate wrist pronation, wrist supination, elbow flexion and elbow extension – A detailed explanation about the techniques used for pattern discrimination can be found in Soares et al. (2003).

The output of the neural network can then be used to update the correct function of the virtual limb.

Virtual Limb

The 3D object (limb) was modeled using 3Dstudio Max® (Figure 2) to represent not only the volume of the limb, but also the bones and their constrained range of motion.

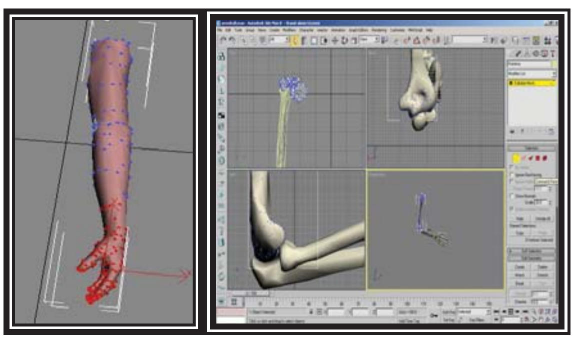

Figure 2: Modeling a virtual limb.

Augmented Reality

The ARToolKit (ARToolKit, 2005) was used to calculate the point of view of the camera in relation to the marker in the real world. OpenGL was used to calculate the virtual coordinates of the camera and draw the virtual limb to generate the virtual scene. Now, the ARToolkit was used to combine the virtual scene with the real one, captured by the camera, and present the result on the desired video output (VR goggles - for immersion - or computer displays) as shown in Figure 3.

Figure 3 – Augmented reality with upper limb prosthesis.

CONCLUSIONS

This paper describes the use of augmented reality to generate a tool for training users on how to control artificial upper-limb prostheses. The proposed system is based on myoelectric signals to control the virtual limb in real-time. The experiments have shown that the virtual prosthesis can respond in real-time to commands of the user. The authors believe that the system has great potential not only as a training tool, but also as a simulation environment to study new designs and new control strategies for prosthetic devices.

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EVALUATING THE USE OF A WALKING DOG AS A REHABILITATION AID: KINEMATIC ANALYSIS

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INTRODUCTION

The benefit of using a rehabilitation dog for assisting people with severe disabilities has long been recognized. Service dogs are known to increase autonomy and facilitate mobility in people with severe neuromuscular disorders such as quadriplegia and muscular dystrophies (Allen and Bloskovich, 1996). Recently, rehabilitation dogs are also being used clinically for gait training (Rondeau, 2007). In patients with acute stroke, observational gait analysis has demonstrated clinical improvements when using a trained dog as an aid compared to using a cane. The purpose of this study was to explore quantitative changes in gait performance in patients with hemiparesis while using a trained dog as a walking aid in comparison to using a cane.

METHODS

3 patients with hemiparesis were recruited from inpatient and outpatients department of the Jewish Rehabilitation Hospital, Laval, Quebec. The patients walked on a 10 m over ground walkway using their usual walking aid (cane/quad cane) and the dog in a random order. The 3 dimensional kinematic data was obtained by a 6 camera Vicon Motion Analysis system (Oxford,UK) through passive reflective markers. The marker placement largely corresponded to the marker placements defined in the PlugInGait model. The marker positions were filtered with the help of a dual pass, 8th order Butterworth Filter. Analog data included force plate measurements and electromyography. The kinematic variables of interest in this study were gait parameters and the Centre of Mass (COM) excursions. The variables were analyzed using the BodyBuilder (Vicon Motion Analysis Systems, Oxford, UK) and MATLAB (Mathworks, USA) softwares. A matched pairs T test was used to analyze the difference in peak-to-peak COM excursions in the two test conditions.

RESULTS AND DISCUSSION

Preliminary analysis of the gait parameters and the peak to peak COM excursions from one subject with left hemiparesis showed that the gait parameters of the patient improved while using the dog as a walking aid. The average comfortable walking speed was 0.56m/s with the cane as compared to 0.94 m/s with the dog suggesting an improved functional mobility with the dog (Figure 1a). The cadence also improved from an average of 70 steps/min to a 100 steps/min with the dog. An increase in the step length and stride length as well as a decrease in the step time and stride time was also observed. The time spent in double support phase reduced from 0.29 s with the cane to 0.24 s with the dog indicative of improved postural stability during gait.

The peak-to- peak COM excursions along the medio-lateral (ML) direction showed a significant difference ($p < 0.05$) between the two test conditions; the excursion being smaller when walking with the dog (16.0 $\pm$6.9 mm) as compared to walking with the cane (39.5 $\pm$9.03mm] (Figure 1). The excursion along the vertical direction also showed a similar trend but the differences across the two test conditions were not statistically significant.

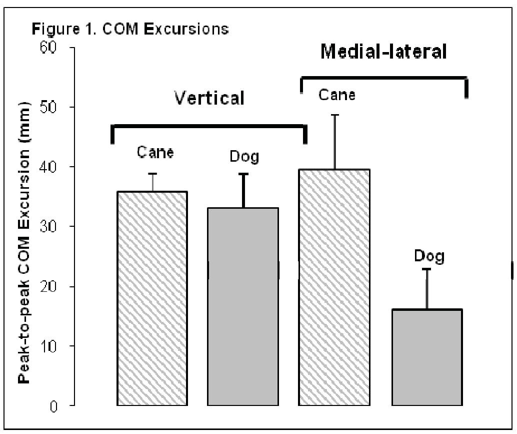

Figure 1. COM Excursions

The average excursion along the vertical direction was 33.22 ($\pm$5.33) mm with the dog and 35.88 ($\pm$3.02) mm with the cane (Figure1).The reduced movements of the centre of mass in both the medial-lateral and vertical directions indicate improved postural control during gait as well as improved biomechanical efficiency.

CONCLUSION

The preliminary gait analysis of one subject using a trained dog as walking aid reveals promising results.

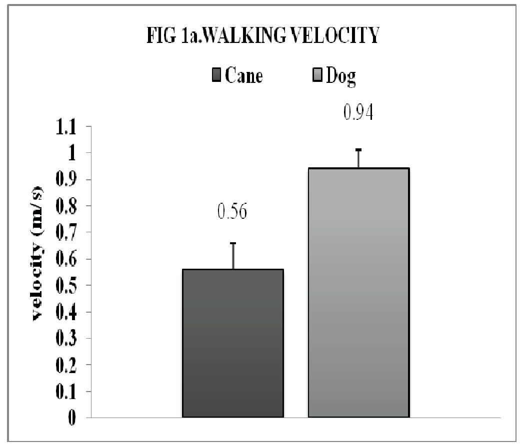

FIG 1a.WALKING VELOCITY

Improvements in gait velocity and cadence as well as double support times were observed. Also, the decreased peak-to-peak COM excursions indicate a shift towards a more normal and efficient gait pattern.

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The authors would like to thank Richard Preuss for his help with the data processing and Michael Trivino for his assistance with patient referral and data collection. We thank Lynda Rondeau for her dedication, training and care of the rehabilitation dog, Cactus, and above all, Cactus, who has diligently provided service to numerous disabled clients in the last 5 years.

EVALUATING THE USE OF REHABILITATION DOG AS A WALKING AID: EMG ANALYSIS

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INTRODUCTION

Postural stability is often compromised following a stroke. A cane or quad cane is conventionally used to increase stability during locomotor rehabilitation. Service dogs have been engaged for some time in rehabilitation (Allen and Blascovich, 1996) but a novel training regime to promote mobility has only been developed recently for neurological rehabilitation (Rondeau, 2004). This study investigates the use of rehabilitation dog as a walking aid on muscle recruitment during gait post stroke.

METHODS

Two participants recovering from a stroke were recruited from the Jewish Rehabilitation Hospital. Both participants normally walked with a quad cane at the time of collection. The rehabilitation dog was an experienced walking-aid dog and was fitted with a custom harness that allowed a comfortable handle height for the human and weight distribution over the back for the dog.

Participants walked across the laboratory with their cane or with the dog. Muscle activity was monitored through electromyography (EMG) recordings from muscles in the lower limbs and trunk using a 16-channel telemetric Telemyo900 system (NoraxonUSA Inc.). Disposable bipolar silver/silver-chloride electrodes were placed bilaterally over the belly of the following muscles: Tibialis anterior (TA), medial gastronemius (MG), vastus lateralis (VL), semi-tendonosis (ST), rectus abdominus (RA), erector spinae (ES), on the paretic side the tensor fascia latae (TFL), and the biceps and triceps muscles of the non-paretic arm using the cane/dog. EMG signals were amplified and band-pass filtered (10-350 Hz) and sampled at 1080 Hz. EMG data was high-pass filtered at 10Hz then full-wave rectified and low-pass filtered (dual-pass 100Hz) and aligned to heel contact. Body and limb movements were captured in real-time with a 6-camera Vicon motion analysis system (Oxfordmetrics, UK) at 120 Hz and used to record the time of heel contact.

In order to compare use of the cane vs. dog, EMG was integrated over the entire step of the paretic limb. Muscle onset was defined as the time when EMG activity exceeded the average +/-3SD of activity during quiet stance. The TA bursts were labeled as “TA1” or “TA2” to represent muscle activity pre- and post-heel contact accordingly.

RESULTS AND DISCUSSION

The amount of triceps muscle activity was significantly higher with use of the cane suggesting increased reliance (Figure 1). The amount of ankle extensor activity is also less when using the dog versus using the cane. The timing of muscle activity for post-heel contact TA burst was significantly delayed during cane use compared to the trials with the dog (Figure 2). Further comparison of the remaining non-paretic muscles that were tested will expand on the understanding of the neural control of locomotion post-stroke while using a rehabilitation dog as a walking aid.

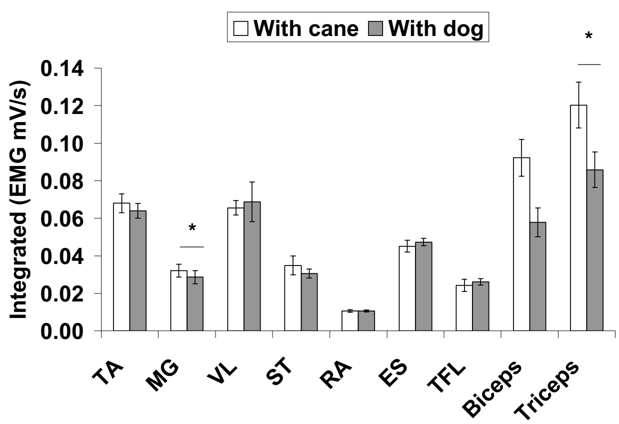

Figure 1: Integrated EMG of the paretic muscles with SE.

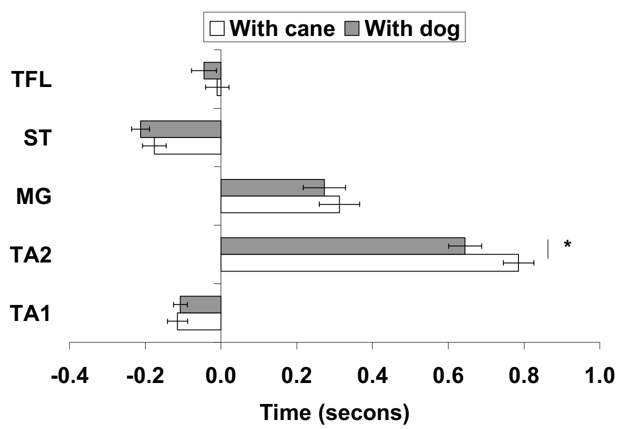

Figure 2: Muscle onset latencies with SE.

SUMMARY/CONCLUSIONS

The use of a rehabilitation dog as a walking aid improves locomotion beyond what is seen with a traditional cane (Rondeau 2004). The reliance on the walking aid, demonstrated through diminished triceps activity, along with diminished MG activity, show improvements in waking with the dog compared to walking with a cane. Further investigations of the muscle activity will illustrate how the neural control locomotion is improved by using the rehabilitation dog as a walking aid.

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Rondeau, L (2007) Utilisation du chien d'assistance physique en physiothérapie avec la clientèle adulte. CRE, Sherbrooke, document interne p.3

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THE EFFECT OF STRENGTH TRAINING ON HAND FUNCTION IN CHILDREN WITH CEREBRAL PALSY

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INTRODUCTION

Children with hemiplegic cerebral palsy (CP) often have impaired hand function and experience problems performing daily activities. This is assumed to be a result of muscle weakness, spasticity and reduced motor control. As a treatment, injection of Botulinum Toxin A (BTX-A) is often applied (Hoare & Imms, 2004). Strength training can also improve strength and reduce spasticity and is often advised next to BTX-A treatment (Fehlings et al., 2000, Gage, 2004). However, most strength training studies have been performed on the lower extremities. In addition, no studies have evaluated the effect of training on hand function compared to other treatments.

The objective of this study was to investigate the effect of strength training and BTX-A treatment and a combination of the two on muscle strength and hand function.

METHODS

Twenty four children (8-18 years old) with hemiplegic spastic CP are included in this study. All the children had a functional use of their upper extremities corresponding to MACS level I or II, and indications for treatment with BTX-A for the pronator teres and/or biceps brachii/brachialis.

The children were divided into 3 intervention groups with 8 children. One group (T) received intensive resistance training, one group (B) received BTX-A treatment and one group (BT) received both intensive resistance training and BTX-A treatment.

The strength training programme consisted of 3 sessions per week in an 8-week period. Free weights were used. The programme was designed as single-joint training for elbow (flexion/extension), forearm (pronation/supination), wrist (flexion/extension) and grip. The dosage was set on the basis of 1RM using free weights, tested three times before intervention: 10RM X 3.

A dynamometer was used to measure strength in the elbow (flexion/extension) and the forearm (pronation/supination). An electronic Grip Force Meter was used to measure grip strength. The Assisting Hand Assessment (AHA) and Melbourne Assessment of Unilateral Upper Limb Function were used to evaluate hand function.

The measurements were carried out one week before and nine weeks after intervention start.

RESULTS

Preliminary results indicate that grip force, elbow flexion, elbow extension and supination strength increase in both training groups and decrease in the BTX-A group.

Hand function as investigated with the Melbourne test and AHA showed a general improvement after treatment, independent of the type of treatment.

SUMMARY/CONCLUSIONS

Upper extremity strength training increases strength, and BTX-A treatment seems to reduce strength in relatively well-functioning children with CP, while both interventions have a similar trend in functional improvement.

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REHABILITATION OF ATROPHIC MUSCLES AFTER TOTAL ANKLE REPLACEMENT, TAR

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INTRODUCTION

Traumatic ankle osteoarthritis usually develops in parallel with severe lower leg muscle atrophy and weakness. In a previous study it was shown that the muscle atrophy considerably altered the spectral aspects of the EMG recorded from the lower leg muscles (Valderrabano et al. 2006). A drop of the mean frequency by around 26 Hz was detected on the injured leg when compared to the healthy contra-lateral one when maximum isometric torque was applied by dorsiflexion or plantar flexion of the foot. Under maximum voluntary contraction one may assume that most muscle fibers get activated and thus fibers contributing lower and higher frequencies to the EMG spectra. The study showed that the EMG intensity recovered during rehabilitation but the high frequency contributions of the EMG did not. This assessment was done on static measurements. In the current study we focused on the muscle activity during walking. Walking is a relatively slow movement and it might be that only parts of the muscle fibers get activated which in turn might be observed in the spectral behavior. The purpose of this study was to test the hypotheses a) that patients with unilateral ankle osteoarthritis produce lower frequency components in the affected leg, AFL, than in the non affected one, NAL, when walking; b) that during rehabilitation the muscle activity pattern of walking does not regain the higher frequency components in the EMG. If these hypotheses prove to be correct, thus even when walking the high frequency components are required in the contra-lateral leg the rehabilitation procedures have to be reconsidered. New training methods similar to those used in athletic training may prove to result in regaining higher frequency components.

METHODS

Fifteen patients were assessed before TAR and after TAR in 3 month intervals up to one year. Beside many other variables EMG was measured for the anterior tibial muscle, TA, the medial gastrocnemius, MG, and on the peroneus longus , PL (Valderrabano et al. 2006). The EMG signals were submitted to a wavelet based time/frequency analysis which yielded the intensity patterns of muscle activation (von Tscharner, 2000). From the intensity patterns the wavelet spectra and the total intensity were extracted and used to assess the rehabilitation of either the intensity or the mean frequency.

RESULTS

The wavelet analysis confirmed hypothesis that the NAL expressed the high frequency components of the EMG while walking and the AFL did not. Thus the activation of the high frequency component in a movement did not require a muscle activity that is close to MVC. During rehabilitation the EMG intensity was mainly recovered indicating the strengthening of the muscle. However, the high frequency components of the EMG did not recover during the rehabilitation period. Thus part of the impairment remained after the rehabilitation period.

DISCUSSION

The lack of high frequency components in the EMG signal may reflect some chronic impairment that was not recovered by the current rehabilitation process. However, the rehabilitation process recovered the muscle mass and with it the EMG intensity thus indicating that a possible arthrogenic muscle inhibition is most unlikely to persist. Up to now one cannot explicitly assign low and high frequency components to certain fiber types and frequency changes can have various causes (Valderrabano et al. 2006). However, there is evidence that during different sequences of a movement predominantly high or low frequencies are expressed in the EMG signal (von Tscharner and Goepfert 2006). Sirca et al (Sirca and Susec-Michieli 1980) and others reported that OA was associated with selective type II fiber atrophy. If one assumes that the high and low frequency components of the EMG are a result of activating one or the other of two groups of muscle fibers then our results would indicate that those fibers contributing the high frequency components did not recover during rehabilitation. This lack of recovery during rehabilitation may impair reflex type of movements and thus affect stability during walking of the patients. Modified rehabilitation methods will have to be tested to see whether these missing high frequency components in the EMG can somehow be regenerated for instance with fast isokinetic high resistance muscle contractions.

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EFFECT OF MULTILEVEL BOTULINUM TOXIN A INJECTIONS ON SURFACE EMG PATTERNS DURING GAIT IN CHILDREN WITH CEREBRAL PALSY

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INTRODUCTION

In children with cerebral palsy (CP), it is thought that abnormal muscle activation is one of the main factors leading leading to deviating gait patterns. To prevent children with CP from walking with knee flexion during midstance from a potential deterioration in their mobility, these children are treated with intramuscular injections of botulinum toxin A (BTX-A) combined with comprehensive rehabilitation [1].

Abnormal muscle activation can be described by surface electromyography (EMG) patterns, which can be quantified during periods of the stride when a muscle should be active (i.e. on-period) or relaxed (i.e. off-period). [2]

This study evaluated the effect of BTX-A on muscle activation based on surface EMG patterns of the rectus femoris, semitendinosus and gastrocnemius medialis muscles.

METHODS

A total of 22 children with CP, hemi- or diplegic , mean age 7.6 (±2.1) years, walking with kneeflexion in midstance participated. GMFCS callisifaction ranged from I-III. After measurements at baseline, an assessment resulting in a specific indication for BTX-A injection for each muscle was established (Scholtes) After that children were randomly assigned to an experimental (n=12) and control group.(n=10) Treatment included comprehensive rehabilitation. After 6 weeks a follow-up measurement was made

Surface EMG was recorded from Rectus Femoris (RF); SemiTendinosis (ST); and Gastrocnemius Medialis (GAM) while children were walking at their comfortable walking speed. Bipolar electrodes were used, placement following the SENIAM protocol [3]. Linear envelopes were calculated using a 2 Hz lowpass filter. Time-normalized envelopes were calculated from 6 strides.[4] From these an on/off quotient was calculated, using predefined periods specific for each muscle [2] , see figure 1.

The change of the on/off-quotient of the injected muscles (RF, ST and GAM) of the intervention group was compared with the change of the on/off-quotient of muscles in the control group that received a treatment indication. The differences in effect between the two groups were analyzed in a linear mixed model analysis, using SPSS 11.5.

RESULTS

See table 1. The on/off quotient decreased after injection with BTX-A; GAM and RF being significant . A typical example is shown in figure 1

Muscle	On/off-quotient of the experimental group			On/off-quotient of the Control group			Effect (95 % CI) on on/off quotient	p-value
	Baseline	Follow-up	Δ	Baseline	Follow-uo	Δ		
RF	1.00	0.69	-0.31	1.03	1.00	-0.03	-0.28 (-0.54 to –0.02)	0.04
ST	1.62	1.23	-0.39	1.54	1.36	-0.18	-0.21 (-0.48 to 0.06)	0.12
GAM	1.14	0.94	-0.20	1.08	1.16	+0.08	-0.28 (-0.47 to –0.09)	< 0.01

Table I. Δ = change from baseline; RF = rectus femoris; ST = semitendinosus; GAM = gastrocnemius medialis.

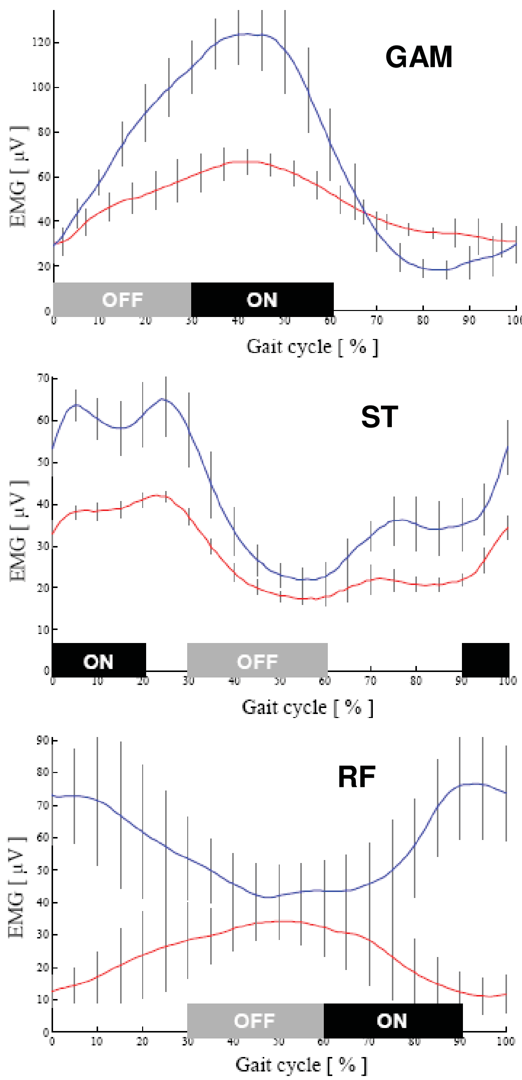

Figure 1. *Ensemble averaged EMG profiles of GAM, ST and RF, before (blue, upper line) and 6 weeks after (red) injections with BTX-A*

DISCUSSION & CONCLUSIONS

The effect of injection of a muscle with BTX-A is known to denervate and therefore paralyse the muscle [5]. Our EMG evaluation confirm these effects in a typical clinical application. Injections with BTX-A showed a non-specific decrease of muscle activation resulting in deterioration of the activation pattern. Especially no decrease of activation in periods of gait where relaxation would be beneficial, was seen. However, other methods of quantification of muscle coordination should be explored. The overall beneficial effects of the BTX-A treatment, are not likely to be the result of a better muscle coordination, but must be attributed to the combined treatment [1,6]

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WALKING IN PEOPLE WITH MYELOMENINGOCELE: MUSCLE ACTIVITY MECHANISM

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INTRODUCTION

Approximately 70,000 people (0.5 to 1 out of 1,000) in the United States are living with myelomeningocele (MMC). MMC is a neural tube defect that involves a disruption of the sensory and motor pathways and causes significant walking problems, such as a significant delay in walking onset, difficulty in ambulation, and cessation of walking in the middle of their life.

Previous studies show muscle strength plays an important role in walking of people with MMC (Chang and Ulrich, 2007a) and external lateral stabilization can improve their walking (Chang and Ulrich, 2007b). However, limited studies examine their muscle activity that may be one of the important mechanisms that affect compensatory walking patterns in people with MMC. The purposes of this study were to investigate muscle activity during walking in people with MMC. By applying the findings of this study, we may be able to design and test more advanced and more efficient treatments. Thus, persons with MMC will be able to move through the world more easily, with improved gait quality, with less pain and fatigue, and with greater independence.

METHODS

We examined 12 people with MMC who could perform at least 4 to 6 independent steps without any orthoses. They walked over ground (barefoot). We used a 6-camera Peak MotusTM motion-capture system, a GAITRite system, Therapeutics Unlimited EMG equipment (with 6 channels), a Cybex isokinetic dynamometer, and anthropometric tools to collect our data. We placed surface electrodes over the following muscles on both legs: tibialis anterior (T), gastrocnemius (medial head) (G), soleus (S), quadriceps (rectus femoris, QR; vastus lateralis, QV), and hamstrings (biceps femoris, H). This study has been approved by the Institutional Review Board of the University of Michigan Medical School.

RESULTS AND DISCUSSION

Overall, we found high co-activations of the antagonist pairs in people with SB (Fig 1). Our results showed significant *negative* correlations between muscle co-activation and muscle strength. The strength of ankle plantarflexors was negatively related to the level of co-activation of T & G ($p = 0.005$), T & S ($p = 0.013$), G & QR ($p = 0.046$), G & QV ($p = 0.026$), and S & QR ($p = 0.029$). The strength of ankle dorsiflexor was negatively related to the co-activation of T & G ($p = 0.002$) and T & S ($p = 0.019$). The strength of knee flexors was negatively related to the co-activation of T & G ($p = 0.004$). The strength of hip adductors was negatively related to the co-activation of T & G ($p = 0.007$). The strength of hip extensors was negatively related to the co-activation of T & G ($p = 0.019$), T & S ($p = 0.033$) and S & QR ($p = 0.009$).

We also found a significant *positive* correlation between the co-activation of T & G and trunk ranges of motion (ROM) in the frontal plane during strike cycle ($p = 0.002$), stance phase ($p = 0.003$), and swing phase ($p = 0.006$). A significant *positive* correlation was also found between the co-activation of QR & H and trunk ROM in the frontal plane during strike cycle ($p = 0.028$) and stance phase ($p = 0.03$). We did not find significant relations between any pairs of muscle co-activation and pelvic ROM in the frontal plane.

As people with MMC have a longer stride cycle compared to healthy people, the prolonged leg muscle activities in this population might be the results of long stride cycles. We normalized muscle burst duration by cycle duration. Our normalized muscle burst duration results suggests that people with MMC did not increase knee extensor burst duration to avoid their crouch gaits during stance phase. Neither did they increase ankle dorsiflexor burst duration to prevent their drop feet during swing phase.

The inverse relations between leg muscle strength and muscle co-activation suggests that people with MMC employ leg muscle co-activation to compensate for their muscle weakness. The positive relations between muscle co-activation and trunk sway in the frontal plane implied that while people with MMC used trunk sway in the frontal plane to help to clear a swing limb, like a two-edged sword, they required more energy to co-active their leg muscles to maintain body stability. Both muscle weakness and instability may induce muscle co-activation adaptation mechanisms during walking in people with MMC.

SUMMARY/CONCLUSIONS

Our results suggest that people with MMC modulate neuromuscular activities to adapt to their muscle weakness and instability. Future researchers may manipulate muscle and sensory gains to test their effects on muscle co-activation in people with MMC by using technologies, such as functional electrical stimulation, or biofeedback.

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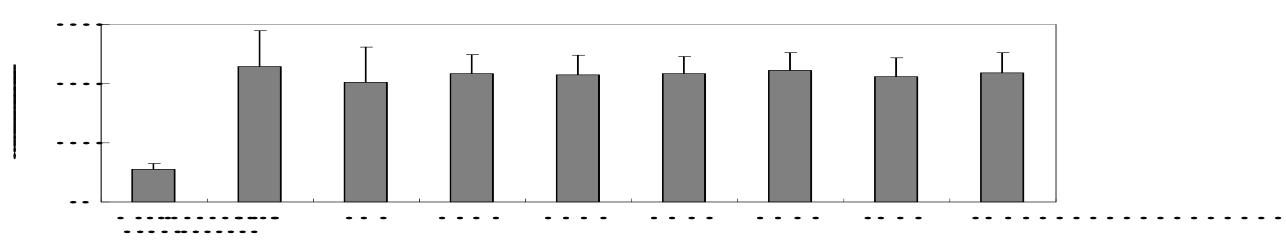

Figure 1: The co-activation percentages for muscle pairs during stride cycle are around 60 to 70 % in people with SB. In contrast, co-activation percentage of T and G for healthy people is 10 to 20 % (Yang et al, 1998).

ABNORMAL ELECTROMYOGRAPHIC SIGNALS IN THE ADULT CEREBRAL PALSY POPULATION

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INTRODUCTION

During routine clinical gait evaluation, abnormal electromyographic (EMG) signals have been noted in patients with cerebral palsy (CP). The abnormal EMG was defined as repeated individual motor unit activation for muscles not involved in the index contraction or muscles that are supposed to be at rest. The abnormal signals are not spastic in nature (they do not represent a response to quick stretch), have a higher frequency (Hz) than clonus, and have a more consistent pattern than fasciculation activity which show a modification in frequency over time (see Figure 1).

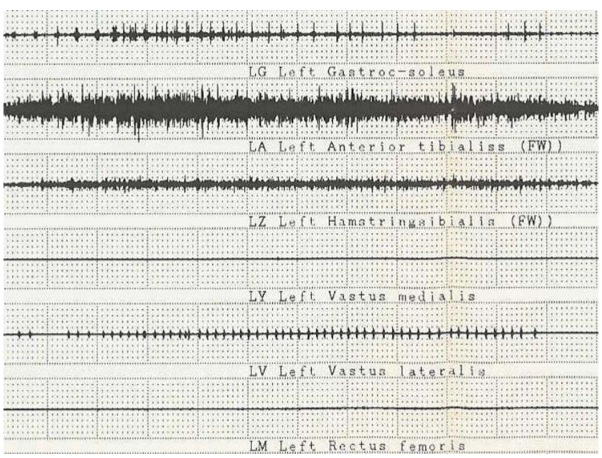

FIGURE 1: EMG signals during voluntary knee flexion

Clinical experience suggests that these abnormal signals are present much more frequently in adults with CP than in children with CP. The objectives of this initial review of clinical data were to determine: 1) the incidence of these abnormal signals within our adult clinical population, 2) if the abnormal signals tend to exist in particular muscle groups, and 3) if the abnormal signals were more frequently present in muscles that had previous orthopedic procedures (lengthenings and/or transfers)

METHODS

Surface EMG signals were captured on strip chart during routine clinical gait analysis as the patient performed isolated voluntary activities while in a seated position (relaxed sitting, hip flexion, knee extension, knee flexion, ankle dorsiflexion, and ankle plantar flexion) and during ambulation.

The muscles captured on EMG were rectus femoris, vastus lateralis, vastus medialis, hamstrings, anterior tibialis, and gastrocsoleus. EMG signals were captured for each of these muscles for every patient bilaterally, for a total of 12 muscles per patient.

A retrospective review of EMG strip chart data was performed on all adult patients seen during the past 10 years to assess the presence of the abnormal EMG data.

RESULTS AND DISCUSSION

There were a total of 120 adult patients seen, 100 had available strip chart data. Fifty seven patients had signs of abnormal EMG on their strip chart data. Abnormal findings were noted in 195 muscles out of the 1200 muscles (12 muscles per patient * 100 patients = 1200 muscles) monitored on EMG.

Table 1: Presence of Abnormal EMG in adults by age group

ABNORMAL EMG	AGE GROUPS				TOTAL GROUP (N = 100)
	18-29 YR (n = 61)	30-39 YR (n = 20)	40-49 YR (n = 11)	50 + YR (n = 8)	
None	54% (33)	15% (3)	18% (2)	63% (5)	43% (43)
Mild Abnormal	31% (19)	50% (10)	36% (4)	25% (2)	35% (35)
Abnormal	15% (9)	35% (7)	45% (5)	12% (1)	22% (22)

Table 1 shows a break down of the findings by adult age group. Presence of abnormal EMG has been categorized as 1) "none" if there were no signs of abnormal data, 2)"mild" abnormal if 1 to 2 muscles had abnormal activity, or 3) "abnormal" if 3 or more muscles had abnormal activity.

Overall, 43% of the patients had no signs of abnormal EMG on their strip chart, 35% had mildly abnormal EMG findings, and 22% had abnormal findings. There was a trend for increased abnormal findings as age increased, with the exception of the oldest group (50 + yrs).

There was no particular pattern of muscles in which the abnormal EMG was present. Presence overall was distributed as following: gastrocsoleus 18.5%, anterior tiblialis 23%, hamstrings 16.9%, vastus lateralis 16.9%, vastus medialis 13.1%, and rectus femoris 11.5%.

Abnormal EMG was noted to be present in most patients during voluntary movements however not typically in the primary agonistic muscle. It was not frequently noted in any of the EMG signals during ambulation probably because the increased muscle contractions would tend to mask the underlying abnormal signal.

It was difficult to assess retrospectively if the abnormal signals were more frequently present in muscles that had previous orthopedic procedures (lengthenings and/or transfers). This was largely because the many of the muscles that had orthopedic procedures were not captured on strip chart EMG data (i.e. posterior tibialis, adductors, iliopsoas). However in muscles that were monitored and had previous surgery there was a low incidence rate 17.4% (34/195).

Additionally, there were a similar amount of soft tissue orthopedic procedures done in the patients that had no abnormal findings (2.7 +/-2.8 procedures per side), as those that had abnormal findings (1.8 +/- 1.7 procedures per side). This suggests that orthopedic soft tissue surgery (lengthenings and/or transfers) may not be related to the presence of the abnormal EMG.

CONCLUSIONS

Clinical interpretation of the presence of these abnormal EMG signals is entirely dependent upon determining their source/cause. If these signals are indicative of a peripheral neuromuscular pathology then this would impact treatment and prognosis.

This retrospective review indicates that the abnormal EMG findings: 1) have a tendency for increased incidence with increased age 2) do not appear to be present in any one particular muscle group/pattern 3) do not appear to be related to previous orthopedic soft tissue procedures (however this needs further analysis due to limitations of data available for this assessment).

THE INFLUENCE OF DIABETIC NEUROPATHY AND THE USE OF HABITUAL SHOES IN LOWER LIMB EMG AND IN VERTICAL FORCES DURING GAIT

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INTRODUCTION

Althought the type of footwear could be considered a factor to prevent diabetic foot ulcers due to its supposed effect in plantar loads redistribution, randomized controlled trials have not provided enough support to the efficacy of therapeutic footwear prescriptions in ulcer prevention (Reiber et al., 2002; Singh et al., 2005). Also, the purchase of therapeutic footwear is restricted to a small part of Brazilian population, since Brazilian healthy system does not pay for it or to its follow up evaluation. Diabetic patients are recommended to always wear shoes to prevent injuries and its further complications. So, our intention with this study was to investigate if vertical peaks and lower limb EMG activity during gait are influenced by the use of the habitual shoes in order to attenuate loads and prevent foot ulcers.

METHODS

45 adults divided into 2 groups: control group (CG; n=21), diabetic neuropathic group (DG; n=24). The peaks and time of peak occurrence of the right vastus lateralis (VL), lateral gastrocnemius (LG) and tibialis anterior (TA) were studied during stance phase. In order to represent the effect of the changes in EMG activity, the first and second vertical peaks and first force peak deflection were determined. Three trials were recorded synchronized in each gait condition (barefoot and shod gait) in 1000 Hz.

The individuals from the both groups wore shoes they used most on a daily basis and in order to reduce variability, all groups wore proportionally the same kind of shoes: sport shoes (30%), loafers (30%), sandals (25%), dress shoes (15%). None of the subjects used therapeutic shoes.

ANOVAs test for repeated measures and Tukey Post-Hoc tests were made to inter-group and inter-conditions comparisons of the EMG and GRF variables (peaks and times of EMG and peaks of GRF) (α=5%).

RESULTS AND DISCUSSION

First vertical peaks between groups were not different in both gait conditions. The use of shoes did not attenuate overload during heel contact in DG and CG subjects. The first peak was significantly higher during shod gait than during barefoot gait in both groups. In despite of that, there were no changes in VL activity among DG subjects as it was observed among control subjects.

Table 1: Vertical GRF variables (mean±SD) (normalized by body weigth) of CG and DG during barefoot (baref) and shod gait.

		CG	DG	p
Fy1	**Shod**	$1.11\pm0.10^{\$}$	$1.13\pm0.07^{\&}$	>0.05
	Baref	$1.05\pm0.09^{\$}$	$1.08\pm0.06^{\&}$	>0.05
	p	0.001	0.004	
Fy2	**Shod**	$1.12\pm0.06^{*}$	$1.05\pm0.06^{*}$	0.001
	Baref	$1.10\pm0.06^{**}$	$1.04\pm0.06^{**}$	0.001
	P	>0.05	>0.05	
deflection	**Shod**	$0.51\pm0.08^{*}$	$0.60\pm0.11^{*\#}$	0.006
	Baref	0.47 ± 0.07	$0.53\pm0.10^{\#}$	>0.05
	p	>0.05	0.01	

$^{\$,\ \&,\ \#}$ represents statistical differences between conditions; $^{*,\ **}$ represents statistical differences between groups (p<0.05).

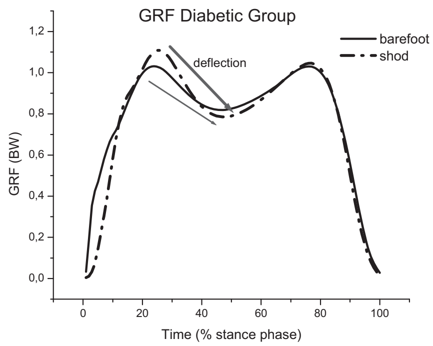

Figure 1: Mean vertical GRF normalized by body weight during barefoot and shod gait of DG. Note the arrows representing the difference between first peak deflections.

Table 2: Time of EMG peak occurrence (mean±SD) (% of stance phase) of CG and DG during barefoot and shod gait.

		CG	DG	p
TA	Shod	6.95 ±2.85	6.45±2.99	>0.05
	Barefoot	5.81±2.47	5.36±2.13	>0.05
p		>0.05	>0.05	
LG	Shod	67.0±4.16*	67.87±4.35**	>0.05
	Barefoot	64.0±4.33*	64.16±5.73**	>0.05
p		0.05*	0.01**	
VL	Shod	15.50±4.28&	14.33±3.73	>0.05
	Barefoot	11.95±3.36&	13.10±2.60	>0.05
p		0.01&	>0.05	

*, ** and ,& means differences between barefoot and shod gait in CG in LG and VL ($p<0.05$).

There were no differences between groups or conditions in the EMG peak values of the muscles studied.

Walking barefoot may be increasing proprioceptive inputs (Shakoor and Block, 2006) among the CG and DG, which allows them to seek strategies for load attenuation, leading to lower values of the first GRF peak. The VL delay on the CG in shod gait explains such higher peak among this group. This delay is not observed on DG. However, the DG GRF curve shows a better load accommodation during shod gait in comparison to barefoot gait, which is confirmed by the statistical difference between the first peak deflection in DG and not in CG (table1). By this, we could assume that shod gait leads to a better load accommodation in DG. Perhaps there is a more adequate knee flexor range of motion to better accommodate load during shod gait.

The lower Fy2 in DG in both gait conditions is not explained by the LG activity, once there was no delay of this muscle in DG. The smaller ankle range of motion presented by the diabetic neuropathic subjects (Mueller et al., 1994, Rao et al., 2006) seems to be the main cause of the deficit of propulsion showed by the lower Fy2. There were no differences in Fy2 caused by the use of shoes. In despite of that, the LG delay in both groups during shod gait, in comparison to barefoot gait, is not changing the propulsion between two gait conditions, but could represent an adaptation to the shod condition.

SUMMARY/CONCLUSIONS

Shod gait leads to a better load accommodation in diabetic neuropathic subjects, independently of muscle activity. There were some differences in muscle activity between shod and barefoot conditions, clarifying the necessity of more biomechanical studies with diabetic population using their habitual shoes, not during barefoot condition.

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HOFFMANN REFLEX FROM THE VASTUS MEDIALIS OBLIQUE MUSCLE DURING TRACTION OF THE LEG

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INTRODUCTION

To investigate a more effective the vastus medialis oblique muscle (VMO) strengthening exercise, we wanted to clarify the neural control of the VMO using electrophysiological methods because generation of muscle tension by voluntary contraction depends on not only the physiologic cross-sectional area of muscle but also the innervating neural drive. We specifically examined the excitability of the spinal neural function corresponding to the VMO using the Hoffmann reflex (H-reflex). Kawano (2004) has reported that the proximal resistance with traction exercise (PRTE for knee extension) is effective during leg-extension for facilitating the VMO contraction. In this method, the leg is pulled inferiorly and internally rotated at the sitting position. Patients and/or athletes then have to extend the knee against the resistance of the proximal part of the leg. We have found that the VMO contraction is facilitated during leg-extension using the PRTE technique. We further wished to identify the dominant factor in the PRTE technique (proximal resistance, internal rotation or traction of the leg). This study therefore investigated the H-reflex from the VMO during traction of the leg as one of the factors of the PRTE technique in order to clarify the excitability of the spinal neural function corresponding to the VMO.

METHODS

Seven healthy male volunteers (23 to 35 years old) participated in this study. The subjects were informed about the experimental procedure and gave their consent.

H-reflex was evoked before, during and traction of the leg using a VikingQuest system electromyography unit (Nicolet Biomedical, Inc.). The subject's experimental posture was lying supine with the knee bent at a 90-degree angle.
The surface stimulating electrodes were applied to the skin over the course of the femoral nerve immediately distal to the inguinal ligament. Stimulus conditions were 0.5 ms square-wave pulse at a frequency of 0.2 Hz. The stimulus intensity that evoked the H-reflex was the threshold of the apparent M wave. The intensity where maximal M wave (Mmax) was recorded was supra-maximal. An Ag/AgCl active electrode for recording the H-reflex and M wave was placed on the muscle belly of the VMO, and a reference electrode was placed immediately proximal to the patella. The ground electrode was placed on the center of the thigh. The H-reflex was recorded 10 times, and the Mmax was recorded 1 time. H-reflex peak-to-peak amplitude was averaged, and the peak-to-peak amplitude ratio of H/Mmax was calculated.

RESULTS AND DISCUSSION

Figure 1 shows typical wave forms for the VMO H-reflex and M wave.

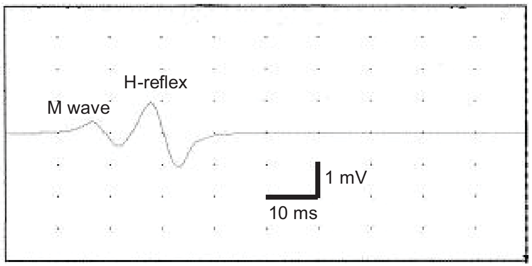

Figure1: This figure shows the M wave and H-reflex evoked by a stimulus intensity of 10.1 mA. Age of this subject was 28 years and the height was 168 cm.

Average values of the amplitude ratio of H/Mmax were as follows: 16.9 ±14.8 % (before), 12.5±13.3 % (during) and 9.9± 11.2 % (after). The amplitude ratio of H/Mmax tended to gradually decrease, but there was no significant effect of traction of the leg.

The decreased H-reflex amplitude during traction of the leg was caused by Ib inhibition of the VMO using the prolonged stretching of the quadriceps tendon and the patellar tendon. However, the cruciate ligaments of the knee (especially the posterior cruciate ligament: PCL), knee fibrous capsule, infrapatellar fat pad and so on were stretched during traction of the leg. Traction of the PCL led to activation of the quadriceps muscle in an animal study of the cruciate ligament reflexes (Krogsgaard MR, et al., 2002). Therefore, if the cruciate ligamentomuscular reflex can be applied to the present study, we postulate that the VMO H-reflex could be facilitated during PCL stretching. However, it is likely that H-reflex amplitudes were inhibited, because the quadriceps tendon and patellar tendons received prolonged stretching during traction of the leg.

H-reflex amplitude did not recover after the traction in this study. It is likely that afferent input from group III and IV fibers affect the alpha-motoneuron pool corresponding to the VMO. Alpha-motoneurons receive synaptic input from group III and IV sensory afferents and it has considered that alpha-motoneurons were inhibited by sensory afferent from group III and IV (Leonard CT., et al. 1994). Free nerve endings exist the knee fibrous capsule and infrapatellar fat pad. Free nerve endings sense the mechanical stress. The parent axon of free nerve ending was group III and IV fibers. And the sensory adaptation of free nerve endings was slow. From above the mention, it might be suggested that H-reflex amplitude did not recover after traction because the excitability of spinal neural function corresponding to the VMO remained inhibiting by the sensory afferent from group III and IV fibers due to the traction stress for knee joint.

CONCLUSIONS

From the result of this present study, it was likely that traction of the leg had the inhibitory influence on the excitability of the spinal neural function corresponding to the VMO. With regard to the inhibitory mechanism, it was considered that the inhibitory afferent input from Ib or III and IV fiber affected the alpha-motoneuron pool corresponding to the VMO.

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ANALYSIS OF THE ELECTROMYOGRAPHIC ACTIVITY AND THE FORCE OF KNEE EXTENSOR MUSCLES BEFORE AND AFTER TWO DIFFERENT ELECTROSTIMULATION TRAINING PROGRAMS

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INTRODUCTION

Neuromuscular electrical stimulation (NMES) is a therapeutic tool commonly used to restore sensorial and motor functions and to evoke muscle strengthening and hypertrophy (HOLCOMB et al., 2000). According to Low and Reed (2001), there are different kinds of NMES currents and they can be separated in low and medium frequency currents. An example of low frequency current is the functional electrical stimulation (FES). On the other hand, Russian current is a medium frequency example. In view of the foregoing, the aim of this study was to evaluate the electromyographic activity and the force of knee extensor muscles before and after two different electrostimulation training programs.

METHODS

Twelve sedentary women (23 ± 3,74 years), with no systemic or orthopedic diseases, gave written informed consent to participate in this study. Approval for the project was obtained from the Ethics Committee on Human Research of the institution where the study was developed (protocol number 07-07/008). Initially, the electrical activity of the vastus medialis (VM), vastus lateralis (VL) and rectus femoris (RF) muscles was obtained by a signal acquisition module, model EMG1000 (Lynx®), with impedance 10^9 Ohms, resolution of 16 bits entry band of ± 5V, and interface with a Pentium III microcomputer. Data acquisition and storage was carried out with Aqdados software (Lynx®), version 7.02 for Windows®.

The subjects remained seated on the Bonet table, with the hip joint at 90° and the knee joint at full extension. Before the electrodes were put into place, the skin was prepared, trichotomized and cleaned with 70% alcohol. To capture the action potentials, 3 simple differential, active surface bipolar electrodes (Lynx®) (20 times gain, IRMC > 100 dB, signal noise rate < 3 μV RMS and inter-electrode distance of 10mm) were used and placed according to SENIAM specifications. The channels were adjusted for a total gain of 1000 times, with band-pass filter of 20-1000 Hz (Butterworth type) and sampling frequency of 2000 Hz. The reference electrode was attached to the anterior tibial tuberosity of the analyzed leg. To measure the force (Kgf) of leg extension a properly calibrated load cell model MM-100 (Kratos®) was used. Signals were collected simultaneously in the 3 electrodes and in the load cell during maximum voluntary isometric contraction (MVIC) for 5 seconds, repeated 3 times and with an interval of 1 minute between them.

After the initial electromyographic evaluation, 3 experimental groups were formed: 1) Control (C), 2) FES electrostimulated group (F) (f=50Hz; T=300us and on/off ratio=5s); 3) Russian current electrostimulated group (R) (f=2500Hz modulated in 50Hz; on/off ratio=5s). For both electrostimulated groups, 4 silicon-carbon (10 X 4 cm), were placed over the belly of quadriceps femoris. The intensity of the electrostimulation was determined by the maximum tolerance of each subject. The electrostimulation programs lasted 5 weeks (3 sessions *per* week, each session lasting 30 minutes). After the end of the training programs each subject was revaluated following the procedures previously described.

Electromyographic signals were processed in specific routines developed in the software Matlab 6.5®, for the analysis of the Root Mean Square (RMS) in μV. Statistical analysis consisted of the Wilcoxon test performed for paired data (intra-group comparisons) and the Kruskal-Wallis test followed by Dunn post-hoc test (inter-groups comparisons). Level of significance was set at 5%.

RESULTS AND DISCUSSION

As to the leg extension force, we verify in figure 1 that there was a significant increase for both stimulated groups after 5 weeks of electrostimulation training programs. Regarding RMS index of VM, VL and RF muscles, there was a significant increase for F and R groups after electrostimulation programs (table 1). However, there was no significant difference between the stimulated groups.

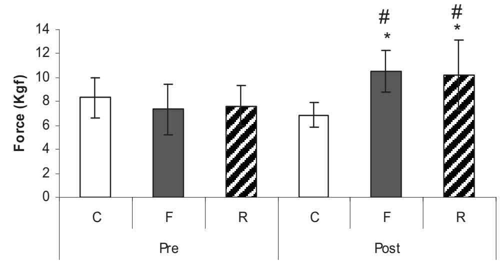

Figure 1: Leg extension force (Kgf) of the different experimental groups under pre and post-training evaluations, n=12. *p<0.05 in relation to the respective group in pre-condition; #p<0.05 in relation to C group in the same condition. Values expressed as mean ± standard deviation (SD).

CONCLUSIONS

Under our experimental conditions, we can conclude that both electrostimulated groups (FES and Russian) had increases of force and RMS index of quadriceps femoris after 5 weeks of training; however, there was no difference between them.

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Table 1: RMS index (μV) of VM, VL and RF muscles of the different experimental groups under pre and post-training evaluations. *p<0.05 in relation to the respective group in pre-condition; #p<0.05 in relation to C group in the same condition, n=12. Values expressed as mean ± SD.

	C		F		R	
	Pre	Post	Pre	Post	Pre	Post
VM	36.82±17.56	59.80±35.84	20.6±10.27	121.13±58.37* #	13.22±1.96	145.39±99,01* #
VL	33.93±23.00	55.32±18.87	15.60 ± 6.21	91.92±18.95* #	15.67±3.57	143.83±71.63 * #
RF	31.67±22.77	37.59±12.74	19.43±9.13	154.22±41.20* #	12.86 ± 4.39	100.17±51.98* #

EFFECT OF INDUCED ISCHEMIA ON ELECTROMYOGRAPHIC FREQUENCY

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INTRODUCTION

Among the surface electromyography (EMGs) parameters, the median frequency is the better to represent the changes into the lack of blood flow in the muscle, therefore it expresses the changes in the conduction velocity, changes in intra-muscular pH, and modification in the recruitment and synchronisation of the motor unit (Allisson e Fujiwara, 2002).

Muscular contractions in ischemic condictions have shown a decrease in the median frequency, due to the changes in the muscle fiber membrane properties (Farina et al., 2005).

The aim of this study was analyse the effect of induced ischemia on the electromyographyc frequency of wrist extensor muscles in healthy women.

METHODS

Thirteen women with median age of 25 years old, healthy and sedentaries, participated in this study.

A load cell MM-100 (Kratos®) was used to determinate the maximal voluntary isometric contraction (MVIC) and the EMG signals were obtained by a signal acquisition module, model EMG1000 Lynx® (São Paulo, SP, Brazil). Imput impedance was 109 Ohms, frequency sample rate of 2000 Hz and cutt-off frequencies were 20-1000 Hz).

Bipolar active surface electrodes, gain of 20, CMMR >100 dB, with simple differential configuration (Lynx® São Paulo, SP, Brazil) were placed with the longitudinal axis of the electrode aligned parallel to the length of the wrist extensor muscles fibers, acording to Cram et al. (1998). The reference electrode was fixed on the lateral epicondyle.

The EMGs was in the following situations: pre-ischemia (normal conditions of blood flow); ischemia (5 min after the blood flow interruption by a sphygmomanometer *Pressure N/C,* inflated in the dominant arm and confirmed by a Doppler Nicolet Vascular Versalab SE) ultra-sound; post-ischemia 1 (immediately after the blood flow back) and post-ischemia 2 (after 10 min from the ischemia starts). In each situations the EMG was registered during 15s of MVIC for three times.

The MATLAB 6.5.1 was used to EMG processing and the median frequency was analyzed in the full period (15s) and window (1-2s, 7-8s e 13-14s). The Friedman test was used with $p<0,05$.

RESULTS AND DISCUSSION

The 5 min of ischemia did not change the EMGs median frequency ($p=0,09$). In the window method, there was a significant increase of the median frequency to the ischemia for post-ischemia 2, in 7-8s period ($p<0,05$).

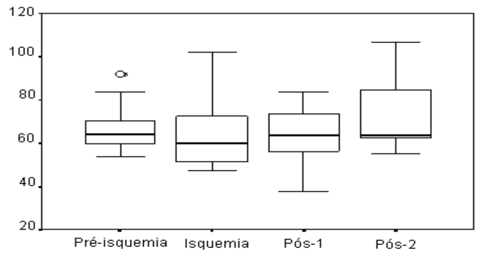

Figure 1: Median Frequency values (Hz) during MVIC pre-ischemia, ischemia, post-1 e post-2.

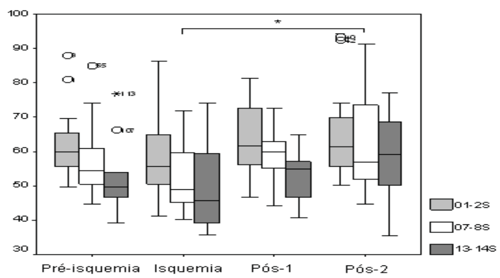

Figure 2: Median Frequency values (Hz) in 1-2s, 7-8s e 13-14s periods, in pre-ischemia, ischemia, post-1 and post-2.

The induced ischemia causes decrease in the velocity of propagation of the motor unit actions potentials (Murthy et al., 2001). However, the results of this study disagree with Zwarts et al. (1987) and Rongen et al., (2002) who observed a decrease in the conduction velocity in the biceps and brachiradial muscles in isometric contractions and 2 min ischemia, respectively. This disagreement can be cause by the predominancy of type I fibers in the muscles of this study (Fugl-Meyer et al., 1992), and in this case, it would be needed a longer time of ischemia to causes changes in the median frequency.

CONCLUSIONS

The 5 min of induced ischemia was not enough to cause a significant change in the median frequency of the wrist extensor muscles in healthy women.

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CONTROLLER OF ULTRASONIC MOTOR FOR REHABILITATION ROBOT

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INTRODUCTION

In rehabilitation, constraint induced movement therapy (CI) was introduced by Taub (Taub et al., 2002) and it achieve an effect. CI forces the use of the affected side by restraining the patient's unaffected arm in a sling. In order to use CI therapy, patients need to be able to extend their wrists and move their arm and fingers. Because the patients have to manipulate something to use their affected arm, in the case of limited hand movement, CI can not be used. However we may measure muscle activities without movement by electromyography (EMG). A robot which supports movement can be made by extracting the information about movement included in EMG and generating suitable power in actuators.

Although there is much information about movement which can be extracted from EMG, we focus our attention on joint stiffness and equilibrium point here. Joint stiffness can be estimated from linear sum of EMG which is rectified and filtered through appropriate low-pass filter (Osu and Gomi, 1999). Equilibrium point can be estimated from filtered EMG of flexor and extensor muscle using artificial neural network (Koike and Kawato, 1994).

We developed a controller for ultrasonic motor (USM) which receives stiffness and equilibrium point, which can be estimated from EMG, and viscosity from PC and controls USM using these parameters. In this paper, we report about this controller and discuss possibilities that this controller can be used for a rehabilitation robot by using USM as joint actuators and adjusting input parameters.

METHODS

To control torque of USM, a lot of methods such as adjusting the frequency, amplitude and phase-difference of two sinusoidal voltage waveforms power parameters have been proposed. Among these methods, we control the impedance of the robot using phase difference of USM.

Figure 1 shows no-load rotation speed of USM when phase difference is adjusted under several different frequencies. This shows that torque of USM changes roughly linearly along with the increase in phase difference because torque is nearly proportional to no-load rotation speed. Therefore, input phase difference ϕ is decided by a following expression:

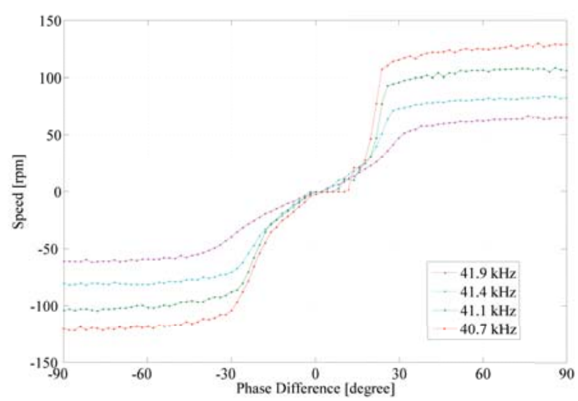

Figure 1: No-load rotation speed of USM versus phase difference under several different frequencies.

$$\phi = -K(\theta - \theta_d) - D\dot{\theta}$$

where K, D, θ_d, θ and $\dot{\theta}$ represent stiffness , viscosity, equilibrium point, angle and angle rate, respectively. Stiffness, viscosity and equilibrium point are input from PC, and angle and angle rate are calculated on the controller from output of the encoder attached to USM.

RESULTS AND DISCUSSION

We developed a controller of USM for controlling impedance by the above-mentioned method (Figure. 2). By receiving the value of stiffness, viscosity, and equilibrium points via USB, this controller calculates appropriate phase difference according to change of angle, and outputs it to USM. These parameters can also be updated while USM is working. This controller allows developers to control impedance of USM by simple programming.

Figure 2: Controller for USM (MSUMC02: Mizoue project Japan Corp.)

A robot arm whose joints are actuated by USM and controlled by this controller will be useful for rehabilitation. For example, if joint stiffness and equilibrium points of a patient are estimated from his/her EMG and USM are controlled using these values while he/she holds the end effector of the robot arm, this robot arm can assist the movement of the patient. This may reduce the patient's stress and motivate him/her to train. Moreover, this system will be useful for control of power-assist system and prosthetic arm using EMG for estimation of force information.

This controller can be conveniently used also for the rehabilitation which does not use EMG. For example, a patient moves a hand along the target trajectory shown visually, holding the end effector of a robot arm, and the system changes equilibrium points of the robot arm along the target trajectory. By changing impedance according to the stage of training, this system will improve his/her ability to move.

CONCLUSIONS

We developed a controller for USM which receives parameters of impedance and equilibrium point and controls USM using these parameters. This controller has possibilities for rehabilitation application.

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EVALUATION PROCESS FOR AGONIST AND ANTAGONIST MUSCLE ACTIVITIES DURING ACTIVE AND PASSIVE WALKING WITH LOKOMAT

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INTRODUCTION

In the recent decade, the effect of robotic-assisted gait training for patient of spinal cord injury (SCI) has been reported. The Lokomat (Hocoma, Switzerland) has been received attention as a robotic-assisted orthosis. This system is comprised of a treadmill and a harness for supporting the body-weight, and a robotic-assisted orthosis. Gait rehabilitation aimed at providing effective stimulation for the central pattern generator. Besides, active motor training is more effective than passive motor training in eliciting performance improvements and cortical reorganization (Lotze *et al*., 2003). Therefore, for incomplete SCI patients who keep some muscle activity, voluntary muscle contraction in the gait rehabilitation is expected to be effective. However, evaluation process of determining patient's voluntary muscle contractions has not yet been clear. In this paper, we verified muscle activities of agonist and antagonist muscles during active and passive walking with the Lokomat.

METHODS

After providing informed consent, one healthy male subject (21.9 yrs. old) with no neurological injuries or gait disorders voluntarily participated in the experiment. We first of all measured the maximum torque at the flexion and extension of the knee joint with the L-FORCE. The L-FORCE is attached at the Lokomat for measuring the muscle force during isometric contractions. The measured maximum torques were 101.4 Nm and 81.1 Nm at the flexion and extension, respectively. Normalizing the measured torque with the maximum torques during a 5-min walking, we set three target torques: 30%, 20%, and 10%. The guidance force of the Lokomat was tested at 100% and 50% at the walking speed of 1.5 km/h. During the Lokomat walking, as the visual feedback, the time-series of the target torque was presented at the front oscilloscope display in order to maintain the knee joint torque as stably as possible.

Total seven trials were tested, including active walking and passive walking at the 100% guidance force without the target torque. We used a wireless measuring system (Myomonitor IV, Delsys) in order not to impede the subjects gait pattern. We recorded surface EMG (SEMG) from the vastus lateralis (VL), biceps femoris (BF), gastrocnemius (GAS), tibialis anterior (TA), and soleus (SOL) muscles, using the active two-bar electrodes (DE2.3, Delsys). The goniometors (ShapeSensor S700, Measurand) were attached at the both legs to measure the knee joint angles. SEMG signals and knee joint angles were sampled at 2000 Hz with 12-bit resolution. Furthermore, the instrument embedded in the Lokomat orthosis recorded the hip and knee joint torques, and angles. These orthosis data were sampled at

2000 Hz with 16-bit resolution. For verifying the measured data at each stride on site, we developed the GUI by MATLAB (R2007a, MathWorks).

We determined each stride cycle using the knee joint angles and estimated a muscle activation period (MAP) in the total gait cycle (0-100%) at every stride. The MAP presents how long each muscle activates at one gait cycle.

RESULTS AND DISCUSSION

In general, the MAP increased with respect to the target torque. At the 100% assisted passive walking, the MAP of gastrocnemius muscle kept 40%, whereas it increased upto 60% at the 100% assisted active walking with the 30% target torque (Figure 1). Voluntary muscle contractions at the active walking with the 30% target torque were more active than passive walking. Besides, averaged MAP of some muscles increased with respect to the target torque and the assist of Lokomat orthosis (Figure 2). At the 100% assisted active walking with the 30% target torque, the averaged MAP of gastrocnemius muscle was 40%, whereas it increased upto nearly 80% at the 50% assisted active walking with the 30% target torque. Due to decrease in the assist, the subject required voluntary contraction in walking.

Accordingly, providing the appropriate target torque and the decrease in assist could promote voluntary muscle contractions. Moreover, MAP evaluated subject's voluntary contraction during active walking with Lokomat. Feed backing the MAP for patients in real-time, patients could know their own muscle efforts and raise motivation in the gait rehabilitation.

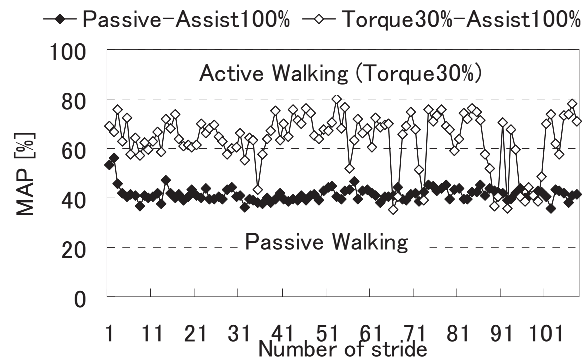

Figure 1: Time-series of MAP at the gastrocnemius muscle during passive and active walking.

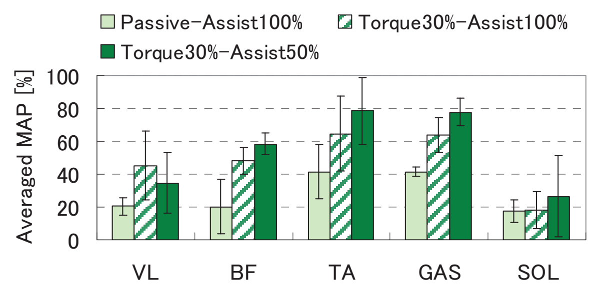

Figure 2: Averaged MAP at the 100% assisted passive walking, 100% assisted active walking with the 30% target torque, and 50% assited active walking with the 30% target torque.

SUMMARY/CONCLUSIONS

We evaluated voluntary contractions of agonist and antagonist muscles during active and passive walking with the Lokomat, using the muscle activation period in the total gait cycle. Averaged muscle activation period of some muscles increased with respect to the target torque and the assist of the Lokomat orthosis. Finally, further discussion is required for the variation in walking speed of the Lokmat (Hidler et al., 2004).

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REDUCTION OF FORCE POTENTIATION INDUCED BY RANDOMIZED FREQUENCY ELECTRICAL STIMULATION: POTENTIAL METHOD FOR THE PREVENTION OF MUSCLE FATIGUE

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INTRODUCTION

Functional electrical stimulation (FES) is one of the useful therapeutic methods to restore sensorimotor functions. FES involves electrically stimulating the neuromuscular system in order to generate a muscle contraction in paralyzed area [1]. However, it has been pointed out that a muscle contraction induced by FES tends to result in a rapid muscle fatigue [2].

It is well known that high frequency with larger amplitude stimulation causes extra contractions beyond the usual tetanic contractions of skeletal muscle within a few seconds of stimulation. For the application of neuroprothesis, such “force potentiation” may be an obstacle in effectively controlling skeletal muscle because the additional muscle torque due to potentiation interrupts linear relationship between stimulus intensity and generated muscle torque. In conventional FES methods, the stimulus parameters, such as pulse width, pulse frequency, and pulse intensity, is kept constant. This may cause certain motor units to become overworked while others remain completely inactive.

The aim of this study was to investigate the effects of FES using randomized stimulus frequency on the extent of force potentiation. Since the randomized stimulus more closely mimics the natural sensory signal during voluntary contraction [3], we hypothesized that the randomized electrical stimulation will yield a smaller force potentiation than the constant electrical stimulation method.

METHODS

Subjects: Twelve able-bodied subjects participated in this study (23-39 years).

Procedure: While subjects sat on a wheelchair, the electrical stimulation was delivered to triceps surae muscle with biphasic, bipolar pulse trains by a conventional stimulator (Compex Motion, Switzerland). A constant frequency stimulation (CON) was set to 40Hz, while the random frequency stimulation (RAN) was specified to have a mean frequency of 40Hz with a +/-15% uniform probability distribution. Both protocols used a 300µs pulse- width, and constant current amplitude. Each CON and RAN stimulation train lasted 20 seconds.

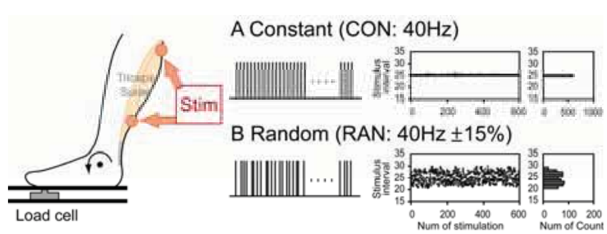

Figure 1: An illustration of the differences between constant (CON) and random (RAN) frequency stimulation protocol.

A series of test trains at varying stimulus intensities was applied to determine the input (stimulation) and output (torque) property. Stimulus intensity was increased until the subject felt uncomfortable or that the onset of cramping was imminent. The order of stimulation was randomized. At the end of the experiment, maximum voluntary force output was determined for each subject to normalize the torque output.

Data recording & analysis: Ankle planter flexion torque induced by FES was recorded by a transducer (LM-1KA, Kyowa, Japan) which was attached on the foot rest portion of the wheelchair. From the quantified FES-induced torque, an exponential relationship between stimulus intensity and torque output was calculated. The exponential index, tau (τ), was used to characterize the input-output dynamics. Paired t-test was used to compare τ between CON and RAN stimulation.

RESULTS AND DISCUSSION

Similar stimulus intensities resulted in decreased torque output during RAN in comparison to CON stimulation (Figure 2). τ value was significantly large in CON than RAN condition (Figure 3), suggesting that randomized frequency stimulation can reduce the extent of force potentiation.

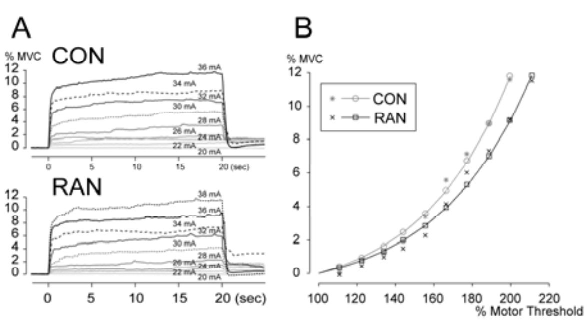

Figure 2 A: Typical example of the force profile during FES. **B:** An example of the exponential curve fits to the force data.

When delivering constant frequency stimulation, the induced muscle contractions are the result of the activation of the same motor units all the time. This causes certain motor units to become overworked while others remain completely inactive. Randomized stimulation may prevent for the recruitment of the same motor units. As the result, the extent of force potentiation was reduced.

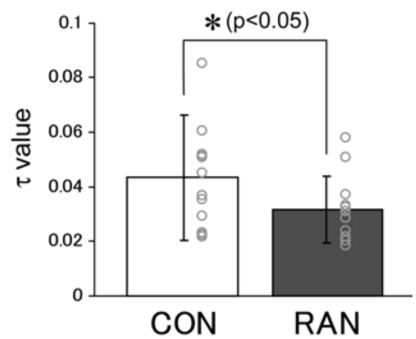

Figure 3: Comparison of τ value between CON and RAN stimulation protocol.

SUMMARY/CONCLUSIONS

We hypothesize that the use of randomized frequency stimulation reduces amount of muscle fatigue. As the result, convincing evidence was found that randomized frequency stimulation reduced force potentiation. This finding may help further efforts to develop a fatigue-resistant FES method

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DEVELOPMENT OF A ROBOTIC ARM FOR UPPER LIMB CONTROLED BY ELECTROMIOGRAPHY IN REAL TIME COMBINED WITH FES, FOR THE RESTARUATION OF THE FUNCTION TO THE UPPER LIMB IN PATIENT WITH SPINAL CORD DAMAGE AND STROKE

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INTRODUCTION

One of the most promising techniques in rehabilitation to recover the functions of the upper limb in peoples with damage in spinal cord and stroke is the used of a neural prothesis and FES.
The electrical activity of the remaning motor unit of the muscles can be used to control a mioelectrical device, as a mioelectrical prothesis or exoskeleton that allows to replace or help a lost movement (Bitzer et al-2006).
The advances in bioengineering have reached more sophistacated prothesis for amputaded and paralytic subjects. The control of this devices required a procesing and classifying mioelectrical signals in real time (Crawford B., Miller. K. - 2005).

METHODS

The research model is classified as applied investigation, since in a first stage a electromechanical system will be develop to help the movement, and in second stage the impact in the process of rehabilitation will be evaluated. The sample corresponds to patients with spinal cord damage of the cervical segment and suffering stroke patients that belongs to Hospital del Trabajador Santiago, Chile. The study group will include five persons (n=5), who must acomplish with the criteria of the investigation and sign an informed assent. The subject must have the following criteria of incorporation:1) Medical Diagnose in stroke or cervical spinal cord damage of minimum of six month of evolution; 2) Medical Stability. The following subjects will be excluded: 1) severe medical pathology that affects functionality of the upper limb. 2) Trofic alteration of the skin evident in the zone of a aplication of electrotherapy; 3) Level of Spasticity over three, according to MAS; 4) Presence of neurogenerative pathologies; 5) Incapacity to understand the information of this research and or following orders.
The study consists of the following variables: Functionality of the upper limb measured, tone of the flexo-extensor musculature of the wrist and elbow, measure with H-reflexes and MAS (Bakheit, A., Maynard, V. - 2003); ROM of wrist, elbow and shoulder, measured by eletrogoniometry; diameter of the muscle fiber measured with ECO.

Materials and Equipment:

The robotic arm is formed by support for an arm and forearm, with a motorized axis aligned to the articulated axis elbow. The HITACHI servo engine of high performance is control with the EMG of the musculature of biceps and triceps brachii.
The EMG is doubled differential Delsys Amplifier and is acquired by a National Instruments card of 32 analog input, 4 analog output, 16 I/O digital and 16-bit, processed in real time by algorithm of control design in the software IGOR PRO 5.04b, de serbo

control will be formed pulse width modulation (PWM).
The control of opening and closing the hand will be activated by FES. The control of FES will depend on the model of movement defined and control by different sensors like: electrogoniomether of the elbow and wrist and accelerometer in the arm and forearm. The constraction control will be in PWM variable between 20 uS and 500 uS square biphasic waveform will be used with a frequency of 40 Hz and evuantually pulse trains to increase the torque.

Evaluation:
The five subjects will have three evaluation: at the beginning, at the end of traning period with the neuroprosthetic in laboratory, and at the end of the intervention, after the period of using at home. This evualtion have the following aspects: neurokinesic that includes wrist and elbow muscle tone (MAS) and ROM of wrist elbow and shoulder; measurement of the level of spasticity (H-reflex); measurement of diameter of the muscle fiber of the flexoextensors of the wrist and elbow.

Procedure:
The subjects were divided in two groups: those who have a part of the grasp function precerved (group A) and those who can not voluntarily contract any muscle below the elbow (group B). The subjects selected (n = 5) will be cited to attend a therapy in a laboratory during 20 sessions, each sessions of 45 minutes, realizing the tasks of the work base with the aplication of the neuroprosthetic.
After this the patient will take the neuroprosthetic to his/her house, continuing with the treatment realizing the DLA with this device.This stage of the research will last for a month .

DISCUSSION
This study is in the stage of development of the device, stimating that in march clinical tests will begin.

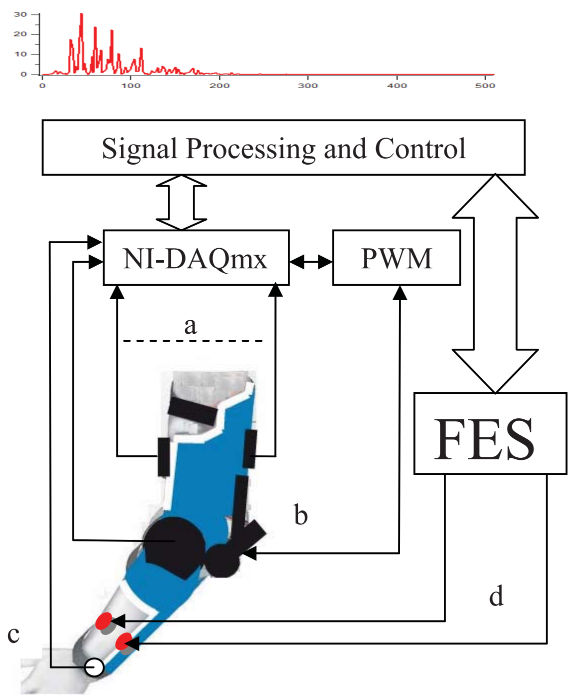

Fig.1: a)Delsys EMG Amplifier b)Servo Engine c) Electrogoniometer d) Electrostimulator

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PATHOLOGIC NORMOREFLEXIA DUE TO COEXISTING UPPER AND LOWER MOTOR NEURON CONDITIONS; TWO CASE REPORTS

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INTRODUCTION

Reflexes maintain many functions in the nervous system, among them, normal musculoskeletal function. Clinicians generally rely upon normoreflexia as one of several findings indicating normal neurological function. The deep tendon reflex (DTR), a stretch reflex, is a simple reflex involving the spinal cord. Normally, it represents a balance between inhibitory and stimulatory inputs. In upper motor neuron (UMN) syndromes, disinhibition or facilitation can produce hyperrflexia. Conversely, lower motor neuron (LMN) conditions inadequately stimulate the reflex arc, producing hyporeflexia.

Cases are often seen in which normoreflexia is observed, not as a sign of neurological health, but rather as the additive sum of two distinctly different and abnormal conditions. We present two cases in which normal reflexes were obtained in patients with known or anticipated LMN dysfunction. These unexpected findings led to the diagnoses of concomitant UMN pathology.

CASE PRESENTATIONS

A 71 year-old female was seen for progressive leg weakness and balance impairments. Medical history was significant for long-standing Type II diabetes and scoliosis. In addition to proprioceptive deficits, physical exam revealed 2+ patellar and Achilles reflexes, Babinski signs, and clonus. Electrodiagnostic studies demonstrated bilateral sensorimotor peripheral neuropathy, consistent with her diabetic history. MRI of the spine demonstrated severe dextroscoliosis and spondylosis with central stenosis.

Case 2 involves a male, age 56, with a 43-year history of Type I diabetes; he presented with worsening pain in the neck and hands. Sensory deficits were noted with a stocking-glove distribution. DTRs at the knees and ankles were normal, with positive Babinski signs. NCV testing revealed sensorimotor peripheral polyneuropathy and bilateral carpal tunnel syndrome. Cervical spine MRI demonstrated mild cord compression via spondylosis and disc disease.

DISCUSSION

DTRs aid clinicians in identifying and assessing disease. Our normoreflexic diabetic, neuropathic patients were found to have abnormal spinal anatomy causing compression/irritation of the spinal cord with associated findings of UMN dysfunction. With myriad causes for hyper/hyporeflexia, the physician must be cognizant that normoreflexia does not automatically indicate normal neurological function and that in the setting of anticipated abnormal reflexes, normoreflexia mandates further evaluation.

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THE EFFECT OF ELECTRODE DISLOCATION ON SURFACE ELECTROMYOGRAPHY AMPLITUDE: IMPLICATIONS FOR AMBULATORY MEASUREMENTS

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INTRODUCTION

The major muscle group of the lower back, the erector spinae, has extensively been investigated using surface electromyography (SEMG) in different fields, including ergonomics, sports and rehabilitation medicine. Although laboratory experiments have contributed to a better understanding of the functioning of this lower back muscle, SEMG measurements outside the lab may reveal relevant information on its use during activities of daily living. For example, this may be relevant for diagnostics and therapeutic interventions in lower back pain.

To allow ambulatory measurement of the lower back muscles, a belt with integrated SEMG electrodes has been developed. A prototype is shown in Fig. 1. A holder with two pairs of 'dry' electrodes (one pair for each side) is fixated on the skin using an elastic belt. The distance between the two electrode pairs can be adjusted individually. Long term measurements have shown that SEMG can be measured with a low artifact rate (7%), a good electrode fixation and an acceptable wearing comfort. The belt is developed in such way, that placement can be performed by the user himself. However, a potential disadvantage may be that the electrodes are not always positioned at the proper position. Particularly amplitude related parameters, such as the root mean square (RMS), may be affected by the induced variability in electrode positioning (Hermens and Vollenbroek-Hutten, 2004).

The aim of the current study was to determine the effect of electrode dislocation on the RMS of the lower back muscles, and to discuss its implications for an ambulatory measurement set-up involving the belt.

Figure 1: Belt with integrated electrode holder.

METHODS

Bipolar SEMG of the erector spinae was simultaneously measured at the recommended electrode site (BC) – determined with respect to bony landmarks by using palpation – and at cranial, distal, and lateral dislocations (AB, CD and EF respectively, Fig. 2) in 16 healthy subjects during five functional tasks (standing, forward flexion, re-extension, unsupported sitting and arm/leg lifting). Five trials were performed and results were averaged. The ratio of the RMS at the dislocated electrode sites divided by the RMS at the recommended site BC (RMS_R) was calculated to show the relative effect of electrode dislocations on the RMS.

RESULTS AND DISCUSSION

The mean values for RMS_R are plotted in Fig. 3. On average, the RMS for EF was 18% lower ($p<0.001$) than for BC. No significant differences were found for AB ($p=0.78$) and CD ($p=0.19$).

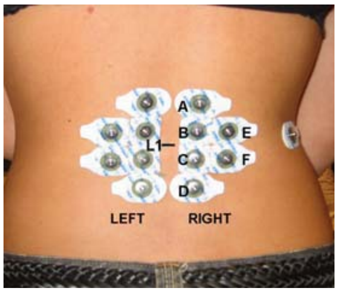

Fig 2. Electrode positioning. The recommended electrode pair BC was placed laterally to the L1 spinal process.

Electrode dislocation in the longitudinal direction seems to have a minor effect on the RMS, whereas dislocation in the medio-lateral direction reduces the RMS significantly. From the first experiences with the belt, placement of the electrode holder in the medio-lateral direction appears to be easy and unambiguous. Furthermore, the distance between the two electrode pairs can be fixated. So, lateral displacement of the belt seems not to be applicable. On the other hand, exact placement at the longitudinal position is difficult, and requires clinical expertise. However, based on the results of the current study, possible longitudinal dislocations will not affect the RMS.

The belt seems to be suitable for ambulatory SEMG measurements, and is currently being used in a study investigating muscle activation patterns in chronic lower back pain patients during daily activities. The knowledge gained in this study might serve as a base for future myofeedback applications.

SUMMARY AND CONCLUSION

The effect of electrode dislocation on the RMS of the low back muscles was studied. Dislocation in the medio-lateral direction reduces the RMS by 18%, whereas longitudinal displacement has a minor effect on the RMS. Based on the results of this study and the first experiences with a prototype belt, ambulatory registration of low back muscle activity seems to be feasible.

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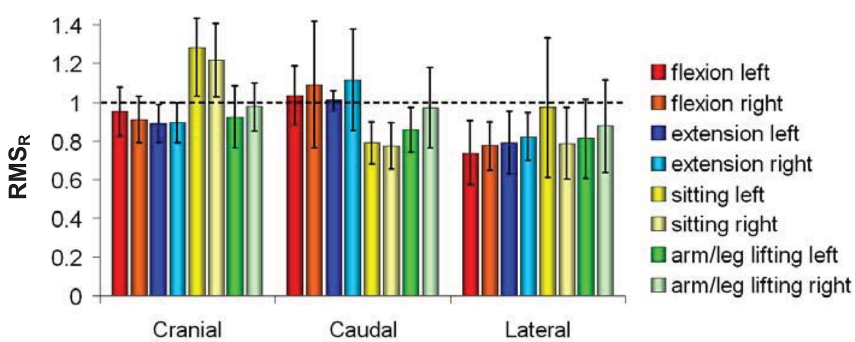

Fig 3. Mean RMS_R for the three simulated dislocations (cranial, caudal and lateral) and standard deviation. Each bar represents the grand average of RMS_R for a specific task and side across five repeated trials per subject over all subjects. For standing, no reliable RMS_R could be calculated due to low RMS values.

ON THE USE OF ELASTIC STRAPS TO IMPROVE GONIOMETER MEASUREMENTS

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INTRODUCTION

Flexible goniometer has been used to record different functional movements. It is a practical, portable and relatively inexpensive device. However, some sources of error, i.e. soft tissue artifact, may decrease its accuracy.

One possible application for the goniometer is to measure knee angles during gait in patients presenting ligament injuries and functional instability. Though, adequate measurement requires precise and accurate equipment and procedures. Considering that tissue artifact can be minimized by the restriction of movement between tissue and goniometer, and there is no available data on this topic in literature, we hypothesize that the use of elastic straps over the goniometer endblocks may decrease soft tissue artifact.

Thus, the objective of this study was to investigate how elastic straps influence knee angles during gait.

METHODS

Sixteen subjects, without any knee disorder, were evaluated. These subjects agreed to participate and signed an informed consent form. Subjects mean age was 30±8 years old, mean height was 177±7 cm and mean mass was 85±10 kg.

A flexible goniometer and torsiometer (M110, Biometrics Ltd, Gwent, UK) were used to record knee angles in three planes of motion (flexion/extension; varus/valgus and medial/lateral rotation).

Subjects walked on a treadmill at 5.0 km/h in two situations: 1. with elastic straps over the goniometer endblocks and 2. without these elastic straps.

Data were descriptively analyzed through calculation of the difference curve between both conditions (condition 1 – condition 2) for each subject. These data were summarized by calculating the mean curve, and 95% confidence curves for each condition, obtained for each instant of gait cycle as mean angle ± 1.96 x standard deviation.

RESULTS AND DISCUSSION

Curves of difference are presented in Figure 1, for the right and left knees.

Figure 1 shows that the confidence area is greater above 0° for flexion/extension and varus/valgus movements. It indicates that the angles obtained with the straps are higher than the angles obtained without elastic straps.

Moreover, the difference curve is higher between 70 and 90% of gait cycle, which is coincident with the swing phase. It is expected that the soft tissue artifact effect during swing gait phase is more evident since lower limb velocity is greater.

Variations were higher for the right knee. It can be related to the lateral dominance of the subjects, since the majority of them had right dominance.

For rotation movement, confidence curves are wider, indicating also greater variability.

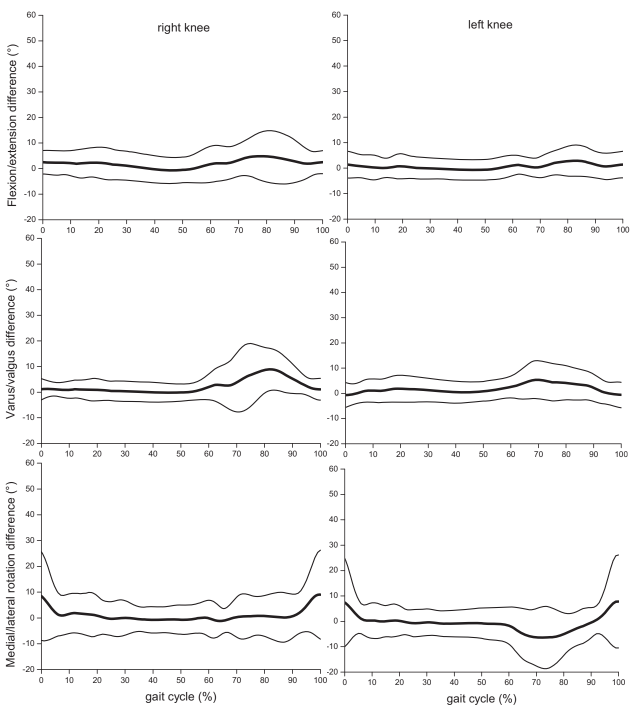

Figure 1: Difference between curves for three planes of motion.

Rowe et al. (2001) have also suggested the use of straps, but they have not presented results from these comparison. The authors argued that elastic straps apply light pressure between the goniometer endblocks and the skin, helding the goniometer in position and reducing the effect of soft tissue movement.

One possible disadvantage of the use of these straps could be the restriction of the joint movement, leading to a decrease in the knee range of motion. To minimize that, the researcher should not apply strong pressure on the straps. Care also needs to be taken during long data collection, in order to avoid the contact between straps and goniometer spring, which could lead to signal interferences.

SUMMARY/CONCLUSIONS

The use of elastic straps is recommended to reduce tissue artifact and achieve accurate measurements of knee movements using a flexible goniometer during gait.

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ACTIVATION OF THE PREFRONTAL CORTEX DURING LEARNING BY MOBILE SOFTWARE

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INTRODUCTION

Recently, because of the underlying software technology the computer can be used for teaching and learning basic subjects. The purpose of this study is to investigate the effect of mobile learning software on the prefrontal cortex and the change of the cerebral activation patterns during learning process.

METHODS

Non-invasive measurement of regional cerebral activity was estimated by measuring the oxygenated hemoglobin and deoxygenated hemoglobin of both sides of the prefrontal region using multi channel near-infrared spectroscopy . 9 healthy high school students volunteered as subjects for the 1st experiment and 2 elementary school students and 1 adult participated in the 2nd experiment. The 1st experiment was a comparison between mobile learning and paper & pencil learning. The software being used is Eitango Target 1900 DS which is Nintendo DS (mobile learning) software aimed at not only elementary school children, but also adults, making it perfect for beginner students of English. It looks like fun. In the case of using Nintendo DS, we write sentences we could hear directly on the screen of DS, not type a keyboard. It has the character-recognition system and a judgement test of the level of English. The 2nd one was a comparison among expert and trainee and beginner for computerized calculation drill machine in which "*Hyakumasu Keisan*" (One Hundred Measure Calculation) is performed as a practice of matrix calculation.

RESULTS AND DISCUSSION

In the 1st experiment, a greater increase in oxygenated hemoglobin was observed in the bilateral prefrontal cortex in 6 out of 9 students during mobile learning than paper & pencil learning (Fig.1). Moreover, after 3 weeks-relearning the prefrontal activation was inclined to be decreased. In the 2nd experiment, an expert student showed prefrontal deactivation and rather activation of parietal lobe. Conversely, trainee and beginner showed respectively bi-prefrontal and left prefrontal activation (Fig.2).

These results confirmed that mobile learning software activates prefrontal cortex in early stage and once skillful learning is well established, prefrontal activation tends to be reduced. Full automaticity marks the performance of the expert. Research on changes in brain activity from beginner to skilled performance has been consistent with this behavioral characterization, showing that a highly practiced skill often requires less prefrontal activation than before practice. In addition, mobile learning software is an advantage that the learner can study with continued interest by adding the game.

Figure 1: Prefrontal activation during dictation of English words.

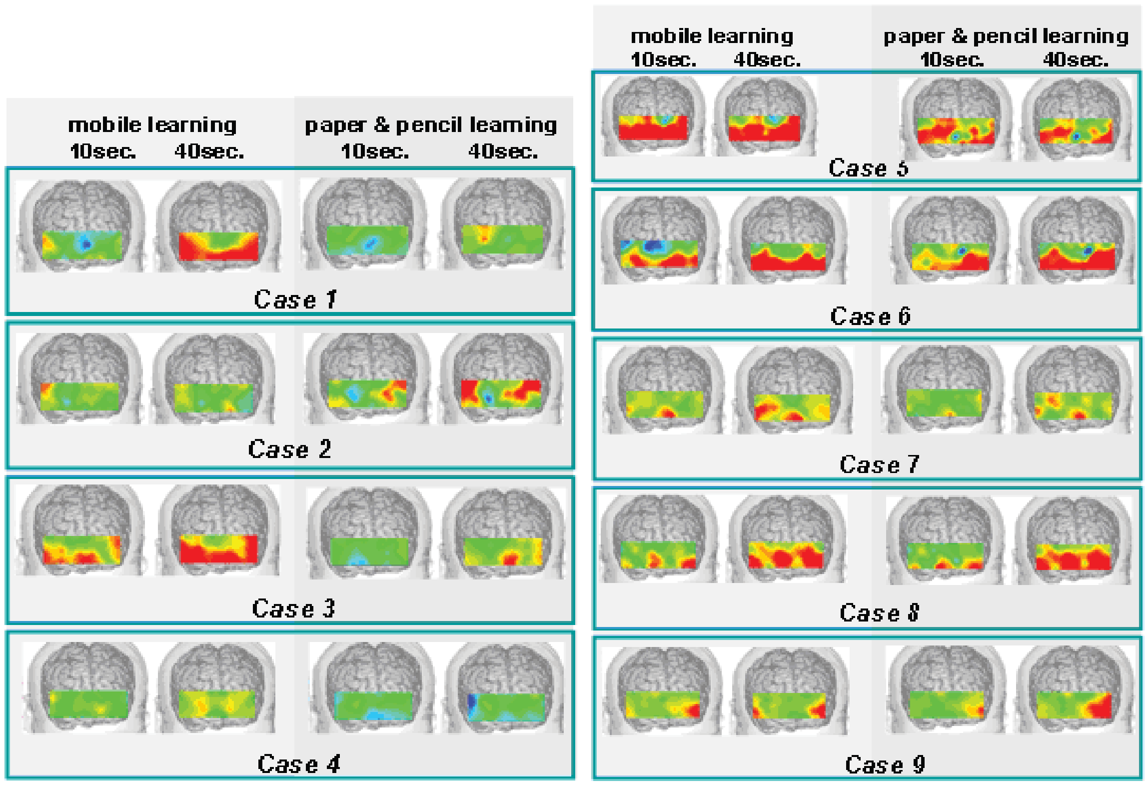
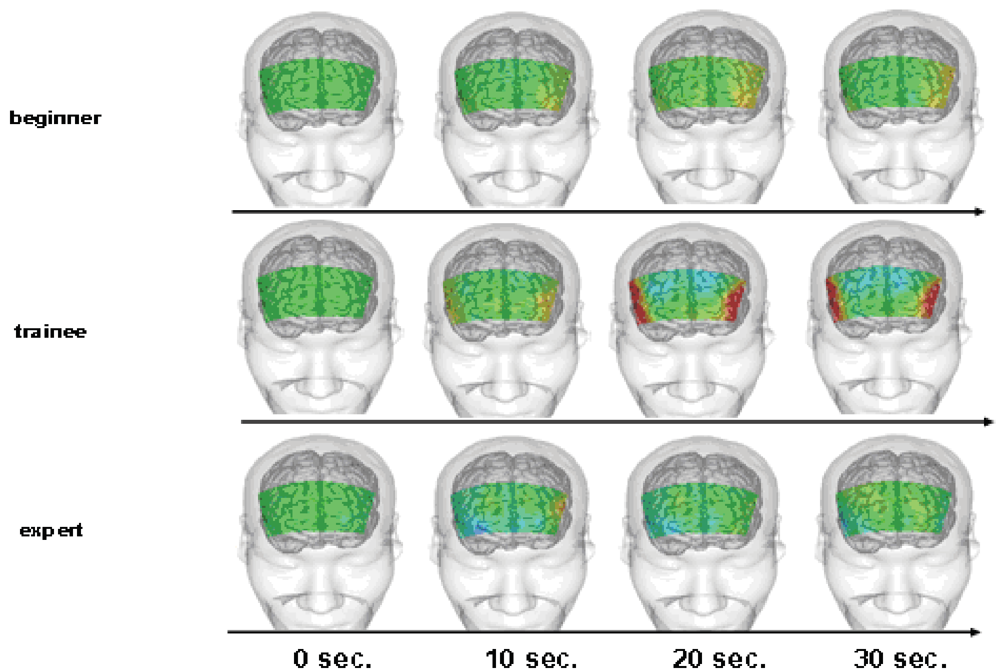


Figure 2: Prefrontal activation during calculation with computerized drill machine.



A METHOD FOR BETTER POSITIONING BIPOLAR ELECTRODES FOR LOWER LIMB EMG RECORDINGS

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INTRODUCTION

Traditionally, sEMG sensors are placed over muscle belly or over the motor endplate zone since it was the easiest location to record robust sEMG signals (Hermes et al, 2000). However, more recent studies (Merletti et al., 2001) clearly indicate that the sEMG pattern over the motor endplate and over innervation zone (IZ) is not very stable and reproducible and substantially alter sEMG variable estimation. In order to identify the IZs of muscles, a new and effective type of electrode has been developed (array electrode). But in many developing countries, like Brazil, the access of this type of technology is not easy and researchers have still been using single pair of electrodes. Aware of the importance of identifying the IZ of muscles in order to capture the best EMG signal quality, but at the same time still using single pair and not array electrodes, the purpose of this study was to identify the best muscle location to acquire sEMG over lower limb muscles using a single pair of electrodes.

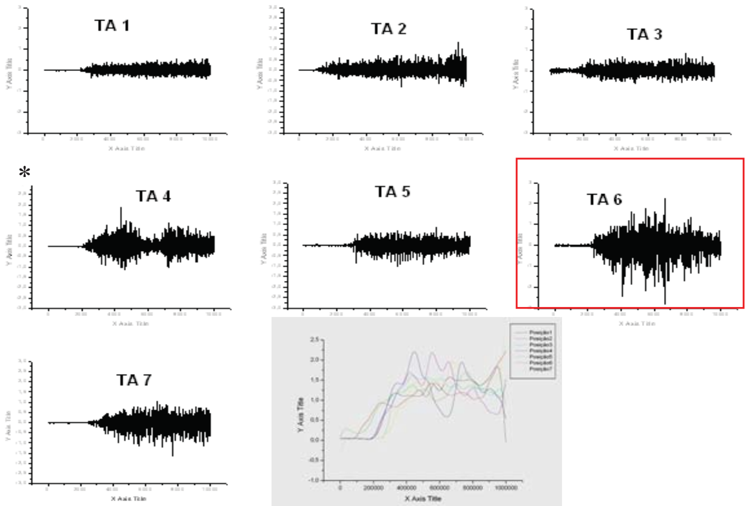
METHODS

Eight healthy females with age between 18 and 25 years (22.7 ±1.6 yr) participated in this study. Myoelectric signal were acquired in a single differential mode from each muscle using circular bipolar adhesive electrodes of Ag/AgCl (ϕ=10mm, inter-electrode distance=25mm). EMG signals were passed through a 10–500 Hz bandwidth filter, amplified (gain=1000), sampled at 2000 Hz. The electrode positions selected of each muscle started with assessing the SENIAM recommended site (Hermes et al, 2000) and then, other positions were located along the length of muscle up and down the SENIAM site with a distance of 25mm until proximal and distal tendon termination according to information available from anatomical atlas and manual palpation during resisted contractions. The motor point was determined by a universal pulse generator and the IZ, in accordance to Rainoldi et al. (2004).

Isometric contractions were acquired and obtained against manual resistance sustained during 5s for each position of electrodes (proximal to distal) in the following muscles: tibialis anterior (TA), vastus lateralis (VL), peroneus longus (PL) and gastrocnemius medialis(GM). Linear envelopes of each position tested for each muscle were determined followed by integral calculus in a 400ms window. The integrated EMG value, the position of the IZ and motor point, qualitative evaluation of the raw signal (density) and linear envelopes were compared in each electrode position. The best electrode location was determined by the following criteria: the site in which the integrated EMG value was higher, far from IZ and motor point (approximately 500 mm), and the site in which the density of raw signal were higher. It was selected the best position of each muscle in each subject related to each muscle length.

RESULTS AND DISCUSSION

VL and PL best site were 66% and 25% of muscle length, respectively, which correspond to SENIAM's location previously described.



Position	Integral
1	68480
2	58868
3	62291
* 4	69313
5	73715
6	**74218**
7	73986

Figure 1: Seven electrode positions of TA, its linear envelopes and EMG integrated values of each position. The best signal was at position 6. *represents SENIAM electrode position.

TA presented the best signal at 47.5% of muscle length, what disagreed with SENIAM location, which recommends the acquisition in 33% of muscle length.GM best site was 32% of muscle length. SENIAM does not have a specific location for its signal acquisition; they just recommend that the location for electrode positioning should be over the most prominent site of the muscle.

Table 1: VL length, best electrode location related to SENIAM recommendation site.

Subjects	VL length	Position of best signal (absolute)	Position of best signal (% VL length)
1	46	SENIAM	66
2	42	2 above	71
3	46	1 below	54
4	44	2 below	77
5	44	1 above	60
6	47	SENIAM	66
7	52	1 below	71
8	46	SENIAM	66
		Average (sd)	66.4 ± 7.1
		median	66

Table 2: PL length, best electrode location related to SENIAM recommendation site.

Subjects	PL length	Position of best signal (absolute)	Position of best signal (%PL length)
1	32	SENIAM	25
2	31	SENIAM	25
3	34	SENIAM	25
4	33	1 below	32
5	32	SENIAM	25
6	32	1 below	33
7	33	SENIAM	25
8	36	1 below	32
		Average (sd)	27.8 ± 3.8
		median	25

SUMMARY/CONCLUSIONS

The SENIAM recommendations for electrodes location in lower limb muscles did not show the position of the better signal in all muscles studied and it's necessary to adopt new positions for tibialis anterior and gastrocnemius medialis in order to guarantee better EMG signal acquisition far from motor point and IZ.

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MUSCULAR EFFECTS OF POSTERIOR CROSSBITE TREATMENT

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INTRODUCTION

Posterior crossbite constitutes a transversal discrepancy, which is observed in 18% of Brazilian children in the mixed dentition (Capelozza Filho, Silva Filho, 1997).

Rapid maxillary expansion (RME) is a widely accepted procedure and is recommended to correct maxillary constriction related to posterior crossbite.

The early posterior crossbite treatment is important to provide symmetrical and normal grow and prevent craniomandibular disorders. However, studies dealing with masticatory muscles in children with posterior crossbite are infrequent, especially regarding the long-term outcome of early treatment. (De-Rossi et al, 2008)

The aim of this prospective study was to analyze the muscular effect of skeletal crossbite treatment in the mixed dentition.

METHODS

The sample consisted of 27 children, of both gender, aged from 7 to 10 years, presenting skeletal posterior crossbite and requiring treatment with rapid maxillary expansion. The treatment was performed with the bonded rapid maxillary expansion appliance, used during 4 months after the active phase. After that period, the appliance was removed and patients wore the removable retention for at least 6 months.

The electromyographic (EMG) activity of right and left masseter muscles (RM and LM) and right and left temporalis muscles (RT and LT) was analyzed before treatment (T1), four months (T2) and 12 months after treatment (T3). EMG activity was evaluated during habitual chewing (10 sec) using the MyoSystem - Br1 electromyographer, with differential active electrodes (silver bars 10mm apart, 10mm long, 2mm wide, 20x gain, input impedance 10GΩ and 130dB CMRR). The EMG signal was analogically amplified with gain of 1000x, filtered and sampled by 12 bits A/D convert board with a 2KHz frequency.

The RMS values of habitual chewing were normalized by RMS of maximum voluntary dental clenching and the data were statically analyzed using the GLM repeated measures analyses (SPSS 10.0 software).

RESULTS

Normalized RMS EMG values of each muscle evaluated at T1, T2 and T3, along with results of statistical analysis are found in Table 1.

Comparing T1 and T2, the EMG activity of RM, LM, RT and LT increased. However, the difference was statistically significant only for RT and LT (p<0,05). At T3, the EMG values were closed from initials ones and there were no statistical differences for any muscles between T1 and T3.

CONCLUSIONS

Considering the specific conditions of this study, it can be concluded that posterior crossbite treatment increased the muscular activity of masticatory muscles in a short-term, however in a long-term the differences were not significant. These findings suggest the association between dental stability and functional adaptation and show the importance of the use of the contention until the muscles have adapted to the new occlusion relationship.

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ACKNOWLEDGEMENTS

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Table 1: RMS normalized average, standard deviation (±) and statistical significance of electromyographic activity during habitual chewing at T1, T2 and T3 (n= 27)

Muscle	T1	T2	T3
Right Masseter	52.19±31.10	95.68±128.54	71.97±46.72
Left Masseter	59.78±32.12	93.88±94.13	67.78±48.61
Right Temporalis	48.85± 23.56*	63.85± 31.97*	60.74±32.69
Temporal Esquerdo	56.95± 23.76†	74.41±43.61†	61.15±29.10

*p=0.038
†p= 0.034

ON THE USE OF EMG AMPLITUDE RATIOS TO ASSESS THE COORDINATION OF BACK MUSCLES

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INTRODUCTION

The electromyographic (EMG) assessment of back muscle coordination in chronic low back pain (CLBP) patients might be flawed when amplitude normalization is performed using EMG reference values requiring maximal voluntary contractions. EMG amplitude ratios (EMG-ratios) between muscles have been used to circumvent this problem. However, other potential confounders [force level, contraction type, subcutaneous tissue thickness (STT) distribution] might influence EMG-ratios. Also, their reliability as well as their sensitivity to gender and back pain has rarely been assessed. The aim of this study was to challenge the use of EMG-ratios for the assessment of back muscle activation patterns in CLBP patients.

METHODS

Healthy subjects (44 men and 13 women) and CLBP patients (57 men) performed three 7 s static ramp extension contractions ranging from 0 to 100% of the maximal voluntary contraction (MVC) while standing in a static dynamometer measuring L5/S1 moments (Larivière et al., 2005). A subgroup of 20 healthy men also performed 5 s step contractions at 10, 20, 40, 60 and 80% MVC. To assess reliability, another subgroup (n = 20 healthy and 20 CLBP men) performed the protocol 3 times, on different days. Surface EMG signals were collected from 4 pairs of back muscles (4 electrode sites : at L5, L3, L1, T10 vertebral levels) (Larivière et al., 2005). STT was also measured at these electrode sites (right side), using a skinfold caliper. EMG RMS amplitude values (250 ms) were computed at each 5% force level from 10 to 80% MVC. Then, relative (EMGRatioR) and absolute (EMGRatioA) EMG-ratios (Reeves et al., 2006) were computed between different electrode sites (L5/L3, L5/L1, L5/T10, L3/L1, L3/T10, L1/T10) and averaged bilaterally. Finally, the corresponding ratios were also computed using the different STT measures (STTRatioR and STTRatioA) to use them as covariates in statistical analyses.

RESULTS AND DISCUSSION

EMGRatioR and EMGRatioA were affected by the force level (10 and 20 %MVC different from 40, 60 and 80 %MVC) and the contraction type (step vs ramp contractions). Post hoc comparisons revealed that in nearly all cases, EMG-ratios at 10 and 20% MVC were alike, as well as EMG-ratios at 40, 60 and 80% MVC. In other words, EMG-ratios significantly changed in the transition from low to moderate force levels. The Pearson correlations between the different EMG-ratios and the L5/S1 extension moment (M_{ext}) during the ramp contractions ranged, on average, from 0.36 to 0.70.

Considering the effect of force level, two methods were used to characterize the relationship between the EMG-ratios and the

M_{ext} (in Nm): (1) slope of the linear relationship with M_{ext} (RatioR_s and RatioA_s), (2) the mean value across the force levels that were over 75 Nm (RatioR_m and RatioA_m). The reliability of these variables ranged from moderate to excellent (ICCs between 0.50 and 0.91).

ANCOVAs were performed to compare 12 men to 13 age-matched women (GENDER factor) and to compare 24 healthy men to 57 CLBP men (PAIN factor), using mass, height, STTRatioA or STTRatioR as covariates. Among the 24 possible EMG-ratios, 14 have shown a significant GENDER effect while only one has shown a PAIN effect. The STTRatio covariates were significant, mainly for RatioR_m. In fact, Pearson correlations has revealed statistically significant association (r = -0.38 to -0.57) between some EMG-ratios and STT-ratios.

Factorial correspondence analysis followed by ascendant hierarchical clustering was carried out to define subgroups among the 24 healthy and 52 CLBP men, using the EMG-ratio variables as predictors. Subgroups (2, 3 or 5) were identified but none of the main clinical variables (pain status, pain intensity, Oswestry disability score) were able to clearly characterize them, though fear avoidance beliefs and SF-12 have shown some small statistical effects.

SUMMARY/CONCLUSIONS

The influence of contraction type showed that EMG-ratios cannot be compared, at least at low force levels (10 and 20% MVC), when computed from ramp and step contractions. EMG-ratios are comparable from each other only at force levels over 40% MVC. It appears that small changes in the activity of the muscles used to calculate the EMG-ratios may have an important effect at low force levels where the signal-to-noise ratio is lower. This might explain the high variability observed in some EMG-ratios at low force levels and why it might not be recommendable to study EMG-ratios at low force levels.

According to the correlation and ANCOVA results, it appears that the distribution of STT under the electrode sites of back muscles is different across subjects and affects EMG-ratios unevenly across individuals. It is thus inaccurate to consider that EMG-ratios circumvent the problem of EMG normalization. It is thus important to adjust the EMG amplitude values for STT attenuation before computing EMG-ratios (Edgerton et al., 1997).

Men and women used different back muscle coordination patterns, as nicely depicted by EMG-ratios. The fact that the PAIN factor was not significant (in general) may indicate that univariate analyses are not well suited to depict different coordination patterns that may exist across different CLBP subgroups. However, multivariate analyses were able to identified subgroups. Even though the available clinical variables were not able to clearly characterize these subgroups, these results suggest that some relevant physiological information associated with the underlying pathophysiology may still be present in EMG-ratio variables.

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EVALUATION OF THE FATIGABILITY OF THE CERVICAL EXTENSOR MUSCLES WHILE DOING THE NECK EXTENSOR ENDURANCE TEST IN PATIENTS WITH MYOGENOUS TEMPOROMANDIBULAR DISORDERS AND HEALTHY SUBJECTS: PRELIMINARY RESULTS

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INTRODUCTION

Some clinical evidence of interconnection between cervical spine and temporomandibular disorders (TMD) has been demonstrated and thus Cervical Spine Dysfunction (CSD) has been associated with TMD. Some current experiments in animals have studied the relationship between the craniofacial region and cervical structures. Yu et al. (1995) found that the irritation of the temporomandibular joint (TMJ) caused an increase in the activity of the masticatory muscles as well as the activity of cervical muscles. In addition, Hellstrom et al., (2002) demonstrated that bradykinin injected into the TMJ, changed the sensitivity of muscle spindles in cervical muscles. These authors concluded that reflex connections between the TMJ nociceptors and the fusimotor-muscle spindle system of the dorsal neck muscles might be involved in the pathophysiological mechanisms responsible for the sensory-motor disturbances in the neck region found in TMD patients.

Neck extensors muscles commonly are tender and painful in patients with TMD. However, evaluation of their function through a fatigability analysis has not been conducted. Therefore, the objective of this study was to determine, through electromyographic evaluation, whether patients with myogenous TMD have greater fatigability of the cervical extensor muscles (midcervical paraspinal muscles [trapezius, capitis, and cervicis, groups]) when performing a neck extensor endurance test when compared to normal control subjects.

METHODS

A convenience sample of 36 healthy females and 30 female patients with myogenous TMD was used. Prior to the electrode application, the subjects' skin was cleaned and shaved when necessary. For the cervical extensor muscles, the electrodes were located over the distal half of the distance between the base of the occiput and the spinous process of the seventh cervical vertebra as described in the protocol used by Falla et al.(Falla et al., 2004). Each subject was in prone position lying on a plinth with their head and neck initially supported over the end and arms alongside their trunk. A strap was placed across the T2 level in order to counter support the thoracic spine. A velcro strap was fixed around the skull at the level of the forehead. An extendable tape was attached to the velcro strap at the point of the subject's eyebrows and hanging at 3 cm from the floor. Thus, this extendable tape acted as pendulum and indicated when the subject had lost the position. A visual feedback indicated to the subject when the position was lost (Figure 1). The subjects were instructed to maintain the position as long as possible, stopping at signs of fatigue or any discomfort.

Figure 1: Cervical extensor endurance test.

EMG data from the cervical extensor muscles was recorded (analog raw signal), and was sampled to 1024 Hz, band-pass filtered between 20Hz-450Hz, and amplified using a gain of 10000. To allow comparisons between subjects, the time course of each EMG variable was normalized with respect to the intersection of the regression line in the fatigue plot. Thus, a fatigue index was obtained for every subject and was compared between patients with myogenous TMD and control subjects.

IGOR Pro5.1 was used to analyze EMG data. The endurance of the neck extensors was also measured recording the time in seconds until the end of the test.

A MANOVA test was used to evaluate the differences in EMG fatigue index and holding time for the cervical extensor muscles while performing the neck extensor endurance test between patients with TMD and control subjects ($\alpha = 0.05$). SPSS Statistical Program version 15.0 was used to perform the statistical analysis.

RESULTS AND DISCUSSION

A MANOVA test determined that there were not statistical differences between subjects with myogenous TMD and healthy subjects neither in fatigue index determined by EMG analysis, nor in holding time ($p>0.05$). Fatigue Index differences and holding time differences have been observed in patients with neck pain when compared with healthy controls. However, in this study these results could not be supported. Probably, the sensitivity of the neck endurance test to evaluate neck muscle fatigue as well as the low level of neck dysfunction and jaw dysfunction presented by the patients in this sample could have contributed to the obtained results. Further research looking at subjects with more severe levels of dysfunction and different types of TMD are needed.

SUMMARY/CONCLUSIONS

Patients with TMD presented not significant differences in Fatigue Index and holding time when compared with healthy controls.

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ELECTROMYOGRAPHIC EVALUATION OF THE PERFORMANCE OF FLEXOR CERVICAL MUSCLES IN PATIENTS WITH TEMPOROMANDIBULAR DISORDERS WHEN EXECUTING THE CRANIOCERVICAL FLEXION TEST (CCFT): PRELIMINARY RESULTS

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INTRODUCTION

Patients with TMD have been shown to have cervical spine dysfunction concomitantly with TMD. However, evaluation of this cervical dysfunction has only been subjectively evaluated through a general clinical examination of the cervical spine (signs and symptoms).

Thus, an objective evaluation of motor activity of the cervical muscles through electromyographic (EMG) assessment of the cervical muscle performance in patients with TMD could clarify the role of the cervical muscles involvement in the symptomatology of patients with TMD.

The aim of this study was 1) to determine, through electromyographic evaluation, whether patients with Temporomandibular disorders (TMD) have an altered cervical muscular performance of the superficial cervical muscles (sternocleidomastoid and anterior scalene) when executing the craniocervical flexion test (CCFTThis page contains information about the abstract submission process and can be used as a model for abstract formatting.

METHODS

A convenience sample of 34 healthy females and 22 female patients with TMD was used. All subjects underwent a clinical examination by a dentist or by one physical therapist (independent of the evaluator who did the electromyographic evaluation) to determine if the subjects met the inclusion criteria. The subjects' skin was carefully prepared by another evaluator who was blinded to the condition of the subjects (evaluator 2). Prior to the electrode application, the subjects' skin was cleaned with alcohol and then electrodes were located on the sternal head of sternocleidomastoid (SCM) and in the anterior scalene as described in the protocol used by Falla et al.(Falla, et al. 2002).

All subjects performed the craniocervical flexion test (CCFT). The CCFT requires the patient to perform the craniocervical flexion movement in five progressive stages. Subjects were instructed to perform a gentle nodding movement (craniocervical flexion) and practice progressive targeting using the air filled pressure sensor. Subjects were trained through five incremental levels with the aid of visual feedback device. Subjects had to maintain the pressure steady on each target for duration of 10 seconds. The five levels were randomized in order. A visual feedback device was located in front of the subject's eyes, so she could see if she has reached the desired level. The subjects repeated this procedure 2 times with a rest period of 1 minute between repetitions to avoid fatigue effect (Figure 1).

Data acquisition was sampled at 1024 Hz, amplified to 10000 (kilogain) and filtered 20-450 Hz for the analyzed muscles. To obtain a measure of EMG amplitude from Sternocleidomastoid (SCM) and Scalenus anterior (SA), maximum root mean square (RMS) was calculated for 3 second for each muscle using IGOR Pro 5.1 and was expressed as a percentage of the EMG activity obtained.

Figure 1: Craniocervical Flexion Test

A three –way mixed design ANOVA with repeated measures (3 independent variables: muscles [SCM, and scalenes], test [5 levels] and groups [control and patients-between subjects]) test was used to evaluate the differences in EMG activity for selected muscles (dependent variable) while performing the craniocervical test under five incremental levels. Paired comparisons using Bonferroni post hoc test were used to evaluate the differences between variables. SPSS 15.0 was used to perform the statistical analysis. The level of significance was set at $\alpha = 0.05$

RESULTS AND DISCUSSION

Ninety five females were screened and 56 were finally included in this study. Thirty four were healthy and 22 had mixed TMD. A three –way mixed design ANOVA with repeated measures determined significant differences in EMG activity across pressure levels and muscles. There were also significant differences between normal subjects and subjects with TMD ($p<0.02$). Bonferroni post hoc test determined that patients with TMD had a significantly higher normalized EMG activity at 22, 24, 28 and 30mmHg when compared with normal subjects ($p<0.05$). In addition, the anterior scalene showed a significant higher activity in patients with TMD when compared with normal subjects at 22, 24, 26, and 30 mmHg Sternocleidomastoid muscle presented significantly higher EMG activity at 22, 24mmHg in subjects with TMD when compared with normals. None of the interactions was statistically significant

SUMMARY/CONCLUSIONS

Based on the results obtained from this study, patients with TMD had a significant greater normalized EMG activity at different pressure levels when executing the craniocervical flexion test compared with healthy subjects. This could indicate that subjects with TMD could have an altered performance of the cervical flexor muscles and thus could contribute to maintain the jaw-cervical dysfunction and craniofacial pain seen in patients with TMD.

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AN ELECTROMYOGRAPHIC STUDY OF PATIENTS WITH MASTICATORY MUSCLE DISORDERS: A RANDOMOZED CONTROLLED TRIAL OF PHYSIOTHERAPEUTIC AND ODONTOLOGIC TREATMENT

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INTRODUCTION

The most common etiologic agent regarding the myogenic TMD is the muscular hyperactivity.This hyperactivity can contribute to internal disarrangements of the TMJ. The muscular hyperactivity can be triggered by malocclusion, postural changes, and emotional stress. The electrical activity of the masticatory muscles in myogenic TMD subjects, associated with miofacial pain, was reported in 88% of the cases analyzed by the author, which was associated with the hyperactivity of an isolated muscle or in different types of combinations.

Several treatment options are proposed to muscular disorder. Among others, utilize the occlusal splint as a therapeutic device to treat muscular disorders resulting from bruxism, with the goal of attaining relaxation of the masticatory musculature. However, only the first author utilize the occlusal splint with a relaxation purpose, thus, not accomplishing the treatment.

There is a substantial relationship between TMD and hyperactivity of the temporal muscle, and the physiotherapeutic treatment (massage) can reduce and eliminate pain and hyperactivity. Nevertheless, the intent of this project is to assess the physiotherapeutic and odontologic approach of massage therapy and occlusal splint in miogenic TMD volunteers through the analysis of the electromyographic trace, comparing pre and post therapeutic bilaterally behavior of the masseter muscle and the anterior portion of the temporal muscle during the bilateral masticatory activity.

METHODS

This study comprises of a randomized, double blind, controlled clinical trial *Research hypothesis*: the treatment associating massage therapy with the muscle relaxant Michigan occlusal splint in more effective to decrease the electromyographic activity of the muscles anterior temporal and masseter bilaterally, and decrease pain in bruxist individuals with TMD.

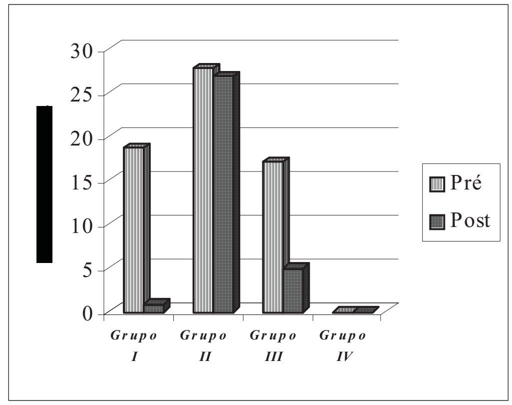
Male and female individuals aging between 18 and 40 years old were assessed. All subjects were submitted to the EMG assessment. The volunteers were divided in four groups. *Group I: 12* young with signs and/or symptoms of myogenic TMD related to bruxism or clenching, also regarded to the inclusion criterion. They were submitted to both EMG exams and massage therapy. *Group* II: 12 young with signs and/or symptoms of myogenic TMD related to bruxism or clenching, also regarded to the inclusion criterion. They were submitted to both EMG exams and occlusal splint. Group III: 12 young with signs and/or symptoms of myogenic TMD related to bruxism or clenching, also regarded to the inclusion criterion. They were submitted to both EMG exams and massage therapy and occlusal splint. Group IV: 12 subjects with normo-occlusion (class I - Angle) and no history of temporomandibular disorder, regarding the inclusion criterion. They were submitted to the EMG exams, but not to the physiotherapeutic and odontologic treatment. Important to mention that only partial results will be presented in this study

To register the electromyographic signals, the channel was calibrated allowing a gain of 2000, with a cut frequency of 10Hz in the high passing filter, and 500Hz in the low passing filter attained through an analogical filter, Butterworth, with two terminals (poles) which presented an acquiring frequency of 1000Hz. The electrodes were positioned in the body of the masseter muscle, and in the anterior portion of the temporal muscle. During the non-habitual masticatory activity the subject placed the Parafilm.

RESULTS AND DISCUSSION

Considering the data obtained and the statistical analysis performed in each studied groups, and noticed that with the techniques applied on the Groups I, II, III and IV, no significant result of EMG activity of all the muscles in study was collected. But, the results point out the difference (p=0,0026) between the AVS scores, which was taken before and after the treatment (analyzed by the ANOVA statistical test).

There was a significant decrease in pain Group I, after Group III, after group II and Group IV don't have difference (Graphic 1). The data collected in this study demonstrated that the massage therapy was efficient in decreasing the pain. Even though it is not possible to draw a comparison with the literature that deals with the effect of massage therapy in patients with bruxism because this parcial results.



Graph 1- Comparison of the pain before and after the treatments measured by Analogical Visual Scale in the all groups studied.

CONCLUSIONS

The physiotherapeutic treatment (massage) and odontologic treatment (occlusal splint) and both treatment together can reduce and eliminate pain. However, the physiotherapeutic and both treatment together can reduce and eliminate pain more.

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OCCLUSAL ADJUSTMENT INFLUENCE ON THE MASTICATORY MUSCLES EMG IN A PACIENT TREATED BY FUNCTIONAL MAXILLARY ORTHOPEDICS – A STUDY OF CASE

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INTRODUCTION

Occlusal adjustment therapy has been used as a treatment modality in dentistry (Okeson, 1998). However, there is no sufficient evidence to support the performance occlusal therapy as a general method for treatment. (Tsukiyama et al., 2001) and there is a need of more investigations about this method.

The aim of this study was analyse the effect of occlusal adjustment on the temporal and masseter muscles EMG recordings in a pacient treated by Functional Maxillary Orthopedics.

METHODS

One woman, 30 years old, healthy and without temporomandibular disorder, Angle Class I, on treatment with Functinal Maxillary Orthopedics, participated in this study.

The occlusal adjustment was based by masseter and temporal muscles digital palpation and articular paper marks, and performed by a trained examiner that did not have any acess to the previous EMG data. The masseter and temporal muscles EMG was registered before, immediately after and two weeks after the occlusal adjustment.

The electrical activity of the right masseter (RM), left masseter (LM), anterior portion of the right temporal muscle (RT) and anterior portion of the left temporal muscle (LT) were obtained by a signal acquisition module, Myosystem BR-1 with 12 channels, 12-bit resolution, and gain of 50 times. maximal biting and slowly closing of the teeth until maximal bitting for 2 seconds were assessed.
bilateral chewing were assessed. Passive surface electrodes were posicioned according Cram & Kasman (1998). The software Myosystem BR-1 3.0 was used to process the SEMG signal, with a frequency sample rate of 2000 Hz and digital band-pass filter of 10-500 Hz. The signals were then analyzed in according to RMS values.

RESULTS AND DISCUSSION

No difference was found imediately after the occlusal adjustment. The temporal muscles potentials were bigger than in the masseter muscles in the first and second register (before and imediately after occlusal adjustment), sugesting a muscular dysfunction according to Berzin (2004).

However, changings in the EMG recordings was observed and suggest a better muscular balance after two weeks.

CONCLUSIONS

This study suggests that EMG can be a useful tool to evaluate the possible muscular balance gotten after an occlusal adjustment.

The occlusal adjustment seems to be positive in patients treated by Functional Maxillary Orthopedics but aparently the effect in EMG register is not imediate and depends on a muscular adaptation. Further researchs are needed supporting these results.

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EFFECT OF HVES IN PATIENTS WITH TMD – ELECTROMYOGRAPHIC EVALUATION

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INTRODUCTION

The treatment for temporomandibular disorder (TMD) consists of guidance, medication therapy and physiotherapy. A resource used in physiotherapy is high voltage electrical stimulation (HVES), which is indicated for analgesia and tissue repair (Stralka et al., 1998).

By EMG it is possible to observe some of the characteristics of patients with TMD (Bérzin, 2004).

The aim of this study was to verify the effect of 10 sessions of HVES on pain and electromyographic activity of the masticatory muscles in patients with TMD.

METHODS

Ten women, with a mean age of 20.45 years (± 2.41), with TMD, according to the Research Diagnostic Criteria for Temporomandibular disorders (RDC/TMD), participated in this study.

After sample selection by means of the RDC/TMD, the volunteers underwent electromyographic exam of the masticatory muscles. This assessment was made at three moments, which were called assessment 1, assessment 2, and assessment 3. Assessment 1 was made soon after sample selection; assessment 2 was made four weeks after assessment 1, and in this period the volunteers received no type of intervention, and were instructed not to begin any medication, dentistry and or physiotherapeutic treatment. Assessment 3 was made 48 hours after the tenth HVES application, so that the results would not be influenced by the immediate effect of the current.

The electrical activity of the suprahyoid muscles (SH), right masseter (RM), left masseter (LM), anterior portion of the right temporal muscle (RT) and anterior portion of the left temporal muscle (LT) were obtained by a signal acquisition module, model EMG1000 Lynx® (São Paulo, SP, Brazil). Five bipolar active surface electrodes were used, with simple differential configuration (Lynx® São Paulo, SP, Brazil).

Three electromyographic signal recordings were made, with duration of 5 seconds, with volunteer with the mandible at rest, that is, lips lightly closed without clenching the teeth.

For HVES application, the equipment *Neurodyn High Volt®* (Ibramed Ltda., Amparo, SP, Brazil). The active electrodes were placed bilaterally on the anterior portion of the temporal muscle (channel 1) and on the masseter muscle (channel 2) with frequency of 10 Hz, twin pulses of 20 microseconds each with an interval of 100 microseconds between them, intensity of over Volts attaining the motor threshold, with positive polarity, during 30 minutes.

RESULTS AND DISCUSSION

Normalization of the electromyographic signal (root square mean - RMS) was performed. Normality of the sample was verified with the Shapiro-Wilk test and the Friedman test was applied.

As the data in the electromyograph from the 1st evaluation were used for normalization, a comparison was made between the 2nd and 3rd evaluations. It was verified that with the mandible in the postural rest position after HVES application, there was a significant reduction ($p<0.05$) in the normalized RMS values for all the muscles evaluated (Figure 1).

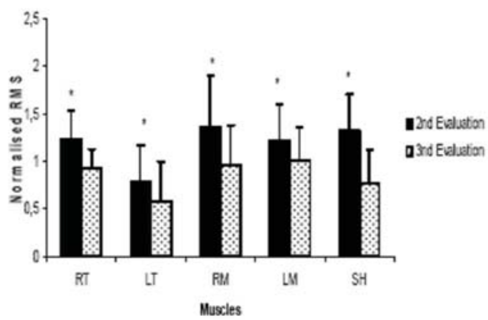

Figure 1: Mean and standard deviation of the normalized RMS values of the right temporal (RT), left temporal (LT), right masseter (RM), left masseter (LM), and suprahyoid (SH) muscles; in the rest situation, before (2nd evaluation) and after (3rd evaluation) the treatment with HVES.

*Was considered for significant differences ($p< 0.05$).

It is believed that the results obtained occurred due to the increase in the blood flow. Moreover, other studies verified increased blood flow in the muscular tissue after HVES applications. (Goldman et al., 2003). One believes that this result is due to pain relief, since Bodéré et al. (2005) found that patients with TMD and myofascial and neuropathic pain, presented greater activity of the temporal and masseter muscles at rest. Moreover, some authors related that individuals with TMD that present pain, have increased electromyographic activity of the masticatory muscles, with the mandible in the rest position (Bodéré et al., 2005).

CONCLUSION

Therefore, by means of the evaluations made in this study, HVES can be considered an efficient method for the treatment of TMD, improving electromyographic activity in these patients.

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DIAGNOSIS CONTRIBUITION OF SURFACE ELECTROMYOGRAPHY FOR TEMPOROMANDIBULAR DISORDERS

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INTRODUCTION

Surface electromyography (SEMG) has been consistently used by clinicians and researchers to assess muscular function. SEMG is considered an important instrument in the evaluation of muscular conditions in Temporomandibular Disorder (TMD) patients (Stohler, 1999; Svensson, 2007). However, it does not seem to be able to determine the presence of dysfunction because it is limited in terms of reliability, validity, sensibility and specificity (Klasser & Okeson, 2006).

The aim of this study was to compare the SEMG of the temporalis anterior and masseter muscles in subjects classified by the Research Criteria Diagnosis for Temporomandibular Disorders (RDC/TMD) Axis I (Dworkin & LeResche, 1992), and to find a measure that could better distinguish between TMD and non-TMD subjects. Once such a measure was found, the objective was to verify SEMG validity to diagnose TMD.

METHODS

Sixty-one female volunteers aged between 18 and 36 (23,3 ± 8,2) participated in this experiment. The volunteers were divided into two groups, namely TMD and control. The TMD group comprised 36 myogenic TMD patients and the control group consisted of 25 non-TMD subjects. The EMG records were obtained using the Myosystem BR-1 equipment with 12 channels, 12-bit resolution, and gain of 50 times. Differential surface electrodes (pure silver, gain 100) were used. The software Myosystem BR-1 3.0 version was used to visualize and process the SEMG signal, with a frequency sample rate of 2000 Hz and digital band-pass filter of 10-500 Hz. The signals were then analyzed according to measures of Root Mean Square (RMS), Integral of Envelope Linear and Median Frequency. Maximal biting and bilateral chewing were assessed.

The results were observed in relation to mean, standard deviation, coefficient of variation and angular coefficient of the linear regression, which were obtained from three windows in each signal during the recorded period. Analysis of Variance (ANOVA) was used to assess the difference between groups, and Discriminant Analysis to test the diagnostic validity of SEMG.

RESULTS AND DISCUSSION

The mean of envelope linear values during dynamic contractions of the temporal muscle was different between groups ($p<0,03$), and the angular coefficient of the linear regression to RMS values was different during isometric ($p<0,04$) and isotonic ($p<0,05$) contractions in the same muscle.

The sensibility values ranged from 80,6% to 84,3%, the specificity values ranged from 66,6% to 76% and predictive positive values ranged from 72,2% to75%%.

According to Dworkin & LeReshe (1992), Chung & Nguyen (2005) and Kitara et al. (2006) temporomandibular disorders is a chronic pain condiction and envolves a multidisciplinary approach, and this characteristics require more than 95% of specificity and more than 70% of sensibility or predictive positive values.

CONCLUSIONS

Based on the findings, the SEMG of temporal anterior muscle is able to differentiate TMD and non-TMD subjects when the mean of envelope linear values during dynamic contractions is observed, and when the angular coefficient of linear regression for RMS values in static and dynamic contractions is examined. Nevertheless, the validity of SEMG as a diagnostic instrument for the TMD condition could not be corroborated in this study.

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IMMEDIATE ALTERATIONS IN SUPRAHYOIDS MUSCLES KINETICS AFTER FUNCTIONAL MAXILLARY ORTHOPEDICS APPLIANCE INSTALLATION – AN ELECTROMYOGRAPHIC STUDY

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INTRODUCTION

The SEMG has been recognized lately by Functional Maxillary Orthopedics as one of good tools that can helps diagnosis and clinical activities, because it permits a correct measurement of muscular kinetics alterations occurred in the treatment of malocclusions associated to muscular dysfunctions (Pedroni et al. 2004). The aim of this study is present alterations in kinetics of suprahyoids muscles by surface electromyography, before and after functional orthopedics appliance installation, correlating these with others diagnosis clinical data.

METHODS

Registers were collected from suprahyoids muscles, bilaterally, by 19 subjects, age 9 to 15, both sexes, with different malocclusions types, in rest mandibular position, in isometrics (stimulated by examinator) and water deglutition in three moments : diagnosis, 8 minutes after Functional Maxillary Orthopedics appliance installation and after gained teeth contact in Determined Area (DA), clinical result preconized by Simões (1985). Surface passive electrodes, bipolars, model 272 NORAXON Inc. were used to collect the signals, connected to a signals conditioner EMG 1000 (Lynx Electronics). Electrodes locations over muscles skin was determined by protocol preconized by Pedroni et al (2004),and utilized by Electromyography Laboratory of Piracicaba Dental College (FOP/UNICAMP), using for each muscle a specific muscular function proof. All recommendations established by ISEK and SENIAM were followed.

RESULTS AND DISCUSSION

On first electromyographic registrations (Diagnosis), the electric activities from the muscles studied don’t present in all subjects, as a normal pattern, as preconized by Faria & Bérzin (1998) and Bérzin (2004).
All the subjects, after installation of Functional Maxillary Orthopedics appliance, independent of malocclusion showed alterations in electromyographic registration suggesting better muscular equilibrium in rest and deglutition. The electromyopgraphic results collected after teeth contact in Determined Area (DA) keep stable for months.
Authors like Graber (1963) and Moss (1975) refereed that functional equilibrium (muscular) is desirable as a malocclusion treatment deal, including as a signal of maintenance of therapeutic results. The data gained by this work are compatible with refereed authors’ opinion.

SUMMARY/CONCLUSIONS

The Surface Electromyography showed as a trustable tool in diagnosis and to follow malocclusion treatment with muscular dysfunction. The Functional Maxillary Orthopedics is an option for the treatment of malocclusion associated to muscular dysfunction.

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COMPARISON OF WATER DEGLUTITION DATA WITH AND WITHOUT FUNCTIONAL MAXILLARY ORTHOPEDIC APPLIANCE , REGISTERED WITH SURFACE ELECTRODES

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INTRODUCTION

The aim of this study is to show suprahyoids muscles reactions registered with surface electrodes after Functional Maxillary Orthopedics appliance installation, comparing with diagnosis data, and if Surface Electromyography is a trustable tool to be used in diagnosis and verification of malocclusion treatment by Functional Maxillary Orthopedics (FMO) clinical results . The results showed that the main objective (muscular action and equilibrated reaction) were gained and viewed immediately; these results are an important part of clinical objectives in malocclusion treatment by FMO.

METHODS

With the approval of Ethics Committee in Research with Human being from Campinas State University, data were collected in 3 times, from suprahyoids muscles, bilaterally, from 19 individuals, both sexes, with different kind of malocclusion , using surface passive electrodes (model 272, NORAXON Inc.), Signal Conditioner EMG1000 (Lynx Electronics) , during water deglutition, for three times: diagnosis, 8 minutes after appliance installation, and after specific clinical result (Pedroni et al, 2004) (teeth contact in Determined Area [DA]) ,Simões 1985). Electrodes location and position were determined following protocol preconized by Electromyography Laboratory of Piracicaba Dental College (FOP/UNICAMP), which uses specific muscular function proof. All recommendation established by ISEK and SENIAM were followed.

RESULTS AND DISCUSSION

All of diagnosis data in deglutition, showed non equilibrated SEMG registration from suprahyoids muscles; just after the appliance installation they moved in bilateral equilibrium direction (normal) (Faria & Bérzin, 1998; Bérzin, 2004).
The results don't depend of malocclusion type, in rest state as in function (water deglutition).
These results were stables for months until the last registration that was done after DA position gained by mouth.
Some articles indicated that the functional equilibrium (muscular) is desirable as an aim in malocclusion treatment, and is considered as a signal of treatment results maintenance (Graber, Moss).

SUMMARY/CONCLUSIONS

These results give support for the conclusions above:

1- SEMG is a trustable tool to be used in diagnosis and to verify the bilateral functional stability of muscular function;

2- Functional Maxillary Orthopedics is an option for the malocclusion treatment with muscular dysfunction.

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EMG PROFILE OF BRAZILIAN WOMEN WITH CHRONIC STRESS AND TEMPOROMANDIBULAR DISFUNCTION

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INTRODUCTION

The prevalence of some clinical pictures of chronic orofacial pain has increased progressively in the last years. Particularly among them are the pain syndromes caused by Temporomandibular Dysfunction (TMD), which makes clinical management difficult due to the wide symptomatology, multifactorial etiology, complicated diagnostic approach and treatment, strong presence of emotional factors and negative impact on personal life (IASP, 1986; Okeson, 1998; Siqueira & Teixeira, 2001).

According the literature, prolonged stress is associated with a state of tension in the masticatory musculature and craniocervical region muscles, common in TMD, and is amenable to measurement by surface electromyography (Schumann et al. 1988; Berzin, 2004). A detailed understanding of the increased muscle activity in individuals with chronic stress can contribute to a more precise and differential diagnosis of the pain symptoms occurring in TMD and moreover to the selection and improvement of efficacious treatment. The aim of this work was to study the electromyographic profile of mastication muscles and craniocervical region muscles in women with chronic stress and pain due to TMD.

METHODS

The study involved 14 women aged 22 to 47 years, with chronic pain due to TMD. An electromyographic examination was performed bilaterally on the following muscles: temporal, masseter, suprahyoid, suboccipital, sternocleidomastoid and trapezius. The patients were tested in a seated position (resting). Ag/AgCl surface electrodes were placed on the skin fixed with a permanent distance of 1 cm between them, connected to an electromyography (Myotronics K6-I Diagnostic System). The calibration of the instrument was 1second/division and 30 μvolts/division. Psychological evaluation was carried out utilizing the Lipp's Inventory of Stress Symptoms for Adults (ISSL) and Goldberg's General Health Questionnaire (GHQ). Statistical analysis were calculated by che-square test and the significance level was 5%.

RESULTS AND DISCUSSION

The results confirm literature findings that allege that there is the electrical activity, in rest, of some muscle groups in individuals with chronic stress and TMD (Semeghini et al., 2001). The chi-square test gives strong indications ($p<0.01$) that the true proportion of activation of the temporal, suprahyoid and trapezius muscles in rest position is significantly greater than that of the non activation of these muscles.

The study also evaluated the frequency and percentage of active muscles in rest position. As illustrated in Table 1, there are no indications of significant differences in the percentages of activation for the muscles studied. According the literature, in rest position, the muscles should not show any action potential , except for small noise potentials of the electrodes, electrical circuits, contamination from the feed current, etc. They can be observed bellow 2 µvolts (Bérzin, 2004). The Figure 1 shows the electrical potentials of all the muscles evaluated over 2 µvolts.

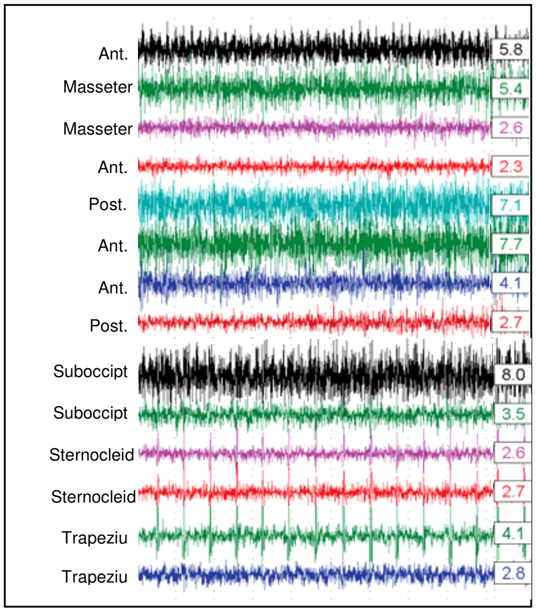

Figure 1: EMG with the jaw in rest position in women with TMD and chronic stress; Left and Right (LR) Temporal, LR Masseter, LR Suprahioid, LR Suboccipital, LR Sternocleidomastoid and LR Trapezius.

SUMMARY/CONCLUSIONS

The study confirms the high electrical potentials of masticatory muscles and craniocervical region muscles, in rest position, in women with chronic stress and TMD. There are no significant differences in the percentages of activation for the muscles studied.

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Table 1: Frequency and percentage of muscles with activation (n:14). Muscles with same letter do not differ according to the chi-square test at a significance level of 5% (n: 14)

	Statistics		
Muscle	Frequency (%)		Chi-square
Temporal	14 (20.59%)		A
Suprahyoid	12 (17.65%)		A
Trapezius	12 (17.65%)		A
Sternocleidomastoid	11 (16.18%)		A
Suboccipital	11 (16.18%)		A
Masseter	8 (11.76%)		A

Chi-square: 1.7059; DF: 5; Pr > ChiSq: 0.8882

EFFECTS ON EXCITABILITY CHANGES IN HUMAN PRIMARY MOTOR CORTEX (M1) ARE VARIABLE DEPENDENT ON CHARACTERISTICS OF FUNCTIONAL ELECTRICAL STIMULATION (FES)

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INTRODUCTION

Functional electrical stimulation (FES) has been widely used as a therapeutic tool for functional recovery of muscle weakness or paralysis. For example, FES is used to treat foot drop during walking and to improve hand grasp in patients with spinal and upper motor neuron lesions. The effects of repetitive FES have been confirmed by previous neurophysiological and clinical studies. FES simultaneously induces not only muscle contractions but also somatosensory inputs. Thus, it is thought that these peripheral afferent inputs may play an import role in plastic changes in the central nervous system; however, the functional properties of FES are still not well known. The purpose of the present study, therefore, was to investigate how the characteristics of FES (cycles per minute, intensity and pulse trains) affect excitability changes in the human primary motor cortex (M1). For this purpose, we recorded motor evoked potentials (MEPs) by transcranial magnetic stimulation (TMS) at rest and during voluntary muscle contraction of the targeted muscle.

METHODS

19 normal subjects, 12 women and 7 men (age 21–30 years), participated in this study. We explained the experimental purpose and procedures to all subjects and obtained written consent. FES was applied to the right flexor carpi radialis (FCR) muscle. FES was applied using an electrical stimulator (Nihon Kohden Co.; SEN-7203) with an isolator (Nihon Kohden Co.; SS-104J). FES conditions are as follows: A) four different FES intensities, 1.3, or 1.5 times the sensory threshold (ST), and 1.1 or 1.3 times the motor threshold (MT). The pulse width is 1ms duration and pulse frequency is 100Hz, B) four different pulse frequencies; 1, 10, 100 and 1000Hz. Each frequency consisted of five trained pulses (pulse width, 1ms; stimulus intensity, 1.3 times ST), C) six different pulse trains; 5 or 50 trains of 10Hz, 50 or 500 trains of 100Hz, and 50 or 500 trains of 1000Hz, respectively. Stimulus intensity is 1.3 times that of ST. In addition, to investigate the effects of voluntary drive, we recorded MEPs of FCR muscles at rest and during target muscle contraction of 5% MVC. Torque was also recorded by strain gauge sensors mounted onto a wrist plate. MEPs to TMS (Magstim Co.; Magstim-200, figure-of-eight coil) from the FCR and the extensor carpi radialis (ECR) muscles were simultaneously recorded before and after applied FES. The intensity of TMS was set to 110–130% of MEP threshold, and TMS was applied after 50ms from the last pulse of each FES applied. Ten to fifteen MEPs were recorded in each FES applied condition. The peak-to-peak amplitudes of all MEPs of

FCR and ECR muscles were measured. MEP amplitude ratios were calculated using MEPs before FES was applied (baseline value). Changes in MEP of FCR and ECR muscles after FES was applied were analyzed using the Friedman test (with Wilcoxon's post hoc test) and α-risk was set to 0.05.

RESULTS AND DISCUSSION

When the FES intensity was higher, MEP ratios of each muscle at rest were significantly enlarged (Friedman test, $P<0.05$, see Fig. 1).

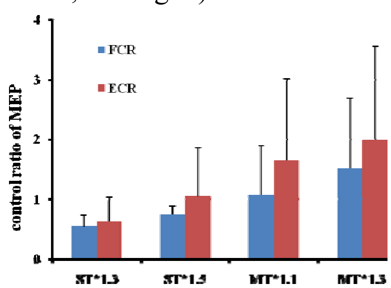

Fig. 1 Means and SDs obtained from pooled data (n=7) at four different FES intensities are represented.

Based on the present results, excitability changes of M1 in both FCR (stimulated) and ECR (antagonist) muscles were enhanced dependent on increasing FES intensities and were not reciprocal. Concerning the effects of pulse frequency, the MEP ratios of both muscles were enhanced dependent on higher pulse frequency, but were not statistically significant. This evidence indicated that a higher pulse frequency might induce temporal summation in M1; that is, increasing train pulse might induce clear excitability changes in M1 of both muscles in the muscle resting state (see Fig. 2). On the other hand, when the stimulated muscle (FCR) was contracted, the MEP ratio of the antagonist muscle (ECR) had an inhibitory effect on M1 (Fig. 3, 100, 1000 z: $P<0.05$).

These findings suggest that the characteristics of the FES condition might influence different modulations on excitability in M1.

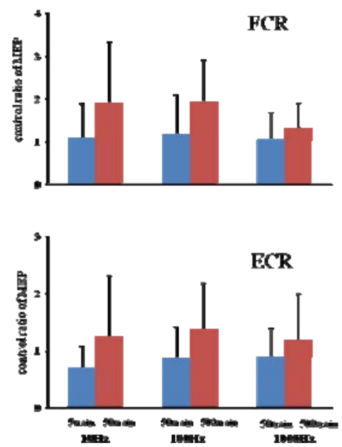

Fig. 2 Means and SDs of frequency effects on both FCR and ECR muscles obtained from group data (n=6) are shown.

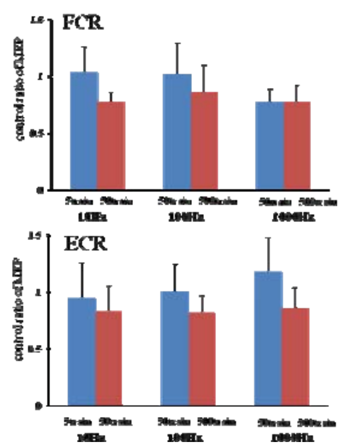

Fig. 3 Means and SDs of frequency effects on both muscles under voluntary contractions obtained from group data (n=6) are shown.

CONCLUSIONS

FES effects on M1 excitability changes vary dependent on the characteristics of FES and muscle conditions (resting or contraction).

THE EFFECTS OF JOINT TRACTION AND POSITION OF UPPER LIMB ON REACTION TIME OF QUADRICEPS FEMORIS

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INTRODUCTION

Joint traction is used to improve the range of motion and muscle strength or to reduce joint pain in physical therapy. Joint traction also acts as a stretch stimulus by elongating muscles and other joint proprioceptors. Previous studies showed increased muscle activities and shortened reaction time by relevant joint traction (Svendsen et al, Kurosawa et al). We found that the joint traction of upper limb increased the amplitude of soleus H-reflex (Koyama et al). These findings suggest that joint traction facilitates the central nervous system and influences between upper and lower limb, reciprocally. Meanwhile, some of the starting limb positions of motion patterns in Proprioceptive Neuromuscular Facilitation (PNF) approach increased soleus H-reflex, and shortened muscle reaction time (Yanagisawa et al, Fujita et al).
It is not certain what influences joint traction or the position of the upper limb facilitates on muscle reaction time of lower limb, and whether an interaction exists between tractions and positions. The purpose of this study was to examine the effects of joint traction and position of the upper limb on the pre-motor time (PMT) and the motor time (MT) of quadriceps femoris.

METHODS

Twenty healthy male subjects participated in this study (mean age (SD): 23.1 (2.2) y, mean height: 171.7 (5.9) cm, mean weight: 66.5 (7.3) kg). This study was approved by the Ethics Board of the Tokyo Metropolitan University, and written consent was obtained from all subjects before testing.
The subjects lay supine on the torque machine (BIODEX), and their left knee was flexed 30 degrees. Bipolar surface electrodes were place on the left rectus femoris muscle, after skin preparation. Then the subject was passively put in the upper limb neutral position and a combined position that was 30 degrees extended, 20 degrees abducted, 70 degrees internal rotated position of the shoulder, respectively, in parallel with joint traction by a pulley. Traction force was set to four types of loads (no-load, 30 N, 60 N and 90 N).
This combined position is used as one of the starting positions of motion patterns in PNF. Eight positions (2 positions × 4 traction force) were tested. The subjects were tested with the left knee at maximum isometric extension with immediate response to a signal sound trigger, while holding each testing position.
Ten trials were measured in each test. Electromyography (EMG) data were sampled at 2,000 Hz and recorded into a personal computer through the amplifier (Maclab) together with extension torque and signal trigger.
Following full-wave rectification, the onset of EMG was identified as the point where the mean of 400 subsequent samples before the trigger exceeded the background level of

activity by 2 SDs. The PMT was measured as the time from the trigger to the onset of EMG, and the MT were measured as the time from the onset of EMG to the onset of the muscle torque.
With the PMT and the MT as dependant variables, two factors (two upper limb positions × four types of traction force) with repeated measure ANOVA were used to identify significant effects. Sheffe's multiple comparison was used as a post hoc test.
All data analysis was performed using SPSS for windows 14.0 (SPSS Inc, Chicago). Level of statistical significance was set at the 0.05 alpha level.

RESULTS AND DISCUSSION

The PMT and the MT on each testing position were showed in Table 1 and 2. With the PMT, There was a main effect on both of traction force and positions, and there was a significant interaction between traction force and positions. With the MT, there was no significant effect on all testing positions.
As the results of multiple comparison, with the neutral position, there were significant differences between 0N-30N, 0N-60N, 0N-90N, 30N-60N and 30N-90N of traction force. There were no differences between each traction force with combined position. With 0N and 30N of the traction force, there were significant shortening of the PMT with the combined position compared with the neutral position.
These results showed that The PMT has been shortened with increase of the traction force, and the combined position produced a significant effect on the PMT with low load of traction force. Though previous studies examined the effect of relevant joint traction, we found the shortened PMT of the lower limb muscle by the joint traction of the upper limb.
It is generally known that the PMT is reflected in the process of the central nervous systems. These results suggest that the joint traction of the upper limb and the combined position might have the facilitated effect on the lower limb by the arousal of the central nervous systems.

Table 1: Average of the PMT (msec (SD))

Position	Neutral	Combined
0N	178.2 (32.5)	166.1 (29.1)
30N	169.8 (28.7)	164.7 (29.6)
60N	162.9 (28.3)	164.8 (31.8)
90N	160.8 (28.0)	163.7 (29.8)

Table 2: Average of the MT (msec (SD))

Position	Neutral	Combined
0N	52.4 (7.0)	53.1 (5.6)
30N	53.2 (6.6)	52.5 (5.5)
60N	52.6 (5.8)	52.6 (5.4)
90N	52.4 (6.1)	52.0 (5.2)

CONCLUSIONS

We examined the effects of joint traction and position of the upper limb on reaction time of quadriceps muscle. The results showed the shortened PMT of quadriceps muscle by the joint traction of upper limb, and showed more shortened PMT with the combined position compared with the neutral position on low load of joint traction. It will be considered that these results were due to the effect of arousal of the central nervous systems.

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THE EFFECT OF INCREASING THE INSTABILITY OF WEIGHT BEARING SURFACE ON THE ACTIVITY OF THE PERIARTHICULAR SHOULDER MUSCLES DURING CLOSE CHAIN EXERCICES

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INTRODUCTION

The shoulder complex relies on varying stabilizing mechanisms, including shapes of the joint surfaces, ligaments and most importantly the periarthicular muscles. The importance of the muscles in the dynamic stability of shoulder has been extensively emphasized in the literature (Labriola et al. 2005).

Close kinematic chain exercises have been suggested to have the best influence on the stabilizing function of the shoulder muscles (Naughton et al. 2005). No investigation has been performed to study the activity of different muscles during these exercises. The aim of the present study is to investigate the changes in the muscular activity of the shoulder muscles at different stability levels of weight bearing surface.

METHODS

Thirthy healthy volunteers participated in this experiment. Surface EMG from Superior & inferior Trapezius , Teres major, Posterior Deltoid, Seratus anterior and long head of Biceps were recorded from dominant side at six different randomly-ordered positions for 10 seconds. The positions were numbered from the most stable position (feet and hands on the ground as position 1) to the most unstable position (hands on a ball at the center of a wobble board and legs on a Swiss ball as position 6). In three positions the feet were on the ground and in the rest a Swiss ball were put under the thighs. The RMS of the recorded EMG from all muscles at all positions was normalized by the position 1.

The data from all positions and all recorded muscles were analyzed by repeated measure ANOVA test. Bonferroni test was also used for pair wise comparison of all positions.

RESULTS AND DISCUSSION

No correlation was found between the level of the stability and the intensity of muscular activity. Surprisingly the intensity of activity of all muscles was at highest during position 1. In fact at all three positions that Swiss ball was applied to the lower quarter the intensity of muscular activity was lower ($P \leq 0.01$) compared to the positions where the feet were on the ground.

Although it is expected that imposing a controlled unstable situation to the shoulder complex would increase the activity of the shoulder muscles, the results of the present experiment does not support this idea. It seems that wrist and the elbow joints were able to absorb the perturbations imposed on the upper limb in this experiment.

Comparing different positions investigated in the experiment, it seems that the amount of weight endured by the upper limb is the decisive parameter for the intensity of the muscular activity. Using Swiss ball under the thighs for example reduces the weight on the upper limb significantly.

CONCLUSIONS

The amount of load on shoulder during CKC exercises seems to be the main factor to design a progressive proprioception exercise program.

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Table 1: The mean normalized EMG activity of shoulder muscles at different positions.

position	**Sup Trapez**	**Inf Trapez**	**Teres M**	**Deltoid**	**Biceps**	**Seratus Ant**
1	100%	100%	100%	100%	100%	100%
2	92%±8%	84%±10%	74%±3%	67%±5%	122%±11%	112%±9%
3	79%±6%	61%±7%	48%±7%	26%±3%	65%±8%	79%±9%
4	75%±6%	64%±7%	55%±5%	44%±6%	54%±6%	89%±5%
5	78%±9%	69%±11%	42%±6%	29%±5%	55%±7%	61%±5%
6	94%±10%	57%±7%	55%±10%	12%±2%	57%±7%	49%±7%

A CASE REPORT OF A PATIENT WITH MYOFASCIAL PAIN SYNDROME: AN ELECTROMYOGRAPHIC STUDY.

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INTRODUCTION

Myofascial pain syndrome (MPS) is a painful condition characterized by the presence of trigger points, local and referred pain, tenderness, referred autonomic phenomena as well as anxiety and depression (Escobar P.L., Ballesteros J., 1989). The pathogenesis likely has a central mechanism with peripheral clinical manifestations. Myofascial trigger points can refer pain to the head and face in the cervical region (Borg-Stein J., 2002).

The therapy for myofascial pain requires enhancing central inhibition through pharmacology or behavioral techniques and simultaneously reducing peripheral inputs through physical therapies including exercises and trigger point-specific therapy (Graff-Radford S.B., 2004).

The objective of this study was to describe the electromyographic (EMG) results of an interdisciplinary treatment in a patient with myofascial pain syndrome.

METHODS

A 55-year-old woman with a 2-year history of chronic pain attended to the neurologist's office reporting daily cervical pain mainly on the right side, irradiating to the temporalis muscles, to the top of the head and a sharp pain going to the back of the eyes. She grits her teeth and it is uncomfortable for her to talk and eat. The primary care management given was the medication venlafaxina 75 mg, and the patient was conducted to a dental and physiotherapeutic evaluation.

The dentist decided for a periodontal treatment because the periodontal inflammatory process conjugates with the algogenic mechanisms that maintain the chronic pain.

The physiotherapist found out that she had limitation on the amplitude of head and neck movements, deficit of motor coordination and lack of body conscience. When the sternocleidomastoid, trapezius, esplenio cervicis, masseter and temporalis muscles were palpated they were very sore, especially in the suboccipital, and temporalis ´s regions (right side). Trigger points were found at the upper trapezius and at the sternocleidomastoid muscle. The patient was conducted to an EMG examination.

The EMG evaluation was done in anterior temporalis, masseter and cervical muscles that were reported as pain regions with active bipolar Ag/AgCl surface electrodes (EMG System Ltda.). EMG signal acquisition was made by 12 channel equipment (Myosystem I / Datahominis Tec. Co.). The analog EMG signal recorded were digitized using 12 bit A/D converter at a sampling rate of 4KHz. After digitalization, the signal was filtered by a digital pass-band of 10-500Hz, visualized and processed by the software Myosystem I version 2.12.

During the signal registering, the patient stayed seated, looking forward and relaxed. The tasks were: mandibular rest position, right unilateral mastication, left unilateral mastication, bilateral mastication and maximal voluntary clenching (MVC).

The algometer was used to get the pressure pain threshold (PPT) in temporalis muscles. To evaluate the stress and the anxiety level, which are psychological characteristics, a questionnaire and IDATE were applied.

Physical therapy based on massotherapy and kinesiotherapy was provided during 24 visits twice a week. The patient practiced exercises daily at home.

RESULTS AND DISCUSSION

In the EMG examination, there was found resting activity in all of the muscles evaluated, mainly in the right side agreeing with Svensson P. *et al.* (2004) that found that jaw muscle pain can be linked to increases in neck EMG activity with the head and jaw at rest. Temporalis muscles presented more activity than masseter muscles, which is commonly observed in dysfunctional patients. During the unilateral mastication, there was found activity between masticatory cycles, where it should have EMG silence (Rodrigues-Bigaton D. *et al.*, 2004). On the MVC it was observed a higher activity on the right side and equilibrium of the temporalis and masseter´s activity.

Clinically there was a significant relief in pain and an improvement of function. The algometric evaluation also showed a relief on the patient's pain (Right temporalis PPT= 2,16 Kgf; Left temporalis PPT= 4,1 Kgf). The PPT values obtained before the treatment were 0,525 Kgf in the right side and 0,41 Kgf in the left side.

Affective and humor alterations seemed to interfere in her pain condition. The IDATE values before the treatment showed a medium level of anxiety which reduced to a low one after that, probably because of the decrease of her pain.

After physical therapy and medications, the patient was submitted to another EMG examination and it was observed a decrease in cervical resting activity of 63,17% in the left side and 87% in the right side. Moreover, there was found an increase in anterior temporalis and masseter muscles activity during bilateral mastication and MCV, which demonstrated the functional improvement achieved and the importance of an electromyographic attendance to demonstrate the treatment efficacy

SUMMARY/CONCLUSIONS

1) Physical therapy associated to medications gave a significant relief in these patient myofascial pain symptoms;

2) The EMG could assist in the diagnosis and demonstrate this improvement observed in an objective way.

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INFLUENCE OF ANKLE POSITION IN THE BIOMECHANICS OF PILATES' FOOTWORK EXERCISES

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INTRODUCTION

The Pilates Method is a system of physical conditioning with applications in the physical therapy practice (Rydeard et al, 2006). The Pilates exercises denominated *Footworks* are directed to the goal of ankle and knee strength and alignment. The *Footworks* are closed kinetic chain exercises done in an apparatus known as *Reformer*. Increased muscle co-contraction leading to joint stabilization is one of benefits attributed to closed kinetic chain exercises in joint rehabilitation (Harter, 1996). The *footwork* exercises are done in three basic positions, with the foot pressed against a bar in the *Reformer*: ankle in dorsiflexion, hip in neutral position, with toes flexed, forefoot on the bar (Arch) or toes extended and heels on the bar (Heel); and ankle in plantar flexion, hips in external rotation and toes on the bar (V position).

The purpose of this study was to investigate the effect of the three different positions of the *footwork* exercises in the ankle range of motion and EMG of tibialis anterior (TA), peroneus longus (PL) and lateral gastrocnemius (LG).

METHODS

Twenty-five healthy subjects (5 males, 31±6 yrs) participated in the study. All subjects had more than 6 months of Pilates training. EMG activity from TA, PL and GL muscles was recorded by means of silver surface electrodes of 10-mm diameter (Medtrace) and an EMG System unit. The signal was amplified by 1,000 factor. The electrodes were placed according to the SENIAM recommendations (Hermens *et al*, 2000). Ankle range of motion was assessed using an electrogoniometer (Biometrics) synchronized with the EMG. The raw EMG data was rectified, normalized by the peak value and the linear envelope was calculated and integrated for the knee flexion and extension phases.

The relationship between the activities of TA and GL muscles was established by the co-contraction index (CI), adapted from Falconer and Winter (1985). The exercise patterns were compared using ANOVA for repeated measures followed by Tukey post hoc test (α=5%).

RESULTS AND DISCUSSION

Results for CI are displayed in table 1. EMG activity are displayed in table 2. The *footwork* Heel had a significantly greater CI value in the extension phase than Arch and V positions. This exercise had also significantly less ROM than other positions (p=0.001) (arch:29.9±5.5º; heel:16.5±7.4º; V position: 33.5±6.8º) and a significantly increased EMG activity for: TA in both extension and flexion phases; GL and PL in the flexion phase. V position had greater ROM and more GL activity in the extension phase than Heel position (marginally significant, p=0.058)

In the beginning of the ankle rehabilitation treatment, when more joint stabilization is needed, Heel position could be employed because it provides increased co-contraction and lower ankle ROM. This exercise could also strengthen PL muscle in the cases of ankle sprain, when specific strengthening of this muscle is necessary (Kaminsk and Hartsell, 2002).

V position could be used in the stages of ankle rehabilitation when one needs an improvement in ROM. This position could also provide strengthening of GL muscle. V position could be more useful than Heel exercise for elderly people that may have a deficit in plantarflexion, although they do not have diminished dorsiflexion strength (Simoneau et al, 2006).

Co-contraction, besides joint stabilization, can be a cause of inefficient movement (Falconer and Winter, 1985). In situations where high dorsiflexion torque could result in impaired platarflexion torque, exercises with less co-contraction – V and arch position – could be best recommended.

SUMMARY/CONCLUSIONS

The *footwork* exercises could be used in the rehabilitation practice for a graded progression. Heel position could be indicated for greater stabilization of foot and ankle complex and Arch and V position exercises for an increasing in ROM and more challenge in stabilization of the foot and ankle complex. In situations where less co-contraction, less strengthening of TA muscle and more ROM are required, exercises Heel and V position should be employed.

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Table 1: Co-contraction Index (CI)

CI	Phase	Arch position	Heel position	V Position	p
TA/ GL muscles	**Extension**	0.75± 0.23	1.07± 0.20*	0.69± 0.24	<0.001[1]
	Flexion	1.09± 0.17	1.17± 0.15	1.15± 0.22	>0.269[1]

Table 2: Muscles EMG activity – Arbitrary Unit

Muscle	Phase	Arch position	Heel position	V Position	p
GL	**Extension**	0.55 ± 0.14	0.51± 0.15	0.57 ± 0.12	0.058[1&]
	Flexion	0.44± 0.16	0.54± 0.19	0.37 ± 0.13	<0.001[1&]
TA	**Extension**	0.35 ± 0.17	0.57 ± 0.15*	0.33 ± 0.16	<0.001[1*]
	Flexion	0.54± 0.14	0.72 ± 0.13*	0.53 ± 0.17	<0.001[1*]
PL	**Extension**	0.60 ± 0.13	0.62± 0.15	0.64± 0.10	>0.639[1]
	Flexion	0.53± 0.15	0.72± 0.15*	0.48± 0.13	<0.001[1*]

[1]Tukey Post hoc test (mean±SD). *Position significantly different. [&]Represents that all positions were different from each other

INFLUENCE OF FUNCTIONAL INSTABILITY IN ANKLE MUSCULAR ACTIVITY DURING A CUTTING MOVEMENT IN VOLLEYBALL PLAYERS

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INTRODUCTION

The ankle sprain is one of the most frequent sport injuries. Injuries most often occur during forward propulsion, jumping or cutting (Lynch et al., 1996). Lateral cutting movements are very frequent in a number of sports activities, particularly in volleyball. Typically, during such a movement the medial side of the shoe sole touches the ground first, producing a large lever relative to the subtalar joint axis (Stacoff et al., 1996).

Individuals with functional ankle instability (FI) usually complain about recurrent sprains during sport activities and report a subjective feeling of the ankle giving way. Thus, the FI is pretty significant for volleyball practice as it interferes with the cutting maneuver, one of the volley skills. Moreover, it is a condition that predisposes individuals to suffer new injuries of the lateral ligament complex of the ankle (Dayakidis and Boudolos, 2006). The alteration of the activities of the fibular muscles and the proprioceptive deficits have been cited as the main related factors to FI (Konradsen and Magnusson, 2000).

The purpose of this study was to compare the muscle activation patterns of selected lower extremity muscles of volleyball players with and without FI performing a side-shuffle cutting movement.

METHODS

33 professional volleyball players were studied: 16 with FI and no evidences of mechanical injury evaluated by the anterior drawer and talar tilt clinical tests (FIG, 20.4±3.7 yr); 17 without FI and no history of lower limbs injuries in the past 6 months (CG, 20.3±3.8 yr).

The tibialis anterior (TA), gastrocnemius lateralis (GL) and peroneus longus (PL) muscles were assessed unilaterally by means of surface electromyography during a side-shuffle cutting movement. The side-shuffle started with the subjects in a crouched position. The subjects shuffled to the side of the evaluated ankle, hitting a force platform after two shuffles. Then, after hitting the platform, the subjects returned, performing the movement as quickly as possible.

Electrodes were placed on the muscle belly, far away from the innervation zone (10 mm diameter, 25 mm inter-electrode distance center to center). EMG signals were analyzed during ground contact, determined by the vertical ground reaction force (GRF). GRF and EMG data were synchronically acquired and sampled at 1000Hz.

We calculated the linear envelopes and they were normalized in time (0-100% cycle of movement) and by the maximal voluntary isometric contraction (MVIC) of each muscle. The magnitude and the time of maximum peak occurrence in the linear envelopes were determined.

Groups were compared using T test for independent samples when normal distribution was present or Mann-Whitney test in case of non-normality (α=0.05).

RESULTS AND DISCUSSION

Results are displayed in Table 1. Subjects with functionally unstable ankles showed a later peak occurrence for gastrocnemius lateralis and a lower peak for peroneus longus muscle during cutting movement.

Neptune et al. (1999) stated that the PL decelerates the rapid supination that occurs in the foot-ankle complex after ground contact during cutting. They found that normal subjects present a high PL activity during ground contact, since the foot-ankle complex remains supinated. The PL is the main evertor muscle of the ankle joint and thus an important stabilizer against excessive inversion. If functionally unstable subjects present a lower activity of PL, they can be more susceptible to suffer ankle sprains.

The later occurrence of GL's peak may be representing a necessity to develop higher propulsion after impact in unstable subjects since the impact absorption probably occurs in the beginning of the ground contact phase and the propulsion phase occurs after that event, later in the movement cycle. If subjects with FI present an altered position during ground contact, for example, the ankle-foot complex will be less efficient, bringing the necessity of a higher muscular activity to produce more propulsion to get off the ground.

CONCLUSIONS

Volleyball players with ankle functional instability present a lower activation of peroneus longus during the cutting maneuver, increasing the chances of suffering an ankle sprain. Besides that, they also delay gastrocnemius lateralis activation that might be representative of an adaptative behavior to the condition of instability.

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Table 1: Linear envelope variables (% of movement cycle) for TA, PL and GL muscles (mean±SD)

Group	**Control group (n=17)**		**FI group (n=16)**	
	Peak magnitude (MVIC)	**Peak timing (%)**	**Peak magnitude (MVIC)**	**Peak timing (%)**
TA	1.3±1.6	70.4±29.2	0.9±0.9	69.2±24.2
PL	2.5±2.1	69.1±5.9	1.7±1.8 *	69.4±7.9
GL	1.2±0.7	54.1±22.9	1.5±0.71.6	69.0±7.0*

* Shows statistical significance ($p<0.05$)

EMG ONSET TIMING DURING A CUTTING MANEUVER IN VOLLEYBALL PLAYERS WITH FUNCTIONAL ANKLE INSTABILITY

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INTRODUCTION

The ankle sprain is one of the most common injuries in athletes, particularly in sports in which participants frequently jump and land on one foot and make sharp cutting maneuvers, like volleyball (Thacker et al., 1999). The most common complication following ankle sprains is functional instability. Functional ankle instability (FI) has been defined as a tendency for the foot to give away after an ankle sprain with no evidence of ligament injuries (Hertel, 2000). Coordination training has been shown to reduce the incidence of ankle sprains, suggesting that altered muscle activation and muscle function might be related to FI (Neptune et al., 1999).

The purpose of this study was to investigate the influence of FI in the onset latencies to ground contact moment of selected lower extremity muscles during a cutting movement in volleyball players.

METHODS

Thirty-three professional volleyball players were evaluated: 33 professional volleyball players were studied: 16 with FI and no evidences of mechanical injury evaluated by the anterior drawer and talar tilt clinical tests (FIG, 20.4±3.7 yr); 17 without FI and no history of lower limbs injuries in the past 6 months (CG, 20.3±3.8 yr).

The tibialis anterior (TA), gastrocnemius lateralis (GL) and peroneus longus (PL) muscles were assessed unilaterally by means of surface electromyography during a side-shuffle cutting movement. The side-shuffle started with the subjects in a crouched position. The subjects shuffled to the side of the evaluated ankle, hitting a force platform after two shuffles. Then, after hitting the platform, the subjects returned, performing the movement as quickly as possible.

Electrodes were placed on the muscle belly, far away from the innervation zone (10 mm diameter, 25 mm inter-electrode distance center to center). EMG signals were analyzed during ground contact, determined by the vertical ground reaction force (GRF). GRF and EMG data were synchronically acquired and sampled at 1000Hz.

The raw EMG signal recorded during ground contact was full-wave rectified and a continuous integration of all data was performed. The integrated EMG (IEMG) and the temporal period selected were then normalized and both the final IEMG value and the selection time were given the value of 1. The normalized IEMG trace was then compared to a reference line with slope equal to 1. The EMG onset latency was defined at the point in time when the distance between the normalized IEMG slope and the reference line was the greatest, and refereed to the instant of initial ground contact (Santello and McDonagh, 1998).

Each muscle onset value was compared using ANOVA within each group to determine if there is any difference among muscle activation. When significant effects were found the Scheffé *post-hoc* test was used. Groups were compared using T test for independent samples (α=0.05).

RESULTS AND DISCUSSION

Results are displayed in Table 1. Besides the muscle onset for TA, PL and GL were similar between groups, there were differences among three muscles onset values within the CG ($p<0.001$) and within the FIG ($p=0.001$). In CG, GL activated earlier (Scheffe test for GL and PL: $p=0.020$; for GL and TA: $p<0.001$). PL activated after GL, followed by TA. Unstable subjects showed the same pattern of activation, but there was statistically significant difference only between TA and GL (Scheffe test: $p=0.008$).

Table 1: Muscular onset activity (mean± SD) for TA, PL and GL (corresponding to the difference between the initial ground contact to the instant of muscular onset activity.

Muscle	**CG (n=17)**	**FIG (n=16)**	**p**
TA (ms)	31.8±36.1	30.8±42.4$^{\&}$	0.471
PL (ms)	50.9±31.6	51.3±39.3	0.489
GL (ms)	81.1±20.9*	76.2±36.1$^{\&}$	0.319
p	<0.001	0.001	

* represents the muscle statistically different from others, $^{\&}$ represents difference between TA and GL.

Players with instability showed an altered onset activity in relation to control players. Santello (2005) affirms that the onset of muscular activation is related to the instant of the expected impact; thus, if the onset activation occurred earlier in relation to initial foot ground contact there would be an increase in pre-landing activity, just like happened with GL among FIG players. The results suggest that the onset of GL activity is closer to the instant of ground contact in relation to what happened in the CG, suggesting a decrease in GL's pre-activity and, though, a decrease in the generation of extensor moments of the foot-ankle complex at initial ground contact (Suda et al., 2007). The GL has an important protection function of the foot-ankle complex before and after foot contacts the ground as their pre-landing activity will increase the joint stiffness before the mechanical upload occurs (Spagele et al., 1999). So, delayed GL onset activation as that one observed in instability subjects, could decrease the protection of the joint complex in players with FI.

CONCLUSIONS

Volleyball players with functional ankle instability present a decrease in gastrocnemius lateralis' pre-landing EMG activity duration, decreasing the protection of the foot-ankle complex before performing cutting movements, increasing the chances of suffering an ankle sprain during the maneuver.

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CONTRALATERAL EFFECTS FOLLOWING UNILATERAL SUBMAXIMAL ECCENTRIC AND CONCENTRIC CONTRACTIONS TRAINING

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INTRODUCTION

Unilateral training has effects on the strength of untrained homologous muscle of the contralateral limb. Greater effects were occurred in the untrained limb following 12 wk maximal eccentric contractions (ECC) training than concentric contractions (CON) training, and this effect was observed in the isometric contraction test as well as that of training mode (Hortobágyi 1997)

Possible causes of contralateral effects can be explained by neural adaptations with bilateral common central drives (Carroll 2006). Neural adaptations are occurred prior to the muscle hypertrophy. In fact, short term unilateral maximal isometric contractions training induced the strength gain of untrained limb (Laegerquist 2006). Unique activation strategies are appeared during ECC (Enoka 1996), and fast twitch motor units are recruited selectively during submaximal ECC. Therefore, submaximal ECC training may change motor units recruitment pattern differently from CON training.

In the commonly performed resistance training and daily life, CON and ECC contractions are achieved by uplifting or downing the weight with submaximal efforts. It is unclear that submaximal CON and ECC training achieved by using the weight have contralateral effects consistent with previous study. In this study, we want to observe training induced changes of isometric forces and median power frequency (MDF) of the electromyogram (EMG). MDF could be affected by motor units action potential shape and firing rate.

METHODS

Five volunteers were randomly assigned to CON training group (2 males) and ECC training group (2 males and 1 female). EMG was recorded from biceps brachii of both limbs using bipolar Ag/AgCl surface electrodes and amplified and filtered 10 – 500 Hz (MEB-2200, Nihon Kohden). Forces were recorded by using the load cell (LTZ-50KA, KYOWA). These data were converted from analog to digital at sampling rate of 4 kHz (Mac lab, AD Instruments) and recorded.

CON and ECC was achieved to rotate right cubital joint from maximal extension/flexion to flexion/extension. Trainings were performed 6 sets of 10 repetitions in a day, 3 times per wk for 3 wk using 10 or 12.5 kg dumbbells. Force and EMG was recorded 3 times during 3 sec voluntary isometric contraction before and after training. To determine MDF, Fast Fourier Transform was calculated from 2048 points EMG data filtered by the Hanning window function. All data were averaged and forces were normalized by pre-training value.

RESULTS AND DISCUSSION

Force was increased following CON training (105 %) and ECC training (101 %) in the trained limb. In the untrained limb, force was increased after ECC training (109 %) but not increased following CON training (99%).

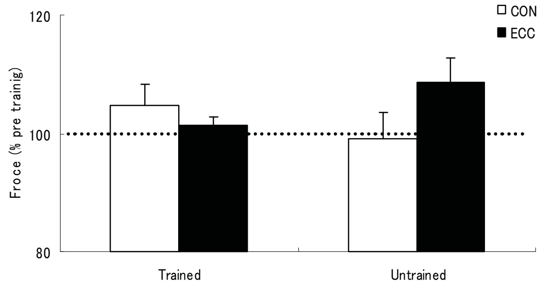

Figure 1: Changes of MVC forces.

MDF was almost same after CON training in the both limbs, on the other hand, was appeared slightly low frequency following ECC trained limb and slightly high frequency in the contralateral limb.

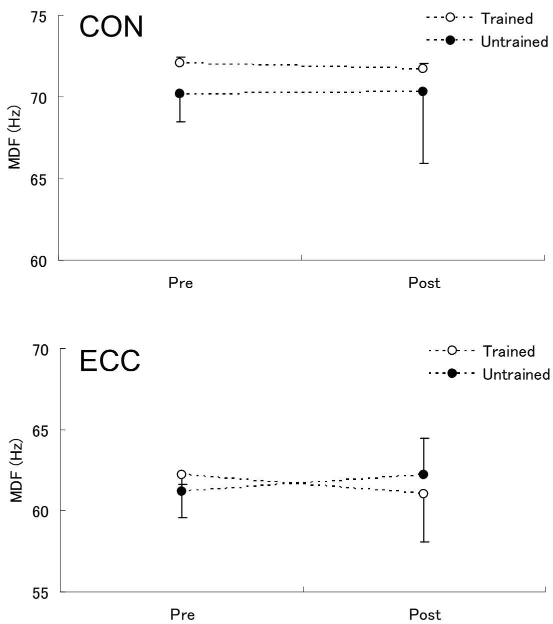

Figure 2: Changes of MDF.

About the force of contralateral limb, submaximal ECC and CON training effects might be similar trends to maximal trainings (Hortobágyi 1997). During submaximal ECC, excitabilities of spinal reflex and corticospinal tract are lower than CON (Sekiguchi 2003), in contrast cortical potential is greater during submaximal ECC than CON (Fang 2001). Consequently, greater corticospinal projections may be inhibited at the spinal level in the trained limb to avoid the reflexive muscle contraction to achieve smooth ECC. Recently, it was reported that excitability changes at the spinal level were not related with contralateral effects (Laegerquist 2006). Therefore, higher common central drives during ECC might not be depressed at spinal level in the untrained limb. These possibilities of differences of neural activations might induce more force increase observed following ECC training than CON training in the untrained limb and different MDP changes between ECC trained limb and untrained limb.

SUMMARY

Statistical power was too low because of the small number of participants in the present study, and we could not find the statistical difference from our results. Although there was statistical limitation to draw the conclusion in this experiment, we observed training effects as follows; 1) force might increase greater following ECC training than CON training in the untrained limb, 2) MDF might be changed differently between ECC trained limb and untrained limb.

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EMG DATA OF THE KNEE MUSCLES CHANGES DEPENDING ON THE DIRECTION OF PASSIVE RESISTANCE DURING THE CLOSED KINETIC CHAIN TASK

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INTRODUCTION

Knee instability about the valgus or varus and internal or external rotation supposed to be associated with serious injuries. Kaneko et al. reported that electromyographic (EMG) activity of the muscles around the knee joint against passive resistance changed depending on the direction of resistance during open kinetic chain task. The purpose of the present study was to examine whether the muscle activities depend on the direction of passive resistance during the closed kinetic chain (CKC) task.

METHODS

Ten healthy young men participated in this study. Each gave his informed consent for the experiment. His one side leg was examined.

The subject stood on the device that we produced, named “KINESTAGE”. His one foot was fixed on the foot-plate. We defined the squatting position at 60 degrees knee flexion as the CKC task. To explore the muscle activity, which changes depending on the direction of passive resistance, kinetic-equilibrating task was adopted. The resistance was applied like as a trapezoid shape, force was gradually increased 6 N per second. The peak force level was set at 42 N for 2 seconds.

Using KINESTAGE, passive force to the subject’s foot was applied from the eight directions [from anterior to posterior (AP), anterolateral to posteromedial (ALPM), lateral to medial (LM), posterolateral to anteromedial (PLAM), posterior to anterior (PA), posteromedial to anterolateral (PMAL), medial to lateral (ML), anteromedial to posterolateral (AMPL)]. Surface electrodes were used to record EMG data from 13 muscles around the hip and knee joints. EMG signals were calculated as average rectified value (ARV). In a respective subject, ARV data recorded during each bout of a direction was normalized (nARV) by ARV, which was maximum value in the single subject during passive moment applied from a certain direction.

RESULTS

The activity of the rectus femoris was found to vary with the direction of passive resistance (Fig 1).

Depending on the direction of passive resistance, the EMG data significantly increased in the rectus femoris, adductor longus, and gracilis during AMPL task (Fig 2). Similarly, the EMG data of the medial hamstring, lateral hamstring, medial head of gastrocnemius, adductor longus, and gracilis increased significantly during ML task (Fig 2, 3). Further, the EMG data of the gluteus medius increased significantly during the ALPM task, and that of the gluteus medius and tensor fasciae latae increased significantly during LM task (Fig 4).

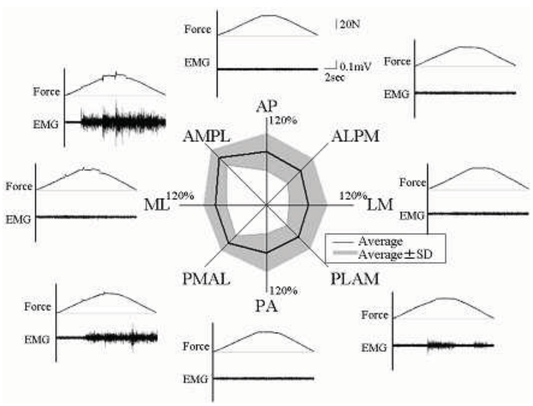

Figure 1: This radar chart shows average ± S.D. of nARV of the rectus femoris for all subjects. The graphs around the radar chart represent the typical EMG and force data.

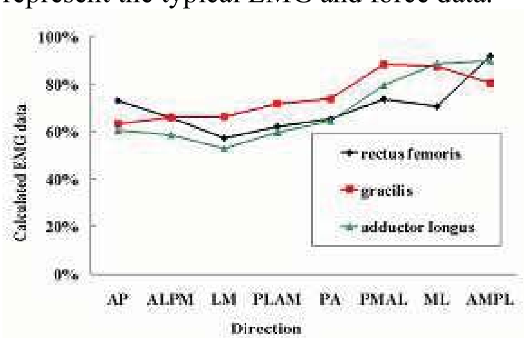

Figure 2: This graph demonstrates significant increases in the EMG data of the indicated muscles during the AMPL task.

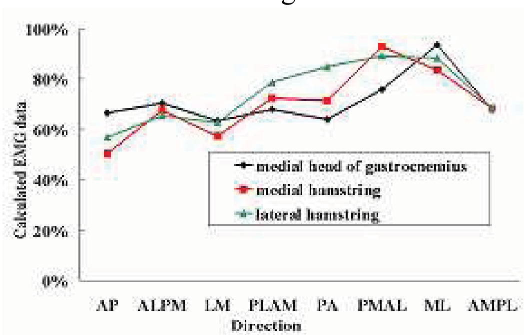

Figure 3: This graph demonstrates significant increases in the EMG data of the indicated muscles during the ML task.

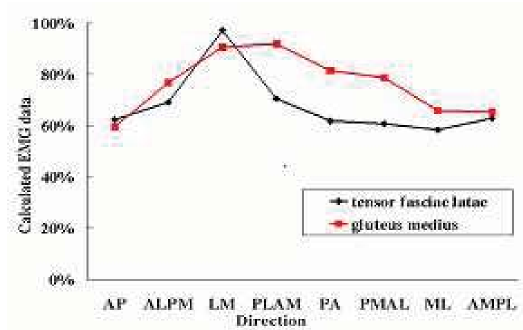

Figure 4: This graph demonstrates significant increases in the EMG data of the indicated muscles during the ALPM and LM tasks.

DISCUSSION

In the present study, we thought that AMPL and ML tasks were modeled on the knee valgus position. In these directions of passive resistance, the rectus femoris, medial hamstring, lateral hamstring, medial head of gastrocnemius, adductor longus, and gracilis activated higher than another direction. This result may suggest that those muscles were capable of preventing the knee valgus position. On the other hand, ALPM and LM tasks were regarded as the knee varus position. In these directions of passive resistance, the gluteus medius and tensor fasciae latae showed higher EMG activity than another direction. These data may suggest that these muscles work to resist the knee varus stress.

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EFFECT OF ACUPUNCTURE TREATMENT IN THE MASTICATORY MUSCLES OF INDIVIDUALS WITH TEMPOROMANDIBULAR DISORDERS

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INTRODUCTION

Acupuncture has become a treatment option in dentistry by promoting anti-inflammatory and analgesic actions. The analgesic action is probably multifactorial and the long-lasting pain relief reported after acupuncture treatment is still difficult to understand. It has been assumed that the inhibitory mechanisms are potentiated for activation of gene-related systems. The purpose of this study was to evaluate the functional characteristics of the temporal and masseter muscles on both sides (RT; LT; RM; LM) of individuals with temporomandibular disorders and to analyze the electromyographic activity of these muscles before and after therapy with ten sessions of acupuncture conducted once a week.

METHODS

All 17 patients, aged between 37–50 years (average 44.2. ± 4.84 years), average weight of 71 ± 9.45 kg, and average height of 1.64 ± 0.07 m) were examined clinically with regard to pain and dysfunction of the masticatory system and VAS scale was applied in each acupuncture session. The inclusion criteria were: the subjective report of temporomandibular joint pain that was primarily muscular in origin, pain for at least six months, and a clinical dysfunctional index in accordance with Helkimo. We excluded people with systemic conditions, arthritis, or a history of TMJ surgery or TMD treatment. The points of needling for the acupuncture were IG4, E6, E7, B2, VB14, VB20, ID18, ID19, F3, E36, VB34, E44, R3, and HN3. EMG analysis was performed before and after treatment using a MyoSystem-BR1 electromyographer with differential active electrodes (silver bars 10 mm apart, 10 mm long, 2 mm wide, 20x gain, input impedance 10 GΩ and 130 dB common mode rejection ratio). Surface differential active electrodes were placed on the skin, previously cleaned with alcohol, bilaterally on both masseter muscles and on the anterior portion of the temporalis. A ground electrode was fixed on the skin over the sternum. The electromyographic signals were analog amplified with a gain of 1000x, filtered by a pass-band of 0.01-1.5KHz and sampled by a 12-bit A/D converter with a 2 kHz sampling rate. The signals were digitally filtered with a pass band of 10 to 500 Hz. RMS data were collected during ten seconds at rest, clenching and during mastication and were normalized to a maximum voluntary contraction (MVC) for four seconds. The results were statistically analyzed using t-test (SPSS- 12.0-Chicago) to compare activation amplitude before and after treatment ($p<0.05$).

RESULTS

There was decreased EMG activity at rest after treatment (averages before RT = 0.19 ± 0.02; LT = 0.24 ± 0.06; RM= 0.19 ± 0.04; LM = 0.20 ± 0.03; after RT= 0.18 ± 0.02; LT = 0.21 ± 0.04; RM= 0.12 ± 0.02; LM = 0.10 ± 0.01), and increased EMG activity during mastication (average before RT = 0.67 ± 0.06; LT = 0.72 ± 0.07; RM= 0.69 ± 0.07; LM = 0.81 ± 0.10; after RT= 0.68 ± 0.05; LT = 0.74 ± 0.09; RM= 0.75 ± 0.09; LM = 0.86 ± 0.13), and clenching (average before RT = 0.67 ± 0.06; LT = 1.04 ± 0.11; RM= 0.17 ± 0.11; LM = 1.37 ± 0.20; after RT= 0.68 ± 0.05; LT = 1.10 ± 0.07; RM= 1.20 ± 0.11; LM = 1.54 ± 0.18). All subjects reported reduced pain after treatment measured by VAS scale (mean before = 8.5 and after treatment 0.0)

CONCLUSIONS

It was verified the acupuncture treatment decreased resting activity and increased activity during mastication.

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ACKNOWLEDGEMENTS

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HAMSTRING MUSCLE CONTRACTION AND TIBIAL POSTERIOR TRANSLATION UNDER ANTERIOR DRAWER TRACTION USING SLING FOR THE PATIENTS WITH PCL INSUFFICIENCY

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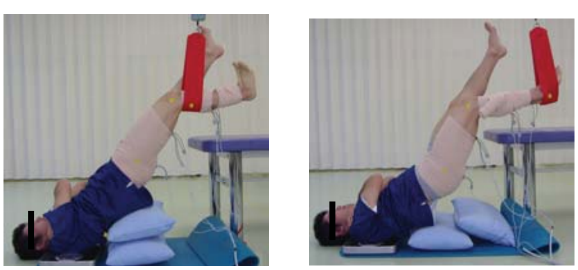
INTRODUCTION

Rehabilitation training of strong contraction of the hamstring muscle or deep knee flexion should be restricted for the patients following posterior cruciate ligament (PCL) reconstruction to avoid over-stress to the graft. It has been shown from our previous study that the average peak torque of hamstring muscle of the patients following PCL reconstruction surgery is much less than that following ACL reconstruction. Therefore, we developed novel hamstring muscle training method using a sling bridge for the patients following PCL reconstruction. The sling bridge exercise was performed by lifting the buttocks from the supine position with the lower leg being raised by a sling. (Figure 1)



Proximal traction Distal traction

Figure 1: Use of the sling bridge

The purpose of this study were to measure the extent of posterior displacement of the tibia on X-rays images and to determine the muscular activities of the legs using the sling bridge.

SUBJECTS AND METHODS

<Experiment 1: Measurement of posterior displacement of the tibia>
The subject was a patient with PCL insufficiency (32–year-old male, 174 cm height, 80 kg weight). Knee flexion angle with the sling bridge was set to 60 degrees. The posterior displacement of the tibia during sling bridge was evaluated by the lateral view of X-ray according to the Gravity Sag View method (step-off; SO, Shino et al. 2000) The difference of traction position in the lower leg was compared between proximal and distal. Traction force on the sling was also measured.

<Experiment 2: Measurement of muscular activities>
Muscular activities of the gluteus maximus (GM), biceps femoris (BF) and gastrocnemius (lateral head; GC) were measured by surface electromyography (EMG) for 5 healthy men. The sling bridge was used under the same condition as in Experiment 1. EMG was normalized relative to the value obtained during maximum voluntary contraction (% MVC).

RESULTS AND DISCUSSION

<Experiment 1; Figure 2>
The SO value of distal traction of the involved side was greater than that of the uninvolved side by 6.7mm. That of proximal traction of the involved was almost similar to that of the uninvolved (0.2mm difference). Traction force on the sling was 346N. These results indicate that the strain on the PCL decrease by proximal traction and the muscle training of proximal traction of sling bridge might be safe for the patients with PCL insufficiency.

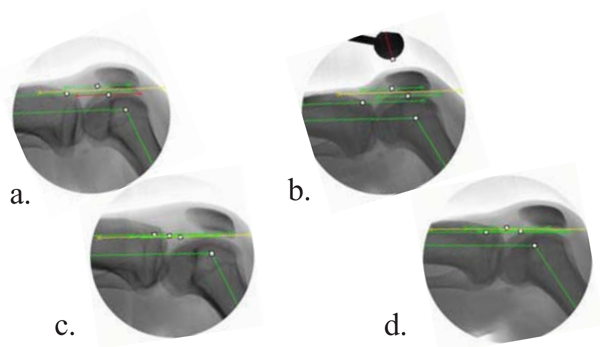


Figure 2: X-ray evaluation of the lateral view of the knee joint during use of the sling bridge.
a : Proximal traction in the involved side. SO 1.9 mm
b : Distal traction in the involved side. SO 9.4 mm
c : Proximal traction in the uninvolved side. SO 1.7 mm
d : Distal traction in the uninvolved side. SO 2.7 mm

<Experiment 2; Table 1>
The %MVC of the BF during use of the sling bridge with proximal traction was 50.2±20.4%. This result suggests that the sling bridge with proximal traction is useful to train a hamstring in the early stage after reconstruction.

Table 1: Mean percentage of maximum voluntary contraction (% MVC) of each muscle when using the sling bridge

	GM	BF	GC
Proximal traction	35.4±33.4	50.2±20.4	2.3±1.4
Distal traction	15.3±6.2	108.3±17.0	9.9±8.5

(mean value ± standard deviation)

SUMMARY/CONCLUSIONS

The extent of posterior tibia displacement by the sling bridge with proximal traction in a patient with PCL insufficiency was smaller than with distal traction and posterior tibia displacement of the involved side with proximal traction was similar to the uninvolved side.
The %MVC of the BF when using the sling bridge with proximal traction was 50.2±20.4%.

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EMG ACTIVITY OF THE MASTICATORY MUSCLES IN COMPLETE DENTURE WEARING INDIVIDUALS

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INTRODUCTION

Studies have found correlations between signs and symptoms of tempor-omandibular dysfunction (TMD) and factors relating to the overall quality of dentures and the occlusal status of dentures. Inadequate maxilo-mandibular contact, absence of retention and overall stability of the dentures may contribute to the development of TMD among complete denture wearers. The aim of this investigation was to study the electromyographic (EMG) activity of the masseter and anterior temporalis muscles in groups of individuals with complete dentures in different structural states.

METHODS

Group 1 was composed of nine subjects who had symptoms of TMD and had worn their prostheses for more than 8 years. Group 2 subjects had worn their prostheses for 3 months and didn't have symptoms. In this study muscle activity was recorded using a Myosystem Br-1 electromyography system with differential active electrodes (silver bars 10 mm apart, 10 mm long, 2 mm wide, 20x gain, input impedance 10 G< and 130 dB common mode rejection ratio). Surface differential active electrodes were placed on the skin, previously cleaned with alcohol, bilaterally on both masseter muscles and on the anterior portion of the temporalis. A ground electrode was fixed on the skin over the sternum region. The analog electromyographic signals were amplified with a gain of 1000x, filtered by a passband of 0.01- 1.5KHz and sampled by a 12-bit A/D converter with a 2 KHz sampling rate. The signals were digitally filtered using a bandpass filter of 10 to 500 Hz in the data processing. Data were collected while subjects wore their complete dentures and performed three ten-second trials of different jaw movements including holding a normal postural position, generating maximal tooth contact with parafilm and during dried grape mastication. The trial data were smoothed using a root means square window of 10 ms duration, and the mean activation amplitude for each trial was normalized by the activity recorded during maximum voluntary contraction induced by maximal jaw clenching. The groups were compared using normalized data as the outcome variables in independent samples t-tests (SPSS- 12.0- Chicago) ($p<0.05$).

RESULTS

The results are presented in Table 1 and revealed that Group 1, our subjects who had TMD and old dentures had significantly more muscle activity in their masseter and temporalis muscles than Group 2, our subjects who had newer dentures and no pain ($p<0.05$).

CONCLUSIONS

Our results suggest that individuals with older dentures and TMD have higher muscle activity at rest and during mastication then individuals with newer dentures and no TMD. Older dentures might have poor maxilomandibular contact which might lead to an increase in the required muscle activation to maintain a resting posture and to masticate.

Table 1. Normalized EMG (%Maximum voluntary activation) means of Groups 1 and 2 measured in right and left masseter muscles (RM and LM), in right and left temporal muscles (RT and LT) during rest position, grape and parafilm mastication.

	Rest				**Grape**				**Parafilm**			
	RM	**LM**	**RT**	**LT**	**RM**	**LM**	**RT**	**LT**	**RM**	**LM**	**RT**	**LT**
Group 1	11.01	8.57	14.22	9.59	43.66	41.48	32.99	32.98	81.44	67.53	49.62	44.89
Group 2	8.27	7.57	7.97	9.14	28.23	28.41	28.64	30.08	35.96	36.86	34.43	34.74

ELECTROMYOGRAPHIC AND BITE FORCE MODIFICATIONS CAUSED IN MASTICATORY MUSCULATURE OF BRAZILIAN WHITE CIVILIZATION WHEN COMPARED TO BRAZILIAN INDIGENOUS CIVILIZATION

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INTRODUCTION

Modern white populations have been frequently surprised by events of masticatory muscular collapse and this may be associated with destructive behaviours related to the lifestyle of this population. It might be useful to investigate the influence of modern lifestyle habits on the functioning of the somatognatic system. The purpose of this study was to compare electromyographic (EMG) data from the temporalis and masseter muscles between male and female genders and to correlate these data to maximum bite force in the molar and incisor regions, compare all results between dentate Brazilian indigenous individuals and white Brazilian individuals.

METHODS

A sample of 82 individuals (41 white individuals and 41 Xingu indigenous), all between 17 and 30 years of age, was used in all three analyses realized in this study. To be included, subjects had to have permanent complete dentition without signs and symptoms of temporomandibular disorders, and acceptable occlusion and facial pattern (type Angle class I) during mastication. The entire sample signed a free and informed consent form, in accordance with Resolution 196\96 by the National Health Control.

All individuals had EMG sampled from the right and left masseter and temporalis muscles using a Myosystem Br-1 electromyography system. The muscles were analyzed during mastication of raisins, peanuts and parafilm, and during the maintenance of postures including rest, cervical protrusion, and holding the mandible in left and right laterality positions. The data were smoothed using root mean square (RMS) windows of 10 ms and were normalized to the activity recorded during maximal clenching. To measure maximum bite force, a digital dynamometer model IDDK (Kratos – Equipamentos Industriais Ltda, Cotia, São Paulo, Brazil) was used, with a capacity of 1000N, adapted for oral conditions. Bit force measures were made at the first molar (right and left) and incisive regions.

All statistical analyses were performed using SPSS according to the specific purposes of this study. In the first evaluation, the white and Xingu samples were compared in terms of EMG amplitudes recorded during the mastication and postural tasks using t-tests. To compare the genders, a subset of each sample was selected. Thirteen males and females from each group were selected and EMG amplitudes recorded during the mastication and postural tasks were compared using t-tests within each group. Finally, 58 individuals were compared on a bite force test using Pearson's Correlations between bite force and EMG. The bite force of right masseter was correlated with EMG of right masseter, bite force of left masseter was correlated with EMG of left masseter, and bite force of the incisive region was correlated with EMG of right and left temporalis muscles.

RESULTS AND DISCUSSION

There was a statistically significant difference in normalized EMG amplitudes between the white and indigenous groups during rest (white = 0.08 ± 0.011, indigenous = 0.04 ± 0.007), chewing parafilm (white = 0.68 ± 0.059, indigenous = 0.49 ±0.036), chewing peanuts (white = 1.00 ± 0.12, indigenous = 0.51 ± 0.37), and chewing raisins (white = 0.69 ± 0.08, indigenous = 0.35 ± 0.038) ($p \leq 0.05$). White individuals showed higher EMG activity at rest and during chewing.

There was no difference in EMG activity of the temporalis and masseter muscles between genders while at rest or during postural movements ($p > 0.05$) but different muscle activation amplitudes were observed between the white and indigenous groups during chewing. White women showed greater right masseter activity than indigenous women and men and white men during chewing raisins (means - white women = 0.65 ± 0.09, white men = 0.58 ± 0.08, indigenous women = 0.45 ± 0.09 and indigenous men = 0.34 ± 0.06) and during peanut chewing. In the left masseter white men revealed greater values (means - white women = 0.77 ± 0.09, white men = 0.81 ± 0.14, indigenous women = 0.56 ± 0.09 and indigenous men = 0.47 ± 0.06), $p \leq 0.05$. The correlation analysis between EMG and bite force data revealed that when bite force increased, EMG decreased in all correlations in the indigenous group (negative correlations between maximum bite force and EMG). The only statistically significant correlation in the white individuals was between maximum bite force at the right molar and EMG of right masseter muscle ($r = 0.415$ to $p \leq 0.05$). In the white individuals, bite force generally increased as EMG increased.

CONCLUSIONS

The results show that indigenous peoples have lower muscle activation during masticatory movements and posture maintenance, which may be protective of the somatognathic system from conditions such as joint and muscle dysfunctions.
The white female subjects had the highest muscle recruitment to perform dynamic activities including chewing. In the indigenous population, there was a negative correlation (when bite force increased, EMG decreased) between bite force and EMG in the temporalis muscle. This differed from the white population where there was a positive correlation between bite force and EMG in all muscles. This suggests that temporalis muscle fibers are more moderately stimulated to increase bite force in indigenous persons. Modern lifestyle might have influenced recruitment of the temporal musculature in posture maintenance and in chewing.

ACKNOWLEDGEMENTS

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IMMEDIATE EFFECT OF THE SELECTIVE NEUROMUSCULAR ELECTRICAL STIMULATION OF THE VASTUS MEDIALIS OBLIQUE MUSCLE

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INTRODUCTION

The Patellofemoral Pain Syndrome (PFPS) is considered the most common knee condition encountered by orthopedic and sports medicine clinicians. PFPS is defined as an anterior or retropatellar knee pain in absence of other associated diseases.

Although the development of PFPS is multifactorial, the reduced motor unit activity of the Vastus Medialis Oblique muscle (VMO) is a commonly accepted hypothesis (Owings et al, 2002). Therefore, it would be desirable to have some way to selectively strengthen the VMO. However, several studies have demonstrated that the VMO muscle cannot be isolated during exercise (Cowan et al, 2002).

The Neuromuscular Electrical Stimulation (NMES) is widely used by physical therapist to increase muscular strength, endurance, working capacity and excitability (Delitto et al, 2002). Few objective data exists to support or refute the use of NMES to selectively strengthen the VMO and to enhance rehabilitation in patients with patellar dysfunctional.

The purpose of this study was to analyze the immediate effect of the NMES on VMO muscle by means of eletromyographic activity of the VMO and vastus lateralis (VL) muscles.

METHODS

Twenty women (mean age= 23± 2 years) with a diagnosis of PFPS participated of this study. Only female subjects were studied because of potential biomechanics differences between sexes. The procedures of the study included eletromyographic analysis of VMO and VL muscles, before and immediately after the neuromuscular electrical stimulation of the VMO muscle (each subject was exercised 10 times). For the NMES, a current generator Neurodim (Ibramed, Brazil) was used. Bearing wave frequency was of 2500 Hz, modulated in 50 bursts/s, with pulse duration of 200 µs, interburst interval of 10 ms. Two self-adhesive electrodes (5 x 10 cm) were attached to the subject involved thigh, on VMO muscle (Figure 01).

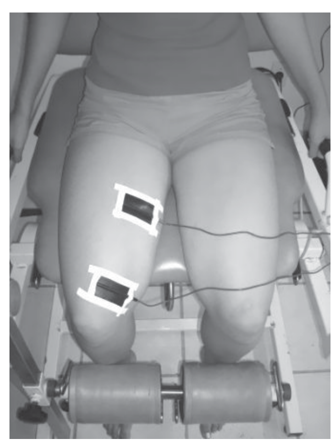

Figure 1: Electrode placement of the VMO muscle.

The applied amplitude was the maximum each individual supported at each session. Subjects performed maximal knee extension on an isokinetic dynamometer at 60 deg/sec. Electromyographic recordings of the VMO and VL were made by using surface electrodes before and immediately after the electrical stimulation. A reference electrode was placed over the fibular head of the untested leg. The detected EMG signals were amplified using EMG System (gain, 1000; bandpass, 20-500 Hz; EMGSystem, Brazil). The transmitted signals were input to an analog-to-digital circuit, digitalized at 1 KHz and stored. The signals of the VL and VMO were full-rectified and integrated. The results were expressed by relative difference VMO/VL. All data were analyzed by using Statistic Package for Social Sciences. The *P* value of 0.05 was accepted as reflecting statistical significance.

RESULTS AND DISCUSSION

The analysis of the data showed a significant increase in VMO activation intensity immediately after the electrical stimulation (p= 0,012), whereas VL activation intensity showed no significant increase (p= 0,92). Moreover, it was also verified a significant increase in VMO/VL relation after the NMES (p= 0,048, Figure 02).

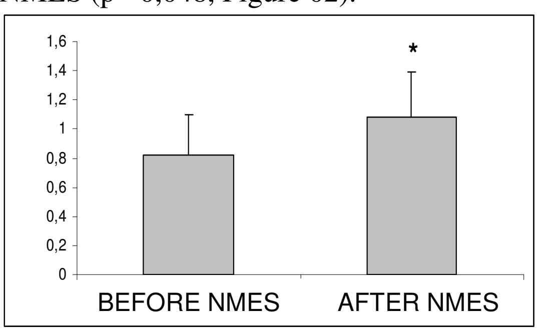

Figure 2: The normalized activation amplitude of VMO/VL muscles, before and immediately after the NMES.

Comparing previous studies, conclusions concerning the use of NMES are difficulted because of inadequate standardization of experimental procedures. It was demonstrated that neural factors were the major influence on strength increase during the application of electrical stimulation. It is also known that the activation of cutuaneous receptors with NMES facilitates motor unit activation of the muscles: electrical stimulation preferentially recruits the fast-twitch, associated with rapid movement (Delitto et al, 2002). Electrical stimulation possibly provided a greater overload to the VMO muscles, as revealed by EMG; therefore, the use of NMES offers a theoretically safe method to selectively strengthen the VMO.

SUMMARY/CONCLUSIONS

These data suggest that Neuromuscular Electrical Stimulation produced a significant modification in VMO/VL relation. Therefore, it may be used as selective strengthening technique of VMO muscle, in subjects with PFPS.

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LOCAL HEAT COMBINED WITH ELECTRICAL STIMULATION TO INCREASE SKIN BLOOD FLOW: A POTENTIAL MODALITY FOR WOUND HEALING

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INTRODUCTION

Electrical stimulation (ES) is a physical therapy modality which has been used to increase skin blood flow (SBF)(Cramp et al. 2002), especially as an aid for the healing of wounds (Petrofsky et al. 2005). Increasing skin blood flow by ES is a proposed mechanism for wound healing, (Petrofsky et al. 2005).

Recently, we have found that in a thermoneutral environment, ES induces only a small increase in skin vasodilatation and by increasing the temperature of the room by 10ºC, the SBF was significantly increased in response to ES (Almalty and Petrofsky, 2007). This may not be the case during local heating, which is controlled by the integrity of the skin blood vessels and sensory nerves. Nothing has been done to investigate the effect of local heating and ES on SBF, which may be a helpful method to improve wound healing. Therefore, the purposes of this study were to investigate how changing local temperature could affect the SBF during ES and to compare and these effects with those obtained with ES during whole body heating.

METHODS

A sample of 33 young healthy males (18-40 years) was divided into two groups: group W (N=15) who received the ES in the warm room protocol, and group L (N=18) who received the ES in the local heating protocol. In the worm room protocol**,** subjects completed two parts. The first part was to assess the response of SBF to ES in a thermoneutral environment (25º±0.5 ºC), and the second in a warm environment (35º±0.5ºC). Two carbonized rubber electrodes were and applied on the anterior aspect of the right thigh. The electrical current was applied for 15 minutes at intensities just below the threshold of muscle contraction. SBF was measured in each of three 15 minute periods: pre-ES, during, and post-ES using Laser Doppler Imager (LDI). In the local heating protocol, the subjects completed three parts. The first part was to assess the response of SBF to ES at a local temperature of 25º±0.05ºC, and at 35º±0.05ºC and 40±0.05ºC for the 2nd and 3rd parts respectively at room temperature 25º±0.5ºC. A secured Peltier system was placed between the two electrodes. The electrical current was applied for 15 minutes and the SBF was measured continuously using a single point scan in the center of the Peltier for three 15 min. periods: pre-ES, during, and post-ES using LDI for each part.

RESULTS AND DISCUSSION

No significant different found of the SBF (measured by flux unit) in the thermoneutral environment, whereas the SBFs were higher ($P<0.01$) during (198.5±83) and post ES (198.7±87.5) compared to pre ES (172.3±73.4) in the warm environment.

ES made no change to the SBF when local skin temperature kept at 25ºC ($P>0.05$). But, when the skin was locally heated to 35ºC, SBFs were higher ($P<0.05$) during

(199.6±103.8) and post ES (229.6±128.0) compared to pre ES (164.9±88.3). Likewise, during 40ºC of local heating, the SBFs were higher ($P<0.001$) during (617.1±433.8) and post ES (639.9±441.0) compared to pre ES (471.2±321.8).

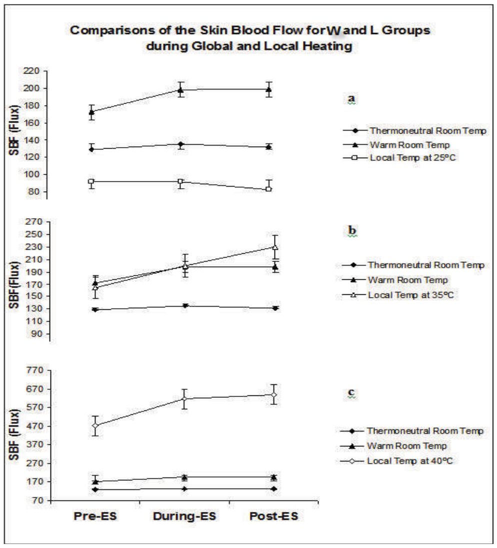

Figure 1: Comparisons of the SBFs response to electrical stimulation in thermoneutral and warm environments with the SBFs at local temperatures of 25ºC (a), 35ºC (b), and 40ºC(c) separately.

Comparing to the thermoneutral environment, the SBF in the thermoneutral environment was higher ($P<0.01$) than the SBF with a temperature of 25ºC of local cooling (Figure 1a). But, the SBFs with skin local temperature at 35ºC and 40ºC were higher ($P<0.01$) than the SBF in the thermoneutral environment (Figure 1a, 1b). Comparing to the warm environment, the SBF in the warm environment was higher ($P<0.01$) than the SBF at 25ºC (Figure 1a). No difference ($P>0.05$) was found between the SBF in the warm environment and the SBF at 35ºC (Figure 1b). The SBF at 40ºC of local heating, however, was higher ($P<0.01$) than the SBF in the warm room (Figure 1c). When the skin was locally heated to 35ºC, the SBF continued to rise post ES (Figure 1b), while the SBF reached a plateau during body heating. In the warn environment, sweating prevented skin temperature from a further increase. But local heating of the skin forces the skin to warm and sweating cannot cool the skin. Furthermore, when the skin was heated locally to 40ºC, the SBF significantly increased by about three fold greater than the SBF during body or local heating to 35ºC, and the maximum capacity of vasodilatation was reached during and post ES. Therefore, local heating is more effective, because to get the same SBF during body heating would necessitate considerable increases in core temperature. To achieve the same effect as local heating to 40ºC, we may need to warm the room to more than 40ºC. Also, it is encouraging to find that when the skin or the room are heated to about 35ºC, the same increase in SBF is obtained, while a threefold increase in SBF response occurs when the skin is locally heated to 40ºC.

SUMMARY/CONCLUSIONS

The use of ES with local heating, however, is a cheaper method and induced the same or better SBF compared to the use of ES in heated room. This led us to assume that the combination of local heating and ES is potentially an effective method to enhance wound healing. Therefore, further studies are needed to confirm this assumption.

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ANALYSIS OF HEART RATE RECOVERY TIME IN EXERCISE OF DEEP TRUNK MUSCLES USING TWO DEVICES

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INTRODUCTION

Presently, the exercise of deep abdominal muscles is more commonly utilized in rehabilitation medicine, as effective exercise of prevention and improvement of low back pain. There are several different types of the method of deep trunk muscles exercise. For example, it is a balance ball exercise and an instability board exercise and etc... There are many reports of effects of these exercises analyzed with electromyography of trunk muscles and magnetic resonance imaging. But the analysis of intensity and circulatory responses in these exercises are unreported. The purpose of this study was to investigate the influence of deep trunk exercises using balance ball and instability board on the circulatory response (heart rate recovery time) in these exercises.

METHODS

Subjects were six males and four females health volunteers with informed consent. The mean for (SD) age, height, and body mass were 22.2 (3.9) years, 166.9 (7.6) cm, and 57.5 (6.4) kg, respectively.
They performed the deep trunk muscles exercise using two kinds of devices. This exercise given to each subject was to keep balancing for twenty seconds with one raised upper extremity and one raised lower extremity, which was on the reverse side of the upper extremity. Each subject conducted the task alternately, right and left sides, and was instructed to continue the exercise for nine times repeated (Figure. 1-a). The exercise 1 (Ex. 1) was using with a balance ball as 600 mm in diameter (*Thera-Band Ex. Ball*, HYGENIC Co. Ltd, USA) , and the exercise 2 (Ex. 2) using with an instability board (*Pelvic board*, Pair-Support Co. Ltd, Japan), as 70 mm height and 250 mm in diameter (Figure. 1-b).
We recorded their heart rate and oxygen consumption throughout this entire trial period by the heart rate monitor (S810, POLAR Co.) and the respiratory analyzing system (AE-300s, MINATO Medical Science Co., Ltd.), and then analyzed influence of exercises to their heart rate recovery time. Their heart rate recovery time was determined as follows, it is when a heart rate after the exercise is recovered within one standard deviation (SD) heart rate at resting (Figure. 2). Exercise intensity was determined by oxygen consumption to metabolic equivalents (METs).
The results were assessed by the student t-test and differences at $p<0.05$ were considered significant. This study was conducted with approval of the Ethics Committee for Safety of Research, Tokyo Metropolitan University in Arakawa campus.

RESULTS AND DISCUSSION

The mean +/- SD values for METs were 2.2 +/- 0.3 (Ex. 1), 2.0 +/- 0.3 METs (no significant difference), respectively. This result was suggested that both exercises intensity were below their anaerobic

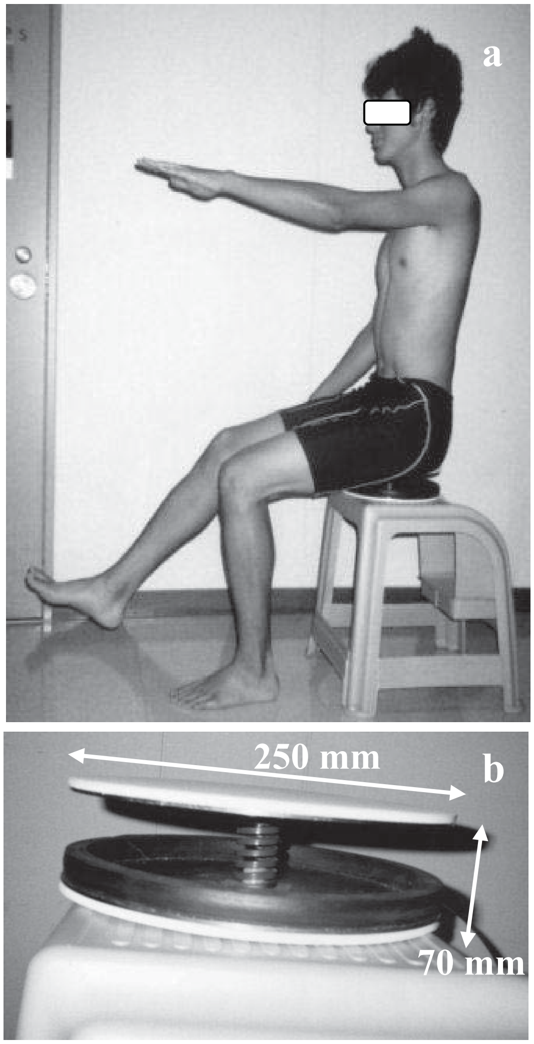

Figure 1: a; Method of deep trunk muscles, reverse side arm and leg elevating and holding in sitting.
b; Instability board using Ex.2 (*Pelvic board*, Pair- Support Co. Ltd, Japan)

threshold (AT). The mean +/- SD values for heart rate in resting and during exercise were 72.3 +/- 7.2 (resting), 87.6 +/- 12.0 (Ex. 1), 85.4 +/- 9.6 (Ex. 2) beats/ min (no significant difference), respectively. The mean +/- SD values for heart rate recovery time were 32.5 +/- 9.6 (Ex. 1), 33.3 +/- 11.7 (Ex. 2) sec (no significant difference), respectively.
These results were suggested that these exercises effects were not difference by using devices, and these exercise had a similar effects on activities of deep trunk muscles, demand for oxygen and autonomic nervous system activities.

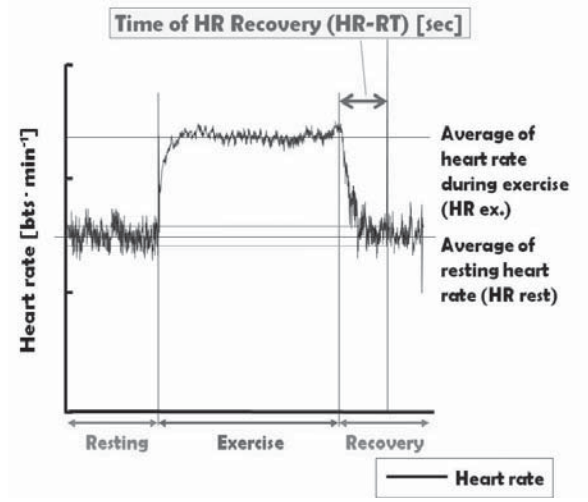

Figure 2: Heart rate recovery time after the exercise was detected as the moment when a heart rate had recovered within 1SD heart rate value during resting period.

SUMMARY/CONCLUSIONS

Physical exercise exerts an influence on increasing and decreasing heart rate. This study reveals that these exercises effects were not difference by using devices.
Deep trunk exercises using an instability board and a balance ball was enabled and equally-effective method on the circulatory response. These exercises were useful in rehabilitation and physical fitness, because these exercises intensity were under the AT level and moderate activities to deep trunk muscles and circulatory responses.

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DIABETIC NEUROPATHY PROGRESSION EFFECTS IN GROUND REACTION FORCES AND IN LOWER LIMB EMG DURING BAREFOOT GAIT

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INTRODUCTION

The purpose of this study was to investigate the influence of diabetic neuropathy and previous history of plantar ulcers, as a sign of progression of the disease, on electromyographical activity of the thigh and calf muscles and on vertical ground reaction forces during barefoot gait. We believe these biomechanical changes caused by diabetic neuropathy will lead to important alterations in the locomotor patterns that can predispose foot ulceration.

METHODS

This study involved 45 adults divided into three groups: a control group (CG; n=16), a diabetic neuropathic group (DG; n=19) and a diabetic neuropathic group with previous history of plantar ulceration (UDG; n=10). An electromyogram (EMG) of the right vastus lateralis (VL), lateral gastrocnemius (LG) and tibialis anterior (TA) were studied during the stance phase. The peaks and time of peak occurrence for TA, VL, and LG muscles were determined. Also, in order to represent the effect of the changes in EMG activity, the first and second peaks of the vertical ground reaction force (GRF) during stance phase were also determined. The data were acquired synchronized in 1000 Hz.

Inter-group comparisons of the EMG and GRF variables (peaks and times of EMG and peaks of GRF) were made using three MANCOVAs with the covariate time of stance phase, once this variable was different among the groups (α=5%).

RESULTS AND DISCUSSION

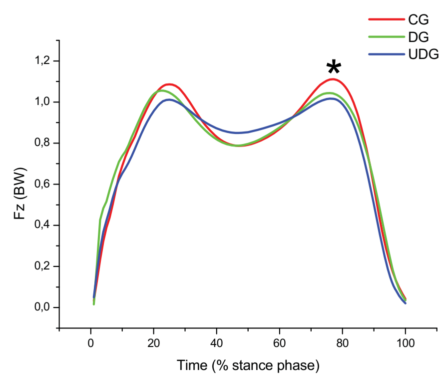

Figure 1 - Mean of GRF curves normalized according to each subjects body weight of the control (CG), diabetic (DG) and ulcerated diabetic (UDG) groups. *difference between CG and UDG independently of the total contact time ($p<0.05$).

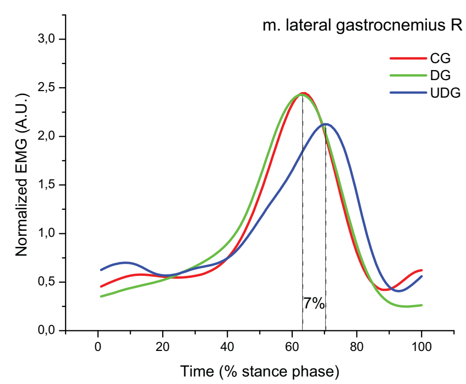

Figure 2 – Mean of linear envelopes of the

LG normalized according to the mean of the Control (CG), diabetic (DG) and ulcerated diabetic (UDG) groups. Note the 7% of delay of UDG.

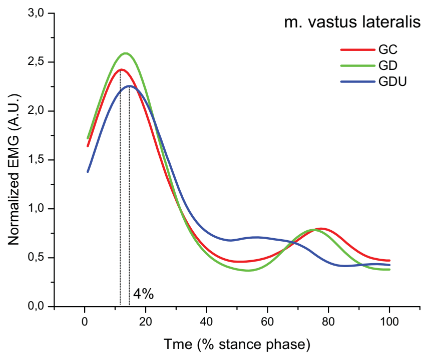

Figure 3 – Mean of linear envelopes of the VL normalized according to the mean of the Control (CG), diabetic (DG) and ulcerated diabetic (UDG) groups. Note there is a delay of 4% of UDG in comparison to CG.

There were no differences in the peak activation of any muscle, but the time of peak occurrence of LG (p=0.04) and VL (p=0.02) were delayed in the UDG compared to the other two groups.

The delay of the VL activation is in agreement with Sacco and Amadio (2003). The knee extensor muscles decelerate the knee flexion during load reception, working as a shock absorber by transferring part of the impact to the thigh muscle mass, but this mechanism maybe impaired in these patients.

Although there was a VL delay at the beginning of stance phase, changes on the first GRF peak that were expected did not occur. When compared to other locomotor tasks (such as walking on inclined walkways), the GRF during gait can be considered a biomechanical variable with low impact (Winter, 1991). Therefore, changes in VL activity are not yet causing many effects on GRF during gait in flat surfaces.

The delayed activity of the LG in the ulcerated diabetics may be indicative of propulsion inefficiency in gait motion. This fact is confirmed by the significantly lower second vertical peak observed among the diabetics who have already suffered foot ulceration. Lower ankle extensor moments in neuropathic subjects have already been described in the literature (Katoulis et al., 1997; Kwon et al., 2003) and it may be due to changes in ankle extensor activity, represented here by the the delayed LG activity during push-off phase.

The propulsion inefficacy could also lead to a longer contact time between foot and ground, which increases the exposure time of the plantar surface to loads. This higher exposure could be a predisposal factor for foot ulceration.

SUMMARY/CONCLUSIONS

The EMG time activation and GRF were influenced by the progression of diabetic neuropathy. These changes could be predisposal factors for foot ulceration.

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ALTERATION IN ACTIVITY PATTERN OF MASTICATORY MUSCLES OF CHILDREN WITH CEREBRAL PALSY

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INTRODUCTION

The basic pattern of mastication is produced by a brainstem central pattern generator that operates under the control of higher centers and that is subject to sensory feedback from the mouth, muscles and joints (Lund & Kolta, 2006). Cerebral Palsy (CP) is a disorder of voluntary movement and posture caused by damage to the developing brain and may have effect on mastication and mandibular function. The study of the electrical performance of masticatory muscles CP patients during mastication can help set the disorders of the oral motor function. This study had the objective of comparing the lateral displacement of the mandible during masticatory muscle activity of children with and without CP.

METHODS

The EMG activity was assessed from 12 normal children (7,91 ± 0,99 years) and 8 spastic CP children (9,5 ± 2,5 years). The activity of the anterior temporalis (AT) and masseter (M) muscles was registered with Myosystem Equipment of 12 channels, with 12 bit of resolution, with a high pass filter of 10 Hz and a low pass filter of 500 Hz and an acquisition rate of 2000 Hz. The activity was bilaterally detected using active differential surface electrodes (Ag) with a fixed inter-electrodes distance of 10 mm. The impedance was 10^{10} Ohms and a minimum Common Mode Rejection Relation of 90 db. The volunteers sit relaxed with arms on their legs, eyes opened, with the Frankfurt occlusal plane parallel to the ground. The skin was throughly cleaned with alcohol and muscle function test was performed before electrode placement. For AT (vertically along the anterior margin of the muscle) and M (2cm above of the external angle of the jaw) muscles the electrodes were positioned on muscle belly (parallel to muscular fibres). The reference electrode was placed over the sternum.

The sEMG amplitude processing used was the Root Mean Square (RMS) measured in µV. For normalization, the EMG potentials were expressed as a percentage of a maximum 1s RMS value obtained across the three repetitions of the isometric contraction for each muscle, and subject. The myoelectric signals were detected during the isotonic contraction (10s) in non-habitual chewing cycle and during the isometric contraction (5s) in maximal intercuspal position. Each volunteer carried out three repetitions with an interval of 1 min between repetitions. A Torque Index was derived from the EMG recording to estimate lateral displacement of the mandible: TI= abs[(ATright + Mleft) –(ATleft + Mright) / (ATright + Mleft) + (ATleft + Mright) X 100]. There TC of 0% when there is no torque during the activity to 100% when a major deviation side of the mandible. The data were not normality distributed (Shapiro - Wilk test) and the MannWhitney test was used to evaluate differences in EMG activity between groups. Analyses were performed using *SPSS*/Windows v. 13.0 with significance level of 5% ($p<0.05$).

RESULTS AND DISCUSSION

Absolute mean values and standard deviations of torque index and results of Mann Whitney test are reported in Table 1. Children with CP showed a pattern of activity of the muscles involved in mastication with greater mandibular lateral displacement.

Assessment of muscular activities during static and dynamic clenching tests could detect functionally altered occlusal conditions (Ferrario, 2000). In healthy subjects, a mandibular rotation on the horizontal axis may be counterbalanced by the actions of ligaments and other jaw muscle forces (Castroflorio, 2004). Other functional changes in the mastication system also can be found as presence of EMG activity in AT and M muscle during the opening phase of the jaw, hyperactivity of the AT muscle on M muscle and functional asymmetry of the stomatognathic system.
Alterations in function of the craniofacial muscles can establish changes in the facial skeleton and in the development of occlusion (Schievano, 1999). Conditions where an apparent good morphological situation in not related to a correct neuromuscular status (Ferrario, 2000). The establishment of alterations in the functional status of masticatory system can guide the multidisciplinary team in the methods and techniques more suitable to promote functional recovery.

CONCLUSIONS

Children with CP showed greater mandibular lateral displacement. Functional evaluation of the masticatory pattern of these children can help establish effective methods of intervention. Although this study has shown significant differences, the results must be considered to be preliminary because the numbers of patients tested were small.

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Table 1. Absolute mean values and standard deviations of torque index of normal children (12) and CP children (8).

Mandibular Contraction	Normal Children	CP Children	p
Isotonic Contraction	5.27 ± 4.12	13.08 ± 15.05	0.02*
Isometric Contraction	2.60 ± 2.88	6.13 ± 7.91	0.16

MannWhitney test: * significant difference between groups ($p<0.05$).

THE CHARACTERISTICS OF MOTION OF HEMIPLEGIC PATIENTS WHILE EATING WITH NON- DOMINANT HAND

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INTRODUCTION

Eating is an essential activity of daily life. Yamane defined the role of eating as (1) an essential activity of daily life (2) nutrition replenishment (3) a way to communicate with others (4) an enjoyment (5) needed three times a day. Eating is also a quality of life need for all individuals, but may be compromised by conditions that impair movement. It is one self-care task many patients can do independently in the recovery stage. However, although patients often can eat independently, they sometimes have difficulties.

The purpose of this study was to compare characteristics of movements (head and arm) of people with right hemiplegia with those of non-disabled people.

METHODS

Fifteen people with right hemiplegia (50～69years old) and twelve non-disabled people (40～67 years old) were invited to join our study. All were right handed.

The subjects were seated at a table where, in front of them, a bowl was placed. We adjusted the height of the table to the elbow of each subject.

The task was to eat yoghurt (90g) in a bowl with their left hands. The total task encompassed grasping the spoon and reaching with the spoon to the bowl, filling the spoon with yoghurt, and transporting the filled spoon to the mouth.

We recorded movement using light reflecting markers and four digital video cameras from four directions (left front, left back, right front and right back). We put reflective markers (diameter 20mm) on the vertex and left wrist. These kinematic landmarks were used to define the dependent variables. We calculated the total movement time, linear, velocity and acceleration with a three-dimensional video-based motion analysis system (Ariel Performance Analysis system; APAS-system, Ariel Dynamics Inc.). We also calculated peak velocity, the time to peak velocity and jerk cost as a measure of smoothness of movement. We compared them between disabled subjects with those of non-disabled subject using Wilcoxon sign ranked test. Significant difference was tested at the $P<0.05$ level.

The research ethics committee at the hospital approved the project. Informed consent was obtained from all subjects.

RESULTS AND DISCUSSION

The time, duration of task and number of repetitions increased in disabled subjects ($p<0.05$). Peak velocity (head and wrist) was lower in disabled subjects, and it was significantly different statistically ($p<0.05$). Peak velocity of head movement appeared first in non-disabled subjects, but appeared late in disabled subjects. Also, the movement of arm and neck appeared simultaneously in non-disabled subjects, but not in disabled subjects. Jerk cost was statistically higher in disabled subjects than non-disabled subjects, suggesting that smoothness of movement of disabled subjects was decreased.

Results indicate that people with right hemiplegia might have difficulty coordinating movement of the arm, head because of impaired control of movement involving complex co-ordination between the arm and head, and that it might be very important and helpful for therapists to address and facilitate neck mobility to eat food easily.

SUMMARY/CONCLUSIONS

We found that the characteristics of motion of people with right hemiplegia were;
(1) Reduced smoothness of neck movement;
(2) Uncoordinated movement between the arm and neck.

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We would like to thank all the subjects, who volunteered to participate in this study and stuffs, who assisted this study. We also thank Professor Shinkoda for his comments.

MUSCULAR ACTIVITY OF SPINAL ERECTORS AND RECTUS ABDOMINAL IN PATIENTS WITH ENCEPHALOPATHY DURING THERAPEUTIC RIDING

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INTRODUCTION:

Non Progressive Chronic Encephalopathy (NPCE), a disease from childhood that attacks the Central Nervous System, has a great notability, and for this reason, needs complementary therapeutic modalities that can favor an adequate evolution. This way, therapeutic riding and hippotherapy are some of the alternative ways for the treatment of these patients. With intuit of quantifying the effects of therapeutic riding, the surface electromyography is an option for the muscular analysis.

OBJECTIVE:

Analyze the activation level of rectus abdominal muscle and lumbar erector muscle, throughout surface electromyography, in patients with chronic encephalopathy, comparing adoption of postures in sole, elastic bed and therapeutic riding.

METHOD:

7 subjects with chronic encephalopathy participated of this study, age between 7 and 30 years old, and they were submitted to an assessment of muscular activation, throughout surface electromyography, in different positions over the horse, on an elastic bed and on sole. For data collection, the electrodes were placed according to SENIAM to capture the activation signals during 30 seconds, when each subject should keep the evaluated posture. The EMG signals were registered with a 4-channel EMG system (Miotool400 USB, Brazil), with a common mode rejection ratio >110 dB and a sampling rate of 2000 Hz by channel. The filtering of the raw EMG was performed with a filter Butterworth type, with a bandwidth of 25–500 Hz, spacing between electrodes fixed in 30mm. Surface electrodes of Ag/ClAg, round, pre gelded and auto adhesive from MEDITRACE**.** The results were analyzed considering the media value and the peak value of activation in each task, according to the analysis of Wilcoxon test and Friedman test, with significance level of 0,05 (5%). (Cram Kasman 1998; Basmajian & De Luca 1985).

RESULTS:

It was observed that orthostatism on elastic bed showed activation of 8,9 μV in rectus abdominal and 26,3 μV in lumbar erectors; on their knees on elastic bed there was an activation of 9,1 μV in rectus abdominal and 23,0 μV for lumbar erectors; on the horse there was a media activation of 13,0 μV for rectus abdominal and 28,6 μV for lumbar erectors; static on the horse showed an activation of 9,5 μV for rectus abdominal and 18,3 μV for lumbar erectors; on their knees on the horse there was an activation of 14,3 μV for rectus abdominal and 36,7 μV for lumbar erectors; gait on sole there was an activation of 18,1 μV for rectus abdominal and 36,4 μV for lumbar erectors.

Lumbar Erectors		Orthostatism	On Knees on Elastic Bed	Seat on Horse Walking	Seat on Horse Static	On Knees on the Horse
Average	On Knees on Elastic Bed	0,594				
	Seat on Horse Walking	0,397	0,433			
	Seat on Horse Static	0,826	0,778	0,149		
	On Knees on the Horse	0,064	0,124	0,109	0,133	
	Gait	0,008	0,001	0,003	0,084	0,730

Rectus Abdominal		Orthostatism	On Knees on Elastic Bed	Seat on Horse Walking	Seat on Horse Static	On Knees on the Horse
Average	On Knees on Elastic Bed	0,861				
	Seat on Horse Walking	0,233	0,133			
	Seat on Horse Static	0,382	0,638	0,004		
	On Knees on the Horse	0,001	0,009	0,198	0,041	
	Gait	0,001	0,026	0,019	0,001	0,730

DISCUSSION:

When comparing the postures in pairs, the postures performed on the horse required more muscular recruiting than the postures performed on elastic bed, and there was a similar recruiting when compared with gait.

CONCLUSION:

The study suggests that for this studied sample the therapeutic riding could be an alternative that favor the muscular activity and motor development in patients with chronic encephalopathy.

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ASYMMETRY INDEX OF MASTICATORY MUSCLE AS A PARAMETER FOR ANALYSIS OF THE TMD TREATMENT OUTCOMES

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INTRODUCTION

The therapy most frequently employed for the treatment of Temporomandibular disorder (TMD) is the occlusion splint (OS). Orofacial exercises have also been applied, with work directed at stomatognathic functions, i.e., orofacial myofunctional therapy (OMT), being included in certain approaches. The literature indicates that evidence is weak and further studies are needed (Michelotti et al, 2005).

Analysis of the asymmetry of muscle pairs on the right and on the left by surface electromyography (EMG) has been performed in order to identify the effects of the use of the OS, but we did not detect any studies using it in order to analyze the effects of the OMT.

The objective of the study was to assess the TMD treatment outcomes based on the asymmetry index (AI) of the mandible elevating muscles.

METHODS

Thirty subjects with articular TMD were randomly divided into 3 groups: 10 subjects treated with stabilization OS (OS group), 10 treated with OMT (OMT group) and 10 were used as controls with TMD (group CTMD). Eight subjects with no signs or symptoms of TMD represented the asymptomatic group (group C). The diagnosis was based on the RDC/TMD (Dworkin, Leresche, 1992). The EMG was recorded using an eight-channel surface electromyograph (Lynx Tecnologia Eletrônica -EMG1000). Active differential surface electrodes were positioned centrally and parallel to the direction of the fiber bundles of each masseter and temporal muscle. A reference electrode was placed on the patient's arm. The clinical conditions investigated were: (1) maximum voluntary dental clench and (2) maximum voluntary dental clench with cotton rolls for 5 secs.

The EMG signals were recorded and later calculated as muscle activity evaluated as root mean square (r.m.s.) of amplitude (µV). The asymmetry index (AI) between muscle pairs was calculated as described by Kibana, Ishijima and Hirai (2002); Saifuddin et al, (2003).
Analysis of variance (ANOVA) was applied for inter-group analysis, followed by the Tukey post-test. The intra-group phase (diagnostic x final) comparison was performed by the *t*-test for paired data. All calculations were made using the Statistica software, with the level of significance set at 0.05.

RESULTS AND DISCUSSION

ANOVA for AI data showed no significant differences in phase D between groups, between muscles, or an interaction effect ($p > 0.05$). In phase F there was a significant difference between groups {F= 5,85, p= 0,002}. The Tukey post-test revealed differences between OMT and OS (p=0.001), and between OS and C. The probability of equality between groups OMT and C increased from 26% to 92% from phase D to phase F.

Comparison of AI between phases: There was a decrease in AI between phases for both muscles in the OMT group, with a significant difference for the masseters ($t = 2.49$, $p < 0.05$). There was no significant difference in AI between phases in the OS, CDTM and C groups for the masseter and temporal muscles ($p > 0.05$).

In the present study, surface EMG was employed to determine the AI between pairs of mandible elevating muscles and to compare groups, as well as to analyze the effect of treatment. The higher the AI, the greater the disequilibrium of EMG activity between sides (Kibana, Ishijima and Hirai, 2002). Asymmetry has been reported to be present also in control subjects, but the indices are significantly higher in individuals with TMD (Alajbeg et al, 2003). In TMD the change in muscle recruitment may be a compensatory mechanisms for pain relief, or asymmetric recruitment may precede the development of the muscle pain symptom (Nielsen *et al,* 1990).

A goal of OMT was to equilibrate the mandible elevating muscles on the right and on the left. Another goal was to keep the functional space free under resting conditions, a fact that might have contributed to the interruption of the constant nociceptive stimuli coming from the occlusal asymmetry between the right and left sides. Phase comparison revealed a significant reduction of AI in the masseter muscle only in group OMT, indicating an improvement of the functional equilibrium between sides.

The increase of AI in the OS group may have been due to the bite deprogrammation caused by splint and consequently to conditioned avoidance of nociceptive stimuli. Alajbeg et al (2003) reported that, after OS treatment the level of temporal muscle asymmetry during clenching increased significantly.

SUMMARY/CONCLUSIONS

(1) the groups with TMD and the control group did not differ significantly during the diagnostic phase although the asymmetry index of group C was lower than for the other groups; (2) there were differences between groups in the final phase; (3) only the OMT group presented a significant difference in AI from the diagnostic to the final phase; (4) AI was useful to define therapeutic goals and conducts and to evaluate and confirm the results in an objective manner.

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COMPARATIVE SUDY OF ELECTROMYOGRAPHIC ACTIVITY OF VENTILATORY MUSCLES DURING MAXIMAL SUSTAINED INSPIRATION EXERCISE, FLOW AND VOLUME RESPIRATORY STIMULATOR

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INTRODUCTION

Evidences suggest that deep inspirations exercises could be so effective as respiratory stimulators (RS) (Oikonnen et al, 1991), which are contradictory in clinical practice when referring to the use of flow respiratory stimulator and volume respiratory stimulator (Azeredo 2000). Surface electromyography (EMG) is a non invasive method and very effective to analyze the respiratory muscular activity and its functions (Duiverman et al, 2004). The objective of this study was to investigate the activity of some ventilatory muscles as scalene (ESC), external intercostals (EI) and rectus abdominal (RA) during the utilization of a flow respiratory stimulator, volume respiratory stimulator and maximal sustained inspiration exercise throughout EMG. EMG MIOTEC model miotool 400 of 4 channels with 14 resolution bits, acquisition per channel of 2000 samples per second, 100x, filter Butterworth high pass 1 polo 0,1Hz and buterworth low pass 2 polo 500 Hz, spacing between electrodes fixed in 30mm. Surface electrodes of Ag/ClAg, round, pre gelded and auto adhesive from MEDITRACE. (Azeredo 2002; Oikkonen 1991)

METHODS

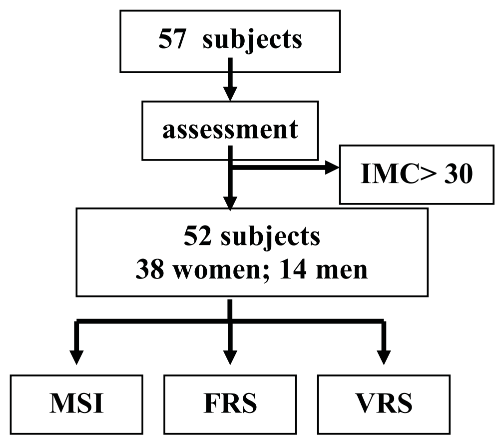

Figure 1: Delineation of the sample
Electromyography data were randomly collected during quiet breathing and pulmonary reexpansion exercises.

RESULTS AND DISCUSSION

It was observed statistically significant differences in the electromyography values for media and peak when comparing maximal sustained inspiration exercises to respiratory stimulators ($p<0{,}001$), however there wasn't significance between flow and volume respiratory stimulator for external intercostals ($p=0{,}672$) and scalene muscle ($p= 0{,}943$)

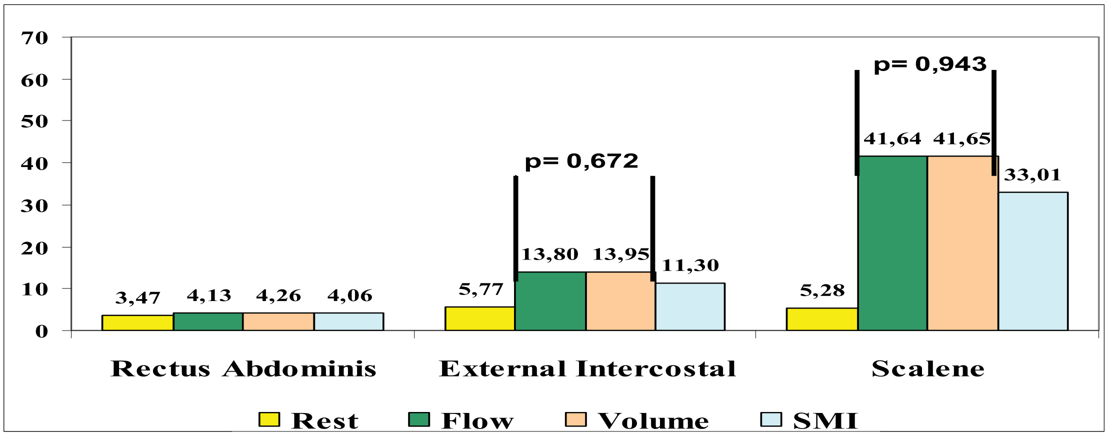

Figure 2: Comparison between analyzed muscles according to electromyographic values of media.

CONCLUSION

Maximal sustained inspiration exercises as well as respiratory stimulators provoked muscle recruiting in the analyzed respiratory musculature, once electromyography activity was higher when the devices were utilized. Rectus abdominals and external intercostals only assist on expiration, so it is not surprising that their activation does not differ much among the tests, whereas scalene are active in inspiration and therefore are more affected by the stimulators.

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LATISSIMUS DORSI AND ITS ROLE IN BREATHING PROCESS

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INTRODUCTION

The legitimacy of this theme is founded in the fact that latissimus dorsi muscle is continuously showed as an appendicular trunk muscle, which function would be hypothetically restricted to shoulder joint and upper limb cingulum, disregarding its possible action over the thoracic wall and consequently participation in breathing process.

OBJECTIVE

The goal of this study was to verify the recruiting of latissumus dorsi during deep inspiration, confirming its possible participation as an accessory breathing muscle.

METHODS

The subjects of this study were young adults, no smoking subjects, not showing any kind of pathological process in the airways. For electromyographic analysis it was used a MIOTEC model miotool 400 of 4 channels with 14 resolution bits, acquisition per channel of 2000 samples per second, 100x, filter Butterworth high pass 1 polo 0,1Hz and buterworth low pass 2 polo 500 Hz, spacing between electrodes fixed in 30mm. Surface electrodes of Ag/ClAg, round, pre gelded and auto adhesive from MEDITRACE**.** It was also utilized a volume inspiration stimulator, in order to facilitate the deep inspiration. Two active eletroctrodes are placed (2 cm apart) approximately 4 cm below the inferior tip of the scapula, half the distance between the spine and the lateral edge of the torso. They are oriented in a slightly oblique angle of approximately 25 degrees (Basmajian & De Luca 1985; Cram & Kasman 1998) The situations the subjects were measured were: (1) subject sitting, breathing freely; (2): subject sitting, breathing with the aid of the stimulator; (3): subject standing, breathing with the stimulator. In the three situations, the volunteer kept the upper limbs along the body, in order to isolate the action of latissimus dorsi over the shoulder joint. This present study disregarded the possible action of this muscle in different postures adopted during breathing with effort and its action in forced expiration, concerning only on its activation during deep inspiration (De Troyer & Estenne, 1998).

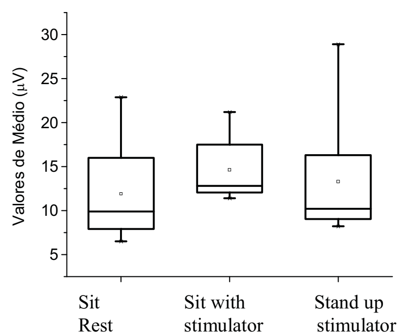
RESULTS AND DISCUSSION

According to results showed below, it was confirmed the action of latissimus dorsi during deep inspiration, specially in sitting position with the aid of a stimulator, what could be evidenced by a huge difference between peak values between the three situations, confirming the recruiting of this muscle in deep inspiration, according to the information in the table and graphic below:

Table 1: Descriptive and comparative analysis of peak value and mean value of latissimus dorsi muscle.

VARIÁVEL	N	MÉDIA	D.P.	MÍN	Q1	MEDIANA	Q3	MÁX	VALOR-P*
SENT_REP_PICO	9	70.82	24.13	46.20	54.30	65.30	84.80	116.00	**P=0.001** (X2=13.56; GL=2)
SENT_INC_PICO	9	92.54	52.22	53.10	58.70	**65.70**	119.20	195.30	
PE_INC_PICO	9	79.60	40.34	43.40	50.60	62.80	105.20	163.30	
SENT_REP_MED	9	11.90	5.39	6.50	8.20	9.90	15.20	22.90	P=0.097 (X2=4.67; GL=2)
SENT_INC_MED	9	14.63	3.51	11.40	12.20	12.80	15.70	21.20	
PE_INC_MED	9	13.30	6.84	8.20	9.60	10.20	12.70	28.90	

*P-value referring to Friedman test for related samples to compare peak and mean values between three situations (values in parenthesis are equivalent to χ^2 statistic and permission length of the test). Significant differences (Wilcoxon test): Peak (Sitting in rest # sitting with the stimulator; sitting with the stimulator# standing with the stimulator)



CONCLUSION

The participation of latissimus dorsi in deep inspiration was confirmed by the results of this study. This way, when an adequate bibliographic revision is done and data collection is made through surface electromyography and through a volume inspiratory stimulator , the study can contribute for the human biomechanic conditions to be enlarged, specially referring to a primordial act, the breathing.

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INFLUENCE OF POSITIONAL RELEASE THERAPY (PRT) ON TRICEPS SURAE MUSCLE DURING GAIT

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INTRODUCTION:

Positional Releasing Therapy (PRT) is an indirect technique, it places the body into a positional of greatest comfort and employs tender points to identify and monitor the sensibility or lesion. For this reason there isn't enough studies that could prove its effectiveness; therefore, triceps surae muscle was chosen, for being an important muscle, superficial, with high potency, and for performing a high muscular activity during gait.

OBJECTIVE:

The goal of this study was to verify the influence of PRT on the activity of triceps surae muscle during gait.

METHOD:

30 health subjects (16 men's and 14 women, with age between 19 – 24 years) without pain were submitted to an evaluation of sensibility points of the muscle. Once the point was found, the subject were submitted to an electromyography analysis of the muscular recruiting of gastrocnemius medialis muscle with a MIOTEC model miotool 400 of 4 channels with 14 resolution bits, acquisition per channel of 2000 samples per second, 100x, filter Butterworth high pass 1 polo 0,1Hz and buterworth low pass 2 polo 500 Hz, spacing between electrodes fixed in 30mm. Surface electrodes of Ag/ClAg, round, pre gelded and auto adhesive from MEDITRACE, and were placed on muscles according to Seniam recommendations. The subjects walked in a plane electric mat during 2 minutes. They were treated with PRT technique during 90 seconds and were submitted to a new electromyography analysis with the same parameters. (Cram Kasman 1998; Ambrogio & Roth 1997). These tender points are located in the superior portion of the medial and lateral heads of the gastrocnemius muscle, and the patient position of treatment is prone. The therapist stands on the side of the tender point, places his or her foot or knee on the table, and supports the dorsum of the patient's foot on the therapist's upper thigh. The therapist produces marked plantar flexion by compressing the calcaneus's cephalad toward the tender point and caudal traction by shifting the supporting thigh away from the tender point.

RESULTS:

In this study, it was used Wilcoxon test. Complementing the descriptive analysis, it was used confidence interval technique for media. Before the performance of PRT technique the muscular group showed a media of muscular activity of 133,1 μV, and after the application the media value decreased to 101,5 μV. The peak media of analysis before PRT was 960,5 μV, and after application was 836,0 μV. The reduction of the activity in percentage was 14,3% before and 12,7% after the technique, with p value was <0,001 for media and peak.

DISCUSSION:

The studies of Martins (2006) and Cyrillo et al (2006) showed a decreased muscular activity after the application of PRT technique, and the same was possible in this present study, where PRT had an important influence in decreasing the muscular recruiting of gastrocnemius medialis muscle during gait.

CONCLUSION:

From these results, it is suggested that when PRT is applied in a right way, it results in a decrease of recruiting and muscular activity of triceps surae muscle during gait, obtaining significant results in values reduction before and after the application of the technique.

KEY WORDS: Electromyography, gastrocnemius muscle, gait, positional releasing therapy

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FORCE AND SURFACE ELECTROMYOGRAPHY PATTERN OF SHOULDER MUSCLES IN WOMEN SUBMITTED TO PATEY MODIFIED RADICAL MASTECTOMY

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INTRODUCTION

Breast cancer is the most common disease reported in Brazilian women and the principal cause of death in developed countries nowadays (Brasil, 2006). Among the modalities of treatment, some involve surgical procedures, which are classified as conservative and radical ones, which are related to Mastectomy and Patey Modified Radical Mastectomy (PMRM). PMRM means the extraction of *Pectoralis Minor* muscle, which can also be accompanied by a lesion of long thoracic nerve, usually observed after surgery in Brazil. Therefore, it is suggested that some biomechanical changes in shoulder and scapular waist must be presented in these patients, which also includes winged scapula (Sakarofas, 2006). However, no previous study was found about the effects of PMRM in shoulder and scapula movements by means surface electromyography (sEMG).

Thus, the aim of this study was to evaluate the sEMG pattern from *Serratus Anterior* (SA), *Upper Trapezius* (UT) and *Deltoideus Medius* (DM) muscles in three simple tasks in women submitted to the PMRM.

METHODS

Eleven patients (60.7 ± 8.03 years old) from the National Cancer Institute, Brazil, were evaluated 48 hours before (T1) and 30 days after (T2) PMRM by means physical examination, and sEMG signals acquisition from SA, UT and DM muscles while performing maximal isometric voluntary contraction (MVC) (Figure 1), which was also qualitatively evaluated through a force graduation scale (0 [low] to 5 [normal]).

The acquisition system was consisted by an electromyography (Model 800C - EMGSystems Ltd., Brazil) and a computer. The sampling frequency and gain were set at 1 kHz and 1000, respectively. Surface electrodes (Ag/AgCl - 3M Korea Ltd., Seoul, Korea; 1 cm of diameter) were placed over the muscles previously cited following SENIAM protocol (Hermens et al., 2000) and Li et al. (2005). The three muscles were evaluated through three positions as presented in Figure 1.

(a) (b) (c)

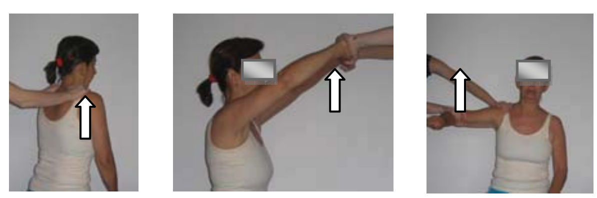

Figure 1: Positions adopted by the patients for evaluating force and collecting sEMG signals from the three muscles (**a.**UT; **b.** SA; **c.** DM). The arrows show the directions of movement.

To compare statistically T1 and T2, it was applied the *Wilcoxon* test ($\alpha = 0.05$).

RESULTS AND DISCUSSION

UT and DM muscles presented an increasing behavior, while SA presented a different tendency. However, there was significant statistical difference (P = 0.009) between T1 and T2 only for SA muscle (Figure 2).

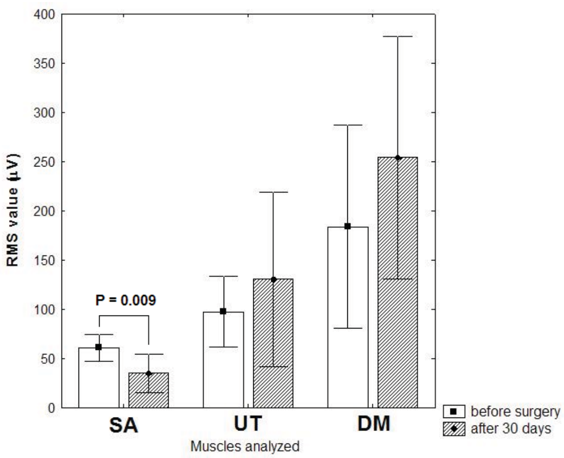

Figure 2: RMS values (Mean ± SD) for the three muscles before (T1) and after (T2) surgery.

There were also no significant statistical differences (P > 0.05) between T1 and T2 for force levels in the three conditions.

The results suggest that the sEMG signal activity from SA muscle, which is innervated by the long thoracic nerve, plus the absence of *Pectoralis Minor* muscle, contributed to the presence of winged scapula in six of eleven patients. The decreasing of sEMG signal activity from SA muscle is ratified by the literature that reinforces this nerve lesion as a common disorder in patients submitted to PMRM (Lhmkuhl and Smith, 1987).

A disability of SA muscle seems to conduct to an increasing of UT sEMG activity when shoulder flexes. This behavior probably occurs as a compensatory strategy of UT of taking over SA and the absence of *Pectoralis Minor* muscle for flexion movement (Ekstrom, 2003).

Regarding to DM muscle, the increasing in its sEMG activity can be related to the unbalance of SA and UT in providing scapula rotation mainly from 90^o of abduction and conducting DM to higher levels of recruitment. When one of these muscles is not able to perform its normal function, DM contracts and helps to rotate the scapula.

SUMMARY/CONCLUSIONS

The preliminary results suggest a possible reorganization in muscles conscriptions of the complex of shoulder and scapular waist provided by the absence of *Pectoralis Minor* muscle as well as a probably incidence of lesion of long thoracic nerve. However, despite this pattern did not seem interfere significantly in patients' force levels, it is not possible to assert for while that daily activities are not impaired.

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EMG OF SPINOCEREBELLAR DEGENERATION PATIENT AND NORMAL SUBJECTS DURING WOLKING

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INTRODUCTION

Spinocerebellar: SCD degeneration patients muscle strength is decrease little by little because of progress of disease. Patients did not sit up, stand up, and walking, when their muscle strength is weakness. It is difficult to distinguish muscle weakness between progress of disease and disuse syndrome. It is important for rehabilitation that patient muscle strength keeps as much as possible. Therefore medical staff must prevent progress of disuse syndrome.
Muscle strength exercise is one of the methods of patient in disuse syndrome. Physical therapist use knee ankle foot orthosis during walking exercise for treating muscle weakness of lower extremity. Gait with orthosis is easy exercise for patient and physical therapist. However, it is unknown that activity of electromyogram during gait with orthosis and without orthosis. The purpose this study was to determin EMG of lower extremity muscle activity during walking with orthosis and without orthosis for SCD patient and normal subjects.

METHODS

Subjects
SCD patient is 80 years old and male. His height is 160 cm and his weight is 60 kg. His on set is 9 years ago, and he did not walk himself recent 3 years. From 2 years ago, he could not move in bed himself. Eight young female subjects participated in this study. They did not have any neuromuscular disease and problem of bones and joints. Their mean age is 24.7 years old, and SD is 3.1. Their mean height is 158 cm, and SD is 4.7. Their mean weight is 51.9 kg and SD is 7.5.
All subjects understood informed consent in this study.

Methods
Electromyography was detected from 5 muscles, vastus lateralis, vastus medialis, rectus femoris, tibialis anterior, and gastrocnemius lateral head. EMG was sampled by bipolar surface electrodes and sampling frequency is 1 kHz. The data was stored by computer for calculating RMS. SCD patient performed kicking during sitting on wheel chair, walking with knee braces, walking with knee ankle foot orthosis, walking without any orthosis. Normal subjects performed walking with and without knee braces.

RESULTS AND DISCUSSION

Figure 1 and table 1 showed that maximum EMG activities of SCD patient's during kicking and gait exercise. Maximum EMG activities of vastus medialis and vastus lateralis during gait without brace were highest than other two gaits. EMG activity of rectus femoris during gait with knee ankle foot orthosis was highest than other two gaits. Kicking exercise was not enough to activate quadriceps muscles. It was better that walking exercise without brace was more activate quadriceps muscles. Knee brace and knee ankle foot orthosis kept these joints position during gait exercise.

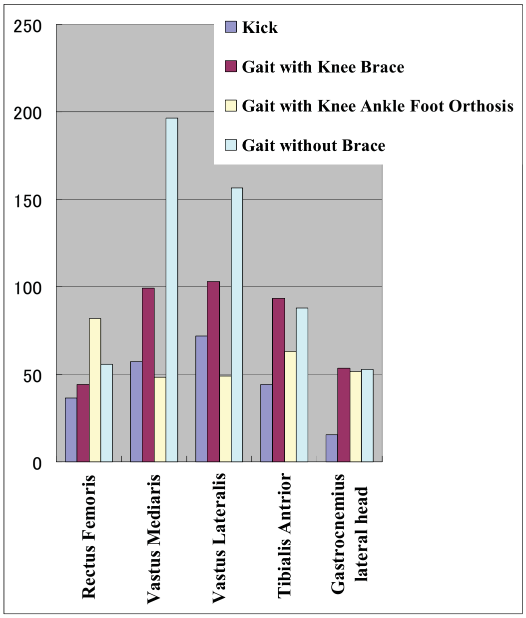

Figure 1: This graph illustrates maximum EMG activity of SCD patient 5 muscles, rectus femoris, vastus medialis, vastus lateralis, tibialis anterior, and gastrocnemius lateral head. Maximum EMG activity was observed during gait without any brace.

Therefore EMG activities were decresed during gait with any brace. During gait with knee ankle foot orthosis, patient could flexd hip joint only, so rectus femoris activity was higher than other muscles. During gait with knee brace, patient must keep his ankle joint to maintain dorsiflexion position, therefore tibialis anterior EMG activity was increased. EMG activities of normal subjects were as same as SCD patient. During gait with knee brace, muscle activity was decreased. It is as same as patient gait with and without barace. Physical therapist must choice the gait exercise without any brace for quadriceps muscle strengthing excericise.

SUMMARY/CONCLUSIONS

The purpose this study was to determin EMG of lower extremity muscle activity during walking with orthosis and without orthosis for SCD patient and normal subjects. EMG was detected from vastus lateralis, vastus medialis, rectus femoris, tibialis anterior, and gastrocnemius lateral head during gait with and with out brace. Maximum EMG activities of vastus medialis and vastus lateralis during gait without brace were highest than other two gaits.

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Table 1: Maximum EMG activity of SCD patient 5 muscles, RF; rectus femoris, VM; vastus medialis, VM; vastus lateralis, TA; tibialis anterior, and GL; gastrocnemius lateral head.

maximau EMG activity	RF	VM	VL	TA	GL
Kick	36.7	57.4	71.9	44.2	15.4
Gait with Knee Brace	44.3	99.3	102.9	93.4	53.4
Gait with Knee Ankle Foot Orthosis	82.0	48.2	49.1	63.2	51.8
Gait without Brace	55.9	196.5	156.5	87.8	52.9

ANALYSIS OF DROP FOOT GAIT PATTERNS IN CHILDREN WITH HEMIPLEGIC CEREBRAL PALSY FOLLOWING AN ANTERIOR TIBIALIS HOME BASED NEUROMUSCULAR ELECTRICAL STIMULATION (NMES) STRENGTHEING PROGRAM

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INTRODUCTION:

The purpose of this study was to determine whether a home based NMES strengthening program for the muscles innervated by the peroneal nerve has a carryover effect on ankle dorsiflexion during swing phase in children with spastic hemiplegia.

METHODS:

The study was designed as prospective and randomized, comparing a treatment group versus a placebo/control group with before and after trial screening.

Twelve subjects with cerebral palsy spastic hemiplegia were recruited to participate in the study. Inclusion criteria required all subjects to have a Type I hemiplegic pattern with only dysfunction at the ankle, and typical motion at the hip and knee. All were independent ambulators and demonstrated a drop foot during swing phase. Inclusion also required minimal tightness of the ankle plantar flexors (0° minimum passive range of ankle dorsiflexion).

Each subject followed a six week home exercise program addressed at strengthening the muscles innervated by the peroneal nerve (see Figure 1). The treatment group (n = 7) performed the exercises with NMES facilitation, the control group (n = 5) performed the same exercises without NMES facilitation. Three dimensional kinematic data, electromyography data (EMG) and clinical measurements of strength and range of motion were collected for each subject during barefoot ambulation prior to and immediately following the 6 week exercise protocol[1].

WEEK #	RATE (pps)	Amplitude (Intensity of Stimulus)	ON-OFF TIMES	ON RAMP (seconds)	ACTIVITY DESCRIPTION
1	25	Pt. Tolerance to sensory level	1:6 2 sec:12 sec	2	Sitting 5 reps, long sitting 5 reps, standing with stool 5 reps, standing in tandem
Return to CCMC PT first day of week 2 to ensure proper electrode placement, and to adjust parameters					
2	35	Pt. Tolerance to muscle contraction	1:5 2 sec:10 sec	2	Sitting 10 reps, long sitting 10 reps, standing with stool 5 reps, standing in tandem 5 reps
Return to CCMC PT first day of *week 3 to ensure proper electrode placement, and to adjust parameters*					
3	35	Pt. Tolerance to muscle contraction	1:2 4 sec:8 sec	2	Sitting 10 reps, long sitting 10 reps, standing with stool 10 reps, standing in tandem 10 reps
4	35	Pt. tolerance to muscle contraction	1:2 4 sec:8 sec	2	Sitting 15 reps, long sitting 15 reps, standing with stool 10 reps, standing in tandem 10 reps
5	35	Pt. tolerance to muscle contraction	1:2 4sec:8 sec	2	Sitting 15 reps, long sitting 15 reps, standing with stool 15 reps, standing in tandem 15 reps
6	35	Pt. tolerance to muscle contraction	1:2 4sec:8 sec	2	Sitting 20 reps, long sitting 20 reps, standing with stool 15 reps, standing in tandem 15 reps

Figure 1: Home exercise program

RESULTS:

Each subject served as his/her own control and therefore paired T-tests were performed to assess significant differences ($p < 0.05$) pre versus post-treatment. Subjects were assessed for changes in sagittal plane ankle motion during the swing phase of gait:
1) peak ankle dorsiflexion during midswing (DF MSW), and 2) ankle dorsiflexion at terminal swing (100% of the gait cycle) (DF TSW).

*Table 1: Comparison of parameters between treatment group and control group. (* indicates significant changes pre versus post-treatment)*

	DF MSW (°)		**DF TSW** (°)		**A.T. burst duration (%)**	
	Treatment Group	Control Group	Treatment Group	Control Group	Treatment Group	Control Group
PRE	-4 ± 6	-5 ± 6	-10 ± 4	-13 ± 3	62 ± 10	71 ± 7
P0ST	-4 ± 6	-10 ± 4*	-9 ± 5	-16 ± 4*	62 ± 14	69 ± 12

The firing pattern of the anterior tibialis was also assessed. The duration of the anterior tibialis burst was measured as a percentage of the swing phase (A.T. burst duration) (see Figure 2).

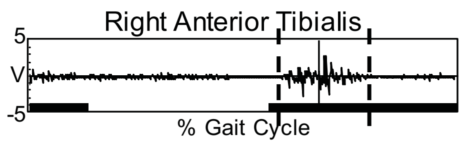

Figure 2: Example of typical anterior tibialis burst duration (A.T. burst duration = ((A.T. $_{off}$ % gait cycle– A.T. $_{on}$ % gait cycle)/(100% - Toeoff % gait cycle))

There were no significant differences between the two groups prior to the participation in the study with respect to mid swing or terminal swing sagittal plane ankle motion (p=0.75 and p=0.31), or anterior tibialis activity (p=0.15)(see Table 1).

There were no significant improvements in dorsiflexion during swing in the treatment group following the NMES program. The control group experienced a significant decrease in DF MSW and DF TSW (see Table 1).

Neither group experienced a change in duration of anterior tibialis recruitment during swing phase (see Table 1).

SUMMARY/CONCLUSIONS:

NMES treatment programs are often recommended for patients with TYPE I hemiplegic patterns with a goal of improving swing phase dorsiflexion and becoming brace free.

The results of this study indicate that there were no significant improvements in the firing pattern of the anterior tibialis or associated improvement in the sagittal plane ankle motion following the 6 week NMES treatment program. Limitations in this intervention/treatment program may be:

1) The exercise program used in this study may be insufficient (time and intensity) to obtain beneficial effects from the NMES treatment.

2) Gait training with NMES may be necessary in order to see changes in ambulatory patients.

3) Ambulation is an activity largely controlled by central pattern generators at the spinal cord level and therefore may not be affected by the isolated training of the anterior tibialis[2].

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ESTIMATION OF CONTROL STRATEGIES ADOPTED BY ELITE FES ROWERS

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INTRODUCTION

Functional electrical stimulation (FES) rowing was initially proposed as an alternative form of total body exercise for paraplegics in order to increase fitness and reduce mortality due to cardiovascular disease (Wheeler 2002). It has developed to a stage where teams regularly compete in national and international indoor competitions, and FES sculling on water has been successfully demonstrated. The effectiveness of FES rowing as exercise is well established (Hettinga and Andrews 2007; Velleren et al. 2007).

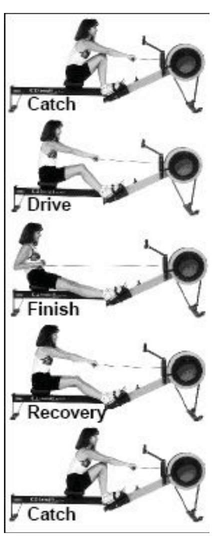

Figure 1: Phases of a complete FES rowing cycle.

The FES rowing cycle, shown in Figure 1, is performed by the coordination of voluntary upper body movements with FES generated movements of the rower's paralysed legs. While athletes have trained themselves to perform very effectively by developing rowing cycles through practice, until recently there has been little experimental data objectively detailing the control strategies they adopt. To address this, we have developed a data acquisition system to measure the many relevant variables and used it to study the performance of an elite FES rower (age = 51yr, bodyweight = 70kg, injury level = T4/ASIA A, time since injury = 6yr, total FES rowing training = 4yr), by phase plane analysis, the results of which we present here.

METHODS

A Concept 2 indoor rowing machine has been modified to provide increased trunk support, constrain leg motion to the sagittal plane and protect knee joints against impact. Shock absorbers are fitted to the seat safety stops in order to dampen impact and assist in momentum transfer from one phase to another (Hettinga and Andrews 2007). A single rower operated push switch controls 4-channels of electrical stimulation to the rower's leg muscles. When the control switch is closed the stimulator activates the quadriceps causing leg extension. Similarly, when open the switch allows stimulation of the hamstrings causing leg flexion.

String (potentiometer) sensors provide seat and handle position data. A National Instruments NI USB-6008 12 bit data acquisition unit, in conjunction with a PC running custom software developed in LabVIEW, simultaneously records seat and handle position data together with the state of the control switch. The software also calculates seat and handle velocity values from the corresponding position data using a Savitzky-Golay algorithm.

RESULTS AND DISCUSSION

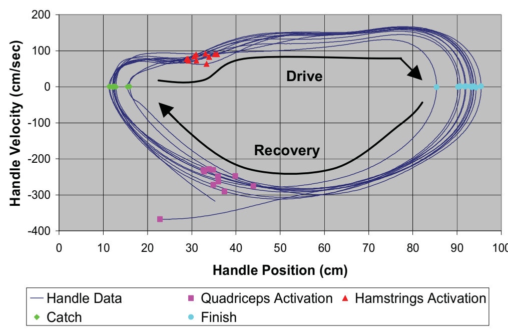

Figure 2: Handle velocity v handle position.

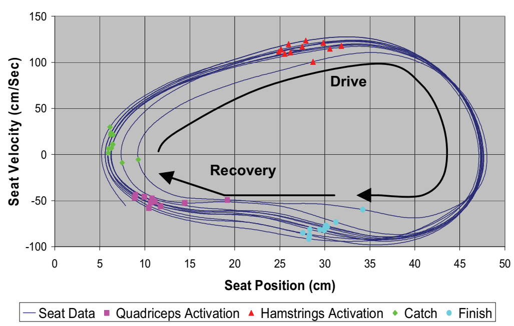

Figure 3: Seat velocity v seat position.

Phase plots of velocity against position for the handle and seat respectively over 10 complete rowing cycles are shown in Figures 2 and 3.

Table 1: Mean time taken and variance of rowing phases

Phase	Mean Time	Variance
Drive	0.811 sec	0.000903
Recovery	0.481 sec	0.000931

It can be seen that the rower switches at regular points on the cycle with very low variance. The curves are smooth with no sudden zeros in velocity. It is also noted that the switching points (i.e. quadriceps and hamstrings activation) occur before the extremes of motion (i.e. catch and finish).

SUMMARY/CONCLUSIONS

The subject appears to use anticipatory control, a learned skill in which the subject continuously predicts the system dynamics and state of fatigue of the stimulated muscle. The legs extend under load applied via the handle. In each stroke the rower loads the legs through the handle force to control the speed of the drive. If too much handle force is applied against the stimulated quadriceps the motion will be sluggish or may stall. As the quadriceps strengthen with use, the rower will impose increasing levels of handle force to regulate the motion. Thus the quadriceps always work under maximal loading. Furthermore, FES activation of quadriceps during late "recovery" will first cause an eccentric contraction (where the quadriceps are lengthening whilst contracting and act like springs) to decelerate the forward motion, then concentric contraction during "drive". Eccentric force actions will generally involve greater force actions than concentric contractions for the same FES stimulus intensity. This pattern of loading may have implications in training the quadriceps muscle properties and may explain long term changes (>1 yr) we have observed.

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ACUTE EFFECTS OF FOOT TAPING ON MUSCULAR RECRUITMENT IN PATIENTS WITH FOOT DEFORMITIES

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INTRODUCTION

Foot deformities such as flatfoot and hallux valgus may affect joint mobility, muscular activity and consequently, postural control and body balance. Association of abnormal structure and mechanics of the foot may also increase the risk of injury (Kaufman et al., 1999; Williams et al., 2001). An increased prevalence of knee soft tissues injuries have been reported as related to low arch in runners (Williams et al., 2001).

Many methods have been employed in attempt to reduce ankle injury associated to flatfoot including footwear adaptation, orthoses, taping and exercises. Although taping procedures have been reported as beneficial to the patient, consensus regarding either the perceived benefits or the mechanism of the effects of ankle taping is lacking in the literature (Wilkerson, 2002).

Nevertheless, beneficial effects of ankle taping have been attributed to the improvement in the joint position awareness when bearing full body weight (Robbins et al., 1995), either due to cutaneous sensory feedback improving joint position perception, described under the nonweightbearing condition (Simoneau et al., 1997), or due to neuromuscular processes characterized by relative increase in muscular activation (Lohrer et al., 1999). We hypothesize that taping can improve function by providing changes in muscular recruitment; therefore we proposed that foot taping applied in subjects with flatfoot and hallux valgus may acutely change the muscular recruitment during gait.

METHODS

Five adult females with clinical diagnostic of flatfoot and hallux valgus were evaluated. Data were obtained from the right lower limb during a barefoot gait in two conditions: with and without antipronation foot taping. For evaluation of changes in muscle activity, the linear envelope of the EMG signal of the gastrocnemius lateralis (GL) tibialis anterior (TA), peroneus longus (PL) and vastus lateralis (VL) muscles were analyzed. Electrodes were placed on the muscle belly, far away from the innervation zone. EMG signal was acquired synchronically with ground reaction force (GRF) and sampled at 1000Hz. The stance phase was determined using vertical GRF data.

EMG data of both gait conditions were compared by ANOVA for repeated measures followed by Scheffe test ($\alpha = 0.05$).

RESULTS AND DISCUSSION

The results are shown in figures 1 and 2 and indicated that after an acute plantar taping application lower limb muscle recruitment was altered. We observed significant increases in PL activity (132.2± 6.6 without tape; and 134.5± 18.5 with tape; $p=0.02$) and VL activity (108.1± 18.6 without tape; and 127.7± 25.2with tape; $p=0.04$) after taping which suggested that acute plantar taping application may work as a sensory feedback and/or provide mechanical advantage for the performance of these muscles.

PL is a potentially critical muscle in preventing ankle sprains injuries as a contributor to the protective mechanism to balance inversion at the heel contact in locomotor tasks. Increased activation of PL and VL may be important to prevent an inversion ankle sprain since the PL everts the ankle and VL activity controls knee extension during mid stance and decelerates the backward swinging of the leg and foot.

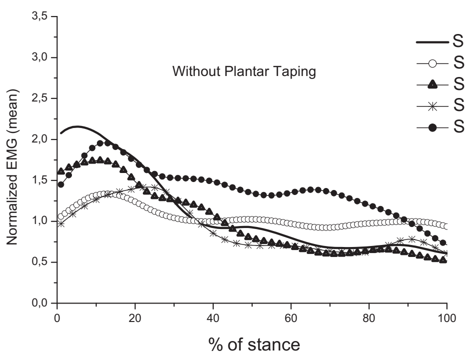

Figure 1: Mean EMG envelopes of VL from all the subjects (S 1, S 2, S 3, S 4, S 5) during stance phase without taping.

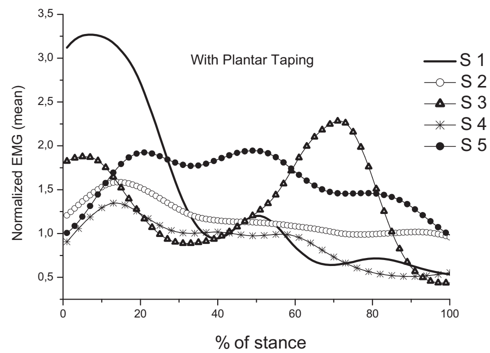

Figure 2: Mean EMG envelopes of VL from all the subjects (S 1, S 2, S 3, S 4, S 5) during the stance phase with plantar taping.

SUMMARY/CONCLUSIONS

Subjects with flat foot and hallux valgus after an acute plantar taping application enhance the vastus lateralis and peroneus longus activation, which play an important role in the stabilization of the foot–ankle complex and contribute to the knee mechanics in locomotor tasks.

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DRIVING EFFICIENCY OF WHEELCHAIR AS HEMIPLEGIC -COMPARE WITH SITTIGN POSTURE-

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INTRODUCTION

The older people tend to come to often use the wheelchair as transportation instead of walking. However, it is seen actually that to drive is difficult for the aged with round back. Moreover, the research concerning the one hand drive that hemiplegic patients with round back is few though there are reports that the driving efficiency at the round back sitting is low compared with a normal sitting. The purpose of this study is to investigate the influence that the round back sitting has given to the driving efficiency of right limbs promoted.

METHODS

Subjects were 15 healthy female (mean age; 22.0±0.7 year) who gave informed consent. Their characteristic were following as: length; 156.3±5.1cm, upper limb length; 49.4±2.7cm, lower limb length; 78.2±2.9cm, Trunk length (acromion-trochanter major); 52.8±1.4cm.
The experiment task of normal and round back sitting postures is surrounding times by both directions (medial, external) as for circumference 10 meters and figure eight combined. Subjects were fixed with trunk brace to make backward pelvic tilt 25 degrees from vertical plane, and then asked to sit center of gravity become 35% forward of the seat.
Driving speed (m/min), driving frequency with each limb (times/min), and physiological cost index (beat/min) were measured by video camera. Tidal volume (TV), Respiratory rate (RR), specific minute ventilation (MV), and specific volume oxygen (VO_2) were measured by erugorespirometer (METMAX, CORTEX Co. Ltd) to analyze the expiration gas.
The round back sitting was compared with the normal sitting by paired t test. Each direction as 10 meter circumference and combined were analyzed by ANOVA respectively. Statistical probability was less than 0.05.

RESULTS

The driving speed as medial, external and figure eight were 42.0±12.1, 38.3±11.0, 39.2±10.6 in the normal sitting, 34.8±10.2, 28.1±8.7, 32.6±7.5 in the round back sitting respectively. While the driving speed on the round back was significantly slower than the normal ($p<0.05$), similarly the driving frequency was increased at these tasks ($p<0.05$).
The physiological cost index were 0.47±0.2, 0.58±0.3, 0.39±0.3 in the normal sitting, 0.61±0.3, 0.79±0.3, 0.62±0.3 in the round back sitting respectively. The index on the round back was significantly higher than the normal ($p<0.05$).
The respiratory rate, MV, and VO_2 on the round back was significantly increased in the figure eight than the normal ($p<0.05$). In addition, the tidal volume became smaller in the external direction (table 1, $p<0.05$). However there were no significant

differences of all the measurements in the comparison of three directions.

DISCUSSION

The wheelchair driving is easy to rotate of the caster and to be done faster, because the center of gravity of normal sitting posture has come to the axis of the big wheel. In contrary it seemed that the friction drag of the caster grew up as body weight hung, because the pelvis reversely moved forward of the seat as for round back sitting posture.

A normal sitting that keeps the movement range of the diaphragm is thought to be position to which it breathes easily. Lin et al [1] reported that the movement of diaphragm is limited the round back sitting posture so that the viscera may push up the diaphragm.

We thought that the activity of the breath muscle with high energy consumption increased to remove the limitation in the diaphragm movement though the tidal volume decreased. And it is thought that the round back sitting decreases the thoracic movement and the upper limb compared with the normal sitting.

SUMMARY

The purpose of study was to investigate the influence that a sitting posture has given to the efficiency of drive as the hemiplegic pattern promoted. At a comfortable speed for three minutes with two sitting posture in directions (lateral, medial, figure eight) were experienced. Parameters were driving speed, frequency, and cardiopulmonary indexes, for example: PCI, TV, RR, VE, VO2, and VCO2. As a results, driving speed quickens up in normal sitting, and number of times by upper limb and PCI have significantly decreased compared with round sitting($p<0.05$). It is thought that a caster rolling resistance decreases, and ischium fixes the origin of hamstrings. On the other hand, it has been understood that driving efficiency has gone wrong in the round sitting. The reason is following as; an increase of resistance, limitation of the diaphragm, the range where the arm moves narrowed.

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Table 1: Comparison of the normal sitting and the round sitting for VE, VO2, respiratory rate (RR), tidal volume (TV)

	external		medial		figure eight	
	normal	round	normal	round	normal	round
VE	302.4±120.8	291.4±101.0	280.4±108.8	286.1±91.9	272.9±69.9	295.5±70.7**
VO2	10.5±4.4	10.3±3.3	10.0±3.8	10.3±3.1	9.7±2.3	10.6±2.3**
RR	26.5±6.2	27.8±4.9	26.4±5.2	28.1±4.8	25.9±5.2	29.0±4.8*
TV	0.60±0.1	0.54±0.1*	0.56±0.1	0.53±0.1	0.55±0.1	0.54±0.1**

(mean ± SD) *: $p<0.05$, **: $p<0.01$

VALIDITY OF MUSCLE FUNCTION TESTS IN THE REHABILITATION MANAGEMENT OF CHRONIC LBP

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BACKGROUND:

Within the rehabilitation management of chronic low back pain (cLBP), assessment of muscle function has been recognized as highly relevant for diagnosing muscle functional impairment, planning of rehabilitation interventions and for documentation of rehabilitation outcome [Cieza et al., 2004]. Thus, dynamometric trunk muscle strength and endurance tests are widely performed in rehabilitation settings. This is, however, done regardless of the sparse and conflicting evidence on the reliability of these measures in cLBP. Moreover, no data seem available on the accuracy (sensitivity and specificity) of these measurements.

This study sought for the first time to examine the accuracy of trunk muscle strength and endurance measurements in cLBP and to test their long term reliability with test protocols that are feasible in rehabilitation settings.

METHODS:

A total of 32 cLBP and 19 healthy controls who matched in age, sex and body mass index (13 matched pairs according to a 2: 1 ratio and further 6 pairs according to a 1:1 ratio) were recruited in this cross sectional study. After familiarization with the dynamometric test protocols, both patients and healthy controls performed four repetitive isokinetic trunk extensions and flexions at 90°/s and three isometric trunk extensions and flexions at 20°, 60°, and 100° per s on a Biodex 2000 dynamometer, respectively. Tests were repeated, if the variability within a series of 3 or 4 consecutive repetitions exceeded 15%. The Biering SØrensen test served to examine back muscle endurance. Therefore subjects were positioned prone over an examination table with the lower extremities stabilized and the trunk being extended beyond the table until the limit of fatigue.

Borg-Category Ratio Scales CR-10 rated participants' feelings and body experience. CLBP patients, who received no therapy, repeated the test protocol after approximately three weeks.

Main Outcome Measures were peak muscle strength (Nm) for the dynamometric isokinetic trunk extension and flexion measurements, and the time to exhaustion (s) for back muscle endurance.

After testing for between group differences, sex adjusted logistic regression analysis served for testing specificity and sensitivity. The respective receiver operating characteristic (ROC) curves were plotted and the area under the curve *calculated* [Zweig & Campbell 1993].

Reliability was investigated based on data from cLBP patients whose measurements were assessed on two different days. Paired t-tests served to examine systematic differences of the outcome measures between the first and the second test day. Intraclass correlation coefficients for repeatability were computed from mixed model analyses, if no changes in the mean were observed [Shrout & Fleiss, 1979]. Repeatability was visualized with Bland-Altman plots [Bland and Altman, 1986].

RESULTS AND DISCUSSION:

Among dynamometric tests, sex adjusted isokinetic trunk muscle measurements demonstrated the best ROC with area under the curve (AUC) values of 0.86 and 0.89, respectively. Reliability testing in cLBP demonstrated highly significant learning effects for the isometric trunk flexion as well as the isokinetic measurements, respectively. Borg-Category Ratio Scale ratings were not associated with the observed changes in the mean. The Biering SØrensen test demonstrated an AUC value of 0.93 for accuracy, which would mean that a cut-off point corresponding to a sensitivity of 0.9 would result in a specificity of 1, and ICC values of 0.59 for long term reliability, respectively.

SUMMARY/ CONCLUSION

In cLBP rehabilitation management, both the isometric trunk flexion and the isokinetic muscle strength measurements are limited to muscle functional diagnostics for treatment planning purpose. Monitoring the treatment outcome with these measures seems problematic. We recommend the Biering SØrensen test for assessing trunk muscle function and rehabilitation outcome monitoring in cLBP.

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AN EVALUATION METHOD FOR CONTROLLING SKILL OF MYOELECTRIC CONTROL HAND

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INTRODUCTION

In Japan, few forearm amputees use a myoelectric control hand (MCH) in daily life (Kawamura et al, 1999). One of reasons is that almost medical staffs don't have experience to train the amputees to utilize the MCH.
For the efficient therapy to utilize the MCH, medical staffs had few experience were required a quantitative evaluation index for a skill to control myoelectric signals that control the MCH,
In this study, we developed an evaluation system for the controlling skill of MCH, and four amputees were evaluated the skill by developed system.

EVALUATION METHOD

MCH utilized widely in Japan requires two input signals (Sig1, Sig2) to control motions (open and close) of MCH.
The MCH is opening when Sig1 is 0.6V or above and Sig2 is under 0.6V. On the other hand, When Sig2 is 0.6V or above and Sig1 is under 0.6V, the MCH is closing.
Sig1 and Sig2 were assigned average rectified electromyogram of forearm extensors and flexors, respectively.
The values of the signals increased depending on the muscles contraction level.
To avoid muscle fatigue and to achieve using the MCH in the daily life, the amputee should contract two muscles separately.
Because the two muscles are antagonist and non-trained amputees often use co-contractions, they are difficult to contract the two muscles separately.
Therapy for using the MCH is done to improve the ability of separate contractions of two muscles.
In this study, this ability was defined as the skill to control motion of the MCH.
Developed method evaluates the skill quantitatively as follow:
At first, motions of the MCH were classified into opening motion and closing motion by the values of the two signals.
Opening motion was defined by each signal where the Sig1 is larger than Sig2. Closing motion was defined by each signal where Sign1 is smaller than Sig2.
Regression coefficient of the signals was calculated by the least squares method that designated intercept as 0.
Explanation value is Sig1 in opening motion, and is Sig2 in closing motion.
The skill in each motion was evaluated by the regression coefficient, which is in inverse proportion to the skill.

MEASUREMENTS

Four trans-radial amputees participated in this study. Informed consent was obtained from all subjects before measurements.
A monitor was put in front of subjects.
Subjects were requested to contract forearm flexor and extensor separately depending on

a circle color on the monitor. This circle color changed each two second in greed and red. Sig1, Sig2 and the circle color were measured.
When the circle color is green, subjects tried to contract only forearm extensor. When the circle color is red, the subjects tried to contract only forearm flexor.
The sampling frequency was 100 Hz. A 16 bit A/D converter with a ± 10V input range was used for the measurement.

RESULTS

Fig.1 illustrates an example of a result of measurement. Fig.2 illustrates an example of a scatter plot.
Table.1 shows the regression coefficients in each subject and each motion as quantities index of skill to contract separately two muscles.

CONCLUSIONS

In this study, we developed a novel evaluation method the skill for controlling the MCH, and levels of the skill in four amputees were evaluated by the developed method.
The quantitative evaluation in the skill by developed method should be useful for efficient therapy to use the MCH.

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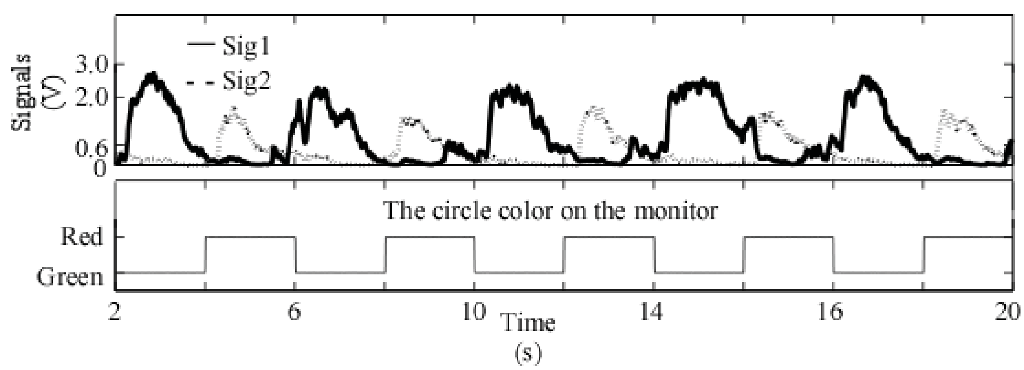

Figure 1: An example of measurement of signals

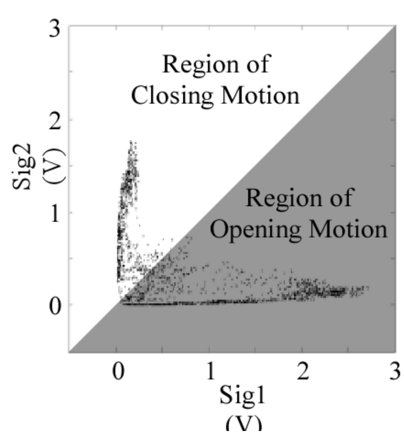

Figure 2: An example of Scatter plot. X-axis coordinated Sig1 and Y-axis coordinated Sig2

Table 1: Regression Coefficients by two signals in each motion.

Subject Number	Opening Motion	Closing Motion
1	0.05	0.01
2	0.11	0.16
3	0.08	0.11
4	0.51	0.19

ELECTROMYOGRAPHIC PEAK ANALYSIS IN SHOULDER STABILIZER MUSCLES DURING PENDULAR EXERCISES

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INTRODUCTION:

Pendular exercise of the shoulder constitutes a traditional exercise in the process of kinetic recuperation of the shoulder. It is a passive exercise in which active or extensive movements should be avoided, as they contribute towards greater activation of the rotator cuff and other dynamic stabilisers of the shoulder, bringing together the acromion and the coraoacromial ligament which could annul the desired effect of the exercise (Kisner & Colby, 1998; Cailliet, 2000).
As it is an extremely unstable joint, the need for stabilisation continues to be fundamental mainly through the action of the rotator cuff, deltoid and long head biceps. Therefore, the aim of this study was to quantify the participation of certain muscle groups that act in the articulation of the shoulder and their contribution during pendular exercises.

METHODS

Thirteen (13) individuals, without any history of shoulder joint injury, were analysed using the electromyographic signal (Miotool 400, Miotec, Brazil) from the brachial biceps and triceps, and medium deltoid during pendular exercise. The movements were: horizontal adduction/abduction of the shoulder and flexion/extension of the shoulder without load and with a 1kg. load. The data were analysed using SAD32 software and the comparisons made using repeated measures variance analysis and *post-hoc* Bonferroni ($p \leq 0.05$).

RESULTS AND DISCUSSION

The anterior deltoid showed greater activation during the movements, reaching 4% of the maximum. However it did not exert significant influence on the objective of the pendular exercise. The level of activation in the other muscles was practically non-existent, being less than 1%. None of the participants in the study had any previous history of joint and/or muscle injury in the region of the shoulder. Therefore, the findings of the present study can only be inferred for healthy people.

CONCLUSIONS

Our findings suggest that pendular exercise without load and with a small load of 1 kilogram produces low activation of the brachial biceps and triceps, anterior deltoid and medial deltoid. Thus, this exercise is shown to be effective in terms of the passivity of the muscular groups analysed in the present study. Though with a very low level of contribution, the deltoid exhibited the highest level of activation when compared with the other analysed muscles, and hence was seen not to specifically influence the pendular exercise. The lack of scientific studies dealing with pendular exercise of the shoulder leads us not confer full reliability to the practice of this exercise. However, the parameters shown in the present study indicate a slight muscular contribution suggesting its clinical application.

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COMPARISON OF THE BRACHIAL BICEPS AND TRICEPS ACTIVATION DURING PEC DECK AND BENCH PRESS EXERCISES WITH THE VALUES OBTAINED DURING THE 1RM TEST IN THE BICEPS CURL AND TRICEPS PRESSDOWN.

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INTRODUCTION

In pec deck and bench press exercises, besides the pectoral muscle, the biceps and triceps brachial muscles also play a role in the execution of the movement. Nevertheless, in the prescription of these exercises, the load imposed on these two muscle groups is unknown, which makes it difficult to precisely identify their level of activation.

Hence, though studies exist that analyse pec deck and bench press exercises, there remains a gap in the literature concerning the quantification of the activation of certain muscles in these exercises, comparing such activation to that found in monoarticular exercises where they are primary motors of the movement.

METHODS

Fifteen (15) male individuals performed a series of three repetitions of the pec deck and bench press exercises at intensities of 20, 40, 60 and 80% of one maximum repetition (1RM). The levels of activation were evaluated using the electromyographic signal (Miotool 400, Miotec, Brazil) form the brachial biceps and triceps muscle groups and the values obtained were compared with those found during the bicep curl and tricep curl exercises at 1RM, respectively. The comparisons were made using repeated measures variance analysis and Bonferroni's post-hoc test. ($p \leq 0.05$).

RESULTS AND DISCUSSION

The results showed an increase of the muscular activity of the brachial biceps and triceps with the increase in the exercise load (Table 1). The maximum level of activity of the brachial biceps was seen at 80% of 1RM in pec deck, attaining 28% of the maximum activation. Similarly, a maximum level of activity of the brachial triceps muscle was seen at 80% of 1RM in the bench press, attaining 44% of the maximum activation. The brachial biceps, besides acting as the primary motor in the elbow flexion movement, also aids in the shoulder joint movements. Its points of origin (coracoid process) and insertion (radial tuberosity) would suggest a role for the short portion of the muscle in the horizontal shoulder flexion (Rash, 1977). In contrast, due to its point of origin (scapula and humerus) and insertion at the extremity of the olecranon (Rash, 1977), the triceps brachial muscle, primary motor in the elbow extension movement, act dynamically during the execution of the bench press exercise. In such case, the activation of these two muscle groups was expected, since the biceps and triceps brachial act as synergists in the pec deck and bench press exercises, respectively.

Furthermore, we observed a non-linear increase in muscular activity with the increment in the load used. These findings are in line with those reported in other studies (De Luca et al., 1982; De Luca, 1985; De Luca, 1997), where the non-linearity of the force/electromyographic signal curve was demonstrated in some muscles. This lack of a quantitative relationship between the two factors impedes us from knowing at what percentage of maximum force the muscle is working during the performance of the exercises at a determined intensity. However, there is, in fact, a qualitative relationship between the force generated by a muscle and the level of electromyographic activity (De Luca et al., 1982; De Luca, 1985; De Luca, 1997; Naito et al., 1998). Accordingly, in the present study, the increases in the electromyographic activity seen with the increase in the intensity of the exercises indicate an increase in the production of force generated by the analysed muscles, though the magnitude of this increase in force cannot be described.

CONCLUSIONS

Our findings suggest that in the pec deck and bench press exercises the brachial biceps and triceps muscles act constantly during the performance of the exercise and increase their activity as the exercise load is increased, and may reach levels of up to 44% of maximum.

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Table 1: Means and Standard deviations of EMG values in Peck deck and bench press exercises, normalized by EMG values obtained in biceps curl and triceps pressdown, respectively.

	20%	40%	60%	80%
Biceps (% max)	7.9 ± 3.6	17.6 ± 6.9	26.9 ± 12.0	44.0 ± 16.2
Triceps (% max)	3.1 ± 1.1	6.5 ± 2.8	12.7 ± 6.5	28.6 ± 16.1

A PRELIMINARY STUDY ON USEFULNESS OF CYCLING WHEEL CHAIR TRAINING FOR THE PATIENTS WITH LOWER EXTREMITIES IMPAIRMENT

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INTRODUCTION

Cycling wheel chair (C-W/C) is a useful tool to provide a chance of physical exercise for the patients with severe impairment in lower extremities. It was reported that most of the non-ambulatory elderly became psychologically active by self-locomotion with C-W/C driving. However, long-term effects of C-W/C driving on functional and physiological change are unknown. In the present study we made a preliminary investigation to answer this question.

METHODS

Five patients with lower extremity paralysis participated in this study. Age and sex, diagnosis, impairment and its level (Brunnstrom stage of paralytic leg in hemiplegia: BS) and time since onset to start of C-WC training (TSO: days) are as below respectively.

Subject A: 53y female, spinal cord injury, incomplete paraplegia (left dominant), 35days. / Subject B: 77y male, cerebral infarction, left hemiplegia (BS4), 38days / Subject C: 82y male, multiple cerebral infarction, left hemiplegia & parkinsonism (BS5), 30days / Subject D: 72y male, cerebral hemorrhage & central type cervical cord injury, right hemiplegia & bilateral upper extremities paralysis (BS3), 88days / Subject E: 87y male, brain stem infarction & dementia, right hemiplegia (BS2), 60days

C-W/C used in this study had a round shaped steering system that realized simultaneous control of both rear wheels.

All of the subjects performed a session of C-W/C training 5days a week adding to standard physical therapy during 4weeks (w). They drove a C-W/C at their own pace on an indoor course 50m in circuit during 3min continuously. One session of C-W/C training in a day included 5times of continuous driving with 1min rest between each driving.

Functional and physiological examinations as below were performed before and 4w after start of C-W/C training: grip strength of the healthy hand (kg), isometric knee-extension strength (IKE) of the healthy leg (kg), maximum walking speed (MWS) in 10m distances (m/min), score of Barthel index (BI), time averaged perfusion (TAP) of middle cerebral artery in the healthy side (cm/sec), oxygen consumption (VO_2) during 3min driving (ml/min). Ultrasound sonography (SonoSite MicroMaxx, SonoSite Co., Japan) was used to measure TAP. The probe was attached to the temporal skull of the healthy side. VO_2 was measured by a portable gas exchange unit (Aerosonic AT-1100, Anima, Japan).

RESULTS AND DISCUSSION

There were clear differences between subject A, B, C (groupH) and subject D, E (groupL) in the level of initial function and the gain evaluated by IKE, BI and MWS

(Table). TSO was also shorter in groupH than in groupL. In these aspects, it is natural that the subjects in groupH showed good recovery of function though there is a possibility that C-W/C training adding to standard physical therapy realized early recovery.

C-W/C training used in this study brought no remarkable improvement of motor function to the subjects in groupL. This result may be based on poor functional level or too late timing of functional recovery. The change of physiological parameters, however, showed a fact that C-W/C training might provide some effects even for the subjects with poor function.
The first evidence of such effects is the change of TAP. TAP could be measured in three subjects (A, B and E). The initial value of TAP in subject A and B was almost same as that of normal control (A:44.2, B:38.2, control:41.9) but subject E showed lower value (21.7). While the value of TAP increased in these three subjects 4w after C-W/C training, subject E showed the most remarkable change and his value reached normal level (A:55.9, B:41.6, E:47.9). TAP in a middle cerebral artery reflects mean volume of local blood perfusion in the cerebrum. It is no wonder that subject E showed low value of initial TAP because his basal disease was an ischemic stroke. The increase of TAP to normal level 4w after training, however, is a considerable phenomenon because location of the lesion in subject E was the brain stem. Therefore the increase of TAP in subject E suggests increase of blood supply to the brain activated by 4w C-W/C training.
The second evidence is the change of peak VO_2. VO_2 could be measured only in two subjects (A and D). At the end of 4w training subject A could drive C-W/C longer than at the beginning (115m→148m) but the peak VO_2 in a steady state during driving was almost same (before:705.2, after:665.8). On the contrary, in subject D, the peak VO_2 4w after training was much lower than the initial level (before:505.4, after:281.3), though the driving distance in 3min was same (before:91m, after:90m). Both subject A and D showed steady level of VO_2 by 1min after start of driving when VO_2 was measured during driving a C-W/C. The peak VO_2 in a steady state implies maximum level of O_2 demand to perform 3min C-W/C driving. The increase of driving distance without change of peak VO_2 in subject A indicates the improvement of muscle strength in lower extremities is a major factor of this phenomenon. On the other hand, the prominent decrease of peak VO_2 without change of driving distance in subject D suggests C-W/C training during 4w improved the level of cardio-respiratory fitness and made it possible to drive more efficiently.

subject		A	B	C	D	E
Grip	before	25.5	26.6	30.5	17.3	10.6
(kg)	after	24.3	31.7	25.3	20.1	9.5
IKE	before	11.6	16.0	10.0	0.0	0.0
(kg)	after	19.0	20.3	10.0	0.0	0.0
BI	before	40	55	35	10	10
	after	85	90	75	15	10
MWS	before	5.6	34.6	29.3	0.0	0.0
(m/min)	after	14.6	83.3	30.0	0.0	0.0

Table: Initial values and gain of function

SUMMARY/CONCLUSIONS

C-W/C has an advantage to give some pleasure of self-locomotion compared to a recumbent type ergometer. Physical load of driving a C-W/C is relatively low but for the patients with poor physical function it may be favorable to improve cardio-respiratory fitness and to provide brain activation. Application of C-W/C training to the patients with various levels of clinical stage is necessary in future.

GAIT ANALYSIS BEFORE AND AFTER ELECTRICAL STIMULATION FOR THE SPASTIC PARAPLEGIC PATIENTS AND HEMIPARETIC STROKE PATIENTS

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INTRODUCTION

Damage of central nervous system caused by a stroke or spinal cord injury usually brings about motor paralysis and increase of muscle tone, spasticity. In lower extremities, moderate level of spasticity often gives some favorable conditions to realize standing up and gait. Excessive spasticity, however, disturbs smooth and coordinated voluntary movement. Therefore, some treatments to control muscle tone are necessary for the patients with excessive spasticity.

There are various methods of medical treatment to inhibit spasticity. Electrical stimulation (ES) applied to peripheral nerve and muscle is one of ways to control muscle tone. ES for the antagonist muscle of the spastic muscle as a target causes decrease of tone in the target muscle via Ia reciprocal inhibition. From the viewpoint of clinical use, ES as a therapeutic tool is very convenient and has much advantage to acquire gait ability through spasticity control.

In this study, we investigated the change of motion parameters during walking in the spastic paraplegic patients and hemiparetic stroke patients before and after application of therapeutic ES with using a three-dimensional motion analysis system.

METHODS

Three spastic paraplegics and three spastic hemiparetics participated in this study. Etiology of paraplegia was unknown in all of the paraplegics and hemiparesis was caused by a stroke. Five were male and mean age was 51.3 ± 12.8 (26~61) years. With regard to the stroke patients, all had left hemiparesis. The level of spasticity evaluated by modified Ashworth scale was more than 1 in knee extension and flexion of paralytic lower extremity in all of the subjects. All of them had not experienced therapeutic ES until participation to this study.

Stimulation electrodes were attached to the muscle belly of the paralytic quadriceps femoris. A portable electrical stimulator (Nodoka, Lintec Co. Ltd., Japan) was used for therapeutic stimulation. Stimulation parameters were pulse width of 0.3msec and frequency of 20Hz. For the spastic paraplegic patients, stimulation was given to right and left leg alternately during 15min with 10sec on and 5sec off interval. For the spastic hemiparetic patients, stimulation was only given to the side of paralysis with same duration and interval as the paraplegics. Three-dimensional motion analysis system (KinemaTracer, Kissei Comtec Ltd., Japan) was used to perform gait analysis for 5m walking before and immediately after 15min stimulation.

RESULTS AND DISCUSSION

As for spastic paraplegic patients, mean velocity of the center of gravity (COG) transfer after ES increased compared to before stimulation. All of the paraplegics showed increase of the velocity after ES. As for hemiparetic stroke patients there was no remarkable difference between before and after ES (figure1).

Compared to before ES, vertical displacement of COG showed a tendency to increase after stimulation in both paraplegics and hemiparetics (figure2).

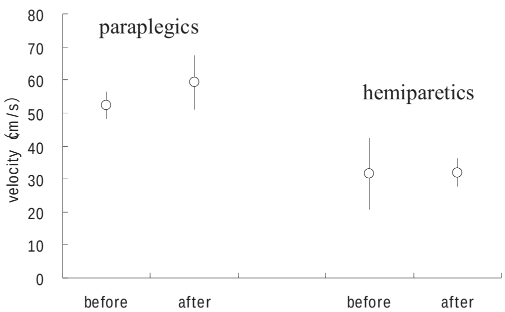

Figure1: Change of the velocity of COG transfer

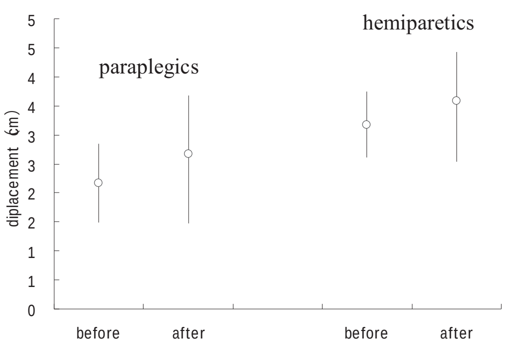

Figure2: Change of the vertical displacement of COG

It was supposed that such a change was caused by increase of the range of flexion and extension angle chiefly in the ankle and knee joint while walking after ES. These results suggest ES may have inhibited spasticity of the muscle around these joints.

Physiological and kinesiological mechanism on the effect of ES providing the change of some motion parameters is not simple. Usually, therapeutic ES for the antagonist muscle brings about inhibition of the target spastic muscle. In the present study, ES for rectus muscle possibly suppressed the tone of hamstrings. Hamstrings have actions as a knee flexor and a hip extensor. Consequently, the hip joint may have been easily bended and it resulted to longer step length. Furthermore, it might be linked with increase of the motion range of the ankle and knee joint.

SUMMARY/CONCLUSIONS

Velocity of COG transfer reflects walking speed. The increase of it in the paraplegics suggests that ES has a possibility to improve walking speed through spasticity inhibition. The increase of vertical displacement of COG after ES was remarkable in the subjects with more prominent spasticity. The effect of ES for muscle tone control appeared during walking may be dependent on the degree of spasticity.

Because of small number of the subjects statistical analysis could not be used in this study. Further examination for more subjects is necessary.

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SURFACE ELECTROMYOGRAPHIC STUDY OF RAMSAY HUNT SYNDROME: A CASE REPORT

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INTRODUCTION

The Ramsay Hunt Syndrome describe an uncommon acute infectious disease caused by herpes zoster and associated with facial palsy that affects the sensory and motor branches of the facial nerve, which may involve many cranial nerves, preferably the 8th (Hunt, 1907). Its prevalence is higher in individuals over 50 years old, and immunocompromised individuals with diabetes (Johnson et al., 1996).

Pain is the most common initial symptoms and is intense, lancinating and burning. After a period of approximately three days can occur a paralysis of the facial nerve homolateral (Ali, 1998; Grossmann, 2003).

The symptoms and signs are an inability to close the eye, to smile, wrinkle the forehead and whistle. A drooping of the face in the affected side. Speech may be mildly slurred. Damage to 8th cranial nerve generates loss to the hearing of the affected side and the patient can report nausea and loss of balance and tinnitus (Da Silva, 1998).

The surface EMG has been used by physicians and dentists as a measure of measuring the degree of commitment of some nerves associated with facial paralysis, as the Ramsay Hunt syndrome. Moreover, the literature about surface electromyography does not present clinical studies further on the subject (Bérzin, 2004).

The aim of this study was to describe through surface electromyography, the electrical activity of Mm masseter and anterior temporalis in a patient with Ramsay Hunt Syndrome.

METHODS

The electromyographic (EMG) activity from anterior temporalis and masseter muscles of a 47 year-old woman with a Ramsay Hunt Syndrome was evaluated.

Recordings were made on 12 channels of simultaneous EMG signal acquisition equipment (Myosystem I / Datahominis Tec. Co.). The analog EMG signal recorded were digitized using 12 bit A/D converter at a sampling rate of 4KHz. After digitalization, the signal was filtered by a digital pass-band of 10-500Hz. Myosystem I software version 2.12 was used to visualize and process the EMG signal.

The tasks were: (1) mandibular rest position for 30 seconds; (2) maximal voluntary contraction (MVC) for 5 seconds and; (3) masticatory activity for 10 seconds.

RESULTS AND DISCUSSION

Although the literature considering to involvement of the 5 th cranial addition, the EMG examination of the patient shows that in habitual mastication, it was observed that the masticatory muscles evaluated acted in

synchronism and satisfactory function with amplitudes within normal limits.
Moreover, masseter presented higher activity in relation to temporalis muscles, suggesting that the motor part of the 5^{th} par skull was not affected by the Ramsay Hunt Syndrome.

SUMMARY/CONCLUSIONS

The surface EMG results of masseter and anterior temporalis muscles of patient with Ramsay Hunt syndrome shows that the motor part of the 5^{th} par skull is preserved.

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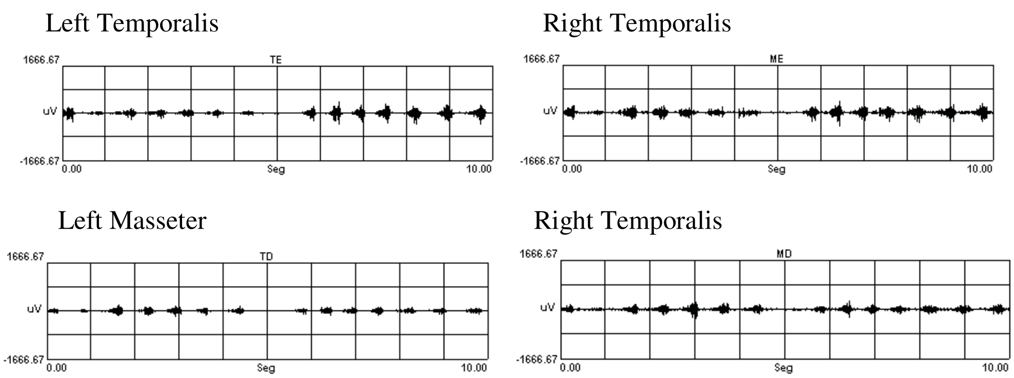

Figura 1: EMG activity during masticatory activity from anterior temporalis and masseter muscles in patient with Ramsay Hunt Syndrome.

NEUROMUSCULAR RESPONSES TO STATIONARY RUNNING AT DIFFERENT CADENCES IN AQUATIC AND DRY LAND ENVIRONMENTS

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INTRODUCTION

Some studies have analyzed the neuromuscular responses in aquatic environment for human gait (Miyoshi et al., 2004; Barela et al., 2006) or therapeutic exercises (Kelly et al., 2000; Pöyhönen et al., 2001). However, these responses are unknown in water aerobics exercises.

The aim of the present study was to analyze the neuromuscular response of young women performing stationary running exercise at different cadences in aquatic and dry land environments.

METHODS

The study sample consisted of twelve young women (22.33 ± 0.57 years), experienced in water aerobics.

Electrodes were placed on the belly of vastus lateralis (VL), biceps femoris (BF), rectus femoris (RF) and semitendinosus (ST) muscles, with a 3-cm center-to-center spacing. Transparent dressing was used to insulate electrodes for the water condition trials.

The electromyographic signals (EMG) were registered with a 4-channel EMG system (Miotool400 USB, Brazil), with a common mode rejection ratio >110 dB and a sampling rate of 2000 Hz by channel.
The filtering of the raw EMG was performed with a filter Butterworth type, with a bandwidth of 25–500 Hz. The EMG data of each muscle were normalized by maximal voluntary isometric contraction (MVC).

The sample performed two test protocols, one land-based and the other water-based, with a two-hour interval between them. The stationary running exercise was executed in each of these environments during 4 min at 3 sub-maximum cadences (60, 80, and 100 bpm) and during 15 s at maximum effort, with a 5 min-interval between each situation.

We used blocked variance analysis, in which the effect of the subject was considered an additional source of variation for the statistical analysis. The data was processed using the SPSS (version 13.0) and R-project programs, with a $p < 0.05$.

RESULTS AND DISCUSSION

The neuromuscular responses showed no significant increase on EMG signal from the VL, BF, RF and ST muscles with higher cadence of execution, except from the sub-maximum cadences to the maximum effort. When comparing the environments, the dry land environment presented significantly greater EMG signal responses from all the muscles at the sub-maximum cadences, except for the ST muscle which presented similar responses in both environments. However, at the maximum effort, all the analyzed muscle groups showed similar responses in both environments (Figure 1).

SUMMARY

In summary, at the cadences used in the present study, the performance of the stationary running exercise in an aquatic environment at sub-maximum cadences presents lower neuromuscular responses than the same exercise performed on land. Yet, at maximum intensities, the amplitude of the EMG signal may present similar muscular activation patterns in the two environments.

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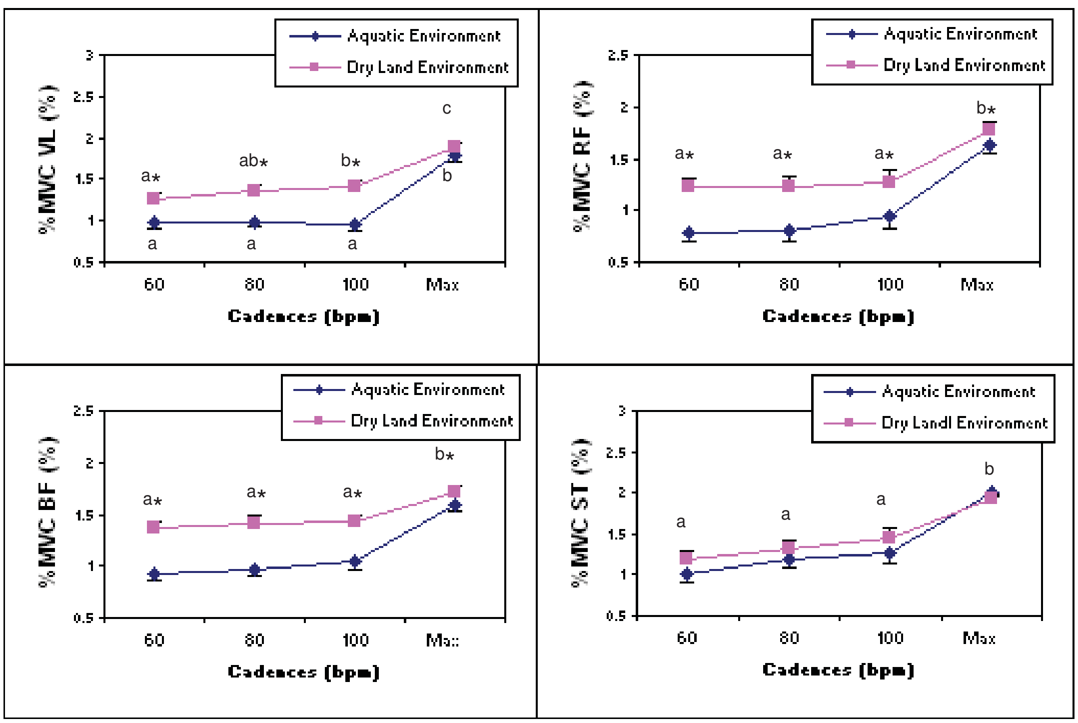

Figure 1 – The EMG normalized by maximal voluntary isometric contraction (MVC) of vastus lateralis (VL), biceps femoris (BF), rectus femoris (RF) and semitendinosus (ST) muscles. Mean values expressed as $\log_{10}$. * indicates significantly differences between aquatic and dry land environments ($p < 0.05$). Different letters indicate statistically significant differences for cadences ($p < 0.001$).

PARASPINAL RECRUITMENT PATTERNS DURING ECCENTRIC AND CONCENTRIC CONTRACTIONS IN CHRONIC LOW BACK PAIN

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INTRODUCTION

SEMG variables seems to be dependent of muscle fiber length (Farina et al., 2001). SEMG spectrum could be associated to neural strategies or geometrical factors (Farina et al., 2004). So, the purpose of this study was analysis the frequency of sEMG during dynamics contractions in healthy and chronic low back pain (CLBP) subjects.

METHODS

Subjects: Paraspinal muscles sEMG were obtained from healthy (n=40) and CLBP (n=7) subjects.
Instrumentation: Single differential electrodes (Delsys Inc.) were used to obtain sEMG. A digital camera was used to capture trunk kinematics.
Protocol Test: Subject performed Flexion-Relaxation Test (FRT): neutral position, flexion, full flexion and extension of trunk (5 s each phase). It was repeated 3 times.
Signal Processing: Trunk flexion (α) and lumbar flexion (β) angles were obtained. To evaluate the muscle recruitment a window of one second whose middle point coincided with 45º of trunk flexion angle (α) for both eccentric and concentric burst was used (figs. 1a, 2a). A Discrete Wavelet Transform (DWT) using Daubechies order 8 was used to transform each segmented burst to time-frequency domain. The intensity in every box of wavelet coefficients was calculated.
Statistical Analysis: Paired Student`s t test and Wilcoxon signed rank test were used to evaluated differences between eccentric and concentric burst of multifidus (L5) and longisimus (L1) muscles.

RESULTS AND DISCUSSION

Healthy Subjects: In multifidus eccentric contractions were characterized by an intermittent sEMG activity (fig 1b.) and concentric contractions by a continuous sEMG activity (fig 1c). Wavelet analysis showed a significance increase in the intensity of frequency band 8-16 Hz ($p=0.0006$) and 16-32 Hz ($p=0.0007$) while concentric contractions had more intensity at 64 and higher frequency ($p=0.001$ and $p=0.0007$, respectively). No differences were found in 32-64 Hz band (fig 3). Similar statistical were found in longissimus muscle.
CLBP Subject*s*: In both muscles eccentric contractions were not characterized by the intermittent activity found in the healthy subjects (figures 2). Wilcoxon signed rank test showed no differences in the intensity of every frequency band in CLBP group (all p values were >0.05) (fig 4.).

SUMMARY/CONCLUSIONS

CLBP shows different recruitment patterns in dynamics contraction when compares to healthy. Probably muscle atrophy and different neural strategies are executed in this movement by CLBP subjects. Caution must be considered with the interpretation of result, geometrical factors affect the contents of sEMG signals in dynamics conditions.

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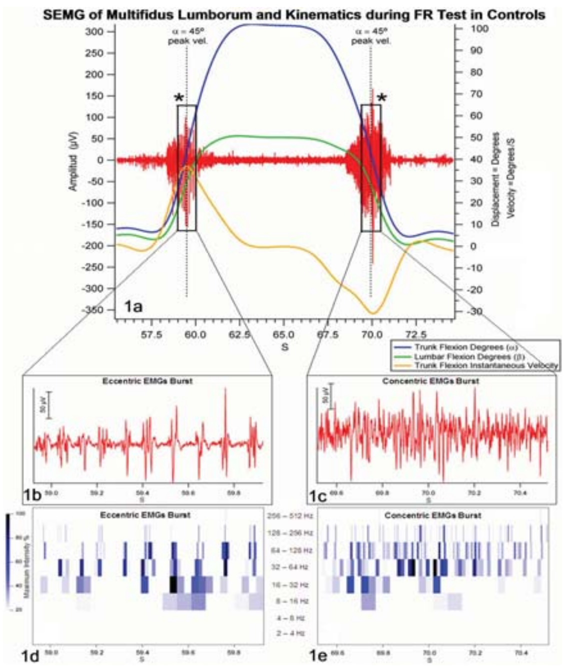
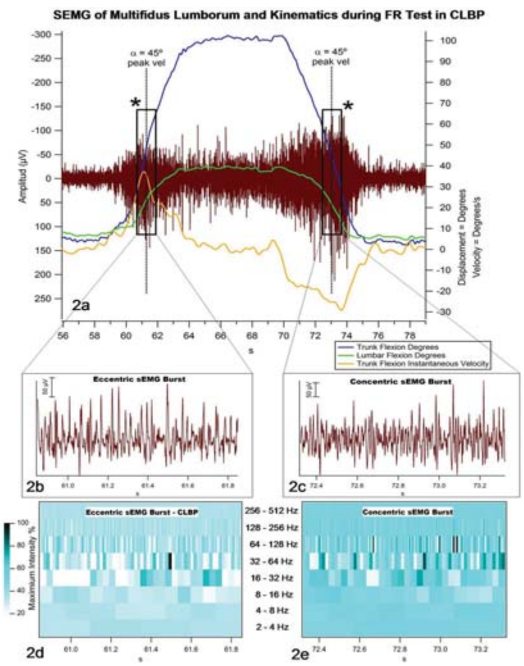


Figures 1a: sEMG and kinematics during a cycle of FRT in a control subject. * = window above which the sEMG was extracted to perform DWT. **1b:** Intermittent activity profile in eccentric phase. **1c:** Continuous activity profile in concentric phase. **1d:** DWT Matrix of sEMG signal during eccentric phase. **1e.** DWT Matrix of sEMG in concentric phase.



Figures 2a: sEMG and kinematics during a cycle of FRT in a CLBP subject. * = window above which the sEMG was extracted to perform DWT. **2b:** sEMG activity during eccentric phase. **2c:** sEMG activity during concentric phase. **2d:** DWT Matrix of sEMG signal during eccentric phase. **2e:** DWT Matrix of sEMG signal in concentric phase.

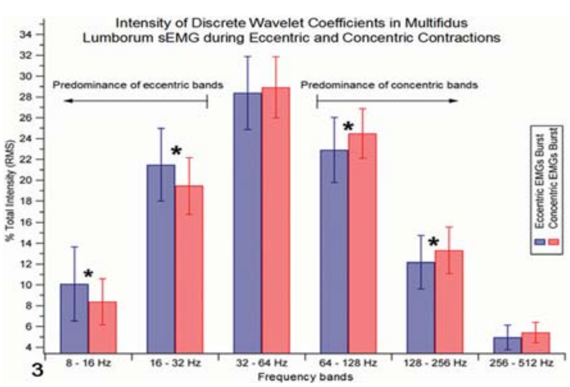


Figure 3: Distribution of frequency bands in multifidus lumborum muscle of the control group. Note the predominance of the different bands of frequency in the histogram. * p<0.05.

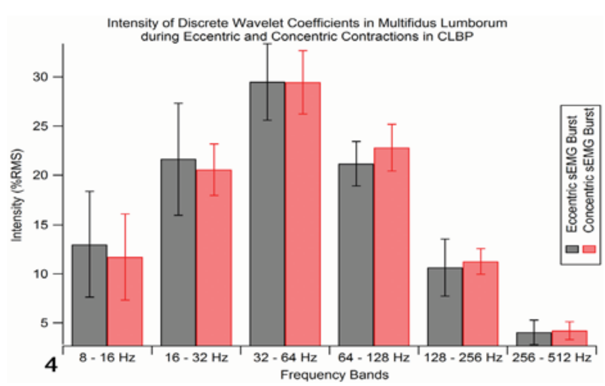


Figure 4: Distribution of different frequency bands in multifidus lumborum of the CLPB group. No differences were observed between frequency bands in this group. * p<0.05.

EFFECT OF THE BASE ON THE ELECTROMYOGRAPHIC AMPLITUDE DURING CLOSED KINECTIC CHAIN EXERCISES FOR UPPER LIMB AND SHOULDER GIRDLE

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INTRODUCTION

Closed kinetic chain (CKC) term refers to exercises in which there is a task involving movement of several joints with the distal extremity fixed on a surface. These exercises are included in painful or unstable shoulder rehabilitation programs since they are considered biomechanically safer and more functional than open kinetic chain exercises.

Studies have suggested electromyography as a tool to evaluate muscle activity during CKC exercises. In addition, some recent studies have compared EMG amplitude during exercises performed with the hand supported on a stable or relatively unstable surface (Behm et al., 2002).

The purpose of this study was to compare EMG amplitude during wall-press, bench-press, and push-up exercises performed with the distal extremity of the segment fixed on a stable and relatively unstable surface.

METHODS

Twenty healthy male subjects (mean age 29±3 yrs) volunteered for this study. All subjects volunteering for this study signed a consent form approved by the institutional review board.

EMG data were collected using surface differential electrodes (two Ag–AgCl bars, 10x2x1mm, with 10mm interelectrode distance, 20x gain, input impedance of 1GΩ and CMRR of 130 dB). SEMG signals and force output were sampled by a 12-bit A/D converter board with a 4kHz frequency, and band-pass filtered at 0.01–1.5kHz. Raw SEMG data were digitally filtered at frequency bandwidth of 10–500Hz and the root mean square (RMS) was calculated.

The first stage involved performing a physical evaluation and determining maximum individual load by calculating average force collected by the load-cell during three repetitions of each exercise.

In the second stage, EMG signals of the following muscles were recorded: the long head of biceps brachii (B), the long head of triceps brachii (T), anterior (AD) and posterior (PD) deltoid, clavicular portion of the pectoralis major (P), upper trapezius (UT), and serratus anterior (S) muscle of the dominant limb during three maximum voluntary isometric contractions (MVIC) in a muscular testing position for manual muscle testing. This was done in order to obtain reference values for RMS normalization. Skin preparation, electrodes position and attachment were done in accordance with SENIAM project.

After a 6-min resting period and after the last MVIC test, each volunteer randomly performed six exercises selected for the study (Fig. 1). Each exercise was repeated three times (6s each contraction) with rest intervals of 2min between contractions to minimize the effects of muscle fatigue. The data needed to evaluate test–retest reliability were obtained seven days later.

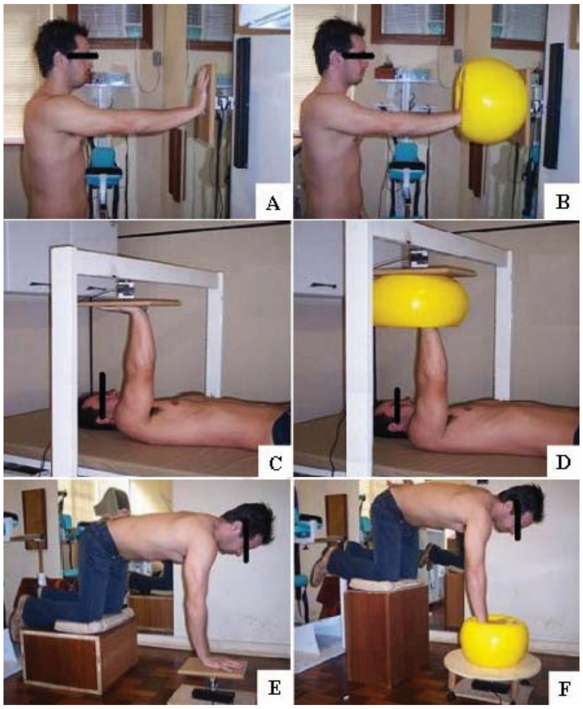

Figure 1: Wall-press, bench-press and push-up exercises performed with the distal extremity fixed on stable surface (A,C, E) and on a *Swiss* ball (B,D,F).

RESULTS AND DISCUSSION

ICC [2,1] values showed excellent intraday reliability for normalized RMS (.78–.99) values for all muscles. However, interday reliability ranged from poor to excellent (.06–.98) for all muscles during exercises on stable surface and *Swiss* ball.

Normalized RMS mean values and standard deviation during the six studied exercises are showed in Fig. 2. Independent of the base of support, none of the studied muscles reached a moderate level of EMG amplitude during the wall-press exercise. This result disagrees with a previous study (Oliveira et al., 2007) that found moderate EMG amplitude in wall-press in an unstable support.

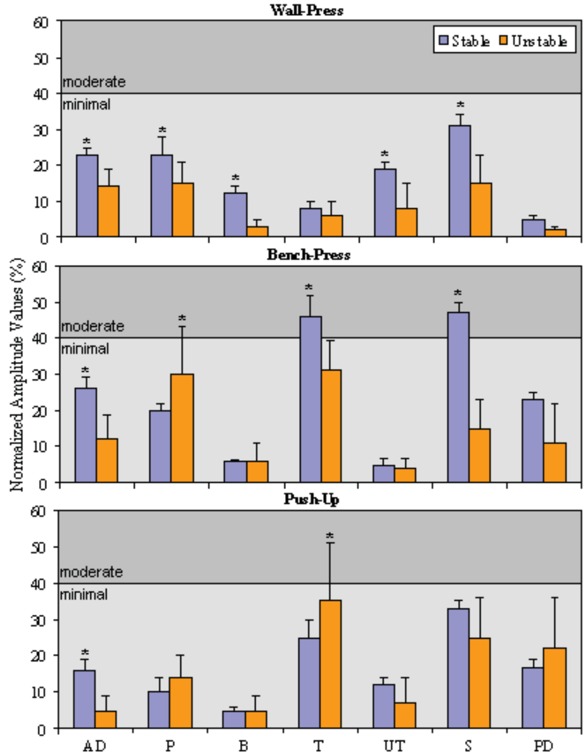

Figure 2: Normalized RMS mean values during wall-press, bench-press and push-up exercises performed with the distal extremity fixed on stable surface and on a *Swiss* ball. (* paired Student *t* test p<.05)

SUMMARY/CONCLUSIONS

This study's results showed that the EMG amplitude of the studied muscles during CKC exercises is different according to the base of support used.

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This study was supported by State of São Paulo Research Foundation (FAPESP) and National Council for Scientific and Technological Development (CNPq).

Track 11

Motor Performance and Sport (MP)

THE EFFECTS OF VARIOUS HIP EXERCISES ON THE ACTIVATION OF HIP MUSCULATURE: AN INVESTIGATION USING INDWELLING FINE-WIRE ELECTRODES

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INTRODUCTION

Strength deficits and/or poor neuromuscular activation of the hip muscles are thought to contribute to disorders of the low back, hip, and knee (Powers, 2003; Leinonen et al, 2000). As a result, various exercises are prescribed to improve strength of the hip muscles. However, it is not clear which exercises are best for activating the key muscles of interest (e.g., gluteus maximus and gluteus medius) while keeping antagonist muscle activity (e.g., tensor fascia latae) to a minimum. It is important for clinicians to know which exercises best recruit the various hip muscles in order to facilitate optimal exercise prescription and progression.

To date, only a few studies have investigated the use of hip exercises for activating hip muscles (Bolgla & Uhl, 2005; Ayotte et al, 2007). However, a limitation of these studies is that surface electrodes were used to evaluate muscle activation. The problem with surface electrodes is that it is difficult to differentiate the signals of the gluteus medius, gluteus maximus, and tensor fascia latae from each other because of electrode cross-talk. Also, the superior and inferior portions of gluteus maximus are considered to have different functions (Lyons et al, 1983), but have not been studied individually during various hip exercises.

Furthermore, we are unaware of any studies that have investigated the activity of these primary hip muscles simultaneously. Therefore, the purpose of this study was to assess the electromyographic (EMG) activity of the gluteus medius (GMED), superior gluteus maximus (SUP-GMAX), inferior gluteus maximus (INF-GMAX) and tensor fascia latae (TFL) using indwelling fine-wire electrodes, to determine which muscle is best activated by each exercise and how well the gluteal muscles are activated compared to the TFL.

METHODS

Eight healthy volunteers participated. Prior to data collection, 50-micron fine wire electrodes were inserted into the GMED, SUP-GMAX, INF-GMAX, and TFL. Electrode placement was confirmed with mild electrical stimulation. The maximum EMG signal amplitude was then recorded (1560 Hz) as subjects performed maximum voluntary isometric contractions (MVICs) for each muscle. Raw EMG signals were band-pass filtered between 20 and 750 Hz. Following electrode placement and MVIC testing, subjects performed the following nine exercises, commonly prescribed to activate hip extensor and abductor muscles: Side-lying Hip Abduction, Bilateral (Bi-) and Unilateral (Uni-) Bridging, Side-lying Clam with an elastic band around the thighs, Hip Hike, Hip Extension in Quadruped on Elbows with Knee Extending (QKE) and with Knee Flexed (QKF), Squat, Step-up. Five repetitions of each exercise were performed and a metronome was used to pace the movements. The mean RMS of the EMG signal was normalized to that of the MVIC, for each muscle. To determine if EMG activity differed among the sampled hip muscles within each exercise, separate one-way repeated measures analyses of variance (ANOVAs), and simple contrast tests, were performed. The alpha level for all tests of significance was 0.05.

RESULTS (Table 1) AND DISCUSSION

ABD: TFL had significantly higher EMG activity than SUP-GMAX and INF-GMAX.
Bi- and Uni- Bridging: SUP-GMAX and INF-GMAX had significantly higher EMG activity than TFL.
Clam: SUP-GMAX had significantly higher EMG activity than TFL and INF-GMAX.
Hip Hike: TFL had significantly higher EMG activity than INF-GMAX.
QKE and QKF: INF-GMAX had significantly higher EMG activity than TFL.
Squat: SUP-GMAX and INF-GMAX had significantly higher EMG activity than TFL.
Step-up: No significant differences were found among muscles.

SUMMARY/CONCLUSIONS

The best exercises for recruiting the SUP-GMAX while maintaining a relatively lower activation of TFL were the Clam, Unilateral and Bilateral Bridging, and the Squat. The best exercises for recruiting the INF-GMAX while maintaining a low activation of TFL were Quadruped Hip Extension with Knee Extending and Knee Flexed, Unilateral and Bilateral Bridging, and the Squat. The GMED was most active during Side-lying Abduction and Hip Hike, but this was also true for TFL. Results from this study should assist clinicians in prescribing appropriate exercises to promote hip muscle strength and/or neuromuscular control.

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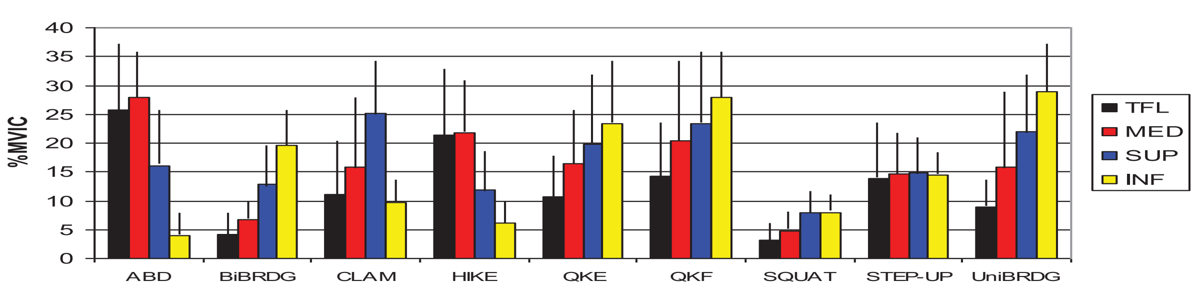
Dr. Lucinda Baker for EMG consultation.

Table 1. Hip Muscle Activity by Exercise
(mean, sd bars)



NEUROMUSCULAR CONTROL DURING ISOKINETIC KNEE EXTENSION IN TOP LEVEL KARATEKA

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INTRODUCTION

Karate is classified among those specialties requiring high technical skills such as a fine control of movement both in static and dynamic conditions, accompanied by a great ability to perform the main technical actions as fast as possible (ballistic contractions). Thus, this work was designed to study the neuromuscular response of knee flexor and extensor muscles during isokinetic contractions in 12 top level male karateka (Age: 30±2 yrs).

METHODS

The surface Electromyographic signal (sEMG) was recorded with an adhesive 4-array electrode from the right vastus lateralis (VL) and biceps femoris (BF) during maximal isometric knee flexion and extensions (MVC) and during three isokinetic contractions at different angular velocities (30°, 90°, 180°, 270°, 340°, and 400°/s). Torque/Velocity (TV) curves was computed for VL and BF muscles. The level of activation of VL and BF muscles while acting as antagonist was quantified through normalized sEMG Root Mean Square value (RMS_{Ant} (%)). The average muscle fibre conduction velocity was also computed for VL (VLCV) and BF (BFCV).

RESULTS AND DISCUSSION

For VL muscle, torque ranged from 314.1±56.8Nm (MVC) to 101±13.5Nm (400°/s,); BF torque values ranged from 168.3±22Nm to 79.4±12.9Nm (MVC and 400°/s, respectively) (Figure 1).

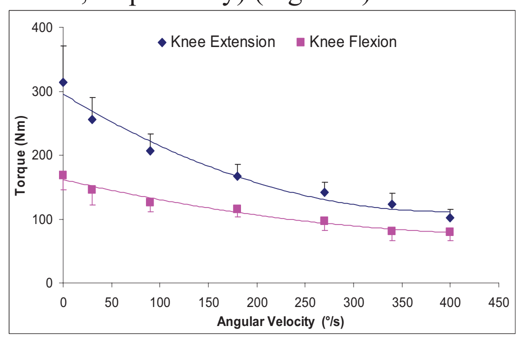

Figure 1. Torque/Velocity curves obtained during knee flexion-extension at different angular velocities. Data are expressed in absolute values.

Normalized T/V curves showed higher values of torque for BF muscle compared to VL at any angular velocity except for 30°/s. MFCV reached its peak at 30°/s in both VL and BF muscles (6.48±0.7m/sec and 3.83±0.52 m/sec, respectively). Afterwards, CV showed a progressive decrease as angular velocity increases for both muscles; a slight increase in VLCV at 400°/s was observed (Figure 2)

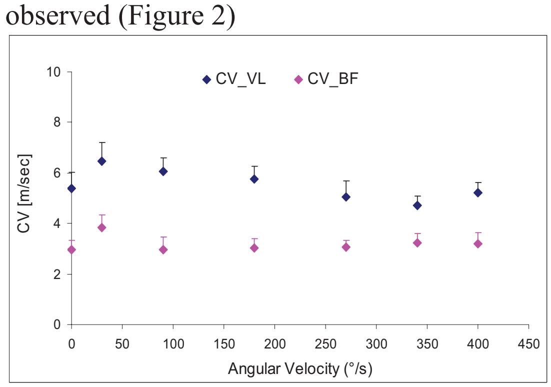

Figure 2. Conduction Velocity (CV) obtained in VL and BF muscles at different angular velocities. Data are expressed in absolute values.

VL-RMS_{Ant}(%) ranged from 2.54% during the MVC to 24.15% at 400°/s; BF-RMS_{Ant} ranged from 1.5% to 12.5% (MVC and 400°/s, respectively) (Figure 3).

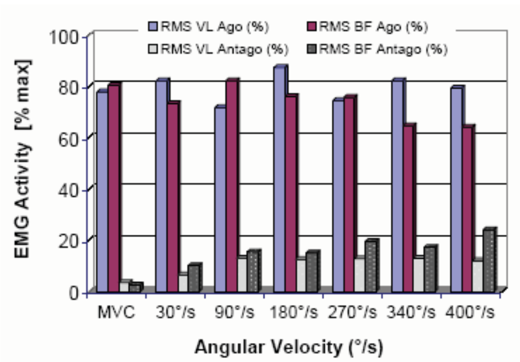

Figure 3. EMG activity of VL and BF during knee flexion-extension at different angular velocities. Normalized RMS data have been reported.

The higher normalized BF torque values obtained in this study may be closely related to the functional specialization developed in these athletes. In fact, the knee flexor action in eccentric condition (higher torque % and lower level of activation) plays a key role in these athletes in order to exert a sort of "braking action" necessary in maintaining static and dynamic postures of the lower limbs during the technical actions. The increased VLCV value observed at the higher angular velocity may be ascribed to a consistent recruitment of fast MUs as it is expected in ballistic exercises.

SUMMARY/CONCLUSIONS

The neuromuscular activation strategy adopted by top level karateka during knee flexion-extension at different angular velocities was assessed in this study. The results obtained support the hypothesis of a peculiar neuromuscular control strategy adopted by these athletes, consisting of selectively activating agonist muscles with a minimal intervention of antagonist muscle groups (low values for BF-RMS_{Ant} and VL-RMS_{Ant}).

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THE RELATIONSHIP BETWEEN OPEN AND CLOSED KINETIC CHAIN STRENGTH OF THE LOWER LIMB AND JUMPING PERFORMANCE IN THE ANTERIOR CRUCIATE LIGAMENT DEFICIENT SUBJECTS

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INTRODUCTION

The ability to jump is a fundamental skill that is required in many sports. Jump tests are widely used to determine an athlete's physical ability, to measure the outcome of a training program, and as a functional measure of an athlete's readiness to return to sport following a sports-related injury. Several studies show that decreased jumping performance accompanies the rupture of anterior cruciate ligament (ACL) of the knee. Researchers have examined several factors that are thought to contribute to jump performance. Several investigations have examined the relationship between muscular strength and vertical jump performance.

Closed kinetic chain (CKC) exercise has become popular in rehabilitation partly due to the belief that it is more closely related to function than open kinetic chain (OKC) resistance. Blackburn and Morrissey (1989) found that lower limb extensor CKC muscle strength is more highly related to jumping performance than knee extensor OKC strength. However, no studies have examined the relationship between OKC and CKC strength of the lower limb extensors and jumping performance in the subjects after an ACL injury. The purpose of the present study was to investigate the relationship between OKC and CKC strength of the lower limb extensors and jumping performance in a group of ACL-deficient subjects and compared them with a group of healthy controls and a group of ACL-reconstructed patients.

METHODS

We studied the isokinetic strength measurement and the jumping performance in a group of ACL-deficient (n=22) patients and compared them with a group of healthy controls (n=16) and a group of ACL-reconstructed patients (n=16). All ACL-reconstructed patients underwent double-bundle anatomical ACL reconstruction using hamstring tendon grafts from the index leg (Yasuda et al. 2004) and were evaluated at 12 months after the surgery. Each subject performed isokinetic strength test for the knee extensors in OKC and for the hip, knee, and ankle extensors in CKC. Isokinetic peak strength for the knee extensors in OKC was measured at 180°/sec using KIN-COM AP (TN, USA). Isokinetic peak strength for the hip, knee, and ankle extensors in CKC was measured at 360°/sec using Strengthergo.240 (Mitsubishi Electric Co., Tokyo, Japan)(Fig. 1). The one-leg vertical jump and the one-leg long jump tests were also assessed. Pearson's correlation analysis were performed to examine the correlation of the isokinetic peak strength per body weight with the height of the vertical jump and the distance of standing long jump in each group.

Figure 1: CKC strength measurement

RESULTS AND DISCUSSION

In all groups, the correlation analysis showed that OKC and CKC strengths were significantly correlated with vertical jump performance and standing long jump performance. However, the correlation between kinetic strength and jump performance were weaker in the ACL-deficient group than in the control and ACL-reconstructed groups (Fig. 2). In the control and ACL-reconstructed groups, lower limb extensor CKC muscle strength is more highly related to jumping performance than knee extensor OKC strength, but CKC strength in the ACL-deficient group is not highly related to jumping performance (Table 1). Demont et al. (1999) reported that the muscular activity in ACL-deficient leg during functional activities was different from that in the normal control subjects. Therefore, the change in muscular activity in ACL-deficient leg during jumping performance may result weak correlation between CKC strength and jumping performance in the ACL-deficient subjects.

SUMMARY/CONCLUSIONS

ACL-deficient subjects have weaker correlation between CKC muscle strength of lower limb extensor and jumping performance compared with normal and ACL-reconstructed subjects.

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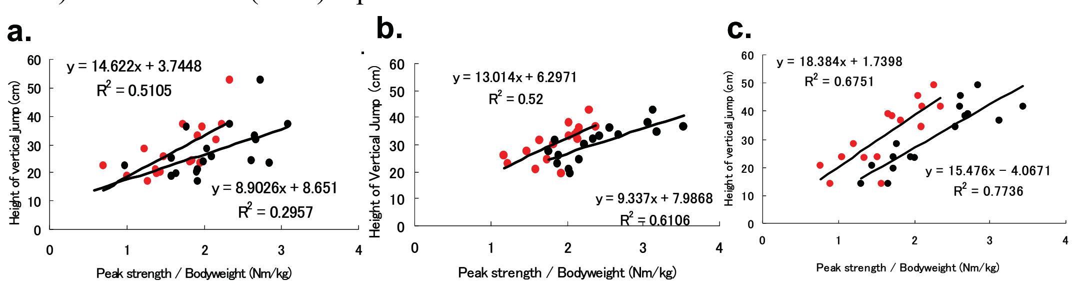

Figure 2: Correction between OKC (red) and CKC (black) peak strengths and the height of vertical jumping performance (a. ACL-deficient, b. control, c. ACL-reconstructed).

Table 1: Correction coefficients between isokinetic strengths and jumping performance

Group	**ACL-deficient**		**Control**		**ACL-reconstructed**	
	OKC	**CKC**	**OKC**	**CKC**	**OKC**	**CKC**
Vertical Jump	0.714	0.544	0.721	0.781	0.822	0.880
Long Jump	0.610	0.520	0.729	0.864	0.741	0.921

INCREASED CO-ACTIVATION OF ANTAGONIST MUSCLES BY FORCE MODULATION TRAINING

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INTRODUCTION

The most common methods for skeletal muscle conditioning are based on the application of a steady resistance. However, as confirmed by several studies on vibration training, the involvement of neuromuscular reflex due to force variations plays also an important role in the muscular conditioning processes [1]. This is mainly due to the stimulation of the muscle spindles and golgi tendon organs, which are sensitive to variations of muscular and tendon length. As a result, several mono- and poly-synaptic reflexes are activated that induce the muscle to compensate for the length variations.

Among the compensation mechanisms, particularly interesting is the reflex induced by mechanical vibrations, which stimulate intensively the muscle stretch sensors. The muscle tries to stabilize the joint by the simultaneous co-activation of agonist and antagonist muscles [1, 2]. This phenomenon, referred to as tonic vibration reflex (TVR), is a complex phenomenon. The spindle afferents seem to stimulate impulses following a polysynaptic excitatory pathway and a presynaptic inhibitory pathway. The first is responsible for the TVR while the latter inhibits vibration-induced reflexes. This is referred to as vibration paradox.

The reflex inhibition is dependent on the vibration amplitude rather than its frequency [3], while the excitatory component is dependent on the vibration frequency. Based on previous studies [2], the maximum fiber recruitment in the vastus lateralis, measured in terms of root mean square (RMS) of the surface electromyogram (sEMG), is obtained for a 30-Hz vibration.

For a better understanding of the physiological processes stimulated by fast force variations, a prototype was built where the force applied to the muscles can be modulated at higher frequency [4]. A preliminary study on biceps and triceps revealed an electrical hyperactivation for a 29-Hz force modulation (FM) [4].

In this paper, the same prototype described in [4] is used to evaluate the effects of a 29-Hz FM on the co-activation of antagonist muscles. This is important as the co-activation of antagonist muscles plays a fundamental role in the joint stabilization and prevention of ligament injuries [5].

METHODS

In order to evaluate the effects of FM on the co-activation of antagonist muscles, we realized and used a prototype for muscular conditioning where the force applied to the muscles, generated by an electromagnetic actuator, can be modulated over time during the exercise. The FM control is implemented in Labview® (National Instruments) on a personal computer.

The measurements were performed during isometric contractions of the biceps at different intensities (baseline force) with and without FM at 29 Hz. Elbow frontal elevation and angle were fixed to 30° and 90°, respectively. The sEMG signal was recorded on the triceps in a bipolar configuration. The inter-electrode distance was 2 cm. The sampling frequency was 2048 Hz. The measurement system was a Mobi8 (TMS International) wireless amplifier with two Ag-AgCl electrodes.

The measurements were performed on 12 volunteers (7 females and 5 males with ages between 20 and 27 years) on the right arm. The baseline force was varied to generate muscular contractions ranging between 20 and 100 % of the maximum voluntary contraction (MVC) with steps of 20 %. The MVC was estimated as the maximum contraction that could be sustained for 10 s.

The recordings were analyzed for a time interval of 6 s, starting 2 s after the beginning of the contraction. Before the analysis, the signal was band-pass filtered between 18 and 200 Hz [6]. The RMS was derived from the integral of the power spectrum over the analyzed time interval. The power spectrum was estimated by Discrete Fourier Transform. To avoid motion artifacts to affect the results, the 29-Hz component was excluded from the power spectrum integration.

RESULTS

The average results for the 12 subjects are reported in Fig. 1. All the measurements are normalized with respect to the RMS estimated for 100 % of the MVC without FM. Averaging over all MVC values, the application of a FM produced an increase of 59 % of the triceps co-activation.

DISCUSSION AND CONCLUSIONS

The co-activation of antagonist muscles during isometric contraction in the presence of FM was studied. The RMS of the sEMG measured on the triceps during isometric elbow flexion task was considered as the measure for muscular co-activation.

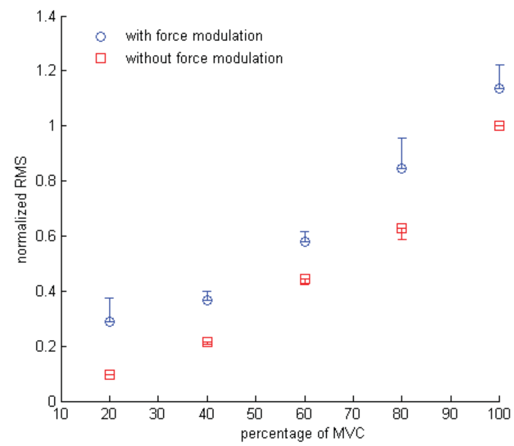

Figure 1. Average normalized RMS value of the sEMG measured on the triceps of 12 test subjects during isometric elbow flexion task at different MVC percentages.

The results showed an increased (59 %) co-activation of the triceps in the presence of FM. FM seems therefore to be an efficient method for training the co-activation of antagonist muscles. This is desirable to improve joint stability and to prevent ligament injuries.

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THE EFFECT OF PATELLAR TAPING ON VMO ACTIVATION DURING FUNCTIONAL ACTIVIES IN HEALTHY SUBJECTS

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INTRODUCTION

Patellofemoral pain syndrome (PFPS) is a common knee condition in adolescents and young adults. A common therapy for PFPS, which lacks strong scientific evidence, is the application of patellar tape in order to realign the patella medially, presumably improving patellar tracking (McConnell, 1986). This has been thought to reduce pain, increase the activation amplitude and/or shorten the activation latency of the vastus medialis obliquus (VMO) muscle, eventually leading to the preferential training of this muscle such that it can stabilize the patella without the need for tape (Crossley, 1996). Patellar taping is also thought to stretch tight lateral structures which may allow the patella to position itself properly (Christou, 2004). No published studies have investigated the efficacy of patellar taping on activation amplitude of the VMO over a wide variety of functional activities. In addition, no studies have compared VMO activation responses to taping using a sex-based analysis. The purpose of this pilot study was to determine whether patellar taping affected the activation amplitude of VMO in healthy subjects during functional activities and to compare these effects between males and females.

METHODS

Seven healthy males and 7 healthy females were recruited to participate. Bipolar surface electromyography (SEMG) electrodes were placed on the thigh of the dominant leg. All activation amplitudes were normalized by each subject's maximum voluntary effort (%MVE) which was induced by taking the highest of three maximum voluntary isometric contractions. Subjects performed seven functional activities, first under a control condition (no tape) followed by the test condition where medial glide patellar taping was applied.

A three-way repeated measures ANOVA was performed with condition (tape or no tape), task (the seven functional activities) and sex (male or female) as factors. Subject was included in the model as a random factor. The interactions of task by condition, sex by condition, and sex by task were also included in the model. Where interactions existed, the data were analyzed using univariate post-hoc analyses by fixing one factor and testing the others (alpha adjusted accordingly).

RESULTS AND DISCUSSION

There was a significant interaction between sex and condition ($p = 0.001$), therefore males and females were analyzed separately. When all tasks were considered together, a lower VMO activation amplitude was seen in both sexes when the tape was applied, although the difference was only statistically significant for females ($p < 0.001$ for females; $p = 0.065$ for males). The post-hoc analysis of the effect of tape by each task showed that males had significantly lower VMO activation in only three of the seven tasks. The females showed lower VMO activation with tape in five of the tasks. The mean %MVE and standard deviations for both sexes are displayed in Table 1.

CONCLUSIONS

This study showed that VMO activation amplitude was actually lower when medial glide patellar taping was applied. This finding is opposite what is commonly believed by clinicians.

One plausible explanation for this result is that we used healthy subjects and not individuals with PFPS. Individuals with PFPS are thought to have an imbalance between the medial and lateral stabilizers of the patella which impairs the tracking of the patella within the trochlear groove during activities requiring knee flexion and extension. Subjects without PFPS presumably have a proper balance of strength between the vastus lateralis and VMO such that medial glide patellar taping might actually inhibit VMO activation through active insufficiency. Alternatively, when a healthy knee is taped, the nervous system may lessen the activation of the VMO in order to maintain the already balanced gliding of the patella (Christou, 2004).
It is also possible that in healthy subjects the knee receives proprioceptive feedback telling it that the patella is already in a good location, and therefore the VMO should not contract.

These findings could potentially influence the way that PFPS is treated in young adults, as, while clinicians prescribe patellar taping to increase VMO activation and promote muscular retraining, the treatment may have the opposite effect. Therefore while patellar taping may still be used to reduce the pain felt by patients, these findings suggest that the treatment is only a temporary solution. This will have to be confirmed through testing a sample of individuals with PFPS. If the results hold true, in order to achieve long term benefits related to VMO training, patellar taping may need to be coupled with separate exercises to specifically target the VMO. Also it has been demonstrated that taping affects males and females differently, which may need to be taken into consideration when treatment programs are being prescribed.

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Table 1: Normalized VMO SEMG for males and females under each condition (no tape vs tape) Asterisks (*) indicate significant differences between the treatment and control conditions..

Task	Female Mean (% ± SD)		Male Mean (% ± SD)	
	No tape	Tape	No tape	Tape
Full Squat	160.95±1.13	113.81±0.500*	101.30±0.535	86.12±0.376*
Half Squat	108.16±0.684	71.91±0.268*	67.16±0.312	58.99±0.239
Sit down in chair	95.56±0.358	73.86±0.186*	57.02±0.262	60.26±0.334
Stand up from chair	131.42±0.716	88.53±0.375*	85.31±0.391	73.39±0.335*
Ascend a step	119.14±0.378	109.23±0.350	87.45±0.453	85.39±0.541
Descend a step	65.57±0.365	37.55±0.254*	37.52±0.210	44.10±0.427
Jump down from ledge	161.91±0.683	148.55±0.792	136.15±0.534	115.00±0.419*

NEUROMUSCULAR RESPONSES TO CONTINUOUS AND INTERMITTENT VOLUNTARY CONTRACTIONS

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INTRODUCTION

It is well known (Bouissou et al. 1989; Brody et al. 1991) that the change in EMG variables during sustained isometric contractions is associated, albeit not exclusively (Vestergaard-Poulsen et al. 1995), with a decrease in blood pH and with the lack of oxygen (Hogan et al. 1994; Russ and Kent-Braun 2003) which is considered as the main factor generating acidosis, in association with the accumulation of H^+ protons as a consequence of the ATP consumption.

A project, supported by the Italian Ministry of Health, was carried out aiming to further highlight the role of oxygen in muscle fatigue. Mechanical and EMG manifestations of fatigue in continuous and intermittent contractions were compared to assess their effectiveness in distinguishing between different athlete phenotypes: endurance vs power trained subjects

METHODS

Thirteen power athletes (PA: weight lifting and martial arts, age: 24.7±4.9 years, height: 175±6 cm, weight: 76.2±11.3 kg) and nineteen endurance athletes (EA: triathlon and cycling) (age: 23.3±6.9 years, height: 175±7 cm, weight: 69.2±8.5 kg) were recruited. The dominant vastus lateralis muscle was investigated during isometric contractions at the leg press.

Subjects performed, in randomized order, one continuous isometric contraction (duration: 30 s; intensity: 90% MVC) and one intermittent isometric contraction of the knee extensor muscle groups (duration: 40 s; 3 s contraction plus 1 s relaxation; intensity: 90% MVC).

Peak and rates of change of force, initial values and rate of change of EMG variable estimates (amplitude, ARV; conduction velocity, CV; and mean frequency of the signal spectrum, MNF) were calculated for the two contraction modalities.

RESULTS AND DISCUSSION

Rates of change of force and of EMG variables were found greater in continuous than in intermittent contractions in the group of athletes as a whole (N=32, Wilcoxon paired test, $p<0.01$).

No difference in force peak was found in both groups in the two contraction modalities. Moreover, force peak was found different between the two athlete groups in both continuous (PA: 182.3±46.8 kg, EA: 123.7±29.3 kg, Mann-Whitney U test, $p=0.0012$) and intermittent (PA: 176.2±32.9

kg, EA: 121.6±23.7 kg, Mann-Whitney U test, p=0.00007) contractions.

No significant differences were found between power and endurance athletes in the mechanical and myoelectric manifestations of fatigue during continuous contractions (force rate of change: EA -0.84±0.37 %/s vs PA -0.97±0.77 %/s, p=ns; CV rate of change: EA -0.43±0.25 %/s vs PA -0.47±0.25 %/s, p=ns), whereas significant differences between the two groups were evident in the comparison of intermittent contractions (force rate of change: EA -0.55±0.24 %/s vs PA -0.76±0.26 %/s, $p<0.01$; CV rate of change: EA -0.13±0.11 %/s vs PA -0.44±0.49 %/s, $p<0.001$).

A possible explanation of the differences between continuous and intermittent contractions is that the one-second rest time between contractions was enough to allow a partial recovery of the membrane properties. Oxygen supply and metabolite wash-out, during the relaxation phase of the intermittent contractions, partially decrease both mechanical and myoelectric manifestations of fatigue.
Furthermore, significant differences between the two groups were observed only for intermittent contractions and were also greater for myoelectric than for mechanical fatigue. These findings highlight the importance of selecting the best exercise modality to compare athletic performances and quantify muscle fatigue.

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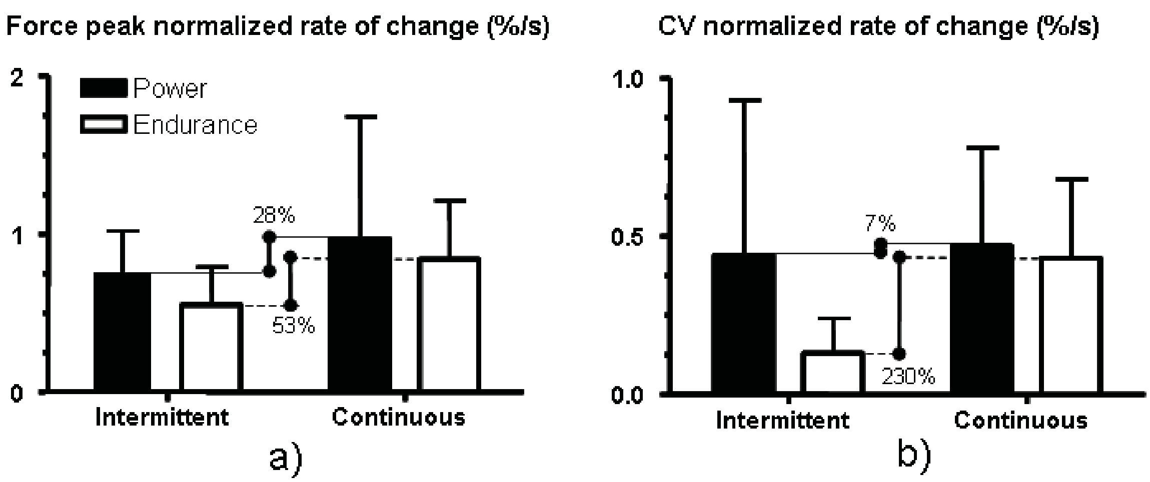

Normalized (with respect to initial values) rates of change of Force peak (a) and CV (b) in continuous and intermittent contractions for both groups. The increment of fatigue in the continuous with respect to the intermittent contraction was greater in the endurance than in the power athletes. Mean SD absolute values are reported.

EFFECT OF POWER ASSIST RATIOS ON PHYSICAL ACTIVITIES DURING CYCLING WITH A TOR UE ASSISTED BICYCLE

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INTRODUCTION

We have studied variations in physical activity and vehicle performance during repetitive cycling trials in the field on a torque-assisted bicycle (TAB) (Kiryu and Yamagata, 2006). Measured data were ECG and surface EMG (SEMG) as biosignals and the speed, cadence, and torque as vehicle data. Estimating the respiratory-sinus–arrhythmia (RSA)-related power from R-R interval time-series (Hayano *et al.*, 1994) and muscular-fatigue-related index from SEMG, we categorized the physical activity into four groups for each trial. The results showed that the torque-assist possibly enlarged the RSA-related power ratio, but was not always effective for preventing development of muscular fatigue.
In this paper, we verify the results based on 50 subjects on the previous experimental field and 23 subjects on the other experimental field. The experimental fields included a steep uphill section near the middle. Then, we verified the relationship between muscle activity at each pedal stroke during climbing and autonomic regulation during rest after climbing, and demonstrated the effect of TABs with different power-assist ratios.

METHODS

Two experimental fields of approximately 870 m and 2100 m long were the circuit courses around buildings (course-A) and in the resort field (course-B), respectively. We divided each circuit into three phases based on the inclination. The lengths of climbing phase were 216 m and 400 m in courses A and B, respectively. An experimental set consisted of four or three consecutive trials with different power-assist ratios and each trial was separated over 5-min. Each participant was asked to keep the pedaling rate as close to 60 rpm as possible by listening to a tone pace maker. We used the power assist ratios of 1:0.5, 1:1, 1:1.5, and 1:2 in the course-A, using a customized TAB. In the course-B, we used a commercial TAB with the power assist ratios of 1:0.5 and 1:1.

We recorded ECG at the chest and bipolar SEMG from the right vastus lateralis muscles using active two-bar electrodes (DE2.1, Delsys). Both the ECG and SEMG were sampled at 2048 Hz with 12-bit resolution. We focused on the RSA-related power ranging from 0.3 to 0.6 Hz in R-R interval time-series and calculated the time-varying RSA-related power ratio, pr_{RSA}. Regarding muscle activity, we estimated the averaged rectified value (ARV) and the mean power frequency (MPF), estimating them with a sliding 100-msec interval every 10-msec in each pedal stroke interval of 200-msec. We divided a contraction into the first and second half and focused on the first half. Then, we obtained the correlation coefficients between ARV and MPF ($\gamma_{\mathrm{ARV\text{-}MPF}}$) at every stroke.

RESULTS AND DISCUSSION

The participants were informed of the risks involved and signed a consent form in advance. Fifty healthy young male volunteers (21.4 ± 1.4 yrs) participated in the course A. In the course B, the participants were 10 healthy young male volunteers (22.1 ± 0.7 yrs), 6 middle age female volunteers (49.2 ± 8.4 yrs), and 7 elderly male volunteers (59.7 ± 6.4 yrs).

We studied the relationship between pr_{RSA} after climbing and $\gamma_{\mathrm{ARV\text{-}MPF}}$ just before the hilltop with respect to the power assist ratio. The pr_{RSA} was significantly enlarged as the power assist ratio increased. On the other hand, $\gamma_{\mathrm{ARV\text{-}MPF}}$ closed to zero at higher power ratios in the course A, but in the course B $\gamma_{\mathrm{ARV\text{-}MPF}}$ enlarged a negative value at 1:1. Since $\gamma_{\mathrm{ARV\text{-}MPF}}$ fluctuated stroke by stroke, we then averaged $\gamma_{\mathrm{ARV\text{-}MPF}}$ every segment that consisted of non-overlapping 10 strokes and classified each segment into a negative or positive segment. Positive and negative $\gamma_{\mathrm{ARV\text{-}MPF}}$ means increasing muscle activity and muscle fatigue, respectively. Based on the contraction pattern for consecutive segments during climbing, we categorized 109 trials into five patterns: fatiguing contractions (FC: 43.1%), gradually fatiguing contractions (GF: 11.9%), alternative contractions (AC: 12.8%), contractions with stable performance (SP: 8.3%), and contractions with unstable performance (uSP: 23.9%).

Figure 1 shows the relationship between contraction pattern and cadence. Actually, the fluctuation of the cadence was small around 60 rpm for SP, whereas the cadence decreased around 50 rpm for others. However, there were trials in which the contraction pattern was FC but the fluctuation of the cadence was small. This was a case in the course B, and participants tried to challenge for regulating the cycling interval as 60 rpm as possible at 1:1 because they were not able to regulate it at 1:0.5. Moreover, alternation of agonist muscles (Knaflitz and Molinari, 2003) was sometimes observed for AC. In this case, muscular fatigue could not be observed due to fixing electrodes on the specific muscle.

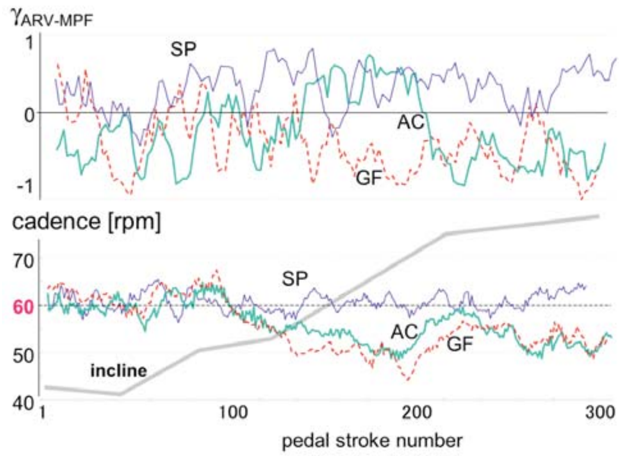

Figure 1: Time-series of $\gamma_{\mathrm{ARV\text{-}MPF}}$ and cadence during climbing with the power assist ratios of 1:0.5 in course-B.

SUMMARY/CONCLUSIONS

For long-term repetitive cycling, we confirmed that a higher torque-assistance ratio enlarged the capacity of autonomic regulation, but did not always support to prevent development of muscular fatigue during climbing. Accordingly, monitoring the autonomic-regulation-related index after climbing and contraction pattern regarding muscular-activity-related indices during climbing would be useful to customize a torque-assisted bicycle for individuals.

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EFFECTS OF EXERCISE-INDUCED MUSCLE DAMAGE ON FATIGUE-RELATED VOLITIONAL AND MAGNETICALLY-EVOKED NEUROMUSCULAR PERFORMANCE

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INTRODUCTION

Knee flexors that function optimally are considered fundamental to the dynamic stabilisation of the knee joint and to the prevention of anterior cruciate ligament (ACL) injury (Johannson et al., 1991). Following unaccustomed high-intensity eccentric exercise skeletal muscle is susceptible to ultrastructural damage (Brockett et al., 2002); functional sequelae include immediate and prolonged loss of muscle force and power of up to 70% and 20%, respectively (Byrne et al., 2004). These types of impairments, and the susceptibility of principally high threshold motor units to damage (Brockett et al., 2002), may substantively compromise the dynamic protective capability of the neuromuscular system. Currently there is limited investigation of the concomitant effects of acute maximal exercise in muscle symptomatic of damage; further impairments to neuromuscular performance capabilities to initiate and muster force quickly may be an antecedent to joint injury. Recent research findings from the estimation of maximum neuromuscular performance capacity by means of magnetic stimulation has shown improved electromechanical delay scores following acute fatigue, and describes a potential compensatory mechanism that may counteract the effects of fatigue-induced impairments during periods of critical threat to the joint system (Minshull et al., 2007).

This study examined the effects of EIMD on fatigue-related knee flexor voluntary and magnetically-evoked neuromuscular performance.

METHODS

Voluntary and magnetically-evoked indices of neuromuscular performance of the knee flexors of the dominant leg of seven males (age: 28.3 ± 7.02 yrs; height 1.84 ± 0.05 m; body mass 83.0 ± 13.9 kg [mean ± SD]) were obtained prior to (pre) and up to 168 hours (168h) following three treatment conditions: (i) an eccentric exercise-induce muscle damage condition (EIMD) of the knee flexors of the preferred leg (performed on the first assessment occasion), in addition, to a static fatiguing exercise task (35s) of the same muscle group, performed on every assessment occasion; (ii) a control condition (CON2) of equivalent duration to the intervention, consisting of the static fatiguing exercise task only; (iii) and a further control condition (CON1) of equivalent duration to the EIMD and CON2, consisting of no exercise. Performance measures were assessed additionally prior to and immediately following each occurrence of the static exercise task (or equivalent period of rest).

RESULTS AND DISCUSSION

An increase in serum CK values ($p < 0.001$), which was most prominent at 72h (93.3 ± 2.76 U/L vs. 1117.9 ± 8.55 U/L, pre- vs. 72h post, respectively), and ratings of perceived soreness ($p < 0.001$) that commenced at 24h and was most prominent at 48h, strongly suggest the presence of eccentric exercise-induced muscle damage. The EIMD condition was also associated with impaired volitional peak force (PF_V) ($F_{[10, 60]} = 4.6$, $p < 0.001$) and rate of force development (RFD_V) $F_{[10, 60]} = 2.8$, $p < 0.01$) that was most prominent at 48h following eccentric exercise (37.5%; 65.2%, of baseline scores, respectively). Volitional electromechanical delay (EMD_V) and all indices of magnetically-evoked performance capacity were maintained during EIMD. The preservation of neuromuscular performance capacity as measured by magnetic stimulation throughout EIMD, alongside the concomitant manifestation of the maximum pain response and impairments to PF_V and RFD_V (48h), may be evidence of vital compensatory strategies to help prevent the sports performer becoming injured on each occasion when they experience muscle damage.

The static exercise task induced fatigue, characterised by an immediate loss to knee flexor PF_V performance of a similar magnitude across both treatment conditions. The additional reductions to volitional contractile capabilities (group-mean 6.3% reduction in PF_V) in addition to the effects of muscle damage (see figure 1) may present substantive challenges to the neuromuscular system to protect the ACL during mechanical loading of the knee joint, especially at knee angles proximal to full extension. However, the acute exercise task was associated with a potentiation of EMD_E (on average 13.0% and 15.7% improvement compared to pre-fatigue values; CON2 and EIMD, respectively) ($F_{[2, 12]} = 8.6$, $p < 0.01$), which suggests a remarkable preservation of neuromuscular capacity under conditions of acute muscle fatigue and in the presence of suspected mechanical disruption to the musculature.

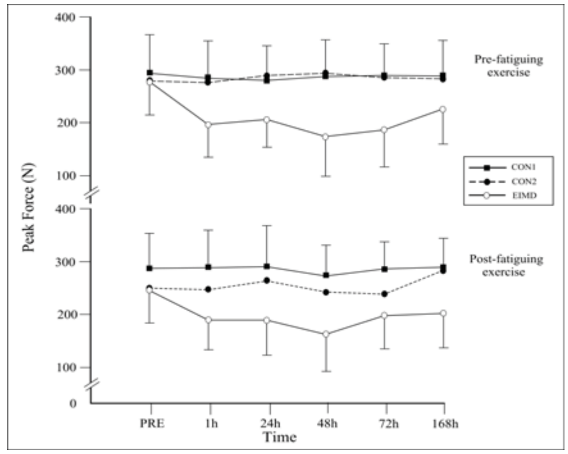

Figure 1: Peak force (PF_V) performance over the three treatment conditions (group mean ± SD).

SUMMARY

At knee angles where key ligamentous structures are under greatest mechanical strain, substantive impairments to the volitional capability of the knee flexors may apparently place the sports performer at increased risk of injury. However, the subsequent deployment of possible neuromuscular compensatory strategies including the ‘down-regulation’ of inhibitory process that may restrict access to the full capacity of motor units during critical periods of joint loading, suggest the potential for improved dynamic protection.

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EFFECTS OF MODE OF FLEXIBILITY CONDITIONING ON NEUROMECHANICAL PERFORMANCE IN MALES

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INTRODUCTION

Enhanced neuromechanical performance capabilities including flexibility, peak force and electromechanical delay are fundamental contributors to successful execution of physical skills and sports performance and may reduce the susceptibility of an athlete to injury (Griffin, 2003). Optimal functioning of the musculature in a timely manner is paramount to the dynamic stabilisation of synovial joints (Minshull *et al*. 2007). Passive and proprioceptive neuromuscular facilitation (PNF) conditioning techniques for the improvement of flexibility have become popular and efficacious. However, the techniques may also provoke either compromised net functional outcomes associated with increased tissue compliance and delayed muscle activation, or alternatively, enhanced performance especially at the extremes of the joint's range of motion because of relatively intense muscle activation during conditioning (Hartig and Henderson, 1999). Therefore the aim of this study was to assess the effects of mode of flexibility conditioning on neuromechanical performance in males.

METHODS

Passive hip flexibility, peak force, and electromechanical delay (volitional and magnetically-evoked) were assessed in eighteen males (age: 20.6 ± 2.1yr; height: 1.78 ± 0.06m; body mass: 71.3 ± 7.8kg [mean ± SD]) randomly assigned into two groups, prior to and immediately after: (i) an intervention condition comprising six weeks of three-times weekly flexibility conditioning of the hip region and knee flexor musculature in the dominant limb involving either proprioception neuromuscular facilitation (PNF) (n=9) or passive exercise (n=9), (ii) a control condition consisting of no exercise during an equivalent time period. The contralateral limb of each participant was used as an additional control throughout the experimental period.

RESULTS AND DISCUSSION

The results showed that passive hip flexibility increased to a similar extent (18.7% mean increase in performance) in both experimental groups (passive: 89.1 ± 5.2 deg vs. 99.6 ± 5.0 deg [11.7% increase in performance]); PNF: 94.4 ± 18.6 deg vs. 120.1 ± 16.4 deg [27.2% increase in performance]; $F_{1,16} = 14.0$; $p < 0.01$). Peak force performance capabilities (309 ± 82N [overall group mean scores]) was not influenced by either conditioning intervention (p > 0.05). Volitional electromechanical delay was substantively increased following passive flexibility conditioning (40.9 ± 3.2 ms vs. 37.0 ± 3.5 ms [10.5% increase compared to baseline]), but increased by a lesser extent by PNF conditioning (38.9 ± 3.3 ms versus 37.7 ± 3.9 ms, respectively [3.2% increase]); $F_{1,16}$ =

16.8; $p < 0.01$; Table 1). Magnetically-evoked electromechanical delay showed similar patterns of change (passive: 22.3 ± 3.2 ms vs. 19.1 ± 2.8 ms [16.8% increase]); PNF: 19.5 ± 4.1 ms versus 18.4 ± 3.9 ms, respectively; $F_{1,16} = 12.9$; $p < 0.01$; Table 1).

While both modes of flexibility conditioning demonstrated efficacy over a six-week intervention period, PNF conditioning in particular had a lesser impact negatively on other important factors in performance contributing to successful dynamic stabilization of synovial joints. Preservation of strength performance and the capability for rapid activation of the musculature in volitional and emergency situations may have been facilitated by the relatively high-intensity muscle activation patterns inherent in PNF conditioning. In contrast, passive flexibility conditioning elicited increased volitional and evoked electromechanical delays.

CONCLUSIONS

In conclusion, while both modes of flexibility conditioning demonstrated efficacy over a six-week intervention period, PNF conditioning in particular had a lesser impact negatively on other important factors in performance contributing to successful dynamic stabilisation of synovial joints. Preservation of strength, sensorimotor performance and the capability for rapid activation of the musculature in volitional and emergency situations may have been facilitated by the relatively high-intensity muscle activation patterns inherent in PNF conditioning. In contrast, passive flexibility conditioning elicited increased volitional and evoked electromechanical delays.

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Table 1. Group mean electromechanical delay (EMD, volitional and magnetically-evoked, ms) associated with the knee flexor musculature (30° degrees of knee flexion) at pre- and post-intervention period assessments. Data are mean (± SD).

		Time			
		Pre		**Post**	
EMD volitional (ms)	**Group**	Experimental period	Control period	Experimental period	Control period
Intervention (preferred) limb	Passive	37.0 ± 3.5	36.7 ± 3.2	40.9 ± 3.2	36.2 ± 3.9
	PNF	37.7 ± 3.9	38.2 ± 3.4	38.9 ± 3.3	38.5 ± 3.4
Control limb	Passive	21.7 ± 4.5	21.6 ± 4.7	22.1 ± 2.6	21.7 ± 4.2
	PNF	39.8 ± 3.9	40.0 ± 3.7	40.4 ± 2.9	40.2 ± 3.8
EMD evoked (ms)					
Intervention (preferred) limb	Passive	19.1 ± 2.8	19.3 ± 3.0	22.3 ± 3.2	19.5 ± 2.9
	PNF	19.5 ± 4.1	18.6 ± 3.9	18.4 ± 3.9	18.6 ± 3.9
Control limb	Passive	21.7 ± 4.5	21.6 ± 4.7	22.1 ± 2.6	21.7 ± 4.2
	PNF	21.4 ± 3.4	21.2 ± 3.7	21.6 ± 3.4	21.0 ± 3.5

DROP LANDING BIOMECHANICS – DOES SUBSEQUENT TASK AFFECT NEUROMUSCULAR MOTOR (PRE) CONTROL STRATEGY?

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INTRODUCTION

In lower limb biomechanical analysis, Landing presents itself as a vital kinetic energy absorption phase. Research as provided us meaningful background information about preparatory landing responses during falling actions, suggesting these are intended to prevent hard impacts upon ground contact, having often been identified by measures of electromyographic (EMG) muscular activity (Liebermann et al, 2007).

Human musculoskeletal system apparently has the inherent ability to choose from several multi-joint neuromuscular motor control strategy options, therefore providing valid multiple possibilities for balance as well as for joint load distribution. Nonetheless all landing phases, within it's large number of intrinsic features of possible influence, introduces an increased injury factor, mainly due to exponentially higher load system exposure (Madigan et al, 2003).

Visual and vestibular information have been assumed to participate in a multiplicity of daily events and activities associated with landing actions. However rarely has preparatory motor control focussed in subsequent task orientation. Hence our main purpose in this study was to assess possible neuromuscular activation strategy changes when preparing for differentiated post-landing tasks.

METHODS

For this purpose we gathered a sample of ten volunteers (5 male and 5 female, aged 21,33±2,15 years).

Experimental testing setup consisted in subjects first performing three 60 cm drop landings onto a Force Plate, and subsequently executing three drop jumps (intending maximal jump height). Subjects were further instructed to land on both feet simultaneously, and also to place their hands on respective hips throughout testing, in order to prevent upper limb negative data influence. Full recovery was waranteed between attempts.

Subjects EMG activation was assessed using surface electrodes (BlueSensor, Medicotest) with bi-polar placement over the dominant lower limb's vastus lateralis (VL), vastus medialis (VM), internal gastrocnemius (GI), external gastrocnemius (GE) and biceps femoris (BF) muscles, in accordance to SENIAM's guidelines. EMG Data collected was recorded at 2000 Hz, using a pre-amplified telemetry system (Glonner), with *Simi Motion* software. Signal was processed using a bandpass [10-600] Hz filter.

Average EMG full wave rectified signal (AvgEMG) was determined as a function of force plate data. Vertical component force curves were plotted and subdivided into pre-activation time (100 ms until ground contact) and time until peak force (time current from ground contact until maximal Fz). Integral data was discarded due to short time EMG recording periods. Data Normalization procedures were also implemented and revealed no differences in results when compared to initial data collection.

Statistical independent samples t-test was determined in order to compare performance means between tasks.

RESULTS AND DISCUSSION

Results indicated no major differences in EMG pre activation before ground contact, apparently suggesting that our central concern when falling from a determined height seems to be, as previously stated in several research papers, musculoskeletal system protection from impact. However when closely observing data from table 1 we may verify that immediately after first ground contact EMG activation differs when preparing the neuromuscular system for distinct subsequent tasks. Hence it appears plausible to suggest that preparation for following goals occurs within different activation amplitudes, and also mostly due to reduced action timing (ballistic tasks), EMG activation appears to be enhanced immediately upon contact, certainly in preparation for following muscle action.

SUMMARY/CONCLUSIONS

In summary our results revealed differences in immediate ground contact EMG muscle activation when in preparation for after-landing tasks. Pre activation, although present, did not differ between groups.

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Table 1: Mean difference between Drop landing and Drop jumping.

60cm	**Drop landing**		**Drop jumping**		**dif**	**t**	**p**
	Mean	SD	Mean	SD			
EMG [mv]	Pre-Activation EMG [100 ms to first contact]						
AvgVL	0,107	0,047	0,081	0,040	0,026	1,563	0,162
AvgVM	0,119	0,039	0,124	0,064	0,006	-0,263	0,799
AvgGI	0,281	0,146	0,266	0,117	0,015	0,395	0,704
AvgGE	0,109	0,077	0,100	0,041	0,009	0,350	0,737
AvgBF	0,071	0,044	0,070	0,040	0,001	0,180	0,862
EMG [mv]	Landing EMG [Time until peak Fz]						
AvgVL	0,411	0,180	0,600	0,252	0,189	-3,610	0,007*
AvgVM	0,364	0,120	0,623	0,185	0,259	-4,668	0,002*
AvgGI	0,109	0,078	0,183	0,114	0,074	-2,095	0,074
AvgGE	0,064	0,049	0,154	0,095	0,090	-3,218	0,015*
AvgBF	0,060	0,026	0,087	0,041	0,027	-1,495	0,179
Ground Reaction Force [N]	Force-Plate Data						
Fz	4739,78	1558,19	4276,36	1402,49	463,43	0,952	0,372
Fz-t	46,67	7,07	48,89	7,82	2,22	-0,555	0,594
Imp	86,02	23,92	96,93	37,08	10,92	-0,981	0,355

* Results were statistically significant at $p<0.05$ level.

MUSCLE ACTIVITY AND ENERGETICS DURING NORDIC WALKING

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INTRODUCTION

Nordic Walking is walking with poles with an active arm push. It is a popular form of exercise in Scandinavia and Germany especially among middle aged and elderly.

Nordic Walking enhances oxygen consumption, energy consumption and heart rate compared to regular walking at the same walking speed. This has been documented both during treadmill walking (Rodgers et al. 1995; Porcari et al. 1997) and field testing (Church et al. 2002; Schiffer et al. 2006). However, the specific causes of these enhancements have not been documented, and remain to be speculations about increased muscle activity in the upper arm muscles during the push off phase of the pole and in the muscles in the forearm in order to grip the pole.

The aim of this study was to explore the causes of the enhanced physiologic responses during Nordic Walking by 1) assessing the muscle activity in six leg muscles and in m. triceps brachii and m. trapezius 2) assessing the mechanical power as calculated from video recordings, and 3) evaluating time curves of kinetic and potential energy levels during a stride cycle.

METHODS

Seven healthy female subjects (mean age 52 yr, range 40-59 yr, mean body height 164 cm, range 150-178 cm, mean body weight 64 kg, range 55-74 kg) participated. The subjects were experienced Nordic Walkers and six of the seven subjects were Nordic Walking instructors.

Oxygen consumption and electromyography (EMG) was recorded during walking and Nordic Walking on treadmill at two walking speeds (1.25 m s^{-1} and 1.67 m s^{-1}). EMG was recorded from six leg muscles (mm. soleus, gastrocnemius, tibialis anterior, rectus femoris, vastus lateralis, biceps femoris) and two upper body muscles (mm. triceps brachii and trapezius). Trials were randomized on both walking mode and walking speed. EMG was averaged over a 2 minute period and expressed relative to maximal EMG (EMG_{max}) obtained during maximal voluntary contractions.

The subjects were recorded on video as they completed ten walking trials at each walking speed (1.25 m s^{-1} and 1.63 m s^{-1}) during walking and Nordic Walking on a level surface. Reflective markers were placed over bony landmarks defining a 14 or 16 segment model for walking and Nordic Walking, respectively.

The instantaneous mechanical energy (E_{mech}) was calculated as a summation of the potential energy, the translational energy, and the rotational energy of all body segments at each video frame. This yields an energy curve of E_{mech} as a function of time. The mechanical work for a stride cycle was calculated as the sum of absolute changes in E_{mech} and the mechanical power was calculated as the mechanical work per second.

Stride length was calculated from video recordings during treadmill walking as the number of strides during a 220 second period divided by the walking distance.

RESULTS AND DISCUSSION

Mean EMG was significantly increased during Nordic Walking at both walking speeds in m. triceps brachii (table 1). However, no differences were seen in mm. soleus, gastrocnemius, tibialis anterior, rectus femoris, vastus lateralis, biceps femoris, and trapezius.

Mechanical power was significantly increased during Nordic Walking at both walking speeds being 19 % and 22 % higher at 1.25 m s^{-1} and 1.63 m s^{-1}, respectively ($n=7$, $p < 0.05$). The increase was largely explained by greater changes in potential energy levels during the stride cycle. This indicates an increase in stride length which was also found at 1.25 m s^{-1} (4.2 %, $p < 0.05$) and a tendency found at 1.67 m s^{-1} ($n=6$, $p=0.059$).

Oxygen consumption was increased by 10.0 % and 8.2 % at 1.25 m s^{-1} and 1.67 m s^{-1}, respectively.

Though an increase in mean EMG is seen in m. triceps brachii, the mean activation is only 9 $\%EMG_{max}$ during Nordic Walking. This and the small size of m. triceps brachii suggest that the increase in EMG alone is unlikely to explain the increase seen in oxygen consumption. Furthermore, the EMG recordings from the leg muscles showed no change in activation level, and therefore the energy consumption of the leg muscles is unlikely to be significantly greater during Nordic Walking.

The increase in oxygen consumption is therefore largely explained by the increased mechanical power. This is caused by a change in movement pattern as seen in the greater changes in potential energy thereby increasing the mechanical work.

SUMMARY/CONCLUSIONS

This study shows that the increase in oxygen consumption can be explained by an increase in mechanical power, while the contribution of m. triceps brachii is minor.

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Table 1: Mean EMG was increased ($n=7$, $*p < 0.05$) in m. triceps brachii during Nordic Walking compared to regular walking at walking speeds 1.25 m s^{-1} and 1.67 m s^{-1} (mean, SE).

	Regular walking		**Nordic Walking**	
	$\%EMG_{max}$	**SE**	**$\%EMG_{max}$**	**SE**
m. triceps brachii				
1.25 m s^{-1}	1.8	0.6	9.0 *	1.1
1.67 m s^{-1}	2.6	0.8	9.4 *	1.5

WAVELET-EMG-ANALYSIS OF THE LEG MUSCLES IN FENCING DURING A FLÈCHE ATTACK

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INTRODUCTION

Fencing is highly demanding on the coordinative skills of athletes. The timing of the muscular activation during a flèche attack is one factor of a successful touch during an attack in a fencing bout. The goal of this study was to analyze the muscular activation sequences of the main leg muscles during flèche attacks and to help to improve training methods in fencing.

METHODS

Kinematics of the whole body [Romkes, 2007] (VICON MX, 240 Hz), ground reaction force during the push-off phase (Kistler, 6000 Hz) and EMG data (Biovision, 6000 Hz, SENIAM-Standard) of the M. Tibialis anterior, (TA), M. Gastrocnemius medialis (GM), M. Vastus medialis (VM), M. Rectus femoris (RM), M. Semitendinosus (HAM) of both legs and of the M. Vastus lateralis (VL) of the front leg was recorded of 7 volunteer male expert fencers. The data of 10 trials/subject were averaged using EMG Wavelet-Transformation (WT-EMG) [von Tscharner, 2000].

The test setup simulates a competition situation. The subject was doing small vertical bouncing jumps on the 2 force plates. The start for flèche attack was given by a visual signal at the target. The distance between the force plate and the target was 2.5 m. The beginning of the forward movement of the center of mass was set as movement start (t = 0 sec).

Figure 1: Athlete performing a flèche attack in the Laboratory for Gait Analysis Basel (Switzerland)

RESULTS AND DISCUSSION

The movement was initialized by lowering the center of mass and shifting it slightly in direction of the target. The actual push-off movement started with an activation of the RF, VM and GM of the rear leg. After taking off the rear foot a short co-contraction of the RF and HAM occur to stabilize the position of the hip and knee joint.

The push-off movement of the front leg started before the muscular activation of the rear leg was finished. It started with the RF, VL, VM, GM and slight co-contraction of the RF and HAM after the foot left the ground. Prior landing all measured muscles of the rear leg were activated, a co-contraction was seen for the ankle joint by the GM and TA and for the hip and knee-joint by the RF, VM and HAM.

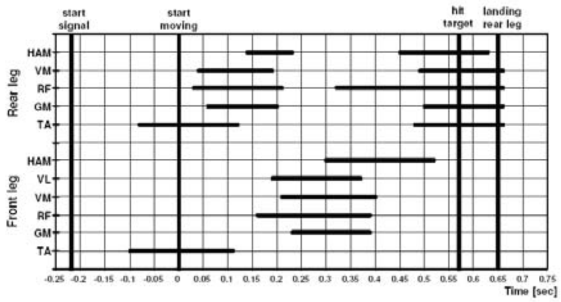

Figure 2: Activation times of measured muscle during the flèche attack (Average of 7 subjects with 10 trials)

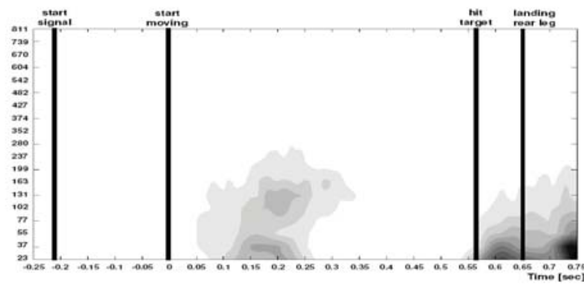

Figure 3: WT-Intensity-Pattern of GM rear leg during the flèche attack.
(Average of 7 subjects with 10 trials, Colour: white: low, black: high intensity)

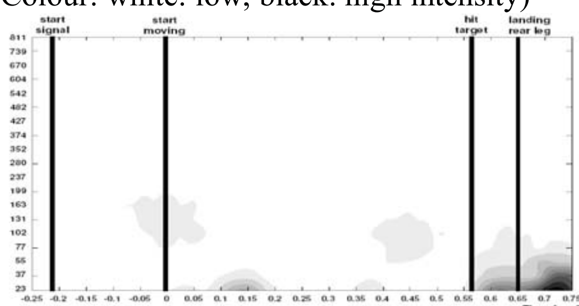

Figure 4: WT-Intensity-Pattern of TA rear leg during the flèche attack.
(Average of 7 subjects with 10 trials; Colour: white: low, black: high intensity)

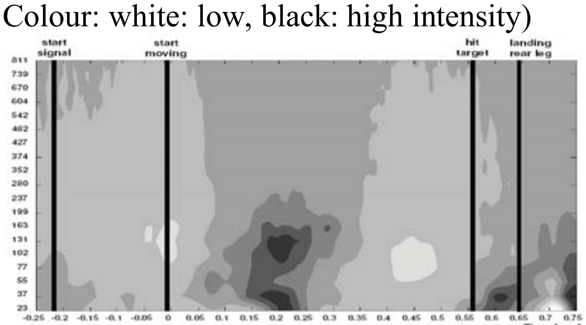

Figure 5: WT-Intensity-Pattern, Difference GM minus TA of the rear leg during the flèche attack. (Average of 7 subjects with 10 trials; Colour: black: GM has higher intensity than TA, White: TA has higher intensity than GM)

The co-contraction of the RF and HAM after take-off of the rear leg is needed to keep the leg in the desired position. While the co-contraction of the RF, VM and HAM and the TA and GM prior to landing is a muscular pre-activation which is needed to control angular position of the joint and to absorb the impact force of the landing of the rear foot after hitting the target. The main landing impact was first absorbed by TA and slightly later by GM and VM, with a clear higher muscular activation.

SUMMARY/CONCLUSIONS

The result of this study shows that the flèche attack is a highly demanding movement for the muscular coordination.

The pre- activation which is known form running [von Tscharner, 2003], can be seen also in the landing after flèche attack. In parallel the co-contraction is part of strategy of positing and stiffening up the joint.

The pre- activation and co-contraction are essential for optimizing the joint loading and preventing injuries.

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ACKNOWLEDGEMENTS

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SURFACE EMG AND HEART RATE RESPONSES IN PATIENTS WITH FIBROMYALGIA SYNDROME

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INTRODUCTION

The development and maintenance of the fibromyalgia syndrome (FMS) is hypothesized to be related to dysregulation of the autonomic nervous system and specifically, an overactive sympathetic system (Martínez-Lavín, 2001). The aim of the present study was to investigate whether surface electromyographic (sEMG) and cardiovascular responses differ between FMS patients and healthy controls during situations with relaxation, nominal rest, and presumed elevated sympathetic activity.

METHODS

Twenty-nine female FMS patients (mean age 52.1 yrs, range 38-66) and 29 age-matched healthy females (mean age 52.7, range 37-64) participated in the study. Body mass ranged from 52-117 kg (mean 74.3 kg) for FMS patients and from 58 to 97 (mean 70.5 kg) for the healthy controls. FMS patients were included if they fulfilled the 1990 American College of Rheumatology Criteria for FMS (Wolfe et al. 1990). The study protocol was approved by the Regional Ethics Committee and carried out according to the declaration of Helsinki.

Four experimental conditions were included: (1) relaxation, i.e. subjects were comfortably seated in an arm chair and watched a cartoon film for 30 min; (2) mental stress test consisting of four 6-min periods alternating between the Stroop test and an arithmetic test; (3) provocation of sympathetic activity during standing by increasing intra-thoracic pressure by a maximal inspiration and thereafter holding the breath for ~15 s with glottis closed; and (4) quiet standing for ~20 s (i.e. nominal rest).

Electrocardiographic (ECG) and sEMG activity from three parts of the upper trapezius (clavicular, descending, and towards scapula), biceps brachii, and middle deltoid were recorded (Myomonitor III, Delsys, US) during the experimental conditions. sEMG and ECG were sampled at 1000 Hz. sEMG were band-pass filtered at 20-450 Hz and root-mean-square (RMS) values were calculated using a 100 ms non-overlapping time window. A bipolar configuration with center-to-center distance of 10 mm was used for all sEMG recordings. Electrodes were placed in standard positions across the chest for the ECG recordings. The QRS complex was detected, and the intervals between the R peaks (R-R intervals) were derived on a beat-by-beat basis. sEMG activity was normalized by the highest sEMG activity obtained during maximal voluntary contractions (EMG_{max}). Trapezius EMG_{max} responses tended to be higher for healthy controls than for FMS patients (p=0.06 for clavicular and descending trapezius).

RESULTS AND DISCUSSION

Trapezius sEMG activity was significantly elevated in FMS patients during TV-relaxation, mental stress test, and increased intra-thoracic pressure (Figure 1). The comparisons remained significant when using non-normalized sEMG activity (i.e. μV) except for the descending trapezius during mental stress (p=0.08) and the clavicular trapezius during increased intra-thoracic pressure (p=0.15). sEMG did not differ between FMS patients and healthy controls during quiet standing.

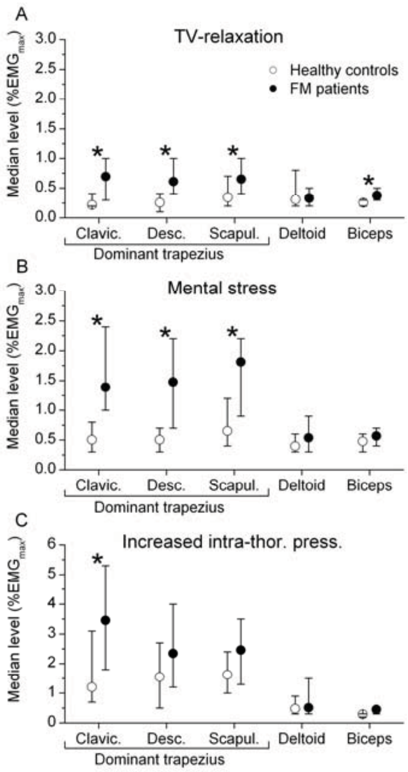

Figure 1: Median sEMG level in trapezius, deltoid, and biceps during TV-relaxation (A), mental stress (B), and increased intra-thoracic pressure (C). Error bars indicate 95% CI. Asterisks indicate significant differences (p<0.05).

Heart rate differed between FMS patients and healthy controls during all conditions except mental stress (Table 1). The lack of difference during the mental stress condition may be caused by a blunted cardiovascular response by the FMS patients, e.g. heart rate differed by no more than ~3 bpm between TV-relaxation and mental stress for FMS patients while this difference was ~8 bpm for healthy controls. Alternatively, the FMS patients did not fully reduce the heart rate during the nominal relaxing conditions.

SUMMARY

FMS patients showed elevated trapezius sEMG activity during a condition with possibility for muscle relaxation and during conditions with elevated sympathetic activation. Moreover, a blunted cardiovascular response during mental stress was indicated for the FMS patients, which is in line with previous studies (Nilsen et al. 2007). The results of the present study indicate an altered reactivity of the autonomic nervous system in FMS patients.

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Table 1: Heart rate (bpm) during the different conditions (mean values with 95% CI).

	TV-relaxation	Mental stress	Increased intra-thoracic pressure	Quiet standing
FMS patients	73 (69-76)	76 (73-79)	77 (75-80)	77 (74-80)
Healthy controls	66 (63-69	74 (72-77)	71 (66-75)	73 (70-76)
p	0.003	NS	0.02	0.04

COMPARISON OF HEALTH-RELATED FITNESS AMONG ADOLESCENTS OF CONTRASTING MATURITY STATUS

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INTRODUCTION

Children of the same chronological age vary considerably in biological maturity status, and individual differences in maturity status influence measures of growth and performance during children and especially during adolescence.

There are contradictory results in previous studies and information on health-related fitness of adolescent females is limited (e.g., Little & Steinke, 1997; Jones, 1949; Malina et al., 1997; Van Lenthe et al., 1996). Therefore, the purpose of the present study was comparison of muscular strength and endurance, cardiovascular endurance, flexibility and body composition among early-, average-, and late-maturing adolescent females.

METHODS

Forty five students in age range 13-15 years were randomly selected from population of adolescent girls in Darab (a town in Iran) in contrasting maturity status (early-, average-, and late maturing) based on age at menarche.

Sport tests such as 1 mile jogging test for evaluation of cardiovascular endurance, sit up test for evaluation of trunk endurance, sit and reach test for evaluation of back and hamstring flexibility, leg press for evaluation of the lower limbs strength, and skin fold thickness, using caliper for evaluation of the body fat percent has been done and gathered data have been compared.

Data was analyzed by one-way analysis of variance and Dunnet and LSD post hoc tests. Significance level was $p<.05$.

RESULTS AND DISCUSSION

The results indicated the adolescents who classified as late-maturing had significantly more relative peak oxygen uptake and muscular endurance compared with those classified as early- and average-maturing ($p<.05$). This finding was not consistent with Malina et al. (1997). Malina et al. didn't observe significant difrences in VO2max among girls of contrasting maturity status.

Furthermore, the results indicated flexibility, muscular strength and body fat percent significantly were higher in early-maturing than late- and average-maturing groups ($p<.05$). These findings were consistent with Little & Steinke (1997) and Van Lenthe et al. (1996), but not with Jones (1949). In Jones's study, early-maturing girls tended to be slightly stronger only early in adolescence. In contrast, in the study of Little & Steinke, girls of contrasting maturity status did not consistently differ in static and explosive strength.

SUMMARY/CONCLUSIONS

Differences among girls of contrasting maturity status are substantially for the reason hormonal changes during puberty and differences in body size and composition. It is recommended that the coaches of endurance women sports for age range 13-15 years (e.g., endurance running or cycling) select girls of late-maturing and in women sports that strength and flexibility are their main need (e.g., disc throwing or gymnastic) select girls of early-maturing.

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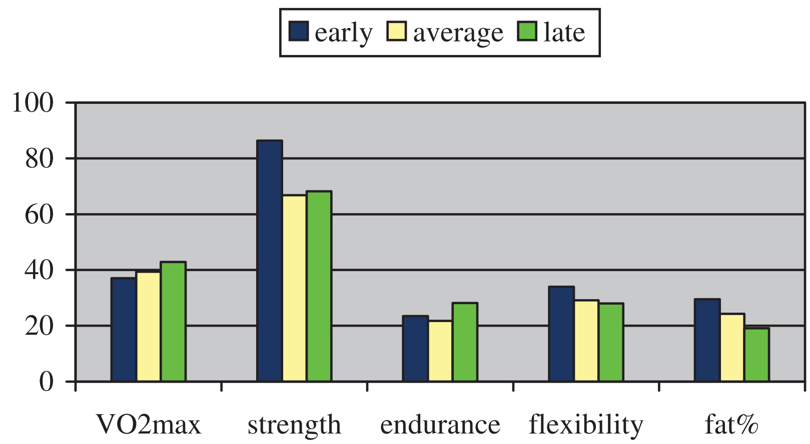

Figure 1: Mean of health-related fitness factors among adolescent females of early-, average-, and late-maturing.

CHARACTERISTICS OF MUSCLE ACTIVITIES IN YOUNG AND ELDERLY TAI CHI PRACTITIONERS DURING TAI CHI GAIT

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INTRODUCTION

Tai Chi (TC) has become a popular exercise for improving balance and postural control, and reducing fall risks in elders. However, we do not yet have a solid understanding of why TC is an effective form of exercise. In particular, we do not know the motor control strategies used during TC practice, and do not know if the motor control strategy changes in people with advanced age. This study was aimed to examine leg muscle activity patterns occurring while TC is practiced by young and elderly individuals. It was hypothesized that elderly people would have shorter duration of leg muscle activations than young people.

METHODS

Six young (aged 21 and 35 years) and six elderly (aged 61 and 84 years) subjects who had been practicing TC for at least three months participated in the study. Subjects were asked to repeat a TC gait (TCG) five times. The kinematics of lower extremity joints and ground reaction forces during TCG were measured using a marker-based motion analysis system (BTS, Italy) and two biomechanical force plates (AMTI, USA) embedded in a walkway. Anthropometric measurements of the subjects were taken and used to estimate joint center locations. Surface electromyography (EMG, BTS, Italy) was measured from six muscles in the left leg (tibialis anterior (TA), soleus (SO), peronaeus longus (PL), rectus femoris (RF), biceps femoris (ST), and tensor fasciae latae (TFL)). The integrated EMG signals, the ground reaction forces and body movements were collected simultaneously at 50 Hz.

The angles of ankle, knee and hip joints in the sagittal and frontal planes were first computed. They were combined with the corresponding muscle EMG to determine the type of muscle action (i.e., isometric, concentric, and eccentric), using the method described elsewhere[1]. The duration of each muscle being active, and the duration of each type of muscle action was then computed for each of the three sub-stance phases: two double stance phases (DS1 and DS2) and one single stance phase (SS), within one gait cycle. Each duration was then normalized by the corresponding sub-stance phase time. A student t-test with a confidence interval of 95% was performed to compare between group differences in the normalized muscle action durations.

RESULTS AND DISCUSSION

The EMG trajectories of six muscles during one gait cycle of TCG are shown in Fig 1. It was found that during SS all muscles except for PL were active for a shorter duration in the elderly group than in the young group (Fig 2a). In particular, the group difference for SO was statistically significant. Also these same muscles had shorter duration of isometric action in the elderly group than in the young group, with significant group differences in TA, RF and ST (Fig 2b).

The shortened duration of muscle action in the elderly group may be attributed by the reduced rigor of their TC practice, such as the amount of knee bend, or the speed of TCG. Indeed, elders had significantly shorter SS time and less knee flexion during SS (0.9±0.3s and 21±12^o, respectively) than the young (1.9±0.3s and 36±9^o, respectively, P<0.00). The shortened duration of isometric muscle action in elderly subjects suggests that elders may either have difficulty maintaining stationary joint positions, or prefer not to maintain stationary joint positions, which is another indicator of the rigor of TC movements.

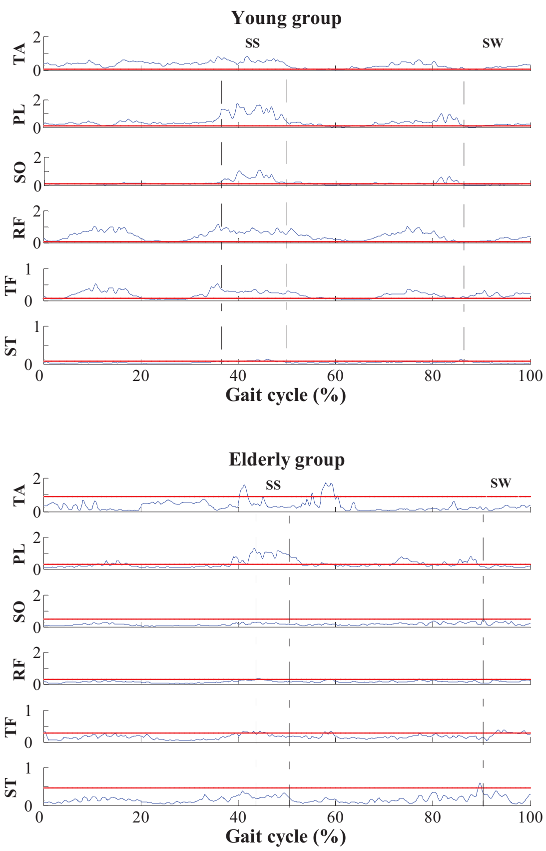

Figure 1: EMG trajectories of six muscles during the stance phase of TCG from one subject. The straight horizontal lines are baseline noise thresholds.

SUMMARY/CONCLUSIONS

The duration of isometric action of leg muscles in elderly TC practitioners was significantly shorter than that of the young during the single stance phase of a TCG. The group differences demonstrate that we cannot simply generalize the effect of TC exercise on young people to elderly people. There is perhaps a direct relationship between the rigor of TC exercise and its outcome effect. This study provides a foundation for future research of the biomechanics of TC to help gain an understanding of why Tai Chi is an effective exercise for improving balance and preventing falls.

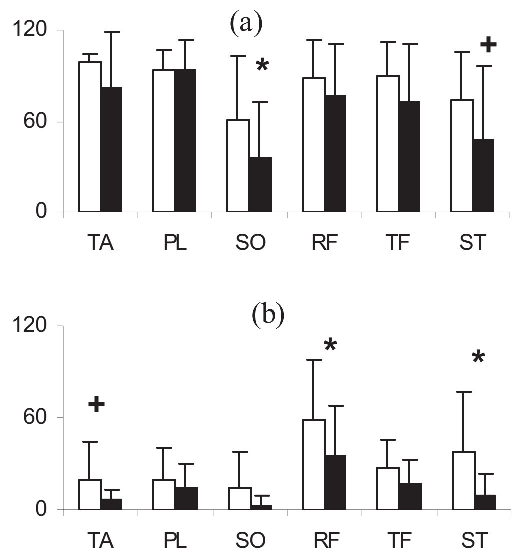

Figure 2: Mean and SD of normalized duration of muscle activation (a) and isometric action (b) in young (open bar) and elderly (filled bar) groups during SS (* and + indicate significant group difference at p<0.05 and p<0.1, respectively).

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IS NEURO-MUSCULAR FUNCTION INFLUENCED BY CREATINE SUPPLEMENTATION?

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INTRODUCTION

Despite oral creatine supplementation has been proven to be effective in enhancing exercise performance, its effect on neuromuscular function is still uncertain.

The present study aimed at verifying whether short-term creatine (Cr) supplementation would improve muscle contractile properties (as assessed by evoked and voluntary contractions), the force-velocity relationship and muscle fatigue during repeated bouts of exercise.

METHODS

16 moderately active men (25±3 years) were assigned to a creatine (CRE) or placebo (PLA) group using a double-blind random design. Subjects assumed either 5g Cr + 15g maltodextrin (CRE) or 20g maltodextrin (PLA) 4 times a day for 5 days.
Before and after supplementation, isometric maximal voluntary contraction (MVC), maximal twitch, force-velocity relationship and repeated dynamic fatiguing contractions were assessed in the elbow flexors. Mechanical parameters were assessed by a KIN-COM dynamometer. Surface electromyographic (EMG) signals were recorded in a single differential mode by means of a 4 electrodes linear array placed on the biceps brachii (BB) muscle.
Mean fibres conduction velocity (CV) was estimated off-line from double differentials EMG signals and used as a parameter of interest.

RESULTS AND DISCUSSION

Peak torque of maximal twitch was 33.46% higher and time to reach the peak torque was 61.29% lower in CRE than PLA ($P<0.05$). Torque-angular velocity curve was improved after Cr supplementation. Mean fibres CV was on average 8.9% higher ($P<0.05$) in CRE at all angular velocities after supplementation (Figure 1). Creatine supplementation did not affect EMG and mechanical parameters during the repeated exercise fatiguing protocol.

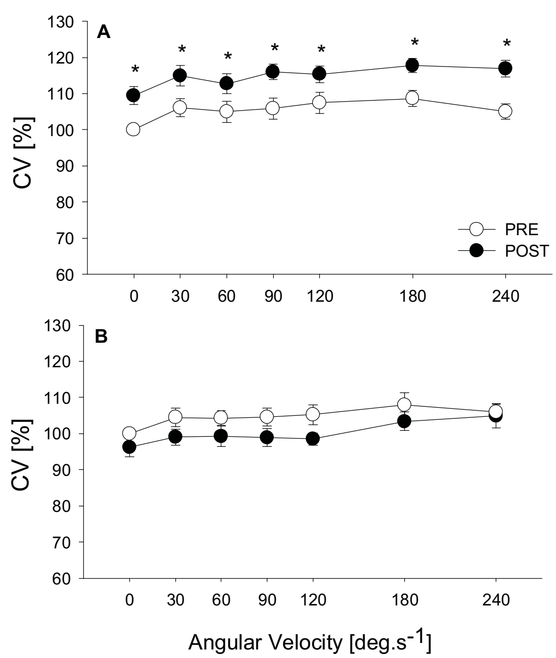

Figure 1: CV values of CRE (panel A) and PLA (panel B) groups during PRE and POST supplementation sessions. Data are expressed in % of CV value obtained during the MVC in the PRE session. MEAN±SE *p<0.05

SUMMARY/CONCLUSIONS

Oral creatine supplementation enhances intrinsic and voluntary contractile capacity of skeletal muscle of young men. This could be related to an increased Ca^{2+} sensitivity and maximum Ca^{2+}-activated force, which are associated to an increase in cellular water content needed to maintain osmolality.

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THE SPATIAL AND TEMPORAL RELATIONS BETWEEN EMG ACTIVITY AND JOINT DISPLACEMENT DURING THE LAT PULL-DOWN EXERCISE

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INTRODUCTION

A number of studies have demonstrated that strength gains are typically specific to the conditions that were used during training (Wilson et al, 2006). To maximize the functional benefits of strength training, therefore, it is critical that the exercises be similar to the tasks in which improvements are intended.

Beginning Movement Load (BML) training was developed by Koyama (1994) to enhance the functional significance of strengthening exercises. BML exercises engage muscles with activation patterns that are more similar to less constrained movements than conventional strength-training activities. One characteristic of BML exercise is that they increase the range of motion of an action by including a rotation about the longitudinal axis of the limb; this is referred to as a dodge movement (Koyama, 1999). The purpose of the study was to assess the spatial and temporal relations between EMG activity and limb kinematics during the lat-pull down exercise with a dodge movement.

METHODS

Six healthy subjects volunteered to participate in the study. The subjects performed the lat pull-down exercise on three machines (A, B and C). The three machines differed in the range of motion (ROM): A – the conventional device used for the exercise; B – one additional degree of freedom (pronation-supination of the forearm); C – two additional degrees of freedom (horizontal flexion-extension of the shoulder). The subjects performed 15 repetitions with a load of 30% of 1RM. The kinematics of the exercise were recorded with two digital video cameras (SONY) that operated at 60 Hz. Five reflection markers were placed on the right side of the subject: wrist joint, elbow joint, shoulder joint, the sternoclavicular joint, and the iliac crest.

The EMG activity of 5 muscles (biceps brachii (BB), triceps brachii (TB), latissimus dorsi (LD), posterior deltoid (DP), and serratus anterior (SA) on the right side of the subject were recorded with bipolar active electrodes (NIHON-KOHDEN, SS-2096) and transmitted (ZB-581G) to a receiver (ZR-550H). The signals were amplified and band-pass filtered (20-500 Hz) (NIHON-KOHDEN, WEB-5500) before being digitized a 2k samples/s.

RESULTS AND DISCUSSION

Performance of the lat-pull down exercise on machine C exhibited the following characteristics: **1)** the vertical displacement of wrist was greatest (A = 44.84±1.33 cm; B = 46.85±1.49 cm, C = 62.0±1.40 cm) ($P<0.05$). **2)** The abduction-adduction ROM of shoulder joint angle was greater (A = 1.28 rad; B = 1.4 rad C = 1.8 rad. **3)** flexion-extension ROM at the elbow joint was smallest (A = 1.45 rad; B = 1.43 rad; C = 0.8 rad. **4)** Significant differences in EMG amplitude included less BB (flexor) and greater TB (extensor), SA (external rotator of scapula), DP (external rotator) and LD (adductor) (Fig 1).

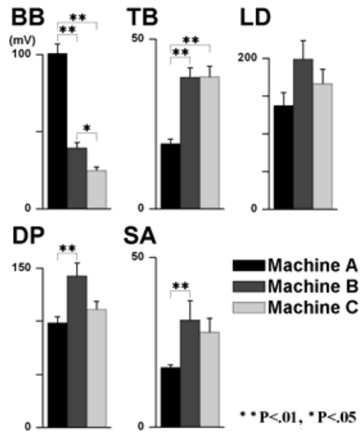

Figure 1: iEMG was calculated for 0.5 s after movement reversal as the load was raised with a shortening contraction on the three different machines.

There were significant differences in the order in which the muscles were activated during the pull-down phase of the exercise on the three machines (Fig 2-left). The performance on machine C involved a progressive proximal-to-distal sequence of activation (SA to TB) that was not required on machine A (DP < the other muscles) and appeared only partially on machine B (SA < the other muscles). These differences in the activation sequence of some of the major muscles involved in the exercise influenced the timing of the peak vertical velocity for the shoulder, elbow, and wrist. Prior to the reversal of the wrist movement, peak vertical velocity of the shoulder and elbow preceded that of the wrist on all three machines (Fig 2-right). On machine C, however, the timing of the peak vertical velocity progressed from the shoulder to the elbow and then the wrist. Such progressions are typically observed during throwing and kicking movements and maximize the speed generated at the end of the kinematic chain (Marshall et al 2000).

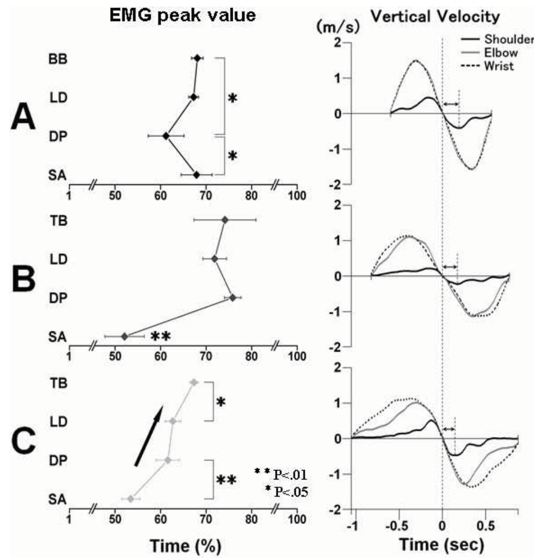

Figure 2: The timing of EMG peak value and vertical velocity of the shoulder, elbow and wrist during the lat-pull down exercise on the three machines. Time 0 is the wrist movement reversal point at right side of panel.

CONCLUSIONS

Beginning Movement Load exercises can improve the association between training actions and functional activities.

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CHANGES IN ISOKINETIC TORQUE RATIO AFTER AEROBIC EXERCISE IS DEPENDENT ON THE VELOCITY OF TESTING AND METHOD OF CALCULATION

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INTRODUCTION

The use of isokinetic test to access muscular balance between knee flexors and extensors (H:Q ratio) have been increasing, and different ways of determining this ratios can be used (Coombs and Garbutt, 2002; Hewett; Myer; Zazulak, 2007). Initially, the concentric H:Q ratio was used, and recently the literature has stated a more functional method using the eccentric hamstring:concentric quadriceps peak torque (Coombs and Garbutt, 2002). The functionality is related to the eccentric hamstring activation during knee extensions in gait and running. H:Q ratio have been widely investigated mainly to prevent injuries, which can be related to sport practice and fatigue (Coombs and Garbutt, 2002). However, there is a lack of information about the effects of endurance exercises in this index. In this way, since running exercises have been related to decreases in torque generation (Denadai et al. 2007), the aim of the present study was to determine the effects of a previous aerobic exercise in the H:Q ratio calculated using concentric and functional methods.

METHODS

Sixteen healthy males (22±1.2 years; 178.3±3.3 cm; 78.2±5.3 kg; VO_2max 40±2 $ml \cdot kg^{-1} \cdot min^{-1}$) performed the following tests on different days: 1) an incremental test in a treadmill (INBRAMED Super ATL, Porto Alegre, Brazil), in order to determine VO_2max (Cosmed Quark PFT ergo, Rome, Italy), and the velocity at onset of blood lactate accumulation (vOBLA) (YSI 2300, Ohio, USA); 2) 5 maximal isokinetic knee flexions and extensions (System 3 PRO, BIODEX Medical Systems, NY, USA), to determine the peak torque, at velocities of -60°.s^{-1} ($E_{ext}60$, $E_{flex}60$) and -180°.s^{-1} ($E_{ext}180$, $E_{flex}180$), and 60°.s^{-1} ($C_{ext}60$, $C_{flex}60$) and ($C_{ext}180$, $C_{flex}180$), in random order; 3) the same isokinetic protocol described previously, performed 15 minutes after a continuous running at 95% vOBLA (8.2±0.2 $km.h^{-1}$, 78.5±3% VO_2max), in which the duration was estimated to a caloric expenditure around of 500Kcal for all subjects. All tests were separated by at least seven days. H:Q ratios pre and post-exercise were calculated by using concentric peak torque ([$C_{flex}60/EC_{ext}60$ = C-60], [$C_{flex}180/C_{ext}180$ = C-180]), and using a functional calculation ([$E_{flex}60/C_{ext}60$ = F-60], [$E_{flex}180/C_{ext}180$ = F-180]). H:Q ratios were compared between pre and post-exercise conditions with a two tailed Students Paired *t* test.

RESULTS AND DISCUSSION

The mean Concentric H:Q ratios (at 60°.s^{-1} and 180°.s^{-1}) were similar to the suggested for muscular balance, which should be above 0.6 (Coombs and Garbutt, 2002) (Figure 1), as well as for F-180°.s^{-1} (1.0 or above). However, Functional H:Q ratios at 60°.s^{-1} did not respond equally to the expected muscular balance, presenting an average ratio below 1.0.

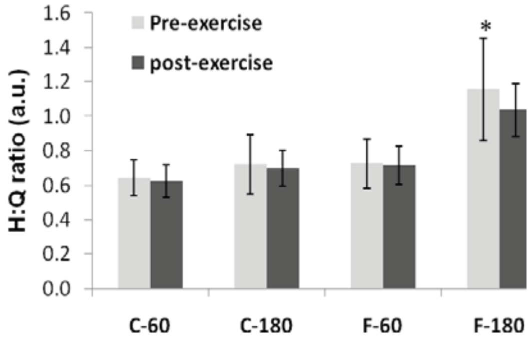

Figure 1. Concentric H:Q ratio at 60°.s^{-1} (C-60) and 180°.s^{-1}(C-180), and Functional H:Q ratio at 60°.s^{-1} (F-60) and 180°.s^{-1}(F-180), calculated before (pre-exercise) and after (post-exercise) running exercise.
* significantly different from F-180 post exercise.

It was found a significant decrease ($p<0.05$) in H:Q ratio after the aerobic exercise only to F-180, however, the post-exercise condition did not reach muscular imbalance (Figure 1). As previously described by others investigations (Leppers et al. 2000; Denadai et al. 2007), the effects of a running exercise in the peak torque seem be predominant during fast eccentric contractions, justifying the absence of effects after the exercise in the Concentric H:Q ratios. Previous investigations (Rahnama et al. 2003), studied the effect of running exercise in the H:Q ratios (concentric and functional), and only the functional H:Q ratio was decreased after a fatiguing exercise. The decreased eccentric torque (at -180°.s^{-1}) after eccentric contractions can be explained by the muscle damage involved during the running exercise, which can decrease the contractile apparatus, particularly in fast twitch fibers (Leppers et al. 2000). Although C-180 was not affected by previous aerobic exercise, the relative H:Q ratio loss for C-180 (-0.84±13.11%) and F-180 (-7.8±13.7%) were good correlated ($r = 0.75$, $p < 0.001$) indicating that the velocity of isokinetic contractions may be an important factor during analysis of muscular balance (Gerodimos et al. 2003). This way, the present data could suggest that fast velocities were more sensitive than slow velocities during the determination of the H:Q ratios. Furthermore, based on the present data and in previous investigations, we could suggest the utilization of fast isokinetic contractions to verify the effects of previous running exercises on the H:Q ratios.

SUMMARY/CONCLUSIONS

H:Q ratio calculated using functional relationship (Eccentric Hamstrings: Concentric Quadriceps contractions) at 180°.s^{-1} can be more reliable to describe the knee muscular balance when the effects of continuous running exercise are being analyzed.

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EVALUATION OF MUSCULAR FATIGUE DURING REPETITVE SKIING TURNS BY REFERRING TO S UAT E ERCISE

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INTRODUCTION

In the field exercise like skiing, it is hard to evaluate the time-series of muscle activity indices estimated by conventional approaches using short-term block sliding in time. We have estimated muscle activity at specific intervals during skiing exercise, using surface EMG (SEMG) signals and the knee joint angles (Ushiyama et al., 2006). Selecting a significant epoch at each skiing turn revealed the explicit decrease in the mean power frequency (MPF) at first several turns and then the small fluctuation in MPF with respect to the number of turns. During repetitive fatiguing dynamic knee contractions, several muscles contribute to prevent the development of muscular fatigue (Pincivero et al., 2006). The purpose of this study is to evaluate the presence of muscle fatigue during repetitive skiing turns by referring to squat exercise.

METHODS

We acquireded SEMG signals and the knee joint angles during skiing exercise, using a wearable and ubiquitous type measurement system. The measurement system, which weight was around 600 g, was composed of a PDA, a PCMCIA type data acquisition card, and a unit (Myomonitor, Delssy) for sensors. The LabVIEW PDA Module (National Instruments) and the CedarFTP for Pocket PC allowed the sampling frequency at 1000 Hz per channel and the FTP client in the PDA. Besides, we developed the GUI by MATLAB (R14SP3, MathWorks) for verifying the measured data at each turn on site.

We measured SEMG at the right vastus lateralis (VL) muscle with the knee joint angles at both legs, using the active two-bar electrodes (DE2.3, Delsys) and the goniometors (ShapeSensor S700, Measurand). To estimate muscular fatigue properly, we estimated the MPF and the averaged rectified value (ARV) of SEMG signals for each epoch determined by the behavior of knee joint angles at each turn. Besides, we calculated the turn interval and the difference in the knee joint angles (DKJA) between the outside and inside legs during 4000 m downhill skiing. The ground was the maximum incline of 20 deg at the first half and the mean incline of 7 deg at the latter half. During the squat exercise with a tube, on the other hand, we asked subjects to try to control squat exercise every 4 sec up to 100 repetitive contractions. Then, we evaluated muscular fatigue at the first and latter halves within a trial and at the period of first 10 turns among trials during skiing exercise, referring to the results during squat exercise. We used the paired t-test at the significant difference level of $p < 0.05$.

RESULTS AND DISCUSSION

Seven healthy male subjects (26.4 $\pm$ 8.1 yrs) participated in this study. We grouped each subject into three groups depending on the accumulated days for skiing: four experts over 100 days; two intermediate experts over 30 day and under 100 days; a beginner under 30 days. We analyzed 1868 skiing turns in 51 trials.

A beginner, who snowplowed throughout ski session, increased the DKJA and the turn interval, but did not show the difference in MPF and ARV. Changes due to muscular fatigue might significantly appear at the tibialis anterior muscle (Ushiyama et al., 2006). Intermediate experts used the parallel turn, and decreased the ARV and the DKJA both within a trial and among trials. Since the decrease in DKJA means an upright posture, this might be a sign of muscle fatigue. Three of experts increased the ARV and decreased the MPF within a trial, presenting a clear behavior of muscular fatigue (Figure 1). However, the DKJA and the turn interval remained almost constant values within a trial. Two of them further presented the similar behavior in ARV and MPF as a function of the trial number, while others did not show the changes. Good skiing performance requires almost constant DKJA and it induced muscular fatigue at the same time.

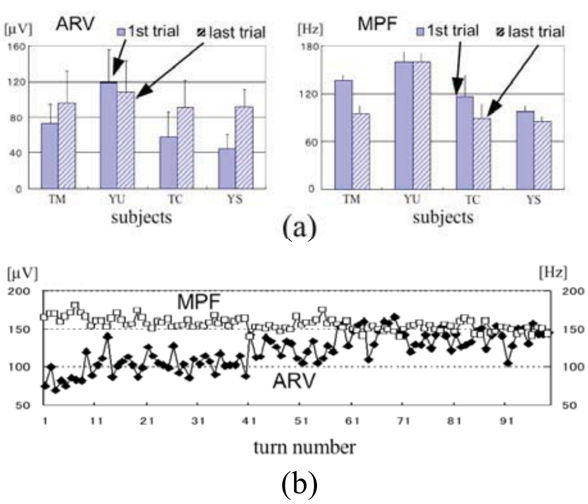

Figure 1: (a) mean ARV and MPF at the first and latter halves within a trial; (b) time-series of ARV and MPF with respect to the turn number.

Two of the three experts showed muscle fatigue both in squat exercise and skiing exercise. One of them showed muscle fatigue in skiing exercise, but did not in squat exercise. Besides, intermediate experts demonstrated muscle fatigue in squat exercise, but did not in skiing exercise. Note that squat exercise and skiing exercise are included in the closed-kinetic chain exercise. These findings could be related to the balance between skill and motivation how to create good performance, by referring to squat exercise in the supplemental experiments. We confirmed the similar presence of muscular fatigue by measuring SEMG signals at the vastus lateral, biceps femoris, tibialis anterior, gastrocnemius muscles. The results showed that several variations of ARV and MPF at different muscles occurred as a function of elapsed time to prevent the development of muscle fatigue by changing agonist muscles or knee joint angles.

SUMMARY/CONCLUSIONS

To prevent the development of muscle fatigue during skiing, alternation of agonist muscles sometimes occurs. Accordingly, several muscles with both knee joint angles should be monitored to evaluate muscular fatigue properly during alpine skiing exercise.

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ADDUCTOR MAGNUS, RECTUS ABDOMINIS, SERRATUS ANTERIOR AND TRICEPS ACTIVITY DURING REVERSE PUNCH (GYAKU ZUKI) EXECUTION

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INTRODUCTION

Reverse punch technique may involved major muscles at proper time to increase momentum and so the resulted impact. Muscles sequence activation from the hip to the arm was summarized by Nakayama (1974). Kinesiology study revealed that punch is not only an elbow extension; shoulder medial rotation also contributes in muscular work. Karate Do initial posture is essential due to the execution of thigh medial rotation and so hip rotation. The purpose of the study is to determine muscular activity between Karate Do practitioners, once muscle preparation of thigh medial and shoulder medial rotations is a long term conditioning training. EMG activity of triceps, serratus anterior, adductor magnus and rectus abdominis was recorded.

METHODS

Sample was constituted by two groups, with three males Karate Do amateur practitioners each. A first group of novice practitioners (NP), body mass of 86.77±21.24 kg and a second group of expert practitioners (BBP), body mass of 74.27±11.98 kg. Sample was selected by practitioner time disposal. After signed informed consent, anthropometry measures and skin preparation were made. EMG activity was recorded using surface bipolar disposable electrodes, model 31118733 (Kendall, USA), placed according to SENIAM Project recommendations. Sample frequency was 2000 Hz. Reference electrode was located in the tibial tuberosity. EMG data was recorded using Miotool and Miograph 2.0 software (Miotec, Biomedical Equipments-Brazil). SAD2 data acquisition software (Engineer Faculty, UFRGS–Brazil) was used for data analyze. Practitioners realized five reverse punches over a bag with a calibrated accelerometer in it (SANT´ANNA, 2007). A switch sensor elbow contacted, regulated vertically by means of physical characteristics of each practitioner, informed the exact start. Last instant was collected by wave impact at the bag. DI 740 series data acquisition device and WINDAQ software (DATAQ Instruments-USA) were used to collect switch sensor and impact data. Foot position was marked in the floor to propitiate a similar stance during execution. Distance between bag and fist was measured for later calculation of mean velocity. SPSS Statistical software was used to verify normality using Shapiro-Wilk Test ($p \leq 0.05$). Groups were compared by Mann-Whitney U test ($p \leq 0.05$) and relations between variables were analyzed by Spearman correlation.

RESULTS AND DISCUSSION

Table 1 resumes EMG data and output variables. Net impact was normalized by body mass and is expressed as gravity acceleration (g) per kilograms. RMS data were normalized by average intensity for each practitioners for further statistic analysis.

Table 1: Elapsed time (msec), Impact (g/kg), mean velocity (m/s) and mean RMS.

	NP	BBP
Time	245±47	217±72
Velocity	3.775±0.677	3.670±0.916
Impact	0.191±0.037	0.233±0.056
Mean RMS (%)		
Triceps	87.90±40.09	64.01±32.54
Serratil A.	116.64±60.89	79.37±34.72
Rectus A.	102.69±46.88	88.08±20.64
Adductor	96.24±43.49	97.12±56.55

Shapiro-Wilk test indicated normality only for a few data. Because of the small sample size (n=30) and high standard deviation, all data was considered non-parametric. Mann-Whitney U-test between groups revealed significance (U, 0.029; $p<0.05$) for normalized impact denoting differences between novice and experts to punch but did not explained EMG activity. Spearman analysis of EMG data, resumed in Table 2 can explained muscles patterns during movement. Correlation between mean velocity and adductor magnus activity suggest evidence that reverse punch as practice in Karate Do begins during pelvic movement induced by thigh medial rotation as an arrangement of posture. With the correlation between serratus anterior, adductor magnus and rectus abdominis activity, we can deduced that movement initiate in the lower limb and has a quick propagation through hip and thorax before arm movement. The experiment did not found statistic evidence of triceps activity and impact but EMG activity of this muscle was higher than the others.

SUMMARY/CONCLUSIONS

Comprehension of Karate Do techniques using EMG is recently. Such studies will explain training effects and physiologic changes due to body adaptation to great impact forces (GRIFFIN, 1990), once net impacts found were of 20 to 30 g.
Instant analysis of RMS peak before and after impact are necessary and will help for further explanations. Increase sample is required to obtain necessary data and explain movement patterns.

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Table 2: Spearman Analysis and correlation coefficients results for main results of the study.

Correlation Coefficient	Mean Velocity (m/s)	Triceps AI	Serratil A. BI	Rectus A BI	Rectus A AI
Serratus Anterior BI	-.110	-.455(*)			
Serratus Anterior AI	-.260	.083	.401(*)		
Rectus Abdominis BI	.271	-.440(*)	.576(**)		
Adductor Magnus BI	-.363(*)	.017	.410(*)	.121	.445(*)
Adductor Magnus AI	.513(**)	-.026	.358	.399(*)	.045

* Correlation is significant at the 0.05 level (2-tailed).
** Correlation is significant at the 0.01 level (2-tailed).
AI and BI= EMG activity after impact and before impact.

DROP LANDING BIOMECHANICS – DOES FALLING HEIGHT IMPLY NEUROMUSCULAR SYSTEM (PRE) ADJUSTMENT?

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INTRODUCTION

In lower limb biomechanical analysis, Landing presents itself as a vital kinetic energy absorption phase. Research as provided us meaningful background information about preparatory landing responses during falling actions, suggesting these are intended to prevent hard impacts upon ground contact, having often been identified by measures of electromyographic (EMG) muscular activity (Liebermann et al, 2007).

Visual and vestibular information have been assumed to participate in a multiplicity of daily events and activities associated with landing actions. However, research also indicates that with an increase in falling height subjects are consequently exposed to higher joint forces on ground impact follow up. Therefore when increasing drop point for landing, we are in fact emphasizing load demands on subject's lower limbs, mainly due to the concurrent necessity for optimal body segment negative impact acceleration management (McNitt-Gray, 1991).

Hence our main purpose in this study was to assess possible neuromuscular activation strategy changes, when preparing landing, for differentiated falling heights.

METHODS

For this purpose we gathered a sample of ten volunteers (5 male and 5 female, aged 21,33±2,15 years).

Experimental testing setup consisted in subjects first performing three 40 cm drop landings onto a Force Plate, and subsequently executing three 60 cm drop landings. Subjects were further instructed to land on both feet simultaneously, and also to place their hands on respective hips throughout testing, in order to prevent upper limb negative data influence. Full recovery was waranteed between attempts.

Subjects EMG activation was assessed using surface electrodes (BlueSensor, Medicotest) with bi-polar placement over the dominant lower limb's vastus lateralis (VL), vastus medialis (VM), internal gastrocnemius (GI), external gastrocnemius (GE) and biceps femoris (BF) muscles, in accordance to SENIAM's guidelines. EMG Data collected was recorded at 2000 Hz, using a pre-amplified telemetry system (Glonner), with *Simi Motion* software. Signal was processed using a bandpass [10-600] Hz filter.

Average EMG full wave rectified signal (AvgEMG) was determined as a function of force plate data. Vertical component force curves were plotted and subdivided into pre-activation time (100 ms until ground contact) and time until peak force (time current from ground contact until maximal Fz). Integral data was discarded due to short time EMG recording periods. Data Normalization procedures were also implemented and revealed no differences in results when compared to initial data collection.

Statistical independent samples t-test was determined in order to compare performance means between groups.

RESULTS AND DISCUSSION

Results revealed, as predicted in other research, differences between biomechanical assessment variables when comparing distinct falling heights. However, besides rather previously documented changes in neuromuscular system activations, often observed in pre-activation, in the present study we also verified interesting changes of motor control management within large thigh muscles. These also represent important strain absorption points, especially with gradual falling height increases. VL and VM muscles presented significant activation amplitude changes, suggesting a vital role, either in preparatory muscle control, as well as in neuromuscular system stabilization mechanics. Force Plate data indicated significantly higher vertical ground reaction forces and lower time length to peak Fz. This last events mainly justified by larger drop times associated with consequent increased center of gravity falling velocity.

SUMMARY/CONCLUSIONS

In summary our results revealed differences in muscle activation amplitude, when comparing different falling heights. increasing amplitude activation in selected muscle groups was observed, revealing that subjects appear to adjust intramuscular activation, for optimized inter muscular fall coordination.

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Table 1: Mean difference between falling heights.

Drop landing	**40cm**		**60 cm**		**dif**	**t**	**p**
	Mean	SD	Mean	SD			
EMG [mv]	Pre-Activation EMG [100 ms to first contact]						
AvgVL	0,076	0,030	0,115	0,044	0,039	-2,227	0,040*
AvgVM	0,083	0,030	0,133	0,039	0,050	-3,193	0,005*
AvgGI	0,281	0,132	0,234	0,051	0,047	0,944	0,360
AvgGE	0,087	0,033	0,091	0,034	0,005	-0,278	0,785
AvgBF	0,071	0,046	0,073	0,034	0,002	-0,106	0,917
EMG [mv]	Landing EMG [Time until peak Fz]						
AvgVL	0,244	0,104	0,411	0,180	0,167	-2,304	0,036*
AvgVM	0,264	0,141	0,463	0,185	0,199	-2,560	0,021*
AvgGI	0,226	0,151	0,149	0,141	0,077	1,147	0,267
AvgGE	0,071	0,045	0,064	0,049	0,007	0,323	0,751
AvgBF	0,063	0,035	0,075	0,052	0,012	-0,578	0,571
Ground Reaction Force [N]	Force-Plate Data						
Fz	3564,15	1132,90	5150,93	1730,94	1586,77	-2,334	0,032*
Fz-t	53,3	5,00	46,00	6,99	7,33	2,601	0,019*
Impulse	80,48	24,37	84,01	23,42	3,54	-0,331	0,745

* Results were statistically significant at $p<0.05$ level.

ELECTROMYOGRAPHY ANALYSIS OF TWO DIFFERENT SQUAT EXERCISES – RELATIONSHIP OF FOOT POSITIONS

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INTRODUCTION

Working out in gyms is one of the most popular types of physical activity in the world. The Squat exercise is very used in lower limbs strength training programs. The Squat exercise involves the hip and knee joints, activating the quadriceps and hamstrings musculature and also involves several neighbor muscles to help the movement (DELAVIER, 2000). The aim of this study was verify, through surface electromyography analysis (EMGs), the quadriceps and hamstrings musculature activation patters on two feet positions squat exercise variations.

METHODS

In this study take part nine men aged between 18 and 20 years old, with working out practice of six months at least, and bodyfat percentage below 20 % and non neuromuscular injury in the last six months. The subjects answered anamnesis, health questionnaires and anthropometrical assessment, using Jackson & Pollock (1978) protocol. The subjects were trained to adapt to specific exercise movement and rhythm. The loads were determined by a sub maximal load test (BOMPA, 2001) to estimate the 80% of maximal repetition. Final position of exercise was limited at 90º of knee flexion (assessment by a goniometer). Was used an electromyographer Miotool 400 (Miotec®), electrodes meditrace™ 200 (Kendall®) and a notebook (Sony®), an iron bar, free weights and digital metronome (Seyko®) to control the rhythm of contractions. Was used a bandwidth of 20-500Hz. Rectus Femoris (RF), Vastus Medialis (VM), Vastus Lateralis (VL) and Biceps Femoris (BF) were analized. Were done two foot positions variations of the squat exercise: medium along (A) and large along (B). Were done 3 series with 10 repetition for each exercise with 3 minutes of rest. The voluntary contraction (MVC) was used to normalize the data and exercises order were randomized. The Root Mean Square (RMS) was calculated. Statistical treatment was done by the SPSS 11.5. Firstly the Kolomogorov-Smirnov test ($p \leq 0.05$) was done. Secondly the Wilcoxon test ($p \leq 0.05$) was done to verify the differences between the two squat exercises.

RESULTS AND DISCUSSION

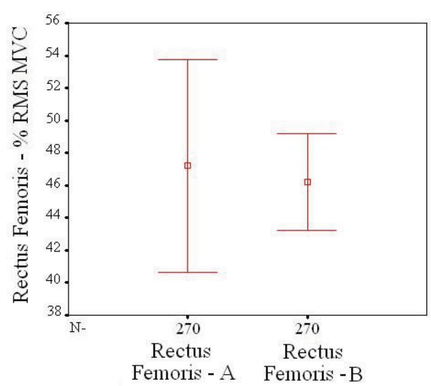

Figure 1: Rectus Femoris (RF) - % RMS MVC.

The Figure 1 shows more activation of (RF) muscle in the exercise (A), however don't show significant differences compared with exercise (B).

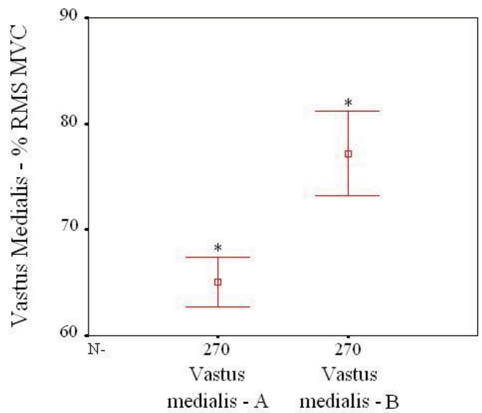

Figure 2: Vastus Medialis (VM) - % RMS MVC.

The Figure 2 shows more activation of (VM) muscle in the exercise (B) and also shows significant differences between both exercises.

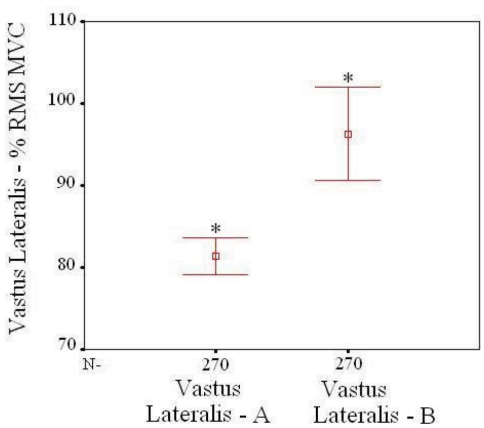

Figure 3: Vastus Lateralis (VL) - % RMS MVC.

The Figure 3 shows more activation of the (VL) muscle in the exercise (B) and also shows significant differences between both exercises.

The Figure 4 shows more activation of (BF) muscle in the exercise (A) also shows significant differences between both exercises.

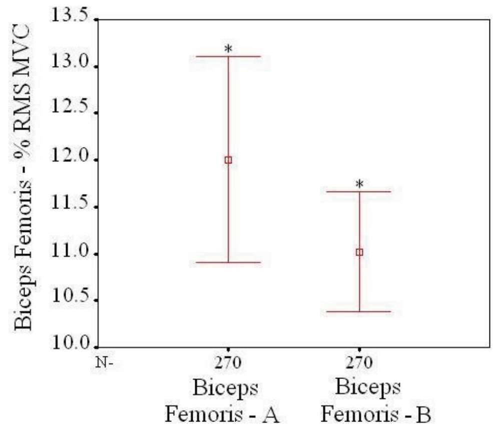

Figure 4: Biceps Femoris - % RMS MVC.

SUMMARY/CONCLUSIONS

The squat exercise had showed an excellent exercise for lower limbs, principaly, for (VM) and (VL) muscles for both variations of exercises. Therefore, it suggested that others studies should be done about this topic, using bigger samples for more relevant results.

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EFFECTS OF DISTANCE OF EXTERNAL FOCUS OF ATTENTION ON THE LEARNING OF BASKETBALL SET SHOT IN ADOLESCENT FEMALES

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INTRODUCTION

Previous studies have shown that motor learning can be enhanced by directing performers' attention to the effects of their movements (external focus), rather than to the body movements producing the effect (internal focus). Furthermore, the advantages of an external focus were enhanced as the distance of the external effect from the body increased (e.g., Maddox, et al., 1999; McNevin et al. 2003). Wulf et al (2000) investigated distance of external focus on learning in golf and suggested that there is an optimal distance.

This effect was not investigated on learning of basketball shot. Therefore, the purpose of this study was to test the hypothesis that increasing the distance between the body and the action effects might further enhance the learning advantages associated with an external focus of attention.

METHODS

For this purpose, 36 right hand and novice females who ranged from 16 to 18 years in age were randomly selected and were divided to three matched groups (far external, near external and internal focus of attention) based on the performance in pretest after instruction.

Then learners were allowed to practice set shot with special feedback provided to the each group during 2 sessions. To the internal focus group (control) was presented feedback about upper limb movements and to the near and far external focus groups were given feedbacks about ball movement during releasing of hand and entering to basket, respectively. Finally, participants performed immediate (in last session) and delayed transfer tests (after one week) without feedback.

Data was analyzed by repeated measures ANOVA and LSD post hoc test.

RESULTS AND DISCUSSION

The results of repeated measure ANOVA indicated the significant differences between groups in immediate and delayed transfer tests. The results of LSD post hoc tests showed that the accuracy of shooting of far external focus group was significantly better than near external focus group in immediate transfer test ($P<.05$). Furthermore, the accuracy of shooting of far external group was significantly better than internal group in delayed transfer test ($P<.05$).

The findings were consistent with Maddox et al. (1999), Mcnevin et al. (2003), and Wulf et al. (2000). They are in line with a constrained action hypothesis that accounts for the relatively poorer learning associated with an attentional focus directed towards effects in close proximity to the body, or towards the body itself.

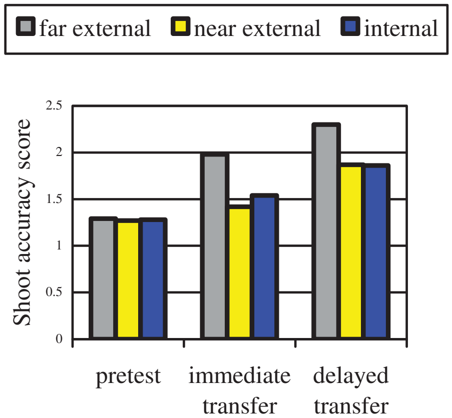

Figure 1: The mean of set shot accuracy in different groups.

SUMMARY/CONCLUSIONS

According to the findings of the present study, it is recommended to basketball coaches that direct his/her learners' attention to the far effects of movements of body for improving and accelerating the learning of basketball set shot with feedback; such as attention to the peak of the shot trajectory or the angle of enter the ball in the hoop.

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Table 1: The mean (±SD) of accuracy of set shot in experimental and control groups

Test / **group**	pretest	Immediate transfer test	Delayed transfer test
Internal focus	1.28 ± .9	1.54 ± .9	1.86 ± .9
Near external focus	1.27 ± .9	1.42 ± .7	1.87 ± .9
Far external focus	1.29 ± 1	1.98 ± .7	2.3 ± .6

MUSCLE FUNCTION INVESTIGATED WITH SURFACE ELECTROMYOGRAPHY IN LATRAL ELBOW TENDINOPATHY

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INTRODUCTION

Lateral elbow tendinopathy or tennis elbow (TE) is the commonest elbow problem in athletes which involves the common wrist extensor origin, particularly the origin of extensor carpi radialis (ECR). At least 40 conservative treatments have been described but the optimal treatment is unknown due to unknown etiology. Muscle imbalance, an unhealthy functional relationship among the periarticular muscles, particularly between agonist and antagonist groups, has become an important topic in the etiology of painful musculoskeletal disorders. We aimed to investigate strength, fatigability, and activity of upper limb musculature to elucidate the role of muscle imbalance in the pathophysiology of TE.

METHODS

Sixteen patients clinically diagnosed with tennis elbow were compared with sixteen control (C) subjects with no history of upper limb musculoskeletal problem. Electromyographic activity (RMS amplitude) and fatigue characteristics (median frequency slope) of five forearm and two shoulder muscles were measured during isometric contraction at 50% maximum voluntary contraction using surface electrodes. Maximum isometric muscle strength was also measured for grip, metacarpophalangeal, wrist, and shoulder on both sides.

RESULTS AND DISCUSSION

All strength measurements showed dominance difference in C (11% for grip, 12% for wrist extension and flexion, and 10–15% for shoulder), but none in TE. In tennis elbow compared to controls, grip (25%), wrist (30%) and shoulder (25–35%) strength were significantly ($p<0.05$) weaker than those in C. Although for most forearm muscles RMS increased with time in both groups, the activity of ECR was markedly reduced ($p<0.05$) in TE group. To consider the importance of kinetic chain principles, we integrated strength and EMG measures in multiple segments of upper limb. A global upper limb weakness was found which needs to be addressed in prevention and treatment strategies for TE. In other words, this suggests that whole upper limb should be included in TE rehabilitation. We were able to identify a muscle activation imbalance (in ECR) in TE that, if not corrected, would result in a widespread imbalance through whole upper limb.

SUMMARY/CONCLUSIONS

We suggest that restoration of normal ECR activity should be a treatment goal in the rehabilitation of TE. Although numerous exercises are proposed for wrist extensor and flexor muscles, future work should address which exercises best restore muscle balance rather than muscle strength per se.

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MUSCLES THAT CROSS THE KNEE JOINT HAVE A PREFERRED ORIENTATION TO RESIST A FORCE APPLIED FROM OUTSIDE EXCEPT FOR EXTENSION AND FLEXION

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INTRODUCTION

The tibiofemoral joint has six degrees of freedom. Based on the proximal attachments of muscles around the knee, each muscle has a certain moment arm length to produce antagonistic joint moment against the rotation and valgus/ varus knee moment applied from outside. It has not been reported which muscle contributes to resistance against multidirectional perturbation applied from outside the knee joint. If this point could be clarified, it would be possible to develop an exercise to stabilize the knee joint against multidirectional perturbation. It may be valuable to produce a rehabilitation program or prevention program for some sports injuries, for example anterior cruciate ligament (ACL) rupture. The present research was carried out to clarify the preferred direction, in which a certain muscle work markedly during a force is applying from various kinds of direction.

METHODS

Ten healthy young men participated in the present study. Each gave his informed consent for the experiment, which was approved by the local ethics committee. His legs on both sides were examined, and the order of the examination was set randomly in each subject.

The subject sat in the front of a haptic device for motor learning that we produced, named “KINESTAGE”. One foot was fixed on the foot-plate, and the knee joint was modulated at a position of 60 degrees flexion (Fig. 1). To explore the muscle activity, which changes depending on the applied force direction, a kinetic-equilibrating task (K-E task) was adopted. In the K-E task, the subject was instructed to produce a certain muscular output to maintain the established foot position following force sense and position sense while the machine was applying certain target force to the foot at any direction, so that the foot position would remain completely still if the task were executed perfectly. The K-E task is thought to be similar to an isometric muscular contraction, with the different being that the condition during the K-E task is unstable. The resistance was applied as a trapezoid shape, with the force being gradually increased at 6N per second (Fig. 2). The peak force level was set at 42N for 2 seconds.

Figure 1: Diagram that a subject is using KINESTAGE.

Using the KINESTAGE, passive force was applied to the subject's foot from the eight directions [from anterior to posterior (A-P), anterolateral to posteromedial (AL-PM), lateral to medial (L-M), posterolateral to anteromedial (PL-AM), posterior anterior (P-A), posteromedial anterolateral (PM-AL), medial lateral (M-L), anteromedial to posterolateral (AM-PL)]. Bipolar surface electromyographic (EMG) activities in 13 muscles were recorded. In the present study, the data recorded in the quadriceps femoris, hamstring, and gastrocnemius are shown as the primary stage report. EMG signals were calculated as the average rectified value (ARV). In a respective subject, ARV data recorded during each bout of a direction were normalized (nARV) by ARV, which was the maximum value in the single subject during passive moment applied from a certain direction.

RESULTS

An example of an EMG recorded from the rectus femoris (RF) is shown in Fig. 1. EMG in RF recorded during ALPM and AMPL task was significantly larger than that recorded during another task. In the vastus medialis (VM) and lateralis (VL), the direction in which EMG was significantly increased was A-P. Not only P-A, PM-AL, PL-AM, but also the M-L and L-M tasks make EMG larger in the hamstring muscles. The preferred direction in the gastrocnemius was AM-PL.

DISCUSSION

Generally, it is supposed that the RF, VM, and VL work to extend the knee joint, so that they must be activated in the A-P task. However, the present study indicates that there is a difference in the preferred direction for the muscular activation among the three extensors. This result suggests that the three extensors have a functional difference depending on the direction from which force is applied from outside. In the hamstring, it was clarified that the extensors work as resistance against force coming from the lateral side.

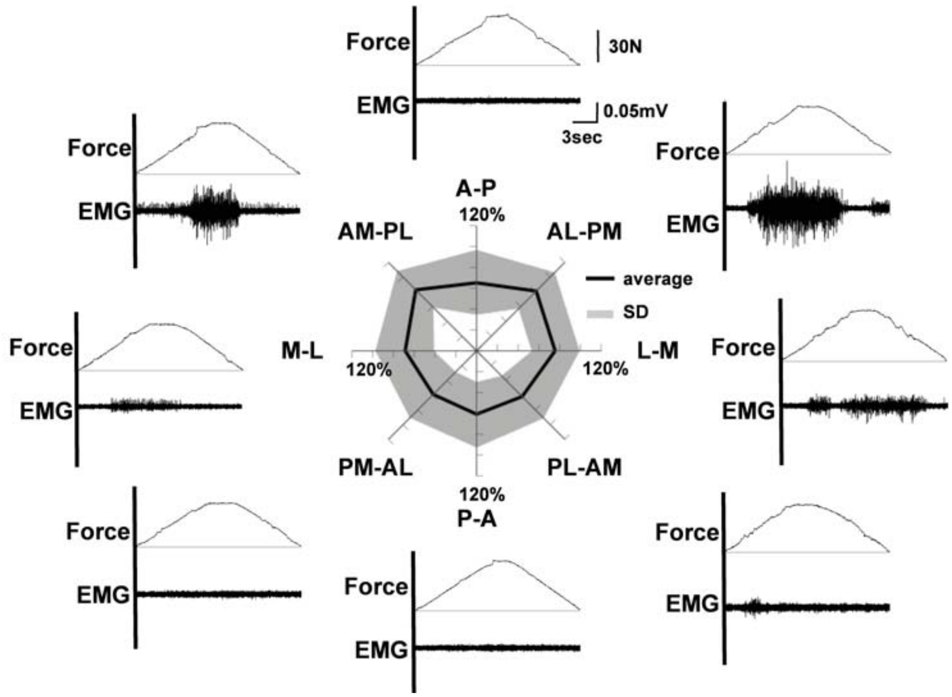

Figure 2: Specimen records of EMG and force curve data in a single subject. EMG was recorded in RF. A radar chart shows average±S.D. of nARV for all subjects.

TEMPOROMANDIBULAR DYSFUNCTION AND SPORT-RELATED FACIAL FRACTURES: A CASE REPORT

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INTRODUCTION

The etiology of temporomandibular dysfunction (TMD) is considered multifactorial. Orofacial injuries sports-related can to be associated with hyperfunction or dysfunction of the masticatory muscles. Currently this dysfunction involves assessment of jaw motion, tenderness of jaw muscles and temporomandibular joints, and joint noises. The purpose of this study was to examine masseter and anterior temporalis muscle activity patterns in one individual with sport related facial fractures.

METHODS

An individual of 23 years practitioner of martial arts suffered several facial fractures (from 15 years) mainly in the region of the left mandible. The main complaints were related bruxism night, pain in the left mandibular region and headache with periods of intense pain around the left side of the face which prevented daily activities such as mastication, work frequency and sporting activities.

The volunteer was submitted to a clinical evaluation and the severity of symptoms was graduated according to the Index of Dysfunction Clinic (Helkimo, 1974).
The activity of the Anterior Temporalis (AT) and Masseter (M) muscles was registered with Myosystem Equipment of 12 channels, with 12 bit of resolution, with a high pass filter of 10 Hz and a low pass filter of 500 Hz and an acquisition rate of 2000 Hz. The activity was bilaterally detected using Medi-Trace Kendall-LTP surface electrodes, with a between-electrodes center-to-center distance of 25mm, connected to pre-amplifiers model PA 1010-VA with 20 times gain. The impedance was 10^{12} Ohms and a minimum Common Mode Rejection Relation of 112 Db.

The volunteers sit relaxed with arms on their legs, eyes opened, with the Frankfurt occlusal plane parallel to the ground. The skin was throughly cleaned with alcohol and muscle function test was performed before electrode placement. The reference electrode was placed over the sternum.

It evaluated the participation of each muscle in Mandibular Rest Position (MRP) (10 s), in Maximum Intercuspal Position (MIP) (5 s) and in Chewing Cycle (CC) (10 s) at the rate of 80 beats/min (metronome). The volunteer carried out three repetitions with an interval of 1 min between repetitions. During the contraction in the MIP and CC, Parafilm “M” material was folded 15 times (1.5cm by 3.5cm) and placed bilaterally on the mandibular first and second molars. To quantify muscle activation was calculated the root mean square (RMS) values and was obtained an average of 3 periods: 10 s in MRP, 5 s in MIP and 3 cycles of contraction in CC.

RESULTS AND DISCUSSION

The Index of Dysfunction Clinic showed that the volunteer presented severe TMD. The figure 1 show the pattern of electrical activity of AT and M muscles, bilaterally. Table 1 show the mean values of RMS.

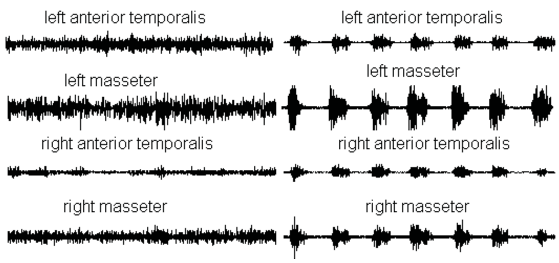

Figure 1: Electromyographic activity during maximum intercuspal position and chewing cycle.

TMD of traumatic origin was found in 79% of children between 4 to 16 years (Bodner & Miller, 1998) and 24.5% of adults between 15 to 70 years and of these, approximately 74.1% had severe dysfunction (De Boever & Keersmaekers, 1996). There is a significant association between bruxism, M muscle pain, headache and limited mouth opening (Solberg et al., 1979). The presence of sensitive points to palpation and night hyperactivity of M were related to the severity of the signs and symptoms of TMD. Subjects with pain in the craniomandibular muscles show significant asymmetry in the use of the AT and M muscles during the intercuspal position, suggesting that these muscles do not work in a balanced pattern during loading (Nielsen et al., 1990). The not coordinated and hyperactive muscles are primary sources of forces of compression and repetitive tension between the structures of the system masticatory (Dawson, 1995).

CONCLUSIONS

The muscular hyperactivity and asymmetric masticatory pattern along with other signs and symptoms of TMD can direct the treatment for a balanced muscle activity and shows be a measure extremely useful to include in the functional examination.

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Table 1: Mean values and standard deviations of RMS (μV) during mandibular rest position (MRP), maximum intercuspal position (MIP) and chewing cycle (CC).

	MRP	MIP	CC
Left Anterior Temporalis	8.32 ± 0.28	183.55 ± 5.88	51.01±3.75
Left Masseter	6.09 ± 1.18	772.24 ± 37.42	171.20 ± 25.14
Right Anterior Temporalis	6.96 ± 1.0	110.12 ± 27.60	38.57 ± 10.60
Right Masseter	6.12 ± 1.35	245.10 ± 42.74	94.45 ± 27.23

ELECTROMYOGRAPHY ANALYSIS OF THREE DIFFERENT TYPES OF CHEST EXERCISES: INCLINED BENCH PRESS, INCLINED DUMBBELL PRESS AND INCLINED DUMBBELL FLY

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INTRODUCTION

Working out is a common practice for athletes as part of training and also nonathletes for health purposes. Some of the working out exercises use several joints; this complicates the identification of which specific muscle group is used for each exercise. Inclined Bench Press (IBP) is associated to chest training, and yet it recruits the muscles of the shoulder. The Surface Electromyography (EMGs) is used to determine the motor unit action potential of muscles (BASMAJIAN & DELUCA, 1985). The aim of this study is to analyze the activation of four muscles: Pectoralis Major (upper portion), Pectoralis Major (lower portion), Deltoideus Mediums and Deltoideus Anterior using EMGs during the three following chest exercises: IPB, Inclined Dumbbell Press (IDP) and Inclined Dumbbell Fly (IDF).

METHODS

The sample consists of four men aged between 20 and 30 years old, with at least one year of working out practices, and bodyfat percentage below 14%. The materials used in body composition assessment were professional stadiometer (Sanny®), digital scale (Plenna®), skinfold caliper (Lange®), anthropometric tape (2.0m; Sanny®), anthropometer (Sanny®) and anthropometric pencil. For the skin preparation, the hairs on the electrodes sites were removed. For data collection an electromyographer Miotool 400 (Miotec®), electrodes Meditrace™ 200 (Kendall®) and a notebook Vaio (Sony®) were used. For the exercises test were used: IBP (Equipment), inclined bench and dumbbells (12-30kg; Righetto®), free weights (20, 10, 2 and 1kg) e digital metronome DM – 11 (Seyko®) to control the rhythm of contractions. The muscles analyzed were: Deltoideus Mediums (DM), Deltoideus anterior (DA), Pectoralis Major upper portion (PMU) and Pectoralis Major lower portion (PML). Three different types of chest exercises were performed by the subjects: IBP, IDP and IDF, and the order of exercises was random. The loads were determined by a load test (BOMPA, 2001). For statistical analysis the software SPSS 11.5 for Windows was used. Firstly the Kolomogorov-Smirnov test ($p \leq 0.05$) was done. Secondly ANOVA for repeated samples with Bonferroni post hoc test ($p \leq 0.05$) was performed.

RESULTS AND DISCUSSION

Data analysis showed more activation of DM during IDP exercise and presents significant differences compared to inclined bench press, as seen in Figure 1. A minor activation of DA for inclined IDF exercise was identified in Figure 2, it also had a significant difference compared to other exercises.

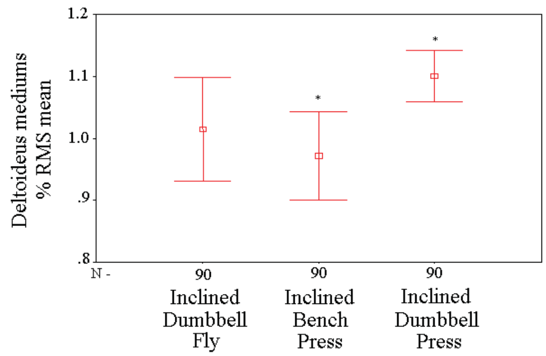

Figure 1: % RMS mean of DM in three exercises.

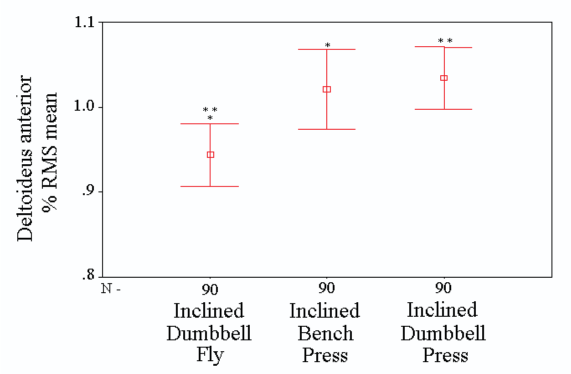

Figure 2: % RMS mean of DA in three exercises.

PML showed great variation and the IBP showed lower variation between the analyzed muscles, as seen in Figure 3. The IBP behavior was not as expected.

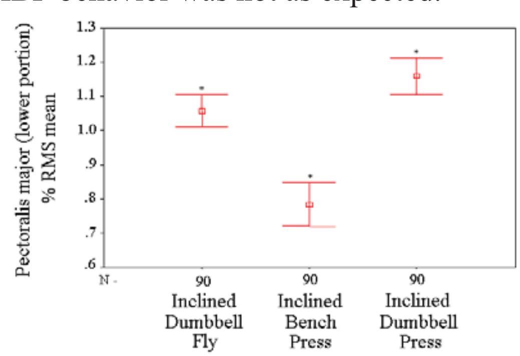

Figure 3: % RMS mean of PML in three exercises.

Figure 4 shows a greater PMU activation for IDP exercise with a significant difference compared to IBP. All muscles analyzed showed greater activation during IDP if compared to the other exercises in this study.

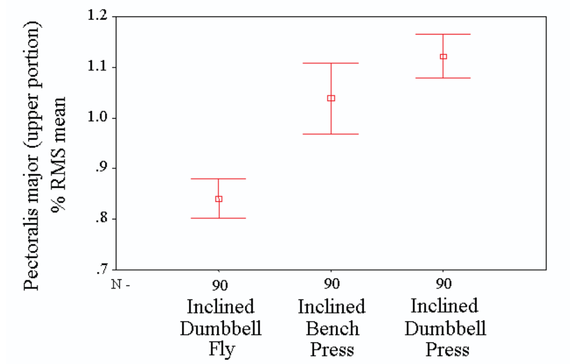

Figure 4: % RMS mean of PMU in three exercises.

This fact can be explained by movement mechanics because of triceps participation. It has been observed that free weights usage causes neighbor muscles to help the stabilization of upper limbs during movement (FLECK & JUNIOR, 2003).

SUMMARY/CONCLUSIONS

It was concluded that IDP exercise presents the greater activation for all muscles analyzed. However, further studies should be carried out with a greater number of significant samples for more faithful results.

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EFFECTS OF A HIGH-REPETITION RESISTANCE TRAINING PROGRAM ON METABOLIC AND NEUROMUSCULAR VARIABLES

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INTRODUCTION

Recently, it has grown increasingly the number and types of group activities in the fitness centers. In general, these activities are conducted as classes with a mean duration of 45-60 min, and have a standard sequence of musics, which determine the type and velocity of movement. Some of these activities have the objective to improve strength and endurance in the major muscle groups and are characterized by a sequence of high-repetition resistance exercises with emphasis in muscle groups of inferior limbs, superior limbs and trunk. However, there are few studies in the literature which have investigated the chronic effects of these exercise sessions on neuromuscular and metabolic responses (O'Connor and Lamb, 2003). The main purpose of this study was to analyze the effects of a high-repetition resistance training program on neuromuscular and metabolic variables.

METHODS

Nineteen healthy untrained women (aged 21.67 $\pm$ 2.06 years, 1.64 $\pm$ 0.1 m, 62.79 $\pm$ 8.9 kg, 26.76 $\pm$ 4.07 % body fat) were were randomly placed into a training group (TG, n = 9) or a control group (CG, n = 10). They were submitted to the following protocols, before and after the training period: 1) One repetition maximum squat test (1RM) and a high-repetition training session (HRTS) for familiarization; 2) Isometric maximum voluntary contraction (MVC) to the knee extensors; 3) Maximal countermovement jump (CMJ); 4) An incremental test until exhaustion in a treadmill to determine the velocity (V_{AT}) and heart rate (HR_{AT}) corresponding to the anaerobic threshold (3.5 mM of blood lactate).

The TG group was submitted to 12 weeks of HRTS of 60 min, called Bodypump® (Mix 46, Les Mills International Ltda). The main characteristics of this session were a particular sequence of musics with different rithms, the use of variable free weights (1–5 kg) with high repetitions per music (42 to 91), and the emphasis on inferior and superior limbs and trunk. To individualize training loads, squat (Music 2 - M2) and split-squat (Music 7 - M7) exercises were performed with initial training intensity of 10%1RM, with an increase of 5% at each two weeks.

To analyze the training effects during this session, the subjects performed two experimental sessions before (Pre) and after (Post) the training period, using the same absolute weights. During this session, blood lactate (LAC) samples and heart rate (HR) measurements were analyzed after M2 and M7. The electromyographic (EMG) signal of vastus lateralis was also recorded immediately after M2 and M7 during a 5 s isometric contraction of knee extensors at 50% MVC. The root mean square (RMS) was determined using EMG data.

The effects of group (CG and TG) and training (Pre and Post) in LAC, HR and RMS were analyzed using ANOVA TWO-WAY, complemented by Scheffé test. The effect of training on 1RM, MVC, CMJ, V_{AT} and HR_{AT} was analyzed using Paired Student *t* Test.

RESULTS AND DISCUSSION

There were no significant difference in body mass and body fat after the training period in TG ($P > 0.05$). CG presented a significant increase in body mass after the training period and higher values of body mass than TG before and after the training period ($P < 0.05$).

There was a significant improvement in 1RM (Pre = 51.6 ± 11.7 kg, Post = 68.33 ± 13.69 kg) in TG ($P < 0.05$). However, there was no significant differences in MVC (Pre = 386.2 ± 80.2 N, Post = 450 ± 86 N) but a significant increase in RMS values (207.47 ± 48.82 µV, Post = 284.59 ± 77.76 µV) during these contractions in TG ($P < 0.05$). There was no significant difference in CMJ in both CG (Pre = 23.83 ± 3.90 cm, Post = 21.95 ± 3.26 cm) and TG (Pre = 25.61 ± 4.04 cm, Post = 27.97 ± 4.20 cm) ($P > 0.05$).

The maintainance of maximal isometric strength and explosive strength (CMJ) seems to indicate that the training effects in muscular strength seems to be evident only in measurement conditions similar to performed during training. The significant increase in EMG may indicate an improvement in neuromuscular characteristics. However, this was not sufficient to promote increase in isometric strength, probably by aspects related to the specificity of the exercise.
In relation to the responses during HRTS in TG, there was no significant differences in LAC at M2 (Pre = 4.43 ± 1.73 mM, Post = 2.86 ± 0.91 mM) and M7 (Pre = 5.52 ± 2.14 mM, Post = 4.2 ± 1.22 mM) ($P > 0.05$). In the same way, there was no significant changes in HR at M2 (Pre = 155 ± 17 bpm, Post = 143 ± 20 bpm) and M7 (Pre = 164 24 bpm, Post = 153 ± 22 bpm) ($P > 0.05$). However, there were lower RMS values during submaximal isometric contractions at M2 and M7 in TG when comparing to CG after the training period ($P < 0.05$). TG presented RMS values significantly lower after the training period at M2 (Pre = 46.1 ± 13%MVC, Post = 31.6 ± 8%MVC) and M7 (Pre = 59.2 ± 11.3%MVC, Post = 39.4 ± 11.2% MVC), possibly explained by increases in strength. The CG presented no changes after the training for all variables. There was no significant difference in V_{AT} and HR_{AT} in both groups ($P > 0.05$).

Conclusion

The strength training protocol using BODYPUMP™ sessions improves muscular strength and decreases the neuromuscular requirement, but seems not be enough to decrease cardiovascular and metabolic demands during training session.

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ELECTROMYOGRAPHY ANALYSIS OF FOUR DIFFERENT TYPES OF ABDOMINALS EXERCISES

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INTRODUCTION

Since WWII, according with Correia & Mil-Homens (2004) EMG studies were generalized by scientist for clinical purpose and kinesiologic studies. Abdominal exercises have been analyzed considering different trunk muscles, as for example, rectus abdominis supra-umbilical and infra-umbilical portion, an external obliquous to show which excersice is more effective during training. (HILDENBRANT et al. 2004; STERNLICHT et al. 2003; STERNLICHT et al. 2005; WILLETT et al. 2001).
The purpose of the study is to analyze conventional, conventional leg supported, reverse conventional and sit up abdominal exercises to verify which one is more effective for abdominal muscles.

METHODS

Sample was constituted by eight subjects (7 men, one female), 22.9±7.2years, 1.75±0.07 m; 77.12±7.57kg body mass and 13.01 ± 5.7 %BF. Surface bipolar disposable electrodes, model 31118733 (Kendall-USA) were used, placed according to SENIAM Project recommendations for muscles analyzed. EMG data were recorded using 4 channel Miotool 400 with sample frequency of 2000 Hz. Data analyze was done by Miographic 2.0 software (MIOTEC, Biomedical Equipments-Brazil) with band-pass 20 to 500 Hz. After filter process, RMS was obtained. Data was normalized using mean value for each channel. Channels were divided as following: channel 1 = 2nd area of rectus abdominis, channel 2 = 4th area of rectus abdominis; channel 3 = external obliquos; channel 4 = femoral rectus. Reference electrode was placed on the tibial tuberosity. All electrodes were placed on the right side of the subject and later performed 3 sets of 10 repetitions per type of exercise randomized.
SPSS 11.5 and Bioestat 4.0 statistical software were used for data analysis.
Firstly, data normality was verify using Kolmogorov-Smirnov ($p \leq 0,05$) test. Friedman with a post hoc test ($p \leq 0,05$) was used later for data interpretation.

RESULTS AND DISCUSSION

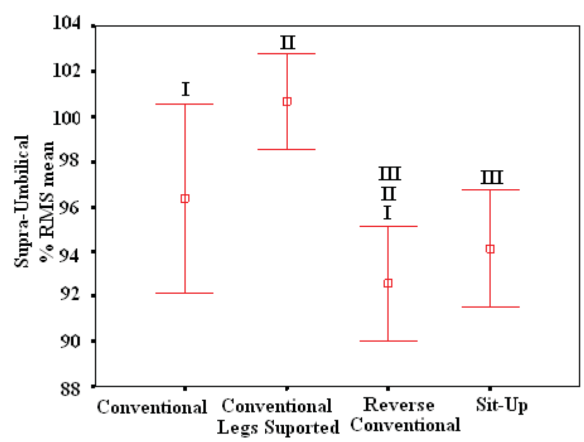

Figure 1: % RMS mean of muscle supra-umbilical.

Conventional and conventional leg supported execises had greater supra-umbilical muscle activation. Hildenbrand et al. (2004) found similar results using appropriate apparatus. Willett et al. (2001) showed that conventional exercise was less effective for that purpose.

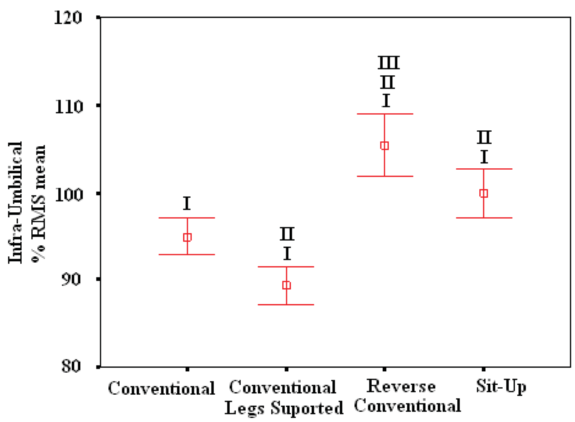

Figure 2: % RMS mean of muscle infra-umbilical

Related to Infra-umbilical exercise activation Willet et al. (2001) found reverse conventional exercise as more effective however, as seem in figure 2, sit-up exercise activated in a similar pattern comparing with reverse conventional exercise.

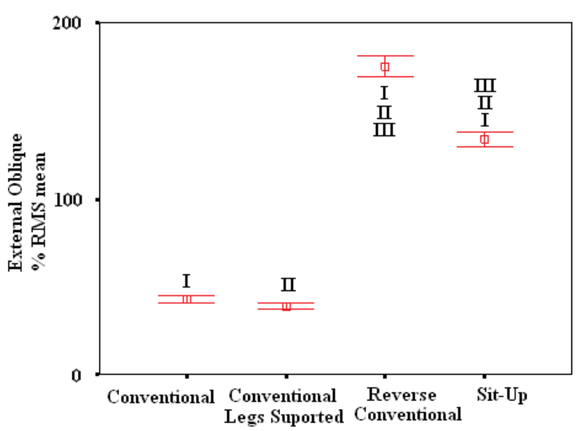

Figure 3: % RMS mean of muscle external oblique.

External oblique greater activation was found during reverse conventional exercise followed by sit-up exercise. Both conventional exercises had lesser activation and corroborate Willet et al. (2001) results.

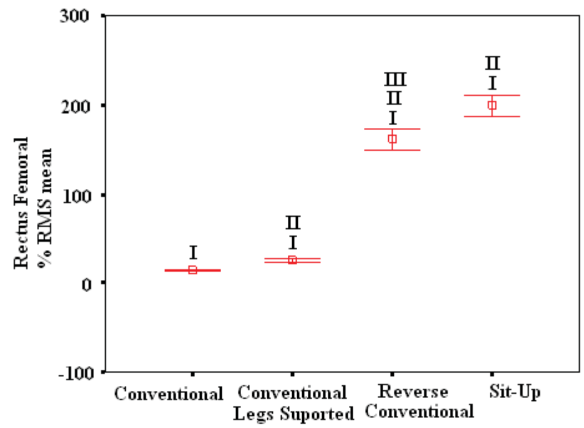

Figure 4: % RMS mean of muscle rectus femoral

Rectus femoralis lesser activation was found during both conventional exercises. This presumes that no hip bend occurs together with trunk bend.

CONCLUSIONS

It was found that conventional exercise highly activated rectus abdominis. By the way, it less activated rectus femoralis.

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Track 12

Movement Disorders (MD)

MOVEMENT ANALYSIS IN EARLY DIAGNOSIS OF A DEVELOPING SPASTICITY IN NEWBORNS WITH INFANTILE CEREBRAL PALSY

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INTRODUCTION

Infantile cerebral palsy (ICP) is a widely used term for a perinatally acquired non-progressive state of functional impairment of the brain. ICP is the most frequent cause of handicap in children and approximately one out of 400 live births is affected. Especially pre-term babies face a much higher risk of suffering from ICP than full-term babies, depending on gestational age. Infantile cerebral palsy causes in most cases a spastic disturbance of the muscle system. To limit these consequences, physiotherapy should start as early as possible. However early diagnosis of a developing spasticity in newborns with infantile cerebral palsy is presently based on visual assessment by the attending physician i.e. subjective impressions. Generally, the diagnosis is governed by observations of the spontaneous motor activity of the baby [Prechtl, H. (1997)]. Thus, there is a distinct need for an objective evaluation of the spontaneous motor activity of newborn babies, which allows an early diagnosis of a developing spasticity based on quantitative data. To this end, a procedure has been developed for the objective evaluation of spontaneous motor activity in newborns [Meinecke, L. (2006)].

METHODS

For retrieving the spontaneous movement in 3-dimensional coordinates a common video based motion analysis system (Vicon 370) has been used. According to the physicians observations a simplified full body model is used, which describes the spatial position of the single body segments (Figure 1).

Figure 1: 3D motion analysis: Marker placement

Those characteristics of movement patterns were identified, which form the basis for visual assessment of a baby's movement and describe the differences between healthy and affected subjects. Algorithms were developed to extract 53 quantitative parameters from the patient's 3D movement data which reflect the characteristics of motion. Out of these 53 parameters an optimal combination of 8 was determined using cluster analysis. Subsequently, a classification procedure was implemented to organize the subjects' movements into homogeneous classes, "healthy" or "at risk" based on quadratic discriminant analysis. The patient and norm collectives consisted respectively of 26 and 66 measurements on preterm and full term newborns.

RESULTS AND DISCUSSION

Figure 2 shows an example of the resulting 3D trajectories. The movement of healthy children, in contrast to that of affected children, is mainly characterized by the smooth and harmonious marker trajectories of the end effectors. As an example, one of the parameters resulting from the cluster analysis that quantifies the smoothness of a child's movement calculates the areas in which the marker trajectory is outside of the standard deviation of the moving average.

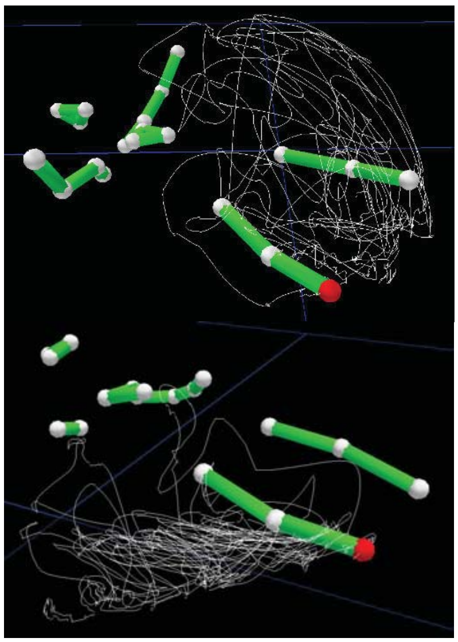

Figure 2: 3D reconstruction of the right foot trajectory for a healthy (upper fig.) and an affected child (lower fig.)

Figure 3 shows this parameter for a part of the right foot's trajectory with its moving average and the standard deviation of the x component. Utilizing the calculated optimal parameter combination, the detection rate of the classification methodology was verified for new and so far unclassified data. From a total of 52 'unknown' measurements in the evaluation database, of which 46 were diagnosed as healthy and 6 as affected, the discriminant analysis with 8 optimized parameters classified 40 measurements correctly. For the evaluation database, sensitivity is 100%, whereas specificity only reaches 73%. The overall detection rate is 77%. Measurements that were already part of the training database were detected with a specificity, sensitivity and overall detection rate of 95% each. The classification algorithm allows a reliable discrimination between healthy and affected subjects based on objective data sets from 3D movement analysis. The overall detection rate is expected to rise with increasing patient and norm collective database size.

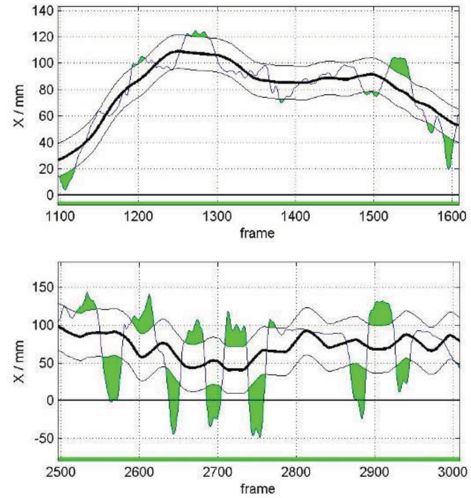

Figure 3: Area outside of standard deviation of moving average, trajectory of the right foot for a healthy (upper fig.) and an affected child (lower fig.)

CONCLUSIONS

Utilizing 3D motion analysis and the aforementioned classification, an objective evaluation of spontaneous motor activity in newborns becomes possible. The methodology presented permits the risk of a developing spasticity to be predicted for newborn babies.

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EVALUATION OF THE ANGLE OF CATCH IN SPASTICITY ASSESSMENT

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INTRODUCTION

Clinical assessment of spasticity is important in treatment of e.g. children with cerebral palsy. A common definition of spasticity is a motor disorder characterized by a velocity-dependent increase in muscle tone, resulting from hyper excitability of the stretch reflex [Lance 1980].

Based on this definition, the SPAT was developed: a clinical spasticity scale as a successor of the Modified Tardieu Scale. The muscle is stretched passively at slow velocity to define the range of motion (ROM) and subsequently a fast passive stretch is performed to detect the angle of catch (AOC), i.e. the angle at which a sudden resistance in the movement occurs, before the ROM is reached [Scholtes 2006]. The AOC is supposed to be the consequence of a sudden appearance of increased muscle activity in spastic muscles in response to the fast passive stretch, which is suggested to be absent in slow passive stretch. However, the relation between the AOC and the sudden increase in muscle activity has been doubted [Pandyan 2005].

This study aims to evaluate whether the AOC is indeed the consequence of a sudden velocity-dependent increase in muscle activity in fast passive stretch.

METHODS

Twenty children diagnosed with spastic CP participated in the study (5-14 years of age, GMFCS range 1-4). The SPAT was performed in three sessions in three leg muscles: medial hamstrings, soleus and gastrocnemius. The tests were performed in a standardized posture.

Two lightweight inertial sensors [MT9, Xsens, The Netherlands] tracked the motion of the proximal and distal segments during the test with a sample frequency of 100 Hz. EMG was recorded from surface electrodes (sample frequency 1 kHz), rectified and lowpass filtered at 40 Hz.

Joint angles were calculated taking the distal with respect to the proximal segment orientations, after determination of the helical axis. From these, joint angular velocities and angular accelerations were calculated. The AOC was defined at the angle where joint angular deceleration was maximal. Mean joint angular velocities and muscle activity were evaluated for the slow and fast passive muscle stretch. Time delay between the onset of muscle activity and the AOC was examined for the fast passive stretch type 3 (i.e. clear catch, blocking further movement). Correlations were calculated between the joint angular velocity at onset of muscle activity, the AOC and the time delay.

RESULTS AND DISCUSSION

Figure 1 shows a typical result of a slow and fast passive stretch of the hamstrings muscle in one subject. Little muscle activity is seen in the slow passive stretch; in the fast passive stretch an AOC is seen due to a sudden increase (burst) in muscle activity.

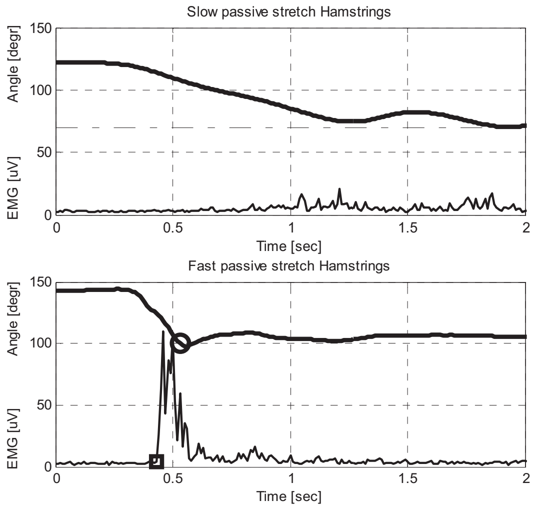

Figure 1: Slow and fast passive stretch of the hamstrings. Thick line: joint angle knee; Thin line: EMG; Dot-dashed line: range of motion; Square: sudden increase muscle activity; Circle: angle of catch.

Mean angular velocities of the slow passive stretch of the hamstrings, soleus and gastrocnemius were 27±18, 8±5 and 6±5 deg/sec respectively. Mean angular velocities in the fast stretch were significantly higher: 246±57, 272±81 and 239±72 deg/sec respectively. 85% of the slow passive stretch tests showed no sudden increase of muscle activity. The amplitude of muscle activity of all muscles in slow passive stretch and of non-spastic muscles in fast passive stretch was significantly lower then the amplitude in spastic muscles in fast passive stretch.

Time delay between the onset of muscle activity and the AOC in fast passive stretch of the hamstrings, soleus and gastrocnemius was 109±27, 32±12 and 34±15 ms respectively. Correlation between time delay and angular velocity at EMG onset was significant for the hamstrings and the soleus: the delay decreased with increasing velocity. Correlation between AOC and angular velocity at EMG onset was low but significant for the hamstrings: in patients with an early AOC a lower angular velocity was required to cause the catch.

SUMMARY/CONCLUSIONS

The results show the sudden appearance of increased muscle activity in fast passive stretch and the lack of this phenomenon in slow passive stretch. The time delay between the muscle activity and the detected AOC is quite consistent per muscle and dependent of angular velocity. The delay is assumed to correspond with the electromechanical delay, however force was not measured.

These results confirm the AOC is the consequence of the EMG burst in spastic muscles and can be detected as a sudden deceleration of the joint movement. Correlation of AOC and angular velocity at onset of EMG suggests that the AOC is related to the velocity threshold of the reflex loop. However muscle length also influences the stretch reflex [Fleuren 2006]. Future studies should therefore apply a range of velocities, to investigate length dependent velocity thresholds in spasticity.

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TAI CHI INTERVENTION IS EFFECTIVE FOR PEOPLE WITH PERIPHERAL NEUROPAHTY

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INTRODUCTION

Peripheral neuropathy (PN) affects many people. There are approximately 20 million suffering from the disease in United States alone. PN can lead to significant balance and mobility problems.

Tai Chi has been successfully used in aging and fragile populations as a rehabilitation tool for balance and mobility. The purpose of this study was to examine the effectiveness of Tai Chi training on PN-related balance and mobility problems.

METHODS

Individuals diagnosed with PN were recruited from the community and then randomly assigned to either Tai Chi or control groups. Tai Chi group practiced Tai for six weeks (three times a week) between the pre and post tests. The control group did not receive any exercise intervention.

Plantar pressure sensitivity (PPS) was assessed at five weight-bearing sites on the right foot sole with a 5.07 gauge monofilament. The number of sites with intact sensation was totaled to produce a PPS score ranging from 0-5.

Mobility was evaluated by the distance of a 6-mininute walk (6MWD) test and the duration of a Timed Up-and-Go (TUG) test. For the 6MWD, participants were instructed to walk as far as possible around two cones placed 30 meters apart down a well lit hallway. Distance covered (m) was recorded. For the TUG, participants sat in a chair with their feet on the floor facing a cone three meters in the front. The time (sec) taken to stand up, walk around the cone, and sat back down was recorded.

Eyes-closed standing balance was assessed by the average velocity (VEL, cm/s) and the area of an ellipse enclosing 95% (AREA, cm^2) of the body center of pressure during quiet stance. The average of two, 30sec trials was recorded.

Isokinetic knee joint strength was examined using a Biodex dynamometer (Biodex Medical, Shirley, NY) at 60 deg/sec. Both knee extensor (KE, Nm) and flexor peak torque (KF, Nm) were computed from five maximal effort trials. Peak torque of the best three trials was averaged and used for analysis.

Two-factor ANOVA with repeated measures was used for data analysis. Tukey post hoc analysis was used if needed. Alpha = .05.

RESULTS AND DISCUSSION

There were 30 participants for each of the Tai Chi and control group. Participants for the Tai Chi group (9 men, 21 women, mean ± SD age = 68.3 ± 10.9 yrs, height = 149.9 ± 8.3 cm, body mass=87.6 ± 25.5 kg) had been diagnosed with PN for 6.0 ± 3.7 years. Participants for the control group (15 men, 15 women, age = 71.0 ± 9.8 yrs, height = 146.6 ± 29.3 cm, body mass=78.6 ± 21.3 kg) had been diagnosed with PN for 6.4 ± 3.9 years. One tailed T-test revealed no difference between the groups for all the measurements mentioned here.

Among the seven measured parameters, there was no group or training difference observed among PPS, VEL and KF. Significance was observed from the following tests: group, pre-post and interaction for 6MWD; group and training effects for TUG; training effects and group X training interaction for AREA, as well as training effects and group X training interaction for KE. Significant interactions for these parameters are presented in the following figures. * indicates significant differences among concerned variables.

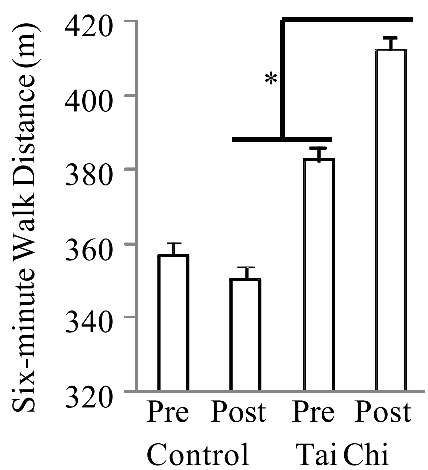

Figure 1. 6MWD after Tai Chi training was significantly greater than that of the pre training in the same group or that of post training with the control group.

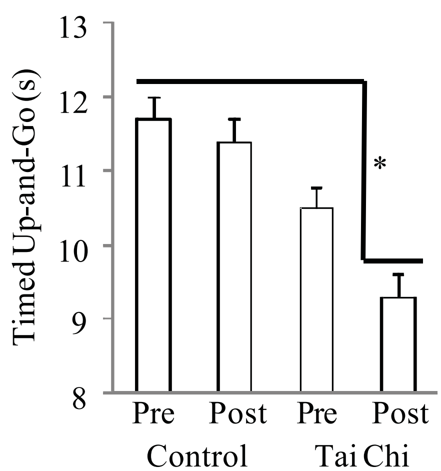

Figure 2. TUG after Tai Chi training was significantly less than the averages of the other three groups.

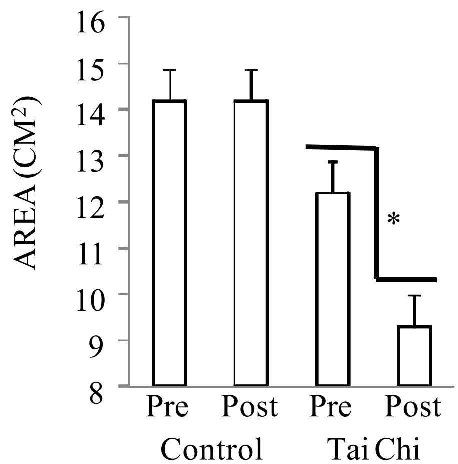

Figure 3. AREA after Tai Chi training was significantly reduced in the training group.

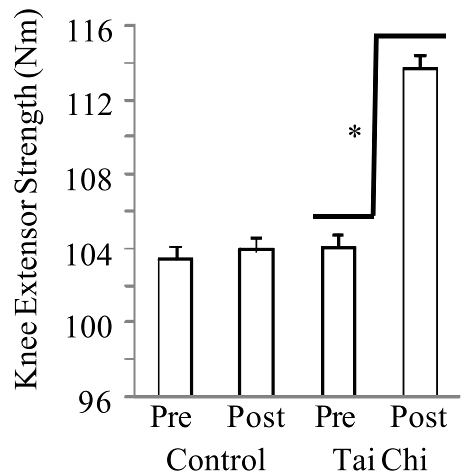

Figure 4. KE increased significantly in the Tai Chi training group.

SUMMARY/CONCLUSIONS

Six weeks of Tai Chi training increased functional performance among people with peripheral neuropathy. They walked longer distance in the 6-minute walk, used less time for the timed up-and-go, reduced center of pressure movement while standing quietly, and strengthened their knee extensors. In contrast, results from the control group suggested that physical performance will not improve and may even decline if left alone. We conclude that Tai Chi can be used as an effective rehabilitation tool for PN.

GAIT DYNAMICS ARE PREDICTIVE OF FUNCTIONAL PERFORMANCE IN OLDER ADULTS AND THOSE WITH PERIPHERAL NEUROPATHY

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INTRODUCTION

Gait variability is often quantified to evaluate the integrity of the human locomotor system. While the *magnitude* of variability is most often examined, the application of non-linear analysis to human walking has recently been employed to examine the *structure* of variability.

It is currently unknown if these properties of gait variability predict locomotion-based physical performance across different populations. Therefore, the purpose of this study was to determine the ability of measures related to the magnitude and structure of variability to predict locomotion-based performance in healthy older adults and those suffering from movement disorder associated with peripheral neuropathy (PN).

METHODS

Twelve individuals with diagnosed PN (mean ± SD age = 67.0 ± 10.9 yr) and 12 aged-matched healthy controls (age = 69.1 ± 10.0 yr) were recruited.

Locomotion-based physical performance was assessed by two common field tests. The 6-Minute Walk (6MW) test was completed with cones placed 30m apart along a lighted hallway. Participants walked as far as possible within 6 minutes back and forth around the cones, and distance (m) was recorded. The Timed Up-and-Go (TUG) test was conducted with the participant seated and a cone placed 3m in front of the chair. The time (sec) taken to stand up, walk around the cone, and sit back down was recorded.

On a separate day, gait variability was assessed during treadmill walking. 6MW distance was used to determine treadmill walking speed. Participants completed 3min trials during which motion analysis (60 Hz) was used to collect sagittal plane hip, knee, and ankle joint angle kinematics.

Stride duration variability (CoVar) was calculated by the ratio of stride duration variability and the mean stride time from 30 strides. Joint angle variability (JTvar) was calculated as the average standard deviation for each joint from the ensemble curves of the same 30 strides (Li et al 2005).

The structure of gait variability was assessed by short- ($?_{ST}$) and long-term ($?_{LT}$) Finite-Time Lyapunov Exponents associated with lower-extremity joint angles. $?_{ST}$ and $?_{LT}$ were computed for each joint from ensemble curves of 100 consecutive strides (Dingwell & Cusumano 2000). JTvar, $?_{ST}$, and $?_{LT}$ values were averaged across lower-extremity joint to produce mean scores.

Potential group differences in locomotion-based physical performance (6MW, TUG), and gait variability (CoVar, JTvar, $?_{ST}$, $?_{LT}$) were assessed with independent-samples T-tests. Pearson's R used to examine the relationships between measures of locomotion-based physical performance and gait dynamics.

Table 1: Group means ± Standard deviation (SE) for Locomotion-Based Physical Performance (i.e., Field Tests) and Gait Dynamics

	Field Tests		Magnitude/Structure of Stride-to-Stride Variability			
	6MW (m)	TUG (sec)	CoVar (sec)	JTvar (deg)	$?_{ST}$†	$?_{LT}$†
CON	530 ± 25	7.0 ± 0.5	2.1 ± 0.1	1.7 ± 0.1	3.0 ± 0.1	0.17 ± 0.01
PN	391 ± 27*	9.5 ± 0.6*	2.9 ± 0.3*	2.6 ± 0.2*	3.1 ± 0.1	0.17 ± 0.01

Note: * p < .05, † units = [<in[dj(i)]>/Stride]*100).

RESULTS AND DISCUSSION

The PN group exhibited reduced locomotion-based physical performance as evidenced by a 26% decrease in 6MW ($p < .05$) and a 36% increase in TUG ($p < .05$) (Table 1). Compared to controls, the PN group exhibited increased magnitude of gait variability (CoVar, JTvar) yet no difference in its structure ($?_{ST}$,$?_{LT}$) (Table 1).

In the Control group, $?_{ST}$ significantly predicted performance in the 6MW ($R = -.71$, $p < .01$) and TUG ($R = .77$, $p < .01$) tests. Individuals with decreased $?_{ST}$ values tended to perform better. Controls with less CoVar also tended to complete the TUG test in less time ($R = .67$, $p = .02$) (Figure 1).

Within the PN group, no measure of the magnitude or structure of variability predicted locomotion-based physical performance.

SUMMARY/CONCLUSIONS

In healthy older adults, the magnitude and structure of variability can predict performance in locomotion-based function. The high correlation between $?_{ST}$ and both the 6MW and TUG suggests that this property may more closely reflect locomotor system integrity.

The breakdown of correlation between measures of gait variability physical performance in the PN group points to the heterogeneity of this population. While gait variability may differentiate between individuals with PN and healthy older adults, additional research is needed to examine other PN-related alterations that lead to decreased locomotor performance.

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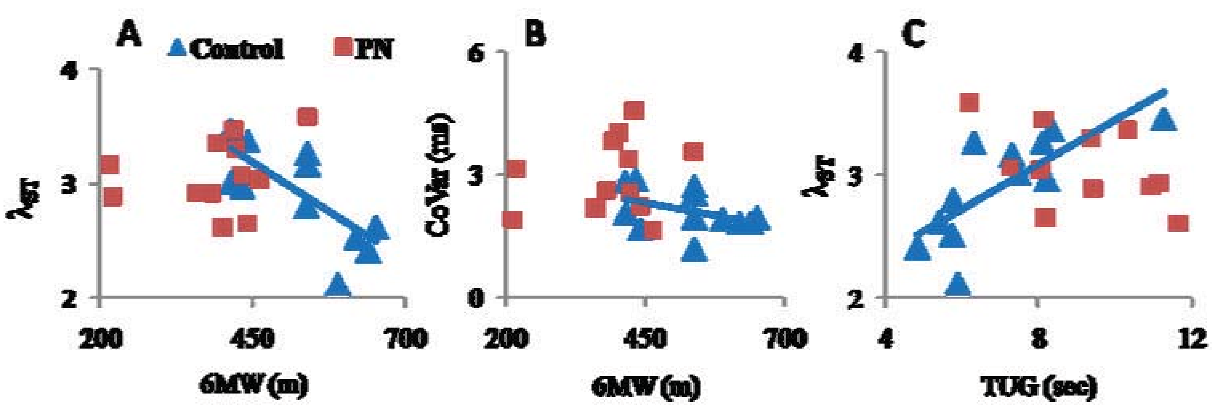

Figure 1 – Scatter-plots illustrating selected variability-related predictors of 6MW/TUG test performance. Linear best-fit lines have been added to indicate significance (Control Group only).

KINAESTHETIC DEFICITS IN FOCAL HAND DYSTONIA: PSYCHOPHYSICAL SENSITIVITY THRESHOLDS AND BRAIN IMAGING DATA

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INTRODUCTION

A dysfunction of the cerebrobasal ganglia loop, as seen in Parkinson's disease (PD) or dystonia, may lead to impaired kinaesthetic sensitivity (Konczak et al, 2007; Maschke, et al, 2003; Maschke, et al 2006). The central origin of the deficit was documented by showing that patients with cervical dystonia or blepharospasm exhibit a reduction in kinaesthetic sensitivity of their non-dystonic fingers (Putzki et al., 2006). This study assessed kinaesthetic deficits in both the affected and non-affected arm of individuals with focal hand dystonia. fMRI data collected during a finger tapping task was then used to quantify the spread of cortical activation and these data were correlated to the afore mentioned kinaesthetic data.

METHODS

Ten participants with idiopathic, task specific focal hand dystonia (FHD) and ten control subjects performed the passive motion detection task.

The passive motion apparatus consisted of an aluminum splint that could rotate horizontally (transverse plane) at one end when driven by a torque motor. The top portion of the sled was covered in a soft foam material for comfort and to minimize tactile cues.

Subjects were seated parallel to the testing apparatus with their forearm completely at rest on the testing apparatus with their shoulder positioned between 70-85º of shoulder abduction. They held a handheld trigger in their non-tested arm, which stopped passive motion once pressed. The splint caused either a flexion or extension movement of the forearm in the transverse plane.

Subjects wore goggles to occlude vision. Each trial was initiated with a verbal precue after which the sled began to move at a preset velocity. Subjects were asked to press the trigger as soon as they detected that their forearm was being moved. Afterwards subjects verbally indicated the direction of forearm motion as either "toward" the body or "away" from the body.

The experiment consisted of 60 pseudo-randomly presented trials for each tested arm. Twelve individual passive motion angular velocities were administered: 0.075°/s, 0.15°/s, 0.30°/s, 0.45°/s, 0.6°/s, 0.75°/s, 0.9°/s, 1.05°/s, 1.2°/s, and 1.35°/s. The direction of the splint varied randomly between flexion and extension and the selected angular velocities were presented in both ascending and descending order to account for any possible order effect.

Surface electromyographic (EMG) measurements were recorded to monitor the activity of the biceps and triceps muscles.

Trials revealing muscle activity were repeated at the end of the testing protocol.

For each trial the subject's judgment of movement direction and the actual direction of the sled were recorded. Detection times were calculated for each trial using the time it took the individual to depress the trigger and the angular velocity for the given trial.

Figure 1. Set-up of the splint apparatus.

In a follow-up session all participants performed a visually cued finger tapping task in a 3 Tesla fMRI scanner (Magnetom Trio, Siemens, Munich, Germany) with an eight channel head coil.

The cortical areas of activation were compared between the FHD participants and the healthy individuals using a previously established dispersion index (Christodoulou et al., 2001).

RESULTS AND DISCUSSION

Previously, we were able to show a deficit in both the directional judgments and detection time data of individuals PD as compared to healthy, age matched control participants. Similar deviations from the healthy group are evident within the FHD data for both detection time and directional judgments. Additionally, the overall trend presented by the FHD group was similar to that of the PD data.

The extent to which the cortical activation of the dystonic individuals deviated from that of the healthy individuals was established using an index which tallies the voxels activated outside of a cortical area defined by the healthy participants. The kinaesthetic sensitivity deficits presented by the individuals with FHD is expected to positively correlate to the amount of cortical dispersion measured during the finger tapping task.

SUMMARY AND CONCLUSION

These data support the idea that the cerebrobasal ganglia loops play an important role in the interpretation of joint position sense. This is important from both rehabilitative and assessment perspectives as deficits in kinaesthesia may be useful in early diagnosis and treatment of pathologies such as PD and FHD.

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IMPROVING MEDICATION TITRATION IN PARKINSON'S DISEASE

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INTRODUCTION

Parkinson's disease affects about 3% of the population over the age of 65 years and more than 500,000 US residents. Current therapy is based on augmentation or replacement of dopamine, using the biosynthetic precursor levodopa or drugs that activate dopamine receptors. These therapies are successful for some time, but most patients eventually develop motor complications. Furthermore, fluctuations in the severity of symptoms and motor complications (referred to as "motor fluctuations") are observed during dosing intervals (Weiner 2006).

Currently available tools for monitoring motor fluctuations are limited. In clinical practice, information about motor fluctuations is obtained by asking patients to recall the number of hours of ON (i.e. when medications effectively attenuate tremor) and OFF time (i.e. when medications are not effective). This kind of self-report is subject to perceptual bias and recall bias. Recent advances in wearable technology (Bonato 2005) make it possible to develop monitoring systems to capture movement patterns associated with motor fluctuations. In routine care, relating this data to the timing and dose of medications would greatly facilitate the titration of medications to minimize adverse symptoms.

METHODS

Twelve individuals were recruited in the study, ranging in age from 46 to 75 years, with a diagnosis of idiopathic Parkinson's. Subjects delayed their first medication intake in the morning so that they could be tested in a "practically-defined OFF" state (baseline trial). They performed a series of standardized motor tasks utilized in clinically evaluating patients with Parkinson's disease. After completion of the baseline trial, subjects took their medications and were subsequently tested using the same procedure every 30 min. During the experiments, we gathered accelerometer data using an ambulatory system (Vitaport 3, Temec BV). For this purpose, uni-axial accelerometers were positioned on the upper and lower limbs. Video recordings were made during each trial so that a clinical evaluation of the severity of symptoms could be performed. Results from the examination of the video recordings were compared with those derived via analysis of the accelerometer data derived as summarized below.

The accelerometer time series were segmented using a rectangular window randomly positioned throughout the recordings. Features were extracted from 30 data segments (i.e. epochs) for each motor task. Five different types of features were estimated from the accelerometer data: the root-mean-square value, the range of amplitude, two frequency-based features (the dominant frequency component between 1 and 10 Hz and the ratio of energy associated with the dominant frequency component to the total energy), two cross-correlation based features (the correlation coefficient derived from pairs of accelerometer time series and the time lag corresponding to the peak of the cross-correlation function), and the approximate entropy. Support Vector Machines (Vapnik 1995) were implemented to predict clinical scores based on accelerometer feature values.

RESULTS AND DISCUSSION

The effect on the prediction error of utilizing windows of different durations to segment the accelerometer data into epochs was assessed. It was determined that a window of 5 s allows one to achieve a prediction error lower than 5 %, while utilizing recordings of only 30 s for each of the motor tasks of interest. Support Vector Machines were implemented and the use of three different kernels was compared. A third-order polynomial kernel was found to be preferable to exponential and radial basis kernels based on the observation that the polynomial kernel provided satisfactory results for a smaller “C”-value. The prediction results were compared across all the motor tasks. Although differences were observed among prediction error values associated with different motor tasks, several motor tasks performed equally well. This observation suggests that the proposed accelerometer features and methodology capture aspects of the movement patterns that are not specific to a given motor task. This further suggests that the proposed analyses could be extended to other motor tasks, possibly including recordings of activities of daily living. Finally, the impact on the prediction error of utilizing different combinations of the feature types was studied. Results indicated that it is possible to reliably predict clinical scores on the basis of three features that are compatible with implementation on a body sensor network: the root mean square value, the data range, and the approximate entropy. Interestingly, the results suggest that the approximate entropy is key to achieving reliable estimates of the severity of all the studied symptoms and motor complications.

CONCLUSIONS

The results of this study suggest that accelerometer data could be utilized to facilitate medication titration in patients with Parkinson’s disease. The development of signal processing and pattern recognition algorithms was performed considering the compatibility of the proposed algorithms with the implementation on a body sensor network. The use of accelerometer features compatible with such implementation was shown to lead to prediction errors of the clinical scores smaller than 5 %.

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AN EMG AND KINEMATIC ANALYSIS OF GAIT WITH RESTRICTED KNEE EXTENSION

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INTRODUCTION

In patients with cerebral palsy, spasticity and muscle weakness lead to gait deviations. If both legs are affected, crouch gait with excessive knee and hip flexion is a frequent problem. Crouch gait in children should be subjected to treatment, as knee and hip flexion angles tend to increase with age due to increasing body weight, ultimately resulting in the loss of independent walking [McNee et al., 2004].

Several factors are known to contribute to crouch gait. In order to look at the effects of crouch gait without any interfering factors, healthy subjects walking in crouch can be used as a model [Van der Krogt et al., 2007; Matjacic and Olensek, 2007].

The purpose of this study was to determine the effect of restricted knee extension during gait in healthy individuals and its influence on muscle activity.

METHODS

Nine healthy adult female subjects underwent a 3D instrumented gait analysis to collect whole body kinematics (VICON 460 motion capture system) and surface electromyograms (SEMG) of selected leg muscles. First the subjects walked with 30° restricted knee extension bilaterally by applying a taping technique (Figure 1A). Then the tape was cut and after at least 10 minutes of rest, recordings continued for normal gait (Figure 1B). The markers and electrodes stayed unchanged between conditions. For each subject and condition data of 5 gait cycles were analysed and averaged. The SEMGs were collected according to the SENIAM guidelines and analysed with a wavelet analysis technique [Von Tscharner, 2000].

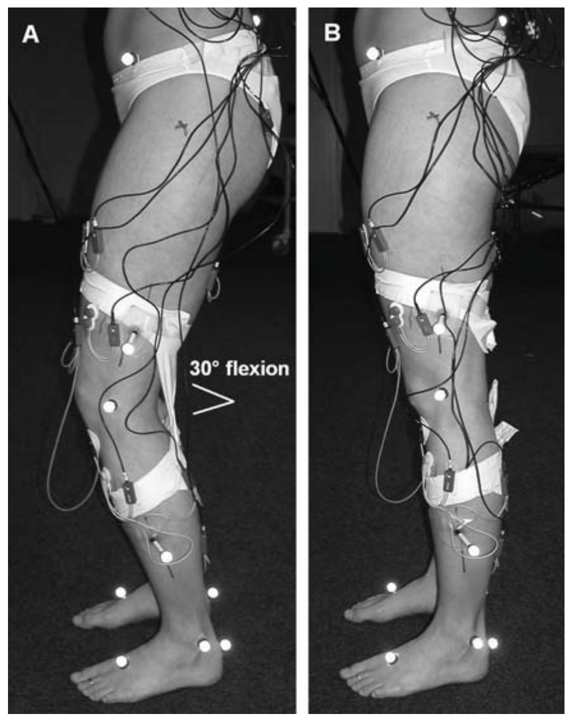

Figure 1: Testing conditions with A) restricted knee extension and B) normal.

RESULTS

The gait pattern with restricted knee extension of 30° resulted in increased anterior thorax and pelvic tilt, increased hip and knee flexion, and increased ankle dorsiflexion compared to normal. Peak knee flexion in swing and spine tilt (relative movement between pelvis and thorax), however, stayed unchanged. The duration of muscle activity over the gait cycle and the normalized amplitude of the SEMGs increased for the majority of muscles with the restricted knee extension gait pattern (Figure 2). Exceptions were for the tibialis anterior, in which the duration of activation did not change between conditions, and for the gastrocnemius medialis where the normalized amplitude decreased with restricted knee extension.

DISCUSSION AND CONCLUSIONS

The main goal of this study was to investigate changes in gait and muscle activity as a result of restricted knee extension. Restricting healthy subjects in their knee extension during gait resulted in substantial deviations from normal gait with compensatory adjustments observed at all levels. It also puts a significant higher demand on muscle activation in order to stay upright. With alterations in pelvic and thorax position towards anterior, the relative angle between these segments in the sagittal plane, however, did not change. The data give us insight into the changes occurring as a result of imposing a single factor that is known to influence crouch gait.

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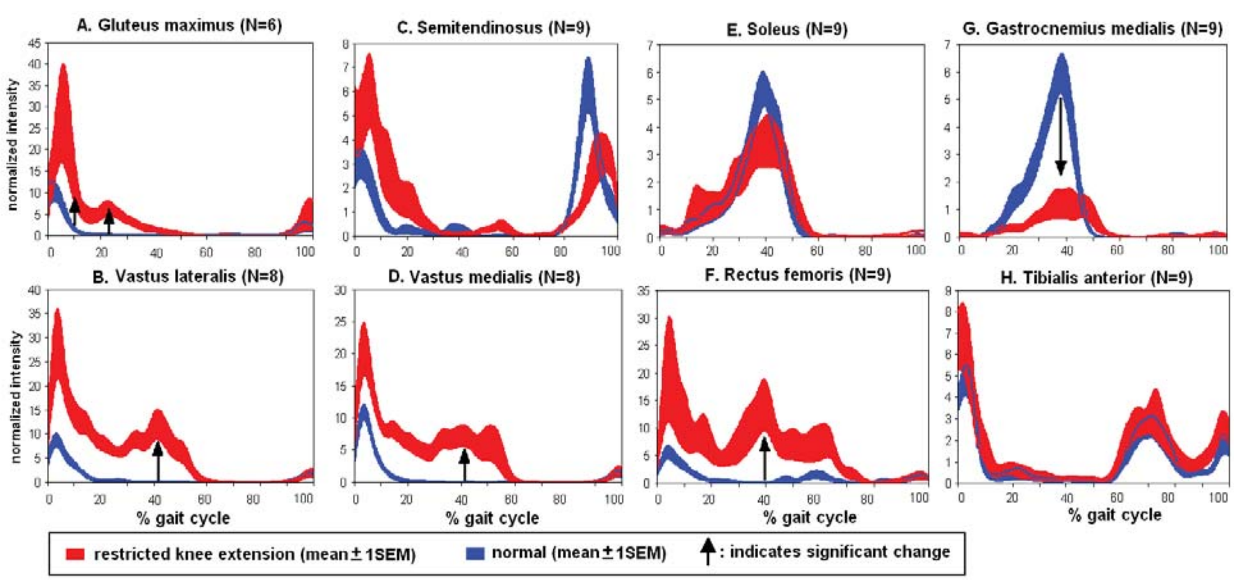

Figure 2: Group results for the muscle activation patterns over the entire gait cycle for all investigated muscles.

AN ELECTROMYOGRAPHIC ANALYSIS OF THE LOWER EXTREMITY DURING GAIT IN KNEE OSTEOARTHRITIS: INVESTIGATING THE EFFECT OF CONTROLLING FOR WALKING VELOCITY

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INTRODUCTION

Studying neuromuscular control and recruitment strategies during gait in individuals with knee osteoarthritis (OA) is relatively novel. Examining medial/lateral muscle group activation is important for the understanding of these factors in individuals with knee OA (Hubley-Kozey et al. 2006, Lewek et al. 2004). Despite these findings, it is difficult to determine if these features of differential activation are characteristic of individuals with a certain severity of knee OA or an artifact of differing spatial and temporal gait characteristics.

The purpose of this gait study was to investigate the overall magnitude of the electromyogram (EMG) of seven lower extremity muscles while controlling for velocity in asymptomatic individuals (ASY), those with mild to moderate knee OA (MOA) and individuals with severe knee OA (SOA) using principal component analysis (PCA).

METHODS AND PROCEDURES

Fifteen ASY, 16 MOA (classified by Kellgren-Lawrence I-III, and functional assessments) and 16 subjects with SOA (tested within one week of total knee replacement surgery) were selected from a large group of individuals who had all completed prior gait analysis. All subjects provided written informed consent. Participants with self-selected gait velocity greater than one m/s from each group were selected. After appropriate skin preparation procedures, circular Ag/AgCl bipolar skin surface electrodes (interelectrode distance 20mm) were affixed over the lateral gastrocnemius (LG), medial gastrocnemius (MG), vastus lateralis (VL),vastus medialis (VM), rectus femoris (RF), lateral hamstring (LH) and medial hamstring (MH). At least five walking trials were completed where subjects ambulated at their self-selected velocity. Three-dimensional motion and forces were measured. EMG signals were collected at 1000 Hz using an AMT-8™ (Bortec, Inc., Calgary, Alberta) EMG measurement system. All EMG waveforms were corrected for subject bias, full wave rectified, low pass filtered (Butterworth, zero lag, 6Hz), time normalized to represent 100% of the gait cycle and amplitude normalized to the respective MVIC activity.

PCA, a multivariate statistical technique was employed (Hubley-Kozey et al. 2006) to extract the waveform features from each muscle group that explained the variance in the overall magnitude of the EMG signal. *PC-Scores* were computed for each original EMG waveform. Analysis of variance models were used to test for group and muscle main effects and interactions. Bonferonni post hoc adjustments were used to test significance at alpha = 0.05.

RESULTS AND DISCUSSION

Demographic and spatial/temporal gait characteristics are shown in table 1. For each muscle group, principal component (pattern) one (PC1) captured the overall magnitude of the waveforms, explaining 63%, 70% and 56% of the variance in the calf, quadriceps and hamstring waveforms respectively.

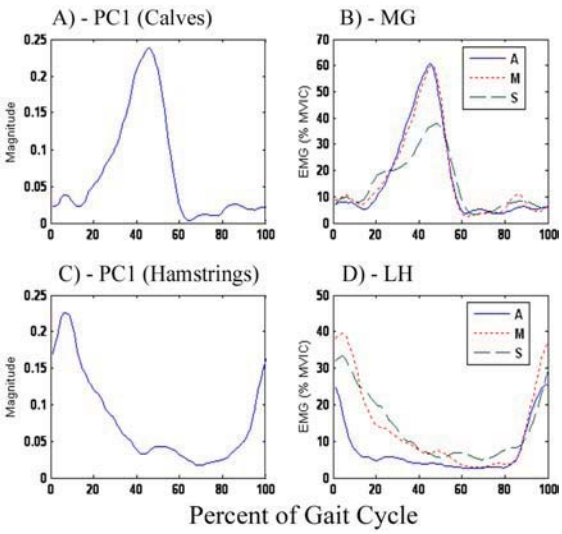

Figure 1: PC1 and ensemble averaged waveforms for MG (A&B) and LH (C&D).

For the ASY group, the MG magnitude was greater than the LG magnitude (p=0.01). This was not found in the MOA (p=0.08) or SOA (p=1.00) group. No group differences were found in the overall magnitude of the LG (p=1.00). The magnitude of the MG was significantly reduced in the SOA group compared to ASY and MOA (p<0.05). This difference is mainly evident during propulsion, a mechanism by which individuals with severe knee OA may reduce medial joint loading despite maintaining a similar velocity (Figure 1-B)).

There were no group differences in the magnitude of VL, VM and RF (p>0.05).

The magnitude of LH was greater than the MH for all groups (p<0.05). The magnitude of the LH and MH was greater for the MOA and SOA groups compared to the ASY group (p<0.001) (Figure 1-D). These findings are consistent with previous literature suggesting that antagonist hamstring activation is occurring during the gait cycle in those with knee OA. These findings also suggest, that the lateral muscles are activated to a greater degree than medial muscles, a possible mechanism to reduce medial joint loading.

SUMMARY/CONCLUSION

This study provides evidence that differences in the magnitude of myoelectric activity during gait occur based on the presence and severity of knee OA and that these differences are occurring independent of gait velocity. Overall, the increased magnitude of LH in both groups with knee OA and a reduction in MG activity unique to SOA subjects are consistent with a strategy to reduce medial knee joint loading.

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ACKNOWLEDGEMENTS

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Table 1: Demographics and Spatial Temporal Gait Characteristics (Mean ± SD)

Group	Age (years)	BMI (kg/m^2)	Velocity (m/s)	Stride Length (m)
ASY	56 ± 5*	25.1 ± 3.5*	1.19 ± 0.06	1.33 ± 0.08
MOA	64 ± 7	31.1 ± 3.9	1.18 ± 0.09	1.35 ± 0.09
SOA	67 ± 9	31.0 ± 4.6	1.14 ± 0.12	1.33 ± 0.11

[* (ASY) < (MOA) & (SOA), p < 0.05]

A NOVEL APPROACH FOR MYOELECTRIC POWERED PROESTHESIS: SIMULTANEOUS AND PROPORTIONAL CONTROL OF MULTI-DOF

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INTRODUCTION

The electromyogram (EMG) recorded at the skin surface contains important information regarding the control strategy of the neuromuscular system, and has been used as a control source of powered prosthetic devices for several decades. The state-of-the-art prosthetic devices controlled by surface EMG employ pattern recognition techniques to map the activities of multi-channel surface EMG to a number of contraction types. The main limitation of this framework is that only one pattern from a set of mutually exclusive patterns can be classified in one decision. In contrast, natural movements of limbs are the result of proportional and simultaneous activations of multiple degrees of freedom (DOF). In this paper, a novel approach is proposed to achieve proportional and simultaneous control of multiple DOF by surface EMG.

A GENERATIVE MODEL OF EMG

In a previous paper [1], a generative model of multi-channel surface EMG is proposed, which is illustrated in Fig. 1. In this model, it is assumed that the activation of each DOF is controlled by a neural primitive, $p_i(t)$. When a certain movement is intended, the primitives are mixed according to the musculature of the joint (mixing matrix S), generating the neural drives to all the muscles involved in the movement, $X(t)$. The activities of the muscles, $Y(t)$, are further mixed due to the volume conductor effect of adjacent myoelectric channels (channel crosstalk), and the resulting signal is the multi-channel surface EMG, $Z(t)$. This model is supported by two well known properties of neuromuscular system: muscle synergy [2] and motor unit common drive [3].

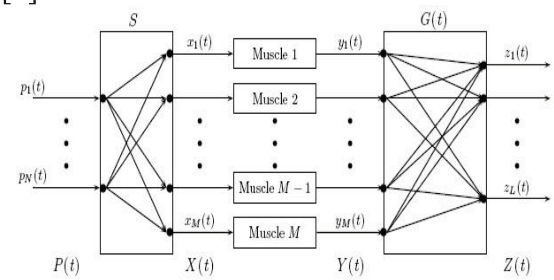

Figure 1. The generative model of multi-channel surface EMG.

It is shown in a simulation study [1], that an estimation of the primitives, $P(t)$, can be obtained from the mean square value (MSV) of $Z(t)$, using an artificial neural network (ANN). In next two sections, the procedure of a preliminary experiment to test the model is discussed, and the results are presented.

METHODS

A six-axis Force/Torque sensor (ATI industrial type Gamma) to measure the force produced at three DOFs of wrist: flexion/extension, adduction/abduction, and supernation/pronation. Eight channels of surface EMG are recorded at the forearm. The experiment setup is shown in Fig. 2. The forces produced and the EMGs collected are displayed in real time as feedback to the subject, who is instructed to perform the following contractions: isotonic contractions at each DOF without activating

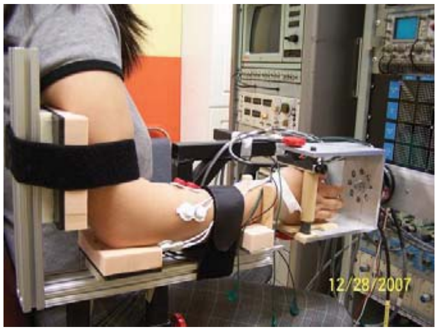

Figure 2. The experiment setup.

other DOFs intentionally; dynamic contractions of each DOF without activating other DOFs intentionally; dynamic contractions of combination of two or three DOFs. A two-layer backprop. ANN is used to estimate the target, *i.e.* the force vector, using the MSVs of the eight-channel surface EMG. The data with isotonic contractions at separate DOFs are used as the training set; dynamic contractions with separate DOFs are used as the validation set; dynamic contractions with combined DOFs are used as the testing sets. The performance of the ANN is presented in Fig. 3, where the estimation of the forces by the ANN is plotted against the measured forces. As a measure of the accuracy of the estimation, the cross-correlation coefficients (CC) of the estimated and the measured forces are

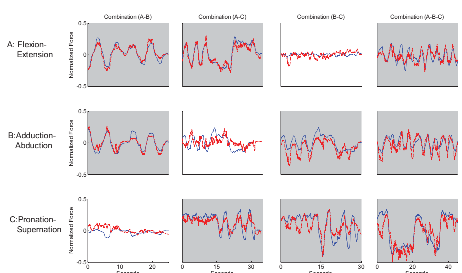

Figure 3. The performances of the ANN. Each row corresponds to one of the three DOFs; each column corresponds to one data set, with 2 or 3 DOFs activated simultaneously (indicated by the shaded plots). The blue solid line in each plot is the measured force, and the red dashed line is the estimated force by the ANN. All horizontal axes are in seconds, and the units of all vertical axes are normalized force.

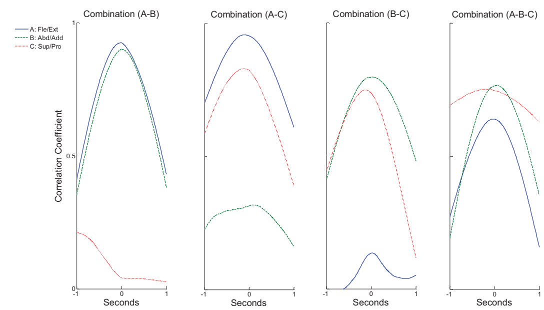

Figure 4. The CC of the estimation and measured forces at all the three DOFs of the four testing data sets.

calculated for each testing data set, and are plotted in Fig. 4. As shown in the figure, the CCs of the DOFs that are intended to be activated are consistently high in all data sets. On the other hand, when a DOF is not intended to be activated, the corresponding CC is low.

CONCLUSIONS

A novel approach toward simultaneous and proportional control of multiple DOFs using multi-channel surface EMG is proposed. The forces produced at wrist in three DOF are estimated from the surface multi-channel EMG recorded at forearm by an ANN. The results are very promising, and further experiments are underway.

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EFFECTS OF UPPER EXTREMITY RESTRICTION ON TRUNK AND PELVIS KINEMATICS WHILE ASCENDING AND DESCENDING STAIRS

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INTRODUCTION

Until recently, the focus of climbing stairs biomechanics research was on normal kinematics patterns of lower extremities with arm swing (Chapdelaine S et al, 2005; Nadeau S et al, 2003; Stacoff A et al, 2003). To date, none of the studies have examined the effects of upper extremity (UE) restriction on trunk and pelvis kinematics during ascending and descending stairs. The purpose of this study was to analyze pelvis and trunk kinematics during ascending and descending stairs with dominant UE restriction in healthy controls.

We hypothesize that having an UE restricted would limit arm swing and cause decreased trunk rotation during gait while ascending and descending the stairs. Decreased trunk rotation would cause decreased pelvic range of motion, resulting in a decrement in normal balance reactions.

METHODS

Ten healthy subjects without upper or lower extremity impairments (5 men, 5 women, and age ranging from 20 to 35 years) were enrolled in this study.

The subjects were instructed to ascend and descend the stairs step over step 3 times at their normal pace in the unconstraint condition; then the shoulder immobilizer placed with the dominant upper extremity in 90 degrees elbow flexion with the shoulder internally rotated and arm across the abdomen using wrist and humeral cuffs.

The subjects then performed the three stair trials again with their dominate arm in the immobilizer. Kinematics of the trunk, pelvis, and limb segments were recorded using an 8-camera VICON M2 system. An Euler angle sequence (Wu et al, 1995) was used to derive the three dimensional joint motions of the trunk and pelvis (Figure 1).

A paired-T test (SPSS 15) was used to examine effect of UE restriction on kinematics of trunk and pelvis during ascending and descending stairs at significance level $P<0.05$.

RESULTS AND DISCUSSION

Unconstrained trials demonstrated pelvic obliquity and pelvic rotation towards the dominant side. These trials also revealed trunk extension during the gait cycle, as well as lateral side bending and rotation of the trunk towards the dominant upper extremity. The constrained trials demonstrated statistically significant ($P<0.05$) a decrease in total joint range of motion in the pelvis and trunk, as well as a decrease in joint angles during ascent and descent in comparison to unconstrained trials (Table 1).

From a clinical standpoint, this study presents a normal standard of gait for the pelvis and trunk while ascending and descending stairs for this age group, with dominant upper extremity restriction. This will assist clinicians to identify situations in which a patient with dominant upper extremity restriction may require an assistive device when ascending and descending stairs. Clinicians will also recognize when to obtain information about the patient's home such as a side rail on a staircase, or an individual to assist them. These results can help us in understanding how to facilitate an individual to return to a normal gait pattern.

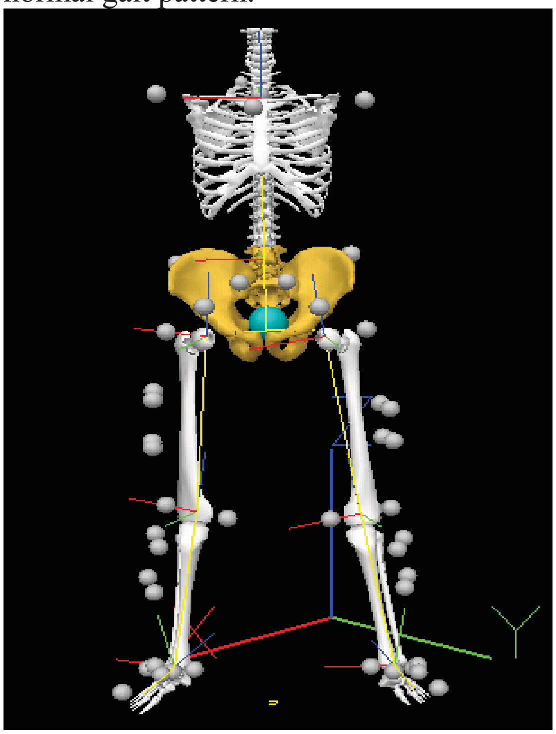

Figure 1: This graph illustrates static musculoskeletal model of lower extremity

SUMMARY/CONCLUSIONS

Our results give a greater understanding of the biomechanical changes imposed on the trunk and pelvis during upper extremity restriction while ascending and descending stairs. Our findings show a reduction in range of motion and a return to neutral in both the trunk and pelvis with upper extremity restriction.

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Table 1: Kinematics of Pelvis and Trunk during Ascending Stairs

Side	**Unconstrained Side**		**Constrained Side**	
Variables (Degree)	**Mean**	**SD**	**Mean**	**SD**
Total Joint Range of Motion Pelvic Obliquity: *	11.183745	4.592232148	9.4304784	2.918114968
Total Joint Range of Motion Anterior/Posterior Rotation: *	5.8592515	2.337510813	4.7640094	1.37479844
Total Joint Range of Motion Lateral Side bending: *	11.060535	4.648037532	9.3756195	2.962730293
Total Joint Range of Motion Trunk Rotation: *	5.6481347	1.980068784	4.3751128	1.078212441

* Significant change between conditions $p<.05$.

ASHWORTH SCORE IS RELATED TO INTRINSIC MUSCLE PROPERTIES

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INTRODUCTION

The Ashworth score (AS) is a clinical measure of joint mechanical resistance, and a common assessment in movement disorders. Besides its attractiveness of simplicity, the relation of the AS to underlying biomechanical properties is unknown. Resistance originates from different passive tissues and/or active responses of the muscle. The latter might reside from increased muscle tone or responding reflex activity. Separation between passive and active contributions enhances significance of the AS and facilitates focused treatment.
The goal of this study is to quantify passive and active reflex contributions of the human ankle joint and relate these to the AS. Accurate estimation of properties was achieved from precise measurements using a haptic robot manipulator and model optimization techniques. With this approach, we emulated the joint rotation trajectories as applied manually during the Ashworth test. Our results indicate that intrinsic muscle stiffness is the only parameter that has high correlation the AS in stroke patients.

METHODS

19 stroke patients (<2 yrs post stroke) and 15 control subjects participated. Ramp and hold ankle rotations were randomly applied: four durations (0.25, 0.5, 1.0 and 2.0 s) and three different amplitudes (33, 66 and 100% of ROM). A slow (120 s) dorsiflexion (DF) torque ramp was applied initially. The angle at the final torque of 15Nm was taken as the max DF angle. The ROM was assigned the difference between the max DF angle and the angle at rest. A single axis robot (Fig. 1) rotated the ankle from three different starting positions. Movement ending was at max DF angle in all cases. From the recorded reaction torque, angle and EMG signals, the parameters of a nonlinear model were estimated. Joint angle and EMG were input to the model. Model parameters were estimated by minimizing the squared error of predicted and measured torque.

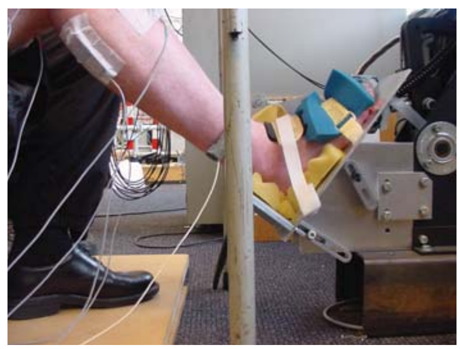

Figure 1: Experimental setup

Intrinsic modeled torque, T_I equals:

$$T_I = I\ddot{\theta} + B\dot{\theta} + K(\theta^*)\theta^*$$

I: foot inertia, B: viscous damping, $\theta^* = \theta - \theta_0$ and:

$$K(\theta^*) = H_{REL} k_1 e^{k_2 \theta^*}$$

the nonlinear stiffness. The amount of force relaxation $(1+k_{rel})^{-1}$ was implemented as a 1st order filter (τ_{rel}):

$$H_{REL} = \frac{\tau_{rel} s + 1}{\tau_{rel} s + 1 + k_{rel}}$$

where s denotes the Laplace operator. The reflexive torque T_R was modeled as:

$$T_R = H_A\left(e_1E_1 + e_2E_2 + e_3E_3 + e_4E_4\right)$$

with $[E_1,\ldots E_4]$ the rectified EMG signals from TA, GM, SL and GL muscles respectively, $[e_1,\ldots e_4]$ are scaling factors and H_A represents a second order filter (critically damped at 2.2 Hz) to describe calcium dynamics. The total modeled equals: $T_{MOD} = T_I + T_R$ and the parameter vector to be estimated was:

$p = [I, B, K_1, K_2, \theta_0, e_1, e_2, e_3, e_{4,}\ \tau_{rel}, k_{rel}]$

RESULTS AND DISCUSSION

Figure 2 shows the parameter reliability: I_a, B and K_1 had least contribution to the torque. A typical model fit is shown in

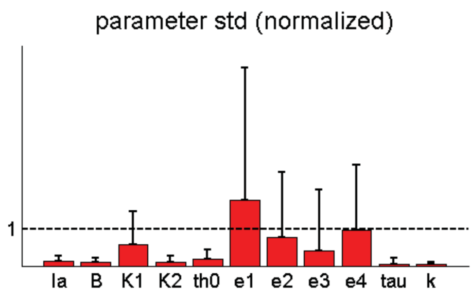

Figure 2: Standard deviation of all parameter fits. Values < 1 correspond to high parameter convergence and relevance.

Figure 3A. Torque Variance Accounted For was larger than 95% in all cases. The exponential stiffness term K_2 showed the most pronounced correlation to the AS (Fig. 3B). ROM co-varied with K_2. Other parameters, or combinations, did not show a clear relationship to the AS or where different between patients and controls.

SUMMARY AND CONCLUSIONS

Stiffness increases with Ashworth score. Reflexive torque was not different between patients and controls. These results can be

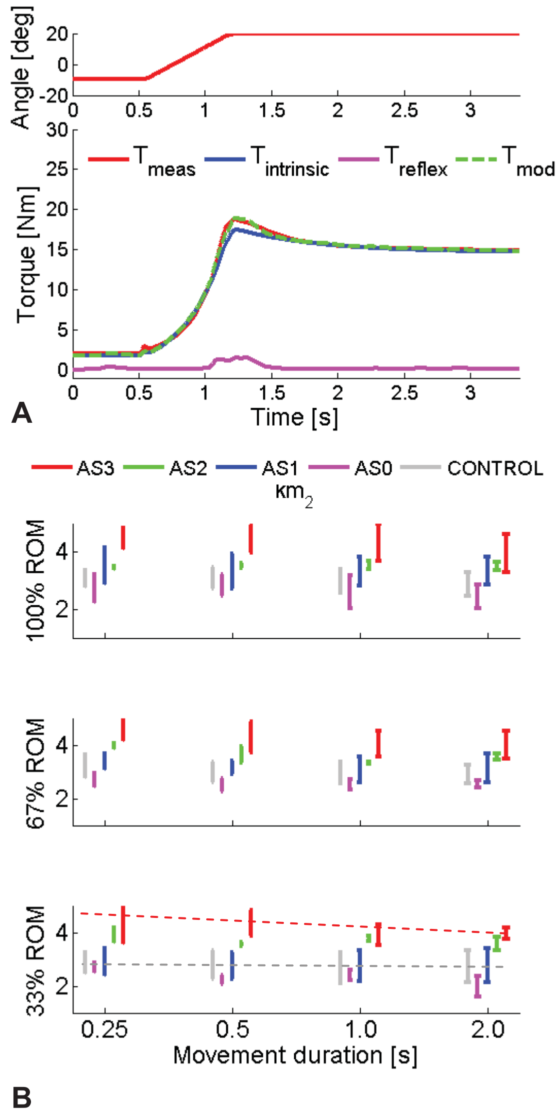

Figure 3: A, Typical example of model fit into its torque components; B, intrinsic stiffness term $\boldsymbol{K_2}$ for all conditions and Ashworth scores (AS0-AS3).

explained from increased stiffness of connecting tissue and/or tendons. A slight velocity effect was also observed (dotted lines) in the stiffness (Fig. 3B) which may indicate to increased tonus, perhaps from reduced cross-bridge turnover.

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AN ELECTROMYOGRAPHIC EVALUATION OF MUSCLE SYMMETRY IN NORMO-OCCLUSION SUBJECTS

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INTRODUCTION

Assessments of morphological symmetry craniofacial have become a part of the usual characterization of both healthy subjects and in patients (Ferrario et al., 1994). The evaluation of functional symmetry of the complex craniofacial usually involves the patterns of movements in the jaw and the activities of the masticatory muscles (Naeije, McCarroll & Weijs, 1989; Ferrario, Sforza & Serrão, 2000). Patterns of contraction of pairs of muscles can be investigated using surface electromyography (EMG), which enables the monitoring of some of the major masticatory muscles (masseter and temporalis). The aim of this study was to analyze the possible existence of asymmetries between the muscles masseters and temporalis anterior on both sides in subjects with normal occlusion and relate the EMG findings of asymmetry with data from the clinical evaluation of the stomatognathic system.

METHODS

Thirty subjects (13 men and 17 women) with an average of 20.77 years, healthy, with normal occlusion, complete and permanent healthy teeth including second molars (at least 28 teeth), with molar and canine into bilateral Angle class I (+ /-1 mm). Data obtained through the following questionnaires: 1. RDC / TMD (Dworkin, LeResche, 1992), 2. Analysis Oclusal (Sheet FORP / USP), 3. Signs and Symptoms (Felício), 4. Fonseca´s Questionnaire (1994). For the electromyographic examination was used Freely EMG (DeGötzen, srl, Milan Italy) and disposable silver/silver chloride bipolar surface electrode (Duo - Trode, Myo-Tronics, Inc.). In order to reduce skin impedance, the skin was carefully cleaned prior to electrodes placement, and recordings were performed 5–6 min later, allowing the conductive paste to adequately moisten the skin surface. Three tests were recording: Test 1: Maximum Voluntary Clench (MVC) with cotton-roll (standardization); Test 2: MVC without cotton-roll (test where the dental relationship is considered), and Test 3: Alternate ‘maximum’ voluntary contractions and relaxations with a 1 Hz frequency (dynamic clench–relax test). The 3 seconds intermediaries of electromyographic waves of the pairs of muscles were compared computing the Porcentage Overlapping Coefficient (POC), Torque Coefficient (TC), Asymmetry Index (ASSIM), Activation Index (ATTIV) and Impact (IMPACT).

RESULTS AND DISCUSSION

EMG potentials of the four analyzed muscles as per cent of maximum voluntary clench on cotton rolls (unit: mV/mV x 100) recorded in the two 3-s tests, as well as the average muscular potentials are reported in Table 1. In the questionnaire applied research of the severity of the signs and symptoms of DTM, the results are in line with expectations for the young and healthy asymptomatic subjects. The index proposed in this work (POC), to quantify asymmetry, has advantages in relation to the index proposed by NAEIJE et al. (1989) and used in many works, because it analyzes the EMG wave whole and not just averages, and better employee at work where there are large EMG variations (FERRARIO et. al., 2000). The subjects of the search have undergone clinical evaluation fonoaudiologyc and for the confirmation of inclusion criteria, which brings the reliability of the sample, resulting in more reliable data. The data found was expected because of the other work (FERRARIO et. al., 2000), in different populations had average values very similar to the ones found in this study. Despite of some aspects of the functions themselves are changed, these changes do not appear to relate to a framework for DTM. Such changes should be probably to other problems such as the oral breath and open bite, aspects observed in the sample evaluated.

SUMMARY/CONCLUSIONS

The results showed that the subjects young adults evaluated had average values of asymmetry within the standards of normality already established for other people. Probably figures are also valid for the Brazilian population, but many other studies with samples should be conducted. Such data could assist in the diagnosis of patients with some type of temporomandibular dysfunction.

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Table 1 and 2: EMG mean potentials as per cent of MVC on cotton rolls, and EMG indices, in 30 healthy young adults (mean ± SD). Porcentage Overlapping Coefficient (POC), Torque Coefficient (TC), Immediate Torque (ITC), Asymmetry Index (ASSIM), Activation Index (ATTIV) and Impact (IMPACT).temp: temporalis muscle; mm: masseter muscle.

	Electromyographic results			
Group	**POC temp**	**POC mm**	**POC mean**	**TC**
Health	86,96 ± 3,20%	86,51 ± 3,33%	86,75 ± 2,48%	8,60 ± 1,08%

	Electromyographic results			
Group	**Assim**	**Attiv**	**Impact**	**ITC**
Health	3,39 ± 6,55%	-2,30 ± 11,84%	101,53 ± 32,15%	0,62 ± 4,11%

APPLICATIONS OF PROTOCOL OF TMD TREATMENT WITH OCCLUSAL SPLINT AND ELECTROMYOGRAPHIC EVALUATION

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INTRODUCTION

Surface electromyography (EMG) can currently be considered a very useful instrument which allows a quantitative assessment of masticatory muscles in patients with temporomandibular dysfunction (TMD) (FERRARIO et al. 2000). The purposes of this study were: to correlate the clinical assessment data before and after the treatment with an occlusal splint for a group patients with TMD, classified according to the RDC / TMD; to compare the results obtained with EMG, before and after treatment with an occlusal splint; to compare the EMG results for this group suffering from TMD and an asymptomatic control group.

METHODS

The electromyographic examination (DeGötzen, srl, Milan Italy) of the masseter (M) and the anterior temporal (AT) muscles was carried out in the first assessment session (Phase 1), after one week (Phase 2) and after five weeks (Phase 3) of treatment, aiming at verifying the stability of the splint and the evolution of the muscular activity in 15 normo-ocllusion subjects (Control Group) and 15 patients (TMD Group). The EMG waves were analyzed using the software, and the following EMG indices were calculated: percentage overlapping coefficient (POC) of the M and TA muscles; torque coefficient (TORS); asymmetry index (ASIM); activity index (ATTIV) and the total electrical activity (IMP). For data expressed at measurement interval levels, nonparametric statistics were adopted, using the Wilcoxon test for the paired data in the intra-group analysis (among the phases). Data at ratio level were analyzed through parametric statistical means: paired data t test for intra-group analysis, independent sample t test for among-group analysis. The significance level was established at 5%.

Figure 1: FARC splint.

RESULTS AND DISCUSSION

After treatment, a statistical significance was found in mouth opening, as well as in the remission in the pain at palpation of a significant portion of the assessed muscles and the TMJ. A significant difference was obtained for the masseter POC and IMP, immediately after the first splint adjustment. When comparing the first phase, without the splint, to the second phase, with the first splint adjustment, a significant difference was observed in the values for masseter POC, ASIM and IMP. There was a significant difference between phase 1, without the splint, and phase 3, with the adjusted splint, for M and TA POC values, ASIM, ATTIV, and IMP. Throughout the entire treatment, there were no significant differences concerning the EMG index in the examinations performed without the splint. There was a statistical significance between the TMD and the control groups in the beginning and in the end of the treatment, with significant differences observed in POC values for both muscles and ATTIV.

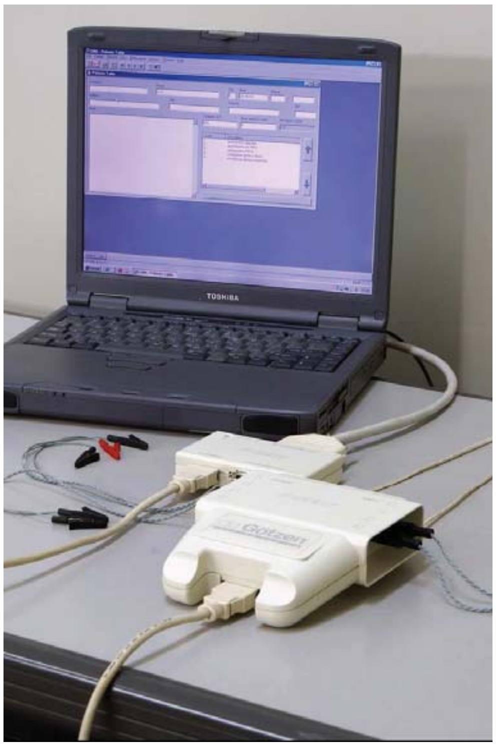

Figure 2: Freely EMG apparatus.

SUMMARY/CONCLUSIONS

The occlusal splint, without provoking permanent changes, proved effective to promote the balance of electromyographic activities during its use, and efficient in relieving the symptoms. The EMG parameters allow its scientific use in identifying neuromuscular unbalance, and as such, this assessment tool allowed an objective analysis and evaluation of the different phases of the traditional treatment for TMD in dentistry, differentiating patients with TMD and asymptomatic individuals.

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MASTICATORY EFFICIENCY IN TMD INDIVIDUALS BEFORE AND AFTER ACUPUNCTURE TREATMENT

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INTRODUCTION

Acupuncture is a therapeutic method which consists of the insertion of small, solid needles, usually made of stainless steel, into specific body points in order to improve health or modify pain states. This technique is shown to be effective in approximately 60-75% of patients with chronic pain.
Aware of the importance that masticatory muscles exercises on the various functions of the stomatognathic system, this study was done with the purpose of analyze the efficiency of masticatory cycles (courses) of the muscles temporal and masseter, right and left (RT; LT; RM; LM).

METHODS

Eighteen subjects with temporomandibular disorders (TMD) were selected to participate in this research before and after therapy with ten sessions of acupuncture conducted weekly, using the electromyographical analysis during mastication.
The points of needling the acupuncture were IG4, E6, E7, B2, VB14, VB20, ID18, ID19, F3, E36, VB34, E44, R3, HN3. This analysis was performed using a MyoSystem-BR1 electromyographer with differential active electrodes (silver bars 10 mm apart, 10 mm long, 2 mm wide, 20x gain, input impedance 10 GΩ and 130 dB common mode rejection ratio). Surface differential active electrodes were placed on the skin, previously cleaned with alcohol, bilaterally on both masseter muscles and on the anterior portion of the temporalis. A ground electrode was also used and fixed on the skin over the sternum region.
The electromyographic signals were analogically amplified with a gain of 1000x, filtered by a pass-band of 0.01-1.5KHz and sampled by a 12-bit A/D converter with a 2 KHz sampling rate. The signals were digitally filtered by a pass-band filter of 10 to 500 Hz in the data processing.
The analysis was collected during Parafilm mastication (10 seconds, three times). The averages of the data were normalized by maximum voluntary contraction (MVC) for four seconds, and the results were statistically analyzed using the independent t-test (SPSS- 12.0- Chicago) during the comparison between groups ($p<0.05$).

RESULTS AND DISCUSSION

There was increased electromyographical activity in mastication: averages before acupuncture RT = 1,08 ± 0,25; LT = 1,38 ± 0,35; RM= 1,36 ± 0,53; LM = 1,09 ± 0,28; and after the therapy RT= 1,37 ± 0,29; LT = 1,56 ± 0,29; RM= 1,41 ± 0,29; LM = 1,19 ± 0,22.

CONCLUSIONS

According to the methodology employed, in experimental conditions described and based on results, it seems reasonable to conclude that occurred more efficiency of masticatory cycles (courses) of the stomatognathic system of subjects with temporomandibular disorder after the acupuncture therapy.

The normalized electromyographic masticatory efficiency data obtained showed a tendency of being greater after the acupuncture treatment.

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EFFECT OF CONGENITAL BLINDNESS ON EMG ACTIVITY OF THE FACIAL MUSCLES

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INTRODUCTION

Blindness is the deprivation or loss (partial or total, transitory or permanent) of vision. The causes of blindness are diverse, from ocular traumas to infection to congenital illnesses. Individuals with congenital visual impairment are seriously deprived of stimulation and therefore subjected to significant alterations in physical (muscular), mental, and social development. Though not being able to observe facial expressions, blind individuals experience alterations in skin musculature. This work aimed to analyze the muscle activity in the orbiculris oris, the orbicularis oculi and frontalis muscles in blind and clinically normal individuals.

METHODS

28 individuals between the ages of 18 and 28 (average age of 21.87 ± 6.55, average weight of 55.29 ± 11.95 Kg and average height of 1.60 ± 0.10 meters) were selected to participate in this study, and were divided two groups. Group 1 (blind) consisted of 14 blind individuals from birth, able to breathe through the nose and with well-formed lips, Angle Class I, chosen from the Ribeirao Preto Blind Persons Association. Group 2 was made up of 14 healthy volunteers, without visual problems. They were matched to Group 1 subjects by gender, age, weight and height. Electromyographic (EMG) recordings were made using the Myosystem EMG System with features including: simultaneous acquisition; common grounding for all channels; low pass filters of 10 Hz to 5KHz; 10 GΩ impedance in differential mode; 12 bits dynamic resolution; amplitude range -10V to +10V; and sample frequency of 2Hkz per channel. For visualization and processing of signals, the Myosystem I version 2.29 program was used, which also allowed for the analog signals to be amplified with gain of 1000x. They were filtered through a bandpass filter of 0.01-1.5kHz.
All subjects performed three trials of functional tasks. For the orbicularis oris muscle the activity were rest, blowing, labial projection and labial compression. For the orbicularis oculi muscles the activities were rest, blinking and forced blinking. For the frontalis muscles the activities were rest, bringing eyebrows closer and raising eyebrows. All data were collected during 10 seconds and the data were smoothed using root mean square (RMS) windows of 10 ms and were normalized to the activity recorded during maximal clenching of each muscle for four seconds. Statistical analysis was performed using t-tests, via SPSS version 12.0 (Chicago, IL).

RESULTS

The results for the orbicularis oris muscle demonstrated that the myoelectric activation was greater at rest among control subjects as compared to blind subjects detailed results are presented in Table 1. The EMG activity of the orbicularis oculi muscle showed no differences in muscle activity between the groups. Detailed results are presented in Table 2. For the frontalis muscle, mean EMG values were greater during rest activities for the control group as compared to the blind group. See Table 3 for details.

CONCLUSIONS

These data suggest the congenital blindness may affect resting muscle activity in the muscles controlling eye movements. The lack of ability to observe others may deprive blind individuals of visual clues which make it possible to learn facial expressions.

Table 1- Electromyographic activity in Group 1 (Blind) and Group 2 (Control) of the orbicularis oris muscle, upper (UOO) and lower (LOO) muscles.

Conditions	Group 1 - Blind		Group 2 - Control		*t* - test
	Average	S E	Average	SE	"p"
Rest					
UOO	0.0881	±0.0259	0.3008	±0.0627	0.004**
LOO	0.1003	±0.0208	0.3055	±0.0759	0.015*
Blowing					
UOO	0.5062	±0.1072	0.4944	±0.0949	0.935
LOO	0.6363	±0.1745	0.6833	±0.1201	0.826
Labial Projection					
UOO	1.5220	±0.2977	1.3564	±0.3315	0.713
LOO	1.4965	±0.3350	2.1830	±0.4073	0.204
Labial Compression					
UOO	0.9765	±0.0501	1.0237	±0.0803	0.622
LOO	1.2539	±0.1522	1.1484	±0.1281	0.601

Table 2- Electromyographic activity in Group 1 (Blind) and Group 2 (Control) of the orbicularis oculi, right (ROO) and left (LOO) sides.

Conditions	Group 1 - Blind		Group 2 - Control		*t* - Test
	Average	SE	Average	SE	"p"
Rest					
ROO	0.2464	±0.0359	0.3335	±0.0707	0.282
LOO	0.3076	±0.0523	0.3444	±0.1008	0.749
Blinking					
ROO	0.2745	±0.0430	0.4999	±0.1071	0.062
LOO	0.3289	±0.0542	0.4165	±0.0661	0.315
Forced Blinking					
ROO	1.0717	±0.0833	0.9519	±0.0703	0.282
LOO	1.0257	±0.0564	1.0550	±0.0947	0.793

Table 3- Electromyographic activity in Group 1 (Blind) and Group 2 (Control) of frontalis muscles, right side (RFM) and left (LFM).

Conditions	Group 1 - Blind		Group 2 - Control		*t* - Test
	Average	SE	Average	SE	"p"
Rest					
RFM	0.3217	±0.0455	0.3319	±0.0579	0.023*
LFM	0.1713	±0.0421	0.1781	±0.0466	0.049*
Bringing eyebrows closer					
RFM	0.5779	±0.0579	0.6279	±0.0745	0.968
LFM	0.5831	±0.1157	0.5415	±0.1293	0.567
Raising eyebrows					
RFM	1.0283	±0.0639	1.0095	±0.0529	0.823
LFM	1.0423	±0.0572	1.0124	±0.0640	0.730

PARAMETERS OF MULTISEGMENTAL MONOSYNAPTIC RESPONSES IN A VARIETY OF LEG MUSCLES IN PATIENTS WITH A RADICULOPATHY

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INTRODUCTION

In patients with marked lumbar radiculopathy and healthy people there were investigated reflex motor responses of some leg muscles bilaterally (m. biceps femoris, m. medial head of the gastrocnemius, m. soleus, m. tibialis anterior) evoked percutaneously by electric stimulation of the spinal cord at the L2-L3-L4 level in the condition of rest. The used technique to elicit multisegmental monosynaptic responses (MMRs) of the specified muscles was similar to that suggested, described and used by a group of authors (G. Courtine, S.J. Harkema, Ch.J. Dy et al., 2007), with the only difference that in the previous studies the stimulation was carried out at the T11-T12 level. However, during our own research it was found out, that reception of motor responses of the muscles of the patients at stimulation at the T11-T12 level is not possible in all the cases even at 80 mA intensity of the stimulus. Furthermore, the stimulation at the L2-L3-L4 level is always accompanied by registration of the evoked responses, the main characteristics of which (the latent period, the threshold strength of current, the amplitude and the shape) allow to suggest that the responses have a similar origin and essence. It proved to be true the amplitudes of the muscle motor responses were suppressed when a conditioning stimulus was delivered 50 ms before the test stimulus.

All individuals gave voluntary written consent to participate in the investigation. Their mean age was 37.5, mean height was 1.82 m (range, 1.70-2.05 m) and mean body weight was 83 kg (range, 60-111 kg). The study was approved by the ethics committee of the University of Physical Education and Sports and conformed to the Declaration of Helsinki.

METHODS

Bipolar surface electrodes with an inter-electrode distance of 2 cm were mounted over 8 leg muscles bilaterally – on the muscle belly midway between the upper and the lower muscle insertion. We placed the cathode over the skin between L2-L3 or between L3-L4 spinous processes and two large anodes bilaterally over the anterior spine of the iliac crest. The optimum site of stimulation (L2-L3 or L3-L4) was located by a hand-held electrode. The site of stimulation was selected where the motor responses could be elicited in all the recorded muscles as symmetrically as possible. When the optimal site of stimulation was detected, the surface electrode was attached to the body.

RESULTS AND DISCUSSION

It was established that the latency of the evoked responses in all the subjects increased with the growing distance between the muscle and the stimulating electrode i.e. corresponded to the expected length of the motor nerve, consequently, it was longer in distal than in proximal muscles (Fig. 1). This observation is well coordinated with the results of the previous studies (G. Courtine, S.J. Harkema, Ch.J. Dy et al., 2007).

The present investigation shows that in patients with the marked radiculopathy some increase of MMRs latent period takes place in comparison with the healthy persons (Fig. 1), that testifies a decrease in speed of passage of an electric impulse in a monosynaptic nervous circuit in the background of the disease.

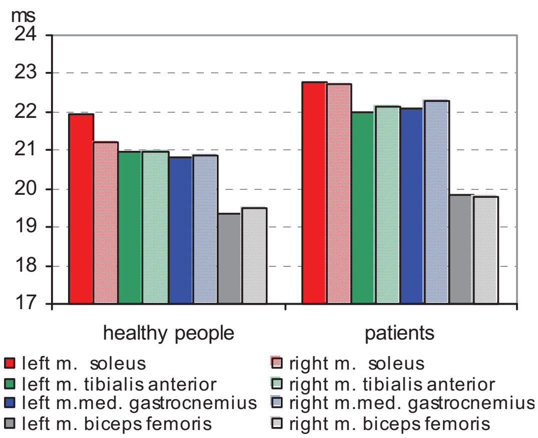

Figure 1: The latent period of MMRs.

The comparative analysis indicated that for all individuals with radiculopathy a significant increase ($p<0.05$) on the threshold of the evoked reflex motor responses takes place (Fig. 2). For example, in the researched group of the patients the average group size of the left muscle soleus threshold made 47.08 mA, and in the healthy individuals it was 32.14 mA (Fig. 2).

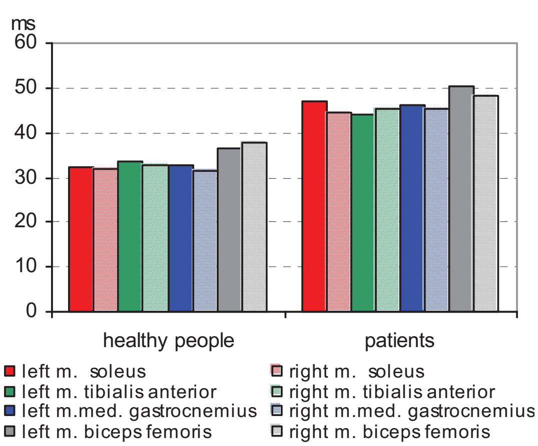

Figure 2: Size of thresholds of MMRs.

The MMR amplitudes of all the muscles studied significantly decreased ($p<0.05$) in the participants with lumbar radiculopathy. For example, the maximal amplitude of the MMRs left muscle soleus in the group of the patients was about 4.64 mV and in the healthy people it was 9.63 mV. It proves that in the background of the considered pathology there is a decrease of the α-motoneuron reflex excitability of both proximal and distal muscles. Besides, the received data allow to suppose that there is a disorder in excitation passing through nerve fibres and deficiency of nerve-muscular transfer in these patients. It is proved by the poly-phase irregular shaped MMRs of the most researched muscles in the patients (Fig. 3).

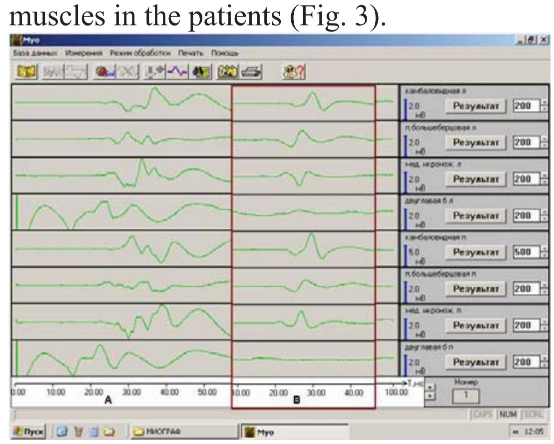

Figure 3: Shapes of MMRs in a patient with radiculopathy (A) and in a healthy subject (B).

CONCLUSIONS

The findings, therefore, allow to suggest a number of peculiarities of voluntary motor activity and, possibly, a change of inter and intra-muscular coordination in the background of the studied disease.

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“PATTERNIZING” STANDARDS OF SIT-TO-STAND MOVEMENT IN CEREBRAL PALSY WITH SPASTIC DIPLEGIA

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INTRODUCTION

Cerebral palsy movements are complex and highly individualistic. In a clinical setting, each physical therapist individually records these movements. However, accumulated data from this method of recording cannot be generalized easily since there is no established evaluation standard. Therefore, it is important to formulate a standard that can evaluate movements of cerebral palsy objectively. This research attempted to obtain objective evaluation index by utilizing “patternizing” standards of sit-to-stand (StS) movement of cerebral palsy with spastic diplegia (SD).

METHODS

Fifty-one children with SD (33 males and 18 females) aged 3 to 18 years old, (*M* 119.7±45.6 months) and ten healthy children (4 males and 6 females) aged 4 to 11 years old, (*M* 86.7±26.7 months) took part in the study. This research study was conducted after having obtained the approval of Tokyo Metropolitan University Research Ethics Committee (05055).

PROCEDURES

In the analysis, first, pictures of subjects' (SD and control group) StS movement were taken from the side with one digital video camera. StS movement procedures were as follows:(1) in the starting position, trunk was extended as straight as possible, and hands placed on knees; (2) sole of the feet to the floor; (3) in sitting posture, hip and knee flexion angle was about 90 degrees; (4) standing by holding onto a handrail; and (5) in standing posture, trunk and knees were extended as straight as possible. All the above procedures were performed in bare feet, and there was no time restriction. Each subject executed this StS movement three times; for the purpose of analysis, one steady movement was selected. Next, these scenes of StS movement were classified into two phases, and the state of the extremities was evaluated by 15 items (Table 1). Based on these 15 items, characteristic movements were identified and recorded by YES or NO. All recorded scenes of StS movement were kept on a personal computer to improve the reliability of analysis, and two dimension operation analysis software (Dual Stream: made by DKH company) was used.

STATISTICAL ANALYSIS

Using SPSS(version15), cluster analysis was conducted. Next, cross tabulation was executed while paying attention to the layered structure in order to understand each group that was classified by the cluster analysis according to the shared characteristics, and the relation between each evaluation item (15 items) and each group was examined by square test and Fisher's direct techniques. The statistical significant difference was assumed to be 5% or less.

RESULTS AND DISCUSSION

Subjects' StS movements were classified into six groups. One group consisted of ten healthy children, and five groups consisted of SD children (A group: n=17, B group: n=21, C group: n=4, D group: n=4, and E group: n=5). Statistical significant difference was observed in 7 out of the 15 items, between A.B and C.D.E groups. Specially, three items (4, 5 and 13) showed YES in A.B groups accounted for more than 70%; whereas the same items showed NO in C.D.E groups accounted for more than70%. That is to say, A.B groups have large forward trunk inclination movement, but C.D.E groups have little of such movement. Statistical significant difference was observed in 3 out of the 15 items between A and B group. One item (5) showed YES in A group accounted for 100%; the same item showed NO in B group accounted for 5%. We assume that the characteristic of A group has larger forward trunk movement; as those of B group has buttocks forward movement. Statistical significant difference was observed in 2 out of the 15 items between C and D.E groups. Two items (6 and 9) showed YES in C group accounted for more than 75%. We think that the characteristic of C group is upper extremities dependence in center of mass movement phase. Finally, statistical significant difference was observed in 2 out of the 15 items between D and E group. Two items (10 and 14) showed YES in D group accounted for 0%; the same two items showed YES in E group accounted for 100%. We think that the characteristic of D group is upper extremities dependence in standing movement phase, and those of E group is similar to healthy children.

SUMMARY/CONCLUSIONS

Utilizing "patternizing" standards of StS movement, SD subjects were able to be classified into five groups based on their shared characteristics of StS movement. We believe that these "patternizing" standards could correspond sufficiently to individuals with SD.

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Table 1: Frame of reference of StS movement

1	Head and trunk extended vertically to the floor.
2	Trunk tilted anteriorly while reaching for the handrail.
3	Trunk stopped after holding the handrail.
4	Trunk tilted more anteriorly after holding the handrail.
5	Acromion located forward from lateral malleolus when trunk tilted most anteriorly.
6	Buttocks moved forward from holding the handrail to lifting off the seat.
7	Knee joint moved forward from reaching the handrail to lifting off the seat.
8	Foot pulled backward while lifting off the seat.
9	Knee joint moved forward after lifting off the seat.
10	Lower thigh tilted anteriorly after lifting off the seat.
11	Bottom of foot bottom grounded completely on the floor after lifting off the seat.
12	Head flexed more than trunk after lifting off the seat.
13	Acromion located forward from lateral malleolus after lifting off the seat.
14	Knee joint moved backward in process to upright positioning.
15	Trunk and lower extremity extended vertically to the floor at standing position.

THE EFFECT OF MANDIBULAR OSTEOPOROSIS ON THE SOMATOGNATHIC SYSTEM: AN ELECTROMYOGRAPHIC ANALYSIS

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INTRODUCTION

Osteoporosis is a disease of bone leading to an increased risk of fracture. Any bone can be affected. In osteoporosis the bone mineral density is reduced, bone microarchitecture is disrupted, and the amount and variety of non-collagenous proteins in bone is altered. The osteoporosis affects trabecular bone volume and bone mineral density in the mandible (Hidebolt, 1997; Lindh, 1997). This disease is defined by the World Health Organization in women as a bone mineral density 2.5 standard deviations below peak bone mass (20-year-old sex-matched healthy person average) as measured by Dual energy X-ray absorptiometry. Osteoporosis is most common in women after menopause, when it is called postmenopausal osteoporosis (Samelson, Hannan, 2006), but may develop in men and premenopausal women in the presence of particular hormonal disorders and other chronic diseases, or as a result of smoking and medications, specifically glucocorticoids.

METHODS

This study compared the electromyographic activity of the right temporal (RT), left temporal (LT), right masseter (RM) and left masseter (LM) muscles during rest, dental clenching, chewing, and while maintaining postural movements (left laterality and right laterality), between 9 individuals with mandibular osteoporosis and 9 individuals without disease with ages ranging between 50 and 70 years. A twelve-channel Myosystem Br-1 electromyographer was used for electromyographic recording. Features included: simultaneous acquisition; common grounding for all channels; low pass filters of 10 Hz to 5KHz; 10 GΩ impedance in differential mode; 12 bits dynamic resolution; amplitude range -10V to +10V. (Figure 1). For visualization and processing of signals, the Myosystem I version 2.29 program was used, which also allowed for the signals to be amplified after digitalization, with gain of 1000x. All signals were filtered through a bandpass filter of 0.01-1.5kHz and sampled through a 12 bit A/D converter with 2kHz acquisition rate. Data were normalized by maximal voluntary contraction of the studied muscles and statistical analysis was executed by means of *t* test, utilizing "Statistical Package for the Social Sciences" software - SPSS - (Chicago, IL).

RESULTS

During maintaining rest, chewing and dental clenching, left laterality and right laterality there was a significant difference between individuals with and without osteoporosis($p \leq 0.05$). Individuals with mandibular osteoporosis showed lower electromyographic activity for all analyzed muscles when compared with individuals without disease.

CONCLUSIONS

The data allow us to conclude that mandibular osteoporosis is associated with reduced electromyographic activity of mastication muscles.

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system and more study is necessary to elucidate these alterations.
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ACKNOWLEDGEMENTS

Financial support FAPESP (06/53563-9)

Table 1: Normalized EMG averages in osteoporosis and control individuals. RM = right masseter, LM = left masseter, RT = right temporalis, LT = left temporalis

	Osteoporosis	Control
Rest		
RM	0,12 ± 0,04	0,24 ± 0,07
LM	0,24 ± 0,08	0,30 ± 0,08
RT	0,19 ± 0,02	0,28 ± 0,10
LT	0,15 ± 0,03	0,24 ± 0,06
Chewing		
RM	0,84 ± 0,18	1,01 ± 0,28
LM	0,78 ± 0,20	0,98 ± 0,23
RT	0,55 ± 0,07	1,12 ± 0,57
LT	0,58 ± 0,08	0,75 ± 0,20
Dental Clenching		
RM	1,38 ± 0,17	1,70 ± 0,45
LM	1,28 ± 0,13	1,61 ± 0,32
RT	1,06 ± 0,10	1,67 ± 0,69
LT	1,07 ± 0,06	1,20 ± 0,23
Left Laterality		
RM	0,20 ± 0,06	0,31 ± 0,09
LM	0,24 ± 0,08	0,29 ± 0,09
RT	0,15 ± 0,01	0,27 ± 0,10
LT	0,15 ± 0,02	0,29 ± 0,06
Right Laterality		
RM	0,14 ± 0,06	0,42 ± 0,14
LM	0,29 ± 0,09	0,34 ± 0,09
RT	0,17 ± 0,01	0,30 ± 0,10
LT	0,20 ± 0,07	0,27 ± 0,07

VARIATIONS IN ANKE PLANTAR FLEXOR FUNCTION IN CHILDREN WITH CEREBRAL PALSY: POSSIBLE IMPLICATIONS FOR TREATMENT OPTIONS

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INTRODUCTION

Botulinum Toxin A (Botox) injections are commonly used in the treatment of spasticity in children with cerebral palsy (CP). The functional outcomes, as assessed by motion analysis and other measures, however, are inconsistent and result in debate about the benefits of Botox for a temporary treatment of spasticity. The reason(s) for inconsistent outcomes are not well understood and as a result Botox is not considered for treatment of spasticity and associated functional problems when it may be a beneficial in certain persons.

A better understanding of how individual muscles function through the use of dynamic electromyography techniques (dEMG) may provide greater insight into how muscles function during movement and in response to quick stretch (during the assessment of muscle tone). In clinical practice, substantial assumptions are made with respect to which muscles contribute to abnormal tone or a spastic "response" (resistance to passive stretch) in the assessment of children with CP. For example, resistance to rapid passive stretch of the knee extensors (during a knee flexion motion) is often considered a result of "quadriceps" over activity. Dynamic EMG data shows that in many cases, that resistance to knee flexion may be a result of rectus femoris activity alone (Õunpuu, 1997). The ankle plantar flexors (medial and lateral head of the gastrocnemius) and soleus may function in the same way. As the plantar flexors are commonly targeted for Botox injection, to address excessive dynamic equinus and resulting toe walking, it would be beneficial to understand if these muscles respond similarly to rapid passive stretch and in walking. It would be reasonable to propose that if muscles are functioning differently, the Botox injection strategy should reflect this variation

METHODS

Dynamic EMG was collected using a 16-channel hardwire system (Motion Lab Systems, Baton Rouge, LA) from the medial and lateral heads of the gastrocnemius (MG and LG) and soleus (SOL) muscles for patients with CP as part of the standard of care for treatment decision-making for both ambulators and non-ambulators. Surface electrodes were used with an inter-electrode distance of 8 mm and were applied bilaterally. Dynamic EMG data was collected during tone assessment, voluntary contractions and gait (if possible). The qualitative assessment of differences in function between muscles was based on differences in activity and timing of the onset and termination of specific EMG signals in relation to others using the raw signal records from a minimum of 3 trials or 3 strides during gait. Cross-talk tests were performed as part of data collection to confirm appropriate electrode placement.

RESULTS AND DISCUSSION

A total of 20 patients (26 sides), with a mean age of 6 +/- 3 years were assessed. Differences were noted among patients in how muscles responded to stretch. In response to rapid passive stretch with the knee at 90°, clonus was noted in MG, LG and SOL in 20/26 sides. In 6 sides clonus was noted in the MG and LG only. In response to slow stretch with the knee at 90°, activity was noted in the MG, LG and SOL in 17/26 sides. In 9 sides, activity was noted in the MG alone or MG and LG. During full passive knee extension, activity was noted in the MG only in 14 of 18 sides.

Example recordings from a non-ambulatory patient, 2+5 years with CP spastic diplegia are included below.

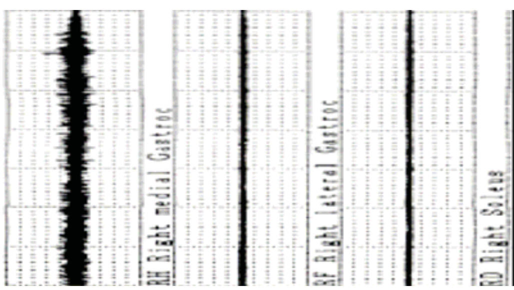

Figure 1: MG, LG and SOL during slow passive ankle dorsiflexion (knee 90 deg).

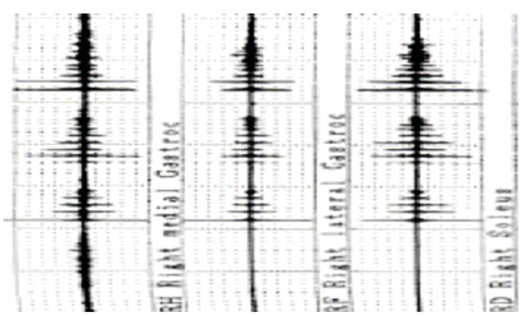

Figure 2: MG, LG and SOL during rapid passive ankle dorsiflexion (knee 90 deg).

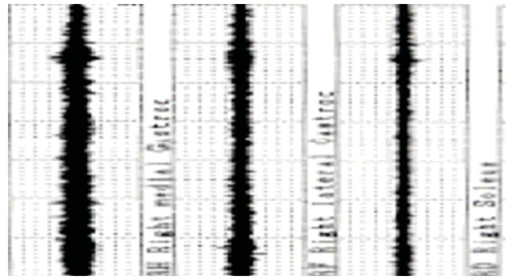

Figure 3: MG, LG and SOL during supported standing.

SUMMARY AND CONCLUSIONS

Preliminary clinical experience suggests that clinicians make assumptions about muscle function that may not be true. There is evidence based on EMG signals that the medial gastrocnemius is “more” spastic than that lateral gastrocnemius in some patients. There is also evidence that the soleus muscle is spastic in some patients and not others. Systematic differentiation of the relative contribution of these three muscles to abnormal tone and movement could lead to more specific injection protocols and ultimately to more consistent and effective treatment and outcomes with Botulinum Toxin A injection.

A major limitation in this preliminary effort was the qualitative evaluation of dEMG differences between muscles. Future efforts need to focus on developing a more viable means of objectively documenting and reporting these differences and ultimately evaluating outcomes of these more specific Botox injection strategies.

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SURFACE ELECTROMYOGRAPHIC STUDY OF MEIGE SYNDROME PATIENT: A CASE REPORT.

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INTRODUCTION

Some brain injuries result in movement disorders such as Meige syndrome (oromandibular dystonia with blepharospasm). This rare syndrome is characterized by orofacial dystonia, including blepharospasm and forced jaw opening, lip retraction, platysma muscle spasm, and tongue protrusion. It primarily affects older adults, with an incidence peak in the 7th decade of life. (Gorlin R.J. *et al*., 2001; Lee K.H., 2007). Facial movements are undoubtedly under the powerful influence of the cerebral cortex and are essential for the appropriate execution of many important functions such as mastication, swallowing, and social interaction, including speech and nonverbal communication (Morecraft R.J. *et al*., 2004).

The purpose of this study was to describe a qualitative evaluation of the surface electromyographic activity of orbicularis oris, masseter and anterior temporalis muscles in a Meige syndrome patient.

METHODS

A 75 year-old woman with a Meige syndrome diagnosed for 30 years was referred to this service for an oral function analysis.

The electromyographic (EMG) activity of the dystonic spasms in the orbicularis oris muscle was obtained by a couple of Beckmann circular passive electrodes of one centimeter diameter. Jaw muscles activity was registered by 05 passive electrodes Noraxon USA Inc. joined to pre-amplifiers. Recordings were made on 12 channels of simultaneous EMG signal acquisition equipment (Myosystem I / Datahominis Tec. Co.). The analog EMG signal recorded were digitized using 12 bit A/D converter at a sampling rate of 4KHz. After digitalization, the signal was filtered by a digital pass-band of 10-500Hz. Myosystem I software version 2.12 was used to visualize and process the EMG signal.

The tasks were: (1) mandibular rest position for 30 seconds; (2) maximal voluntary contraction (MVC) for 5 seconds and; (3) masticatory activity for 10 seconds.

RESULTS AND DISCUSSION

The patient was a 75 years old female, concording with the literature which describes a 2:1 female prevalence predilection.

Clinically, there were observed spontaneous, repetitive, nonrhythmic symmetric dystonic spasms, first involving the orbicularis oculi muscles and after affecting the muscles of the lower face and jaw.

During electromyographic examination in mandibular rest position, all of the investigated muscles: superficial masseter, anterior temporalis and orbicularis oris were involved in the spasms and their activity amplitudes were similar to masticatory cycle amplitude in the habitual mastication (Figure 1). The affected muscles demonstrated involuntary and unlike normal firing of muscle electrical activity, agreeing with what was described by Lee K.H. in 2007.

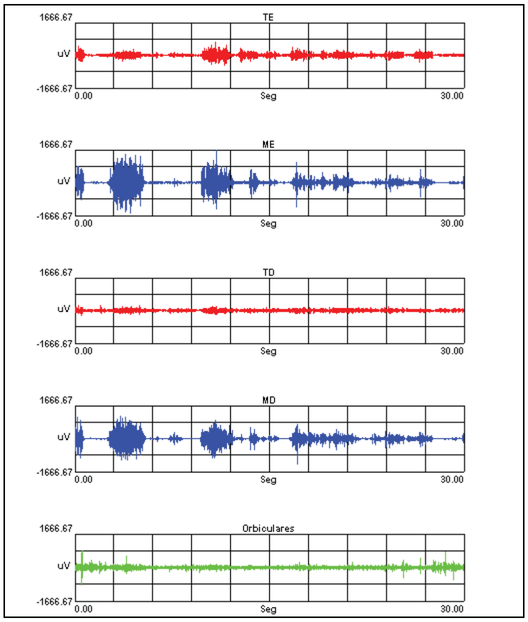

Figure 1: EMG recordings of the anterior temporalis, masseter and orbicularis oris muscles during mandibular rest position. Note the muscular spasms caused by the Meige Syndrome.
TE= left anterior temporalis; ME= left masseter; TD= right anterior temporalis; MD=right masseter; Orbiculares= orbicularis oris.

In habitual mastication, it was observed that the masticatory muscles evaluated acted in synchronism and satisfactory function with amplitudes within normal limits. Moreover, masseter presented higher activity in relation to temporalis muscles. The orbicularis oris muscle participated in the habitual mastication hyperactively.

At the end of the MVC task the masticatory muscles presented a decrease of the amplitude (Figure 2), probably because of the continuous activity observed in the mandibular rest position, caused by the syndrome.

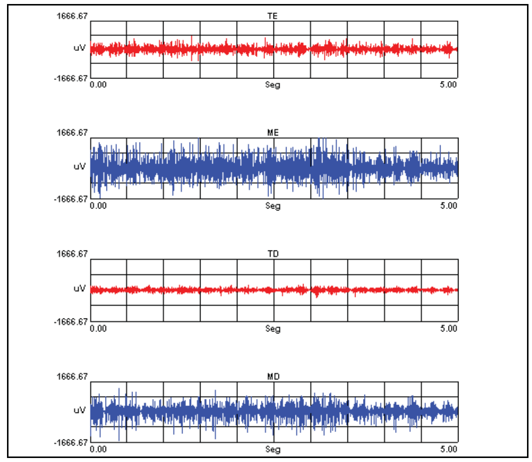

Figure 2: EMG activity during MVC from anterior temporalis and masseter muscles.

The treatment of this condition relies mainly on botulinum toxin injections (Blitzer A. *et al*., 1993). The patient responded excellently to this treatment, although she had to interrupt it because of financial issues. She is currently complaining about muscle exhaustion.

SUMMARY/CONCLUSIONS

1) Although the orofacial muscles exhibited the spasms, the masticatory function was not affected in this Meige´s syndrome patient;

2) Electromyography revealed to be an useful tool in this movement disorder evaluation.

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PAIN REDUCTION IMPROVES BALANCE CONFIDENCE IN PEOPLE WITH PERIPHERAL NEUROPATHY

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INTRODUCTION

Peripheral neuropathy (PN) is a growing epidemic that affects nearly 20% of U.S. citizens aged 75-85 (Franklin et al.1990). This disease is classified by the progressive deterioration to sensory nerves. It is often accompanied by allodynia, or the perception of pain from non-noxious stimuli (Boulton et al 2004).

Individuals with PN also exhibit decreased self-reported daily physical activity. Chronic pain negatively impacts many psychological factors, and often leads to depression and poor quality of life. However, it is currently unknown if decreased confidence in performing normal activities of daily living is associated with PN-related pain.

The purpose of study was to 1) examine the relationship between PN-related foot pain and activity-specific balance confidence, and 2) examine the potential for acute reduction in foot pain to influence confidence in balance within this population.

METHODS

Participants with physician-diagnosed PN were recruited from the community. Initial screening was completed to ensure the presence of foot pain. Self-reported foot pain was measured by a Visual Analog Scale (VAS) for pain. The scale ranged from 0-10 with 10 being the worst pain possible. Participants were included if they reported pain ranging from 3-8.

Participants completed three days of testing separated by one week. Balance confidence was assessed by the modified Activities-specific Balance Confidence (ABC) questionnaire (Filiatrault et al. 2007). Respondents self-rated the degree of confidence in their balance when performing 15 different activities of daily living. A four-category response format with descriptive anchors was used (0, not at all confident; 1, slightly confident; 2, moderately confident; 3 very confident). Responses were summed to produce an ABC score ranging from 0-45.

Following completion of the VAS pain scale and the ABC, one of three topical analgesic treatments was applied to the participant's feet. A double blind, placebo-controlled format was employed. Treatment order was randomized using a Latin-square method such that participants received each treatment exactly once.

Foot pain level (VAS) and balance confidence (ABC) were reassessed 30 minutes following treatment application. The same variables were also assessed once per hour for the following eight hours using a palm pilot and the Purdue Momentary Assessment Tool (PMAT, Bangstate, Inc.).

Pain levels and ABC scores were analyzed using two-factor (Treatment X Time) ANOVA with repeated measures. Pearson product correlations (R) were used to examine the relationship of pain and balance confidence. Tukey's post hoc analysis was used wherever appropriate.

RESULTS AND DISCUSSION

Included participants (20 men, 22 women, mean ± SE age = 70.8 ± 1.4 yrs, height = 151.6 ± 1.60, body mass = 79.4 ± 3.3) had been diagnosed with PN for 6.4 ± 0.7 years.

While additional main effects and interactions were observed, only time effects are presented here. Pain level was significantly affected across time ($F_{9,36}$ = 29.85, $p < 0.01$). A two point reduction in pain was observed from pre-test (i.e., 0hr) to 1hr (treatments were applied at 0.5hr). Significant pain reduction was present until 5hr, upon which time sensation of pain gradually rose to pre-test levels by 9hr (Figure 1).

ABC scores also changed within the 9hr study period ($F_{9,36}$ = 2.59, $p < 0.01$). ABC significantly increased from 0hr to 1hr, and remained elevated between hours 3-5hr (Figure 1).

Correlation analysis revealed a significant negative correlation between pain and ABC (Pearson R = -.2762, $p < 0.0001$). However, detailed inspection of the graph (Figure 1) revealed that the inverse relationship between pain and ABC only occurred at 0-1hr and again at 6-9hr. Between these time points (i.e., 2-5hr), pain and ABC changed in parallel.

SUMMARY/CONCLUSIONS

The presence of PN-related foot pain negatively impacts confidence in balance needed to complete normal activities of daily living. Interestingly, this inverse relationship may only hold for relatively high self-reported pain levels (i.e., VAS pain scale > 3.5). Future research is needed to fully explore this relationship.

An acute reduction in foot pain significantly increases one's confidence in balance. The inclusion of a topical analgesic should therefore be considered in both the management PN and the design of future intervention studies for this population.

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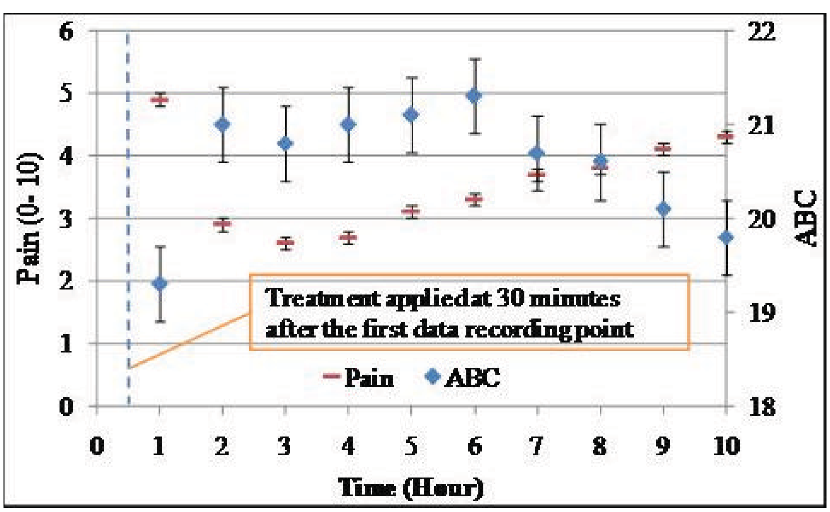

Figure 1 – Pain and ABC levels across nine continuous hours of data collection. Following treatment, pain level decreased and ABC scores increased. By 9hr, both pain level and ABC scores returned to close pre-test (0hr) levels.

ELIMINATION OF PAIN BY PASTING METAL PARTICLES ON DOMAIN OF SKIN DETECTED PAIN FOR DISEASE OF HIPBONE AND KNEE

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INTRODUCTION

It has been known long time that complete elimination of pain of joint, nerve and so on is much difficult. Our report will show that the pain is eliminated by pasting metal particles on domain of skin which pain is detected due to push with finger or tool. The effect was evaluated psychologically with use of VAS (Visual Analog Scale). In the report, both of procedures and effect of elimination of pain are explained and the reason is estimated.

METHODS

Domain of pain is detected by pushing skin with use of finger or metal stick shown in Fig 1. Number of subjects for disease of (a) herniated disk, (b) hipbone nerve (sciatic nerve) and (c) those for knee joint are each same 80. An average age for three diseases is around 50 years old. All the subjects received informed consent. The subjects are divided to two groups in each disease; Metal particles are pasted on the domain detected pain on skin for half of subjects (*Metal Group*), and the non-metal particles are used for other half of subjects (*Control Group*). Succession of medical treatment for three days for disease of herniated disk is carried out in order to effect of elimination of pain by metal particles. Experimental procedure is as follows:

(1) Questionnaire of VAS before medical treatment.
(2) Search of domain of pain on skin with use of metal stick (Figure 1).
(3) Pasting metal particles (Figure 2) for Metal group, and non-metal particles for Control group (Figures 3 and 4)
(4) Medical treatment by stimulation of low frequency and massage for ten minutes.
(5) Questionnaire of VAS after the treatment.

Figure 1: Metal stick for searching pain. Tip is cone type and bottom is flat with 7mm diameter, and the stick is 23cm long.

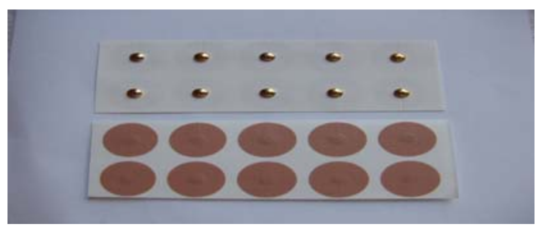

Figure 2: Example of magnetic metal particles is 5.5mm diameter and 1.5 height. Upper part is pasted on skin. Distance between two metal particles is set between 8 to 12 mm.

Figure3: Detected region is marked.

Figure4： Metal particles shown in Figure 2 are fixed on skin complained of pain.

RESULTS AND DISCUSSION

Effect of treatment of metal particle for diseases(i.e., Metal group) is shown in Tables 1. VAS value before treatment in each disease is different, since subjects in Metal and Control groups are different. But, VAS value of Metal group after treatment significantly decreases compared with that of Control group for each disease. The degree of the effect reveals for disorder of hipbone and knee joint compared with disorder of herniated disk. These results denote effect of treatment of metal particles for one day. As for Control group, VAS value decreases after treatment. The cause is considered to be medical treatment of stimulus of low frequency and massage. As for the effect of continuous treatment for three days, the result of disorder of herniated disk is shown in Table 2. The metal treatment for Metal group after three days denotes dramatic effect.
All the subjects tell that the pain is eliminated completely. Moreover, sensation of the painless continues for a long period. The fact shows that increase in the number of treatment of metal particles demonstrates more elimination of the pain comparing with lower number of the treatment .
The elimination of pain is considered to be (ｉ) function of muscles connected joint and (ii) domain on skin showing pain. The lower function of muscle shows lower extension and contraction. The phenomenon looks like used rubber band. The muscle becomes hard. The influence gives damage to the neuron operated the muscles and the signal of pain reaches to upper control system, then pain occurred. It is considered that use of metal particles on the skin gives thermal stimulus. Nerve roots of pain and thermal signals are different. The thermal signal controls the pain signal. The mechanism is known as gate control theory. The pain occurred from joint or disk projected on wide domain of skin. The phenomenon looks like damage area when ball hits on body: That is, the damage shows not point but area. Therefore, a plate with metal particlese is pasted on skin shown in Figure 4 in order to reduce or eliminate signal of pain.

Table 1: VAS before and after treatment

(a) Disorder of herniated disk

Group	Before	After
Metal	53.6±22.8	21.9±21.2*
Control	45.5±15.1	36.8±25.6

(b) Disorder of hipbone nerve

Group	Before	After
Metal	58.7±13.7	17.3±13.4*
Control	56.5±12.1	38.4±12.6

(c) Disorder of joint of knee

Group	Before	After
Metal	65.8± 8.4	14.3±12.0*
Control	63.5±10.4	45.6±10.3

Mark * means significant difference before and after treatment by 5%

Table 2: VAS in treatment for three days

(a) Disorder of herniated disk

Group	Before	After
Metal	53.6±22.8	2.3±20.4*
Control	45.5±15.1	34.3±27.7

SUMMARY/CONCLUSIONS

It is found that metal particles pasted on domain of skin detecting pain come to elimination of pain for diseases of joints.
The effect is proved by VAS. Increase of number of treatment denotes more remarkable degree of elimination of the pain.

ELECTROMYOGRAPHY AS IT PERTAINS TO THE STUDY OF MUSCULAR MOVEMENT WITHIN OCCUPATIONAL THERAPY

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INTRODUCTION

The purpose of this study was to analyze simultaneous electromyographic activities of the muscles long radial extender of the carpus, ulnar flexor of the carpus, and number interosseous II, during the manipulation of different objects: glass, fork, knife and hair brush as well as the effectiveness of electromyograph (EMG) as an instrument of evaluation. The goal of the present study was to understand the coordination and execution of hand movements using electromyography as means finding a new way of evaluate the upper-limbs.

METHODS

The participants were ten healthy right-handed persons without motor problems (7 men, 3 women) performing predetermined movement according to established protocol.

Electromyography (EMG) was recorded from three muscles: long radial extender of the carpus, ulnar flexor of the carpus and number interosseous II.

The electromyograph equipment was made by EMG System of Brazil Ltd. It is composed of differential double electrodes, a band pass filter at 20 to 500 Hz, with a common mode rejection ratio of 80 dB and a subsequent amplification of 20 times; each electrode has 25 mm^2 contact area placed at 10 mm apart with 12-bit A/D converter with sampled frequency of 2.0 kHz for each channel.

The recommendations from the International Society of Electrophysiology and Kinesiology (ISEK) regarding electromyography applications were followed.

RESULTS AND DISCUSSION

The results show that the electromyographic signal of three muscles respectively, related with dynamic movements during the grasping and manipulation of different materials. Small variation in muscle activities was observed.

This suggests that the variation has a direct correlation with the object’s weight and extent of movement according to studies realized by Johansson et al (1992).

According to formation from the curves obtained it can be said that one major force is required to overcome the inertia from the still object.

Figure 1 shows the electrical muscular activities of different objects (cup, fork, and hair brush).

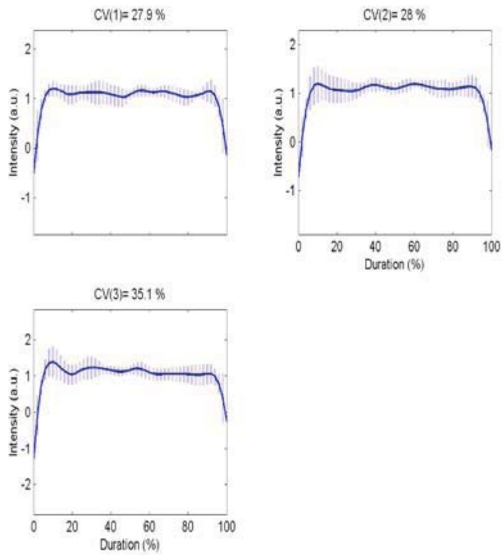

Figure 1: Shows the electrical muscular activities during grasping: long radial e**xtender of the carpus, C**V(1); number interosseous II - CV(2); m. Flexor Ulnar of the Carpus – CV(3).

SUMMARY/CONCLUSIONS

We concluded from the analyzed sample and the muscles' electrical activities which were present during the grasping, that it is possible to recognize the standard of activation in the electrical signal through electromyograph (EMG) during the specific task.

The behavior of the electrical activities from the muscles studied in different tasks remained in the same synergistic correlation, independently of the material's shape. It confirms the muscles' main function during the dynamic movement.

Electromyography (EMG) has proven to be an efficient tool to study upper limbs, and can be used for clinical evaluation in patients with musculoskeletal harm.

The information obtained through electromyography (EMG) of surface is results relevant and can be used as effective strategies of treatment in rehabilitation.

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NEUROPATHIC PAIN EFFECTS PHYSICAL FUNCTION

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INTRODUCTION

Chronic diffuse polyneuropathy, more commonly known as peripheral neuropathy (PN), affects approximately 20 million Americans (108^{th} Congress, 2005). PN is marked by chronic deterioration to peripheral nerves that leads to both positive (i.e., painful) and negative (i.e., numbness) symptoms (Boulton et al. 2004).

As little to no lower-extremity strength declines are observed, movement disorders secondary to PN are believed to arise primarily as a result of reduced somatosensory feedback. Although painful symptoms occur in over one quarter of cases, this symptom has been overlooked as a possible factor leading to reduced physical function. The purpose of this experiment was to determine the relationship between foot pain and various measures of physical function in the PN population.

METHODS

Individuals with physician-diagnosed PN were recruited from the community. Included participants underwent assessments of foot pain, plantar pressure sensitivity, and physical function.

Self-reported foot pain (PAIN) was recorded on a Visual Analog Scale for pain. The scale ranged from 0 - 10, with 10 being the worst possible pain. Plantar pressure sensitivity (PPS) was assessed at five weight-bearing sites on the right foot sole with a 5.07 gauge monofilament. The number of sites with intact sensation was totaled to produce a PPS score ranging from 0-5.

A battery of tests was then completed to assess several components of physical function. Locomotion-based physical function was evaluated by the distance of a 6-mininute walk (6MW) test and the duration of a Timed Up-and-Go (TUG) test. For the 6MW, participants were instructed to walk as far as possible around two cones placed 30 meters apart down a well lit hallway. Distance covered (m) was recorded. For the TUG, participants sat with their back against a chair and feet on the floor with a cone three meters in front of the chair. The time (sec) taken to stand up, walk around the cone, and sat back down in the chair was recorded.

Eyes-open and eyes-closed standing balance was assessed by the average velocity (VEL, cm/s) and the area of an ellipse enclosing 95% (AREA, cm^2) of the body center of pressure during quiet stance. The average of two, 30sec trials was recorded.

Isokinetic knee joint strength was examined using a Biodex dynamometer (Biodex Medical, Shirley, NY) at 60 deg/sec. Both knee extensor (KE, Nm) and flexor peak torque (KF, Nm) were computed from five maximal effort trials. Peak torque of the best three trials was averaged and used for analysis.

Pearson product correlations (R) were used to examine the relationships between PN symptoms (PAIN, PPS) and the aforementioned physical performance outcomes (6MW, TUG, VEL, AREA, KE, KF). Significance was set to $p < .05$

Table 1 – Sensory and Pain Correlations (Pearson) on Physical Performance Tests

			Eyes-Open		Eyes-Closed		Leg Strength	
	6MW	TUG	VEL	AREA	VEL	AREA	KE	KF
PAIN	-.46*	.41*	.13	.09	.31	.13*	-.21	-.19
PPS	-.04	-.002	-.32	-.35*	-.32	-.32*	-.28	-.23

Note: * $p < .05$

RESULTS AND DISCUSSION

Participants (20 men, 22 women, mean ± SE age = 70.8 ± 1.4 yrs, height = 151.6 ± 1.6 cm, body mass=79.4 ± 3.3 kg) had been diagnosed with PN for 6.4 ± 0.7 years.

PAIN was significantly correlated to performance in both locomotion-based measures of physical function. Specifically, those with increased PAIN tended to perform worse in both the 6MW and TUG tests (Table 1). Conversely, the degree of sensory loss (PPS) was not correlated to performance in either of these measures.

Both PAIN and PPS accounted for a significant amount of variance in selected measures of standing balance performance. Increased PAIN was related to increased eyes-closed AREA and to a lesser extent, eyes-open VEL. Decreased PPS was related only to increased eyes-closed AREA. Neither PAIN nor PPS was related to either measure of leg strength.

SUMMARY/CONCLUSIONS

The primary symptoms of PN, namely foot pain and numbness (PPS), are predictive of physical function. While PPS is associated with reduced standing balance, pain more closely predicts common measures of locomotion-based physical function. Future studies should thus consider self-reported foot pain as a critical factor influencing physical function in this population.

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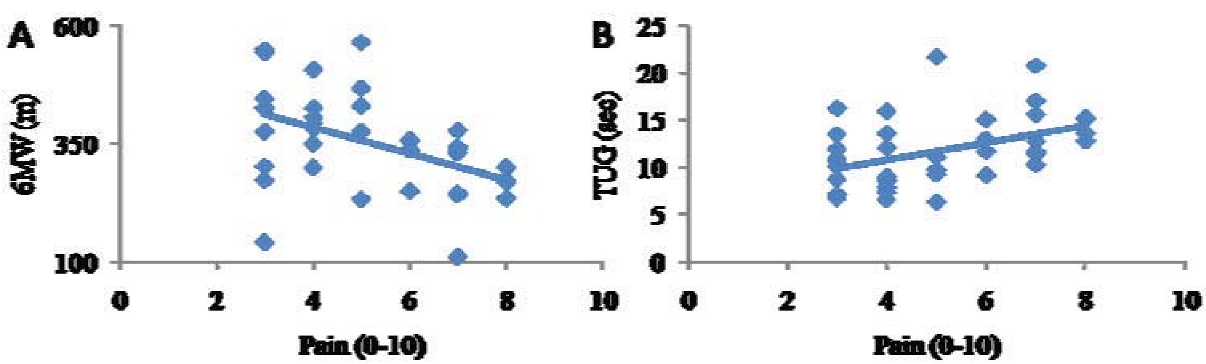

Figure 1. Scatter-plots illustrating the relationship between self-reported foot pain and selected measures of physical function. Linear regression lines are indicated with the solid lines in the scatter-plots.

GAIT CO-ACTIVATION PATTERNS IN TERMINAL OSTEOARTHRITIS KNEE PATIENTS

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INTRODUCTION

The neuromuscular patterns used by knee osteoarthritis (OA) patients during gait seem to evolve with the progressive nature of the disease. These patterns, cohort differences and surgery intervention changes have been addressed using waveform analysis procedures principal component analysis-PCA (Astephen, 2005), and co-activation (Lewek, 2005) to reduce the volumus gait cycle analysis data. While these procedures are successful in reducing the multi-dimensional data available through gait analysis they are not effectively address the phasic nature of walking. The current study uses co-activation assessment on a phasic breakdown of the gait cycle in an attempt to identify phase specific neuromuscular patterns that are related to knee osteoarthritis. Our hypothesis to be tested is that differences in the neuromuscular patterns between terminal OA patients and controls are gait cycle phase specific. This characterization of the phasic co-activation pattern differences can be used in the assessment of the rehabilitation process after total knee replacement procedures and potentially in all stages of osteoarthritis and other knee pathologies.

METHODS

This is a cross-sectional cohort contrast study. We compared the muscle co-activation patterns between a group of OA patients (N=15) with terminal unilateral knee OA (scheduled for total knee replacement surgery) to age and gender matched control group (N=10). A surface electromyography system (Motion Lab Systems, Inc.) was used to measure muscle activation during gait at a self selected pace. The MA-300 pre-amplified EMG signals were sampled @ 1200 Hz using a 12 bit A/D and a 20 – 350 Hz band pass filter. Each muscle's rectified signal was low passed at 10 Hz (linear envelope, Figure 1) and was magnitude normalized (%) to 5 trials' peak amplitude (Ricamato, 2005). The co-activation index (*CI*) for each muscle pair/phase was calculated as the ratio of the activation sum:

$$CI = \frac{EMG_{LA}}{EMG_{HA}} \times (EMG_{LA} + EMG_{HA})$$

This method uses both timing and magnitude of the relative EMG signal of of muscle pairs for each phase (Lewek, 2005). The Co-activation of the vastus lateralis-biceps femoris (VL-BF), vastus medialis-semitendinosus (VM-ST), rectus femoris-biceps femoris (RF-BF), rectus femoris-semitendinosus (RF-ST), vastus lateralis-gastrocnemius (VL-GA), vastus medialis-gastrocnemius (VM-GA), and the rectus femoris-gastrocnemius (RF-GA) were calculated for three phases of the gait cycle (GC): (1) loading response and midstance (LR & MS), (2) terminal stance and preswing (TS % PSW), and (3) swing (SW). Student's t-test (independent) was used to determine differences between groups.

RESULTS AND DISCUSSION

The control and OA group characteristics are: age (52.7 ± 3.4 vs. 61.5 ± 4.4 years), height (1.69 ± 0.06 vs. 1.68 ± 0.09 meters), weight (75.3 ± 13.6 vs. 95.9 ± 19.0 kg) and gender (66 vs. 50 % females). The walking speed of the control and the OA subjects was 1.22 ± 0.11 and 0.92 ± 0.16 m/s, respectively ($p = 0.003$).

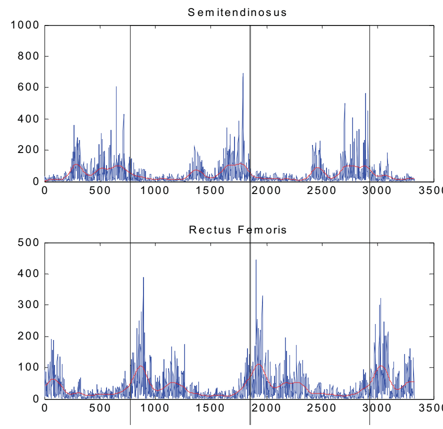

Figure 1: EMG data during gait are shown. Activation (mV) of the semitendinosus (top) and the rectus femoris (bottom). Vertical lines represent heel strikes.

The knee flexion range of motion during loading response (LR) was significantly less for the OA group 9.5 ± 6.4 vs. the control group 15.5 ± 1.6 degrees. Generally, while the OA group's CI values were larger than the control's for the stance phase of the gait cycle (Table 1), only during the initial 50% (LR & MS) they were found significant. No differences were found between OA and control groups in co-activation for all the muscle pairs in the later part of the stance (TS & PSW) and the swing phases. At the loading response of the gait cycle the rapid weight transfer on the OA leg; and during the midstance the stabilization of the OA knee require reduction of knee range of motion under higher load conditions. This is achieved by antagonistic co-activation of the knee musculature.

SUMMARY/CONCLUSIONS

Terminal OA patients display different EMG co-activation patterns than cohort controls. These results suggest that the altered neuromuscular patterns are gait phase specific during the loading response-midstance phase. Subsequent investigations using the phasic co-activation process may prove useful especially on post total knee replacement surgery, earlier in the OA disease process, and on other knee pathologies (ACL or patellofemoral pain).

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Table 1: Co-activation Indices (CI) of muscle pairs (mean ± SD) for the control and the OA groups. Data for each gait cycle (GC) phase are included. († significance at .01)

GC Phase	**LR & MS**		**TS & PSW**		**SW**	
Group	**Control**	**OA**	**Control**	**OA**	**Control**	**OA**
VL-BF	27.2 ± 5.3	59.7 ± 4.7†	12.2 ± 3.8	21.0 ± 3.4	42.9 ± 5.1	30.6 ± 5.1
VM-ST	22.2 ± 5.7	45.5 ± 5.1†	13.1 ± 5.2	23.0 ± 4.7	44.3 ± 6.4	32.8 ± 5.7
RF-BF	27.8 ± 5.5	56.4 ± 4.9†	13.9 ± 3.0	16.2 ± 2.7	41.7 ± 5.2	28.5 ± 4.7
RF-ST	22.0 ± 5.5	45.5 ± 4.9†	12.7 ± 3.6	17.6 ± 3.2	41.6 ± 4.5	31.9 ± 4.0
VL-GA	19.3 ± 5.0	51.6 ± 4.7†	10.0 ± 4.8	20.1 ± 4.3	12.3 ± 2.9	16.7 ± 2.6
VM-GA	20.7 ± 6.1	45.8 ± 4.9†	9.7 ± 4.2	19.2 ± 3.5	12.7 ± 2.8	17.2 ± 2.5
RF-GA	24.0 ± 5.6	48.4 ± 5.0†	10.3 ± 5.0	21.6 ± 4.5	12.9 ± 2.5	16.9 ± 2.2