ISEK

Tenth Congress of the International Society of Electrophysiology and Kinesiology

June 21-24, 1994
The Omni Hotel at Charleston Place

CHARLESTON,
SOUTH CAROLINA
USA

Abstract Book

Edited by Dr. Richard Shiavi and Dr. Steve Wolf

Abstract Book

Tenth Congress of the International Society of Electrophysiology and Kinesiology

June 21-24, 1994

Dr. Richard Shiavi and Dr. Steve Wolf
Editors

Chairmen's Letter to Delegates

Dear Colleagues,

We are pleased to provide you with this book which comprises the proceedings of the Tenth Congress of the International Society of Electrophysiology and Kinesiology (ISEK). The general program mirrors the purpose of the society which is to promote, teach, and disseminate information in the area of movement control through fostering knowledge in sciences including, but not restricted to, electromyography and kinesiology. The delegates represent a broad range of professions and include clinicians, physicians, scientists, physiotherapists, engineers, and others who have interest in pursuing research in electromyography and kinesiology and their applications to the clinical environment and improving the quality of life.

The scientific program consists of the keynote lecture, a plenary lecture, two perspective sessions, three minisymposia and eighty submitted presentations. The abstracts of the submitted presentations are printed in this proceedings. This year we have used a slightly different format. In order to provide some of the material in a more archival nature, the keynote and plenary lectures and the minisymposia will be printed in a special issue of the society's journal, *Journal of Electromyography and Kinesiology*. Also some information was provided in a more interactive format through the two perspective sessions. The perspective sessions are similar to roundtable discussions but encourage participation of the other attendees. Two leaders act as principal discussants and make a five- to ten-minute presentation on a topic. Then the floor is open to the delegates for comments and discussion. The results of these sessions are recorded and the discussions will be summarized in future issues of the Society's newsletter.

This year we are very fortunate to have available to us exhibitors who manufacture or sell measurement equipment and software that have a particular interest to the delegates of the Congress and to the members of the Society in general. A summary of their services and products are listed in the proceedings as well. In addition we are fortunate to have Butterworth-Heinemann Ltd. provide an exhibit. As many of you know, this company prints the Society's journal. We encourage all of you to visit these exhibitors and our publisher. They are here to assist you!

The organizers would like to thank all of those who played a role in planning, organizing, and publicizing this congress, especially The Conference Table, which is the Congress Secretariat. As you know, much time and energy is expended in creating a congress and all of its attendant events. The list of those who performed various roles are listed in the following pages.

CONGRESS ORGANIZERS

Richard Shiavi, Biomedical Engineering
Vanderbilt University, Box 6117B
Nashville, Tennessee 37235

Steve Wolf, Rehabilitation Medicine
Emory University School of Medicine
Atlanta, Georgia 30322

ISEK Council

Gerald Zilvold - President
Enschede, The Netherlands

Richard Shiavi - Vice President
Nashville, Tennessee USA

Paul Anderson - Secretary
Baltimore, Maryland USA

Lars Arendt-Nielsen - Treasurer
Aalborg, Denmark

Jan Clarijs
Brussels, Belgium

Yukio Mano
Nara, Japan

Antonio Pedotti
Milano, Italy

Gunter Rau
Achen, Germany

Steve Wolf
Atlanta, Georgia USA

Congress Secretariate

Judith Hendrick
The Conference Table
P.O. Box 398
Charleston, South Carolina USA

Program Committee

Paul Anderson - program chairperson
Baltimore, Maryland USA

Arthur Sherwood - exhibitor chairperson
Houston, Texas USA

Brigitta Jann - accompanying persons chairperson
Atlanta, Georgia USA

Lars Arendt-Nielsen
Aalborg, Denmark

Richard Baratta
New Orleans, Louisiana USA

Carlo Frigo
Milano, Italy

Jan Pieter Clarijs
Brussels, Belgium

James Collins
Boston, Massachusetts USA

Keith Hayes
London, Ontario Canada

Hermie J. Hermens
Enschede, The Netherlands

Carl G. Kukulka
Rochester, New York USA

Yukio Mano
Nara, Japan

Roberto Merletti
Boston, Massachusetts USA

Lars Oddsson
Boston, Massachusetts USA

Philip Parker
New Brunswick, Canada

Gunter Rau
Achen, Germany

Serge Roy
Boston, Massachusetts USA

Richard Segal
Atlanta, Georgia USA

Lynn Snyder-Mackler
Newark, Delaware USA

Gary Soderberg
Omaha, Nebraska USA

Dick Stegeman
Nijmegen, The Netherlands

Willemien Wallinga
Enschede, The Netherlands

Steve Wolf
Atlanta, Georgia USA

Gerald Zilvold
Enschede, The Netherlands

Exhibitors and Sponsors

The International Society of Electrophsiology and Kinesiology and the Organizing Committee of this tenth congress are very grateful to the exhibitors who are present and the sponsors who contributed financial support. They are listed below.

Please visit the exhibitors. A description of products and services with complete addresses of the sponsors and exhibitors can be found on the pages following the abstracts section.

Exhibitors

Advanced Medical Technology
Boston, Massachusetts

Ariel Dynamics, Inc.
Trabuco Canyon, California

B & L Engineering
Santa Fe Springs, California

BTS - Bioengineering Technology & Systems
Woodmere, New York

Butterworth-Heinemann Publishing
Oxford, England

Kistler Instrument Corporation
Amherst, New York

Noraxon USA, Inc.
Scottsdale, Arizona

Tekscan, Inc.
South Boston, Massachusetts

Sponsors

Butterworth-Heinemann Publishing
Oxford, England

Dataq Instruments
Akron, Ohio

Kistler Instrument Corporation
Amherst, New York

Vanderbilt University
Nashville, Tennessee

Special Congress Issue of JEK

These titles will appear in the special congress issue of JEK.

Mechanisms of Muscle Fatigue - Roger Enoka

From the Research Laboratory to the Medical Clinic - Gerald Zilvold

Double Tasks as a Tool in Movement Analysis - Theo Mulder

Protocols for Diagnosis and Movement Analysis in CP Children - Arnand Nene

Technical Specifications for Movement Analysis - Hermie Hermans

Muscle Compartmentalization and Relevance to Therapy - Richard Segal

EMG Pattern Analysis - Ed Ciaccio

Muscle Pain and Motor Performance - Lars Arendt-Nielsen

Muscle Pain - State of the Art - Lars Arendt-Nieslen

Experimental Models to Induce and Assess Muscle Pain - Peter Svensson

Exposures for Work Related Muscle Skeletal Disorders - Hanne Christensen

Transcranial Magnetic Stimulation: Principles and Applications - Arthur Sherwood

Fundamental Aspects of Magnetic Stimulation - Reza Jalinous

Clinical Applications of Transcranial Magnetic Stimulation - Yukio Mano

Central Motor Conduction Studies in Patients with Spinal Cord Injuries - Keith Hayes

Transcranial Magnetic Stimulation as a Tool for Studies of Human Neurophysiology - Arthur Sherwood

Schedule of Scientific Sessions

Wednesday - Session A - 1000-1200 - Cypress Room

Processing EMG

Chairpersons - Hermie Hermans and Serge Roy

MYOELECTRIC CONDUCTION VELOCITY ESTIMATION ON VASTUS LATERALIS
J Y Hogrel, J Duchene, J F Marini
Universite di Technologie de Compiegne; Compiegne, France

CHANGES IN THE EMG POWER SPECTRUM WITH INCREASING FORCE IN HAND MUSCLES
P Noel, M Bilodeau, A B Arsenault, E Thibault, D Gravel
University of Montreal and Montreal Rehabilitation Institute; Montreal, Canada

SURFACE AND WIRE EMG CROSS-TALK AMONG NEIGHBORING MUSCLES
M Solomonow, R Baratta, B Zhou, M Bernardi
Louisiana State University Medical Center; New Orleans, LA

DESCRIPTION OF ELECTROMYOGRAPHIC LINEAR ENVELOPE FOR SYNERGY ANALYSIS IN ISOKINETIC TEST
M Cesarelli, M Pisaniello, F Pizza
University of Naples; Naples, Italy

A NEURAL NETWORK PATTERN RECOGNITION APPROACH TO SYNERGY ANALYSIS OF ELECTROMYOGRAPHIC DATA
C A Tucker, H J Yack, S C White
State University of Buffalo; Buffalo, NY

QUANTIFICATION AND GRAPHICAL PRESENTATION OF MUSCLE COORDINATION PATTERN
Jiping He, Yang Wang
Thomas Jefferson University; Philadelphia, PA

EMG AS A MEASURE OF MOTOR CONTROL IN MAN: TOWARD A BASIS FOR QUANTIFICATION
A M Sherwood, W J Eaton, W B McKay, N F Kharas,
Baylor College of Medicine; Houston, TX

EFFECTS OF MUSCLE FIBER TYPE AND SIZE ON EMG MEDIAN FREQUENCY
S H Roy, E J Kupa, S C Kandarian, D J DeLuca
Boston University; Boston, MA

Wednesday - Session B - 1000-1200 - Dogwood Room

Gait Posture and Balance

Chairpersons - Paul Anderson and James Collins

EMG LATENCIES IN SEVEN POSTURAL MUSCLES FOLLOWING POSTURAL PERTURBATION IN YOUNG ADULTS
G V Smith, P A Anderson
University of Maryland; Baltimore, MD

EMG AND ANGULAR MOTION OF THE KNEE JOINT IN THE WALKING GAIT ANALYZED BY USE OF THE FAST FOURIER TRANSFORM
B R Brandell, C L Rice
University of Saskatchewan; Saskatoon, Canada

AGE-RELATED CHANGES TO OPEN-LOOP AND CLOSED-LOOP POSTURAL CONTROL MECHANISMS
J J Collins, C J De Luca, A Burrows, L A Lipsitz
Boston University; Boston, MA

THE EFFECT OF PRESSURE ON SPINAL CORD EXCITABILITY AND PARAMETERS OF GAIT INITIATION
D Brunt, J A Robichaud, R Annan
University of Florida; Gainesville, FL

THE EFFECTS OF CRYOTHERAPY ON SELECTED BALANCE PARAMETERS
B Pfleger, T Whittaker, JE Lander
Life College; Marietta, GA

COMPARISON OF ACROSS-SUBJECT EMG PROFILES USING SURFACE AND MULTIPLE INDWELLING WIRE ELECTRODES DURING GAIT
R A Bogey, J Perry, E L Bontrager, J K Gronley
University of California; Sacramento, CA

QUANTIFICATION OF HUMAN GAIT WITH THE PRESENCE OF BACK PAIN
L Arendt-Nielsen, T G Nielsen, P E Pedersen, J H Svarre
Aalborg University; Aalborg, Denmark

A NOVEL APPROACH TO THE ESTIMATION OF MUSCLE ON-OFF TIMING DURING GAIT
P Bonato, T D'Alessio, M Knaflitz
University of Rome; Rome, Italy

Wednesday - Session C - 1330-1530 - Cypress Room
Minisymposium
From the Research Laboratory to the Medical Clinic
Convenor - Gerald Zilvold

I: ***Introduction***
Gerald Zilvold
Rehabilitation Center, Enschede, The Netherlands

II: ***Double Tasks as a Tool in Movement Analysis***
Theo Mulder
Rehabilitation Center; Enschede, The Netherlands

III: ***Protocols for Diagnosis and Movement Analysis in CP Children***
Arnand Nene
Rehabilitation Center; Enschede, The Netherlands

IV: ***Technical Specifications for Movement Analysis***
Hermie Hermans
Rehabilitation Center; Enschede, The Netherlands

Wednesday - Session D - 1330-1530 - Dogwood Room

Muscle Fatigue

Chairpersons - Lars Oddsson and Steve Wolf

CHANGES IN ANKLE STRETCH REFLEX WITH FATIGUE INDUCED BY SPONTANEOUS HOPPING
B Maton, A LePellec, E Golomer
Universite de Paris Sud; Orsay, France

MUSCLE FATIGUE MANIFESTATIONS IN THE FOLLOW UP OF PATIENTS AFFECTED BY DUCHENNE MUSCULAR DYSTROPHY
M Knaflitz, G Balestra, C Angelini, M Cadaldini
Politecnico di Torino; Torino, Italy

BRAKING PROCESS ADAPTATION TO MUSCULAR FATIGUE IN HUMAN RAPID AIMED FOREARM MOVEMENTS
S Le Bozec, M Gentil, J Zouhri
Universite de Paris Sud; Orsay, France

LOCALIZED AND GENERAL FATIGUE INDUCED BY PROLONGED LIFTING TASKS
P Capodaglio, M Buonocore, E M Capodaglio, G Bazzini
Medical Center of Rehabilitation; Montescano, Italy

FATIGUE DURING LOW-INTENSITY CONTRACTIONS OF THE TRICEPS SURAE MUSCLE
J H van Dieen, P Heijblom, M H B Helmes, H H E Oude Vrielink
IMAG-DLO; Wageningen, The Netherlands

INFLUENCE OF DIFFERENT CONTRACTION LEVELS ON SURFACE EMG SPECTRAL PARAMETERS OF BACK MUSCLES
L Oddsson, S H Roy, J E Giphart, M Emley, J A Levins, C J De Luca
Boston University; Boston, MA

EMG AMPLITUDE IN THE LUMBAR SPINE EXTENSOR AND FLEXOR MUSCULATURE DURING MAXIMAL AND SUBMAXIMAL CONSTANT FORCE CONTRACTIONS
M B Frazer, R W Norman
University of Waterloo; Waterloo, Ontario, Canada

POST FATIGUE FORCE AND EMG IN FES ACTIVATED MUSCLES
J Mizrahi, D. Seelenfreund, A Aviram, E Isakov
Loewenstein Rehabilitation Center; Raanana, Israel

Thursday Session A -1000-1200 - Cypress Room
Biomechanics and Motion Analysis
Chairpersons - Carlo Frigo and Richard Baratta

COMPUTER AIDED ASSESSMENT OF RESPIRATORY SYSTEM FUNCTION
A Aliverti, P Carnevali, G Ferrigno, A Pedotti
Politecnico di Milano; Milano, Italy

METHODOLOGICAL ASPECTS OF KINEMATIC ANALYSIS OF PEN MOVEMENT DURING HANDWRITING USING THE ELITE SYSTEM
A Campisi, M Cincera, M Rabuffetti, A Pedotti
Centro di Bioingegneria; Milano, Italy

MULTIPARAMETRIC ANALYSIS OF LIPS AND JAW KINEMATICS IN NORMAL AND STUTTERING SUBJECTS
M Redolfi, G Ferringno, A Pedotti
Centro di Bioingegneria; Milano, Italy

A THREE-DIMENSIONAL ANALYSIS OF RELEASE PARAMETERS IN ELITE NETBALL SHOOTING
S Miller
Cardiff Institute of Higher Education; Cardiff, Wales

GAIT FEATURE CHARACTERIZATION THROUGH MOMENT/ANGLE RELATIONSHIP
C Frigo, L M Jensen
Centro di Bioingegneria; Milano, Italy

THREE DIMENSIONAL CHARACTERIZATION OF MUSCLE LENGTH, LOAD AND VELOCITY
R V Baratta, M Solomonow, G Nguyen, R D'Ambrosia
Louisiana State University Medical Center; New Orleans, LA

PROSTHETIC MASS AND INERTIAL MEASUREMENT PILOT STUDY
R D McAnelly, B Rogers, V Faulkner,
University of Texas Health Science Center; San Antonio, TX

A PRELIMINARY KINEMATIC ANALYSIS OF NORMAL BALANCE
P A Anderson, G V Smith, G Alon
University of Maryland; Baltimore, MD

Thursday Session B -1000-1200 - Dogwood Room
Surface EMG and Movement
Chairpersons - Gary Soderberg and Roberto Merletti

A FIR NEURAL NETWORK FOR MYOELECTRIC SIGNAL CLASSIFICATION
K B Englehart, B S Hudgins, M Stevenson, P A Parker
University of New Brunswick; Fredericton, New Brunswick, Canada

NEUROMUSCULAR PARTITIONING OF THE HUMAN LATERAL GASTROCNEMIUS: LEVELS OF ELECTROMYOGRAPHIC ACTIVITY DURING ISOMETRIC TASKS
R L Segal, L S Cheng, P A Catlin, S L Wolf, A W English
Emory University School of Medicine; Atlanta, GA

STRUCTURE AND FUNCTION OF THE ABDOMINAL MUSCLES DURING PREGNANCY AND THE IMMEDIATE POST-BIRTH PERIOD
W L Gilleard, J M M Brown
University of Sydney; Lidcombe, NSW, Australia

MOMENT ARM AND MUSCLE LENGTH EFFECTS ON EMG PRODUCTION IN HUMAN GASTROCNEMII MUSCLES
M R Nourbakhsh, C G Kukulka
Tehran University; Tehran, Iran

WRIST JOINT RESTRICTION: EFFECT ON FUNCTIONAL UPPER LIMB MOTION DURING PERFORMANCE OF THREE FEEDING ACTIVITIES
J E Cooper, E Shwedyk, A Quanbury, J Miller, K Nemeth
University of Manitoba; Winnipeg, Manitoba, Canada

FINGER REACTION ANALYSIS WITH JAPANESE LANGUAGE ENTRY KEYBOARD
Y Okada, S Katsumi
Nippon Bunri University; Ooita, Japan

PERTURBATION OF HUMAN MASTICATORY EMG-ACTIVITY BY NOCICEPTIVE LASER STIMULI
P Svenseson, L Arendt-Nielsen
University of Aarhus; Aarhus, Denmark

INFLUENCE OF FORCE OUTPUT AND MUSCLE LENGTH ON ERECTOR SPINAE EMG MEDIAN FREQUENCY
A F Mannion, P Dolan
University of Bristol; Bristol, UK

Thursday Session C - 1330-1530 - Cypress Room
Noninvasive Identification and Characterization of Single Motor Units by EMG (1330-1430)
Chairperson - Gunter Rau

I: Principle and Basic Physiological Applications
Gunther Rau
Institute for Biomedical Engineering; Aachen, Germany

II: First Clinical Applications in Diagnosis of Neuromuscular Disorders
Catherine Disselhorst-Klug
Institute for Biomedical Engineering; Aachen, Germany

III. Discussion

EMG Pattern Analysis - (1430-1530)
Ed Ciaccio, Columbia University New York City; New York, NY
Chairperson -Gunter Rau

Thursday Session D - 1330-1530 - Dogwood Room
Spasticity, Paresis, Tremor and EMG
Chairpersons - Lars Arendt-Nielsen and Carl Kukulka

COMPUTER-CONTROLLED INSTRUMENTATION FOR DYNAMIC MEASUREMENTS OF SPASTICITY IN THE LOWER LIMBS OF MS-PATIENTS
L De Wolf, L Janssens, G Nuyens, W De Weerdt, A Spaepen,
P Ketelaer, J Carton
Groep T Institute of Engineering; Leuven, Belgium

A COMPARATIVE STUDY OF PHYSIOLOGICAL TREMOR IN NORMAL AND RHEUMATOID ARTHRITIC SUBJECTS
D M Halliday, B A Conway, W L Ng, J R Rosenberg
Univ. of Glasgow; Glasgow, Scotland

ADVANCES IN HYBRID SYSTEMS FOR PARAPLEGIC WALKING
H J Hermens, G Baardman, H M Franken, P H Veltink,
H B K Boom, G Zilvold
University of Twente; Enschede, The Netherlands

DIFFERENT THERAPEUTIC ELECTRICAL APPROACHES TO DIFFERENT KINDS OF SPINAL CORD LESIONS (SCL) SPASTICITY
R Prati, S Visconti, V. Alfieri
Rehabilitation Center; Lonato, Brescia, Italy

COMPENSATION OF LOSS OF POSTURAL MUSCLE ACTIVITY IN SPINAL CORD INJURED PEOPLE
H A M Seelen, Y J M Potten
Institute for Rehabilitation Research; Hoensbroek, The Netherlands

COMPARISON OF VOLUNTARY MOVEMENTS AND LOCOMOTION IN SPASTIC CEREBRAL PALSY
M E Johanson, S R Skinner,
Shriners Hospital; San Francisco, CA

BASIC FEATURES OF GAIT IN PATIENTS AFFECTED BY HIP DYSPLASIA: PRE AND POST OPERATIVE MULTIFACTOR ANALYSIS
C Frigo, C Romano
Instituto Orthopedico; Milano, Italy

TREMOR AND EMG MEASUREMENTS OF LOCALIZED MUSCLE FATIGUE IN VDU WORK
G Sundelin, L Burstrom, M Hagberg
University Hospital; Umea, Sweden

Friday Session A- 0815-0945 - Cypress Room
Minisymposium
Muscle Pain and Motor Performance
Convenor - Lars Arendt-Nielsen

I. Muscle Pain - State of the Art
Lars Arendt-Nieslen
Aalborg University, Lab. for Motor Control; Aalborg, Denmark

II. Experimental models to induce and assess muscle pain
Peter Svensson
Royal Dental College; Aarhus, Denmark

III. Exposures for work related muscle skeletal disorders
Hanne Christensen
National Institute of Occupational Health; Copenhagen, Denmark

IV. Discussion

Friday Session B- 0815-0945 - Dogwood Room
Minisymposium
Transcranial Magnetic Stimulation: Principles and Applications
Convenor - Arthur Sherwood

I. ***Fundamental Aspects of Magnetic Stimulation***
Reza Jalinous
The Magstim Company, Ltd.

II. ***Clinical Applications of Transcranial Magnetic Stimulation***
Yukio Mano
Nara Medical University, Nara, Japan

III. ***Central Motor Conduction Studies in Patients with Spinal Cord Injuries***
Keith Hayes
The University of Western Ontario; London, Canada

IV. ***Transcranial Magnetic Stimulation as a Tool for Studies of Human Neurophysiology***
Arthur Sherwood
Baylor College of Medicine; Houston, TX

V. Panel Discussion: The Future of Transcranial Magnetic Stimulation

Friday Session C - 1015-1200 - Cypress Room
The Knee
Chairpersons - Lynn Snyder-Mackler and Richard Segal

NEUROMUSCULAR KNEE JOINT CONTROL DURING DOWNHILL WALKING
M Kuster, G A Wood, S Sakurai
University of Western Australia; Perth, Australia

DYNAMIC BALANCE CHANGES FOLLOWING ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION
M A Finley, P A Anderson
University of Maryland; Baltimore, MD

ESTIMATION OF ACTIVATION PATTERNS FROM SURFACE MYOELECTRIC SIGNAL IN TKR PATIENTS DURING GAIT
M G Benedetti, P Bonato, F Catani, T D'Alessio, M Knaflitz,
A Battistine, A Leardini
Instituto Ortopedico Rizzoli; Bologna, Italy

EMG ACTIVITY EXHIBITED BY ACL DEFICIENT KNEE OF HIGH LEVEL SKIERS USING A CUSTOM-MADE KNEE BRACE DURING DOWNHILL SKING
M Lamontagne, G Nemeth, K S Tho, E Eriksson
Karolinska Hospital and Department of Sports Orthopaedics Surgery;
Stockholm, Sweden

A NEW TECHNIQUE TO IMPROVE REHABILITATION AFTER ACL-RECONSTRUCTION USING EMG OF THE VASTUS MEDIALIS MUSCLES
Th WiBmeyer, J Sterk, J Boos, L Kinzl
University Hospital; Ulm, FRG

ANALYSIS OF SIX DEGREES OF FREEDOM KNEE KINEMATICS OF ANTERIOR CRUCIATE LIGAMENT INJURED AND UNINJURED KNEES DURING LOCOMOTION
L Q Zhang, R Shiavi
Department of Biomedical Engineering, Vanderbuilt University; Nashville, TN

ELECTROMYOGRAPHIC ANALYSIS OF HABITUATION PROCESSES IN TREADMILL WALKING
J J Chen, I S Hwang, J J Liou, T C Huseh
Institute of Biomedical Engineering; Tainan, Taiwan

Friday Session D - 1015-1200 - Dogwood Room
Motor Unit Action Potentials
Chairpersons - Willemien Wallinga and Philip Parker

PRECISE ESTIMATION OF FIRING RATES OF MULTICHANNEL SURFACE ELECTOMYOGRAPHY: A BISPECTRAL APPROACH
J J J Chen, Y R Lin, T S Lin, Y N Jenp
National Cheng Kung University; Tainan, Taiwan

RANK-ORDERED REGULATION OF MOTOR UNITS
Z Erim, C J DeLuca, K Mineo
Boston University; Boston, MA

ACCURATE AND ROBUST ESTIMATION OF AVERAGE FIRING RATE OF SURFACE EMG INTERFERENCE SIGNAL
C T M Baten, H J Hamberg, H J Hermens
Roessingh Research & Development BV; Enschede, The Netherlands

KOHONEN MAPS FOR CLUSTERING MOTOR UNIT ACTION POTENTIALS
G Balestra, A Capra, E Pasero
Politecnico di Torino; Torino, Italy

REPRESENTATION OF MOTOR UNITS IN THE SURFACE ELECTROMYOGRAM

K Roeleveld, D F Stegeman, H M Vingerhoests, A S V Oosterom
University Hospital Nijmegen; Nijmegen, The Netherlands

MOTOR UNIT RECRUITMENT STRATEGY OF THIGH ANTAGONIST MUSCLES IN STEP-WISE AND RAMP INCREASING ISOMETRIC CONTRACTIONS

M Bernardi, M Solomonow, R V Baratta
Louisiana State University Medical Center; New Orleans, LA

APPLICATION OF THE METHOD OF SPECTRAL-STATISTICAL ANALYSIS EMG IN THE OBJECTIVE DIAGNOSIS AND FOR EVALUATING THE INFLUENCE OF MEDICAL PREPARATIONS IN THE TREATMENT OF PARKINSON'S DISEASE

O E Khutorskaya
Russian Academy of Sciences; Moscow, Russia

Friday Session E - 1330-1545 - Cypress Room

A Kaleidescope of Topics I

Chairpersons - Jan Clarijs and Dick Stegeman

ELECTROMYOGRAPHIC POWER SPECTRUM ANALYSES FROM DYNAMIC AND STATIC CONTRACTIONS

H Christensen, K Sogaard, B R Jensen, L Finsen, G Sjogaard
National Institute of Occupational Health, Copenhagen; Denmark

THE EFFECT OF ANKLE JOINT VISCOELASTIC PROPERTIES ON THE DYNAMIC RESPONSE OF THE CAT'S MEDIAL GASTROCNEMIUS MUSCLE

B H Zhou, R V Baratta, M Solomonow, R D'Ambrosia
Louisiana State University Medical Center; New Orleans, LA

LIGAMENTO-MUSCULAR REFLEX ARC IN THE SHOULDER

M Solomonow, C Guanche, T Knatt, R Baratta
Louisiana State University Medical Center; New Orleans, LA

MECHANICAL AND ANATOMICAL EQUILIBRIUM IN CYCLISTS

J P Clarijs, I Zinzen, P Van Roy
Vrije Universiteit Brussel; Brussels, Belgium

MOVEMENT-RELATED CORTICAL POTENTIALS AND THE SELECTION OF MOVEMENT

P Praamstra, D F Stegeman, Th. Oostendorp, M W I M Horstink, A R Cools
University of Nijmegen; Nijmegen, The Netherlands

THE EFFECTS OF SELECTED PARAMETERS OF MUSCLE ENERGY TECHNIQUE ON LUMBAR PARAVERTEBRAL MUSCLE ACTIVITY IN HEALTHY INDIVIDUALS

L Ash, R Bechtel, G Ford, M Kelly, M Marks, T Radov, S Strunk
University of Maryland; Baltimore, MD

FAR-FIELD POTENTIALS IN THE NEUROMUSCULAR SYSTEM

D F Stegeman, K Roeleveld, S L H Notermans
University Hospital of Nijmegen; Nijmegen, The Netherlands

ELECTROMYOGRAPHICAL FEATURE OF HUMAN REACTION ON INFLUENCE OF NON-IONIZING AND IONIZING RADIATION
E A Andreeva, O E Khutorskaya
Russian Academy of Sciences; Moscow, Russia

Friday Session F- 1330-1545 - Dogwood Room
A Kaleidescope of Topics II
Chairpersons - Keith Hayes and Yukio Mano

INFLUENCE OF BODY POSITION ON MOTOR EVOKED POTENTIALS OF THE LOW BACK MUSCLES
M A Lissens, A Decleir, G G Vanderstraeten
University Hospital of Ghent; Belgium

CERVICAL MUSCLE OVERUSE AS A FACTOR IN CHRONIC HEADACHE; A COMPARISON OF HEADACHE PATIENTS VS NORMAL SUBJECTS
S J Middaugh, J J Halford, W G Kee, J A Nicholson
Medical University of South Carolina; Charleston, SC

LONG TERM ASSESSMENT OF TRAPEZIUS MUSCLE LOAD USING AN AMBULATORY DEVICE
J Sandsjo, T Oberg, R Kadefors
Lindholmen Development; Goteborg, Sweden

ACTION CURRENTS OF FAST AND SLOW MUSCLE FIBRES
Henk Wolters, Willemien Wallinga, Dirk L Ypey, Herman B K Boom
University of Twente; Enschede, The Netherlands

LONG LATENCY EFFECTS OF TRANSCRANIAL MAGNETIC STIMULATION OF THE MOTOR CORTEX ON H-REFLEXES IN LATERAL GASTROCNEMIUS
D L Wolfe, K C Hayes, P J Potter
University of Western Ontario; London, Canada

MOTOR EVOKED POTENTIALS IN THE LOWER LIMBS OF NORMAL AND SPINAL CORD INJURED SUBJECTS
B A Conway, M Delargy, F Proudlock
University of Strathclyde; Glasgow, Scotland

CENTRAL MOTOR CONDUCTIVITY IN AGED PEOPLE
Y Mano, T Nakamuro, K Ikoma, T Sugata, S Morimoto,
T Takayanagi, R Mayer
Nara Medical University; Nara, Japan

H-REFLEX OF THE VASTUS MEDIALIS MUSCLE; A NEW DIAGNOSTIC TOOL IN THE TREATMENT OF RUPTURES OF THE ANTERIOR CRUCIATE LIGAMENT
Th WiBmeyer, P J Hulser, T Kutter, L Kinzl
University Hospital; Ulm, FRG

Abstracts

MYOELECTRIC CONDUCTION VELOCITY ESTIMATION ON VASTUS LATERALIS

J.-Y. Hogrel°, J. Duchêne° and J.-F. Marini*
° Université de Technologie de Compiègne, URA CNRS 858, Compiègne, France
* Université d'Aix-Marseille II et CNRS UPR 418, Marseille, France

INTRODUCTION

Since the last decade, surface electromyography (SEMG) has been used to tentatively provide useful information on action potential conduction velocity (CV). It is now well known that SEMG parameters are sensitive to several physiological and methodological factors, particularly with respect to electrode location (Sadoyama et al., 1985; Roy et al., 1986; Hägg, 1993) and to contraction level (Andreassen and Arendt-Nielsen, 1987). Moreover, these two factors concomitantly act on CV estimate because of volume conductor modifications with varying contraction level. These combined effects have not yet been investigated on vastus lateralis. Their study may give rise to methodological choices in order to obtain a reliable and consistent CV estimate.

METHODS

The experiments were carried out on one sedentary subject and consisted of a longitudinal study on the effects of electrode location on CV estimate with respect to varying contraction level. We assumed that measures on other subjects would lead to the same qualitative results.

The subject was asked to perform six short isometric contractions (3 s-duration) at 1, 20, 40, 60, 80 and 100% of the Maximal Voluntary Contraction (MVC). Besides, a myotatic reflex was induced to evaluate the volume conductor influence on CV estimate. Twelve reflex responses were measured for each location.

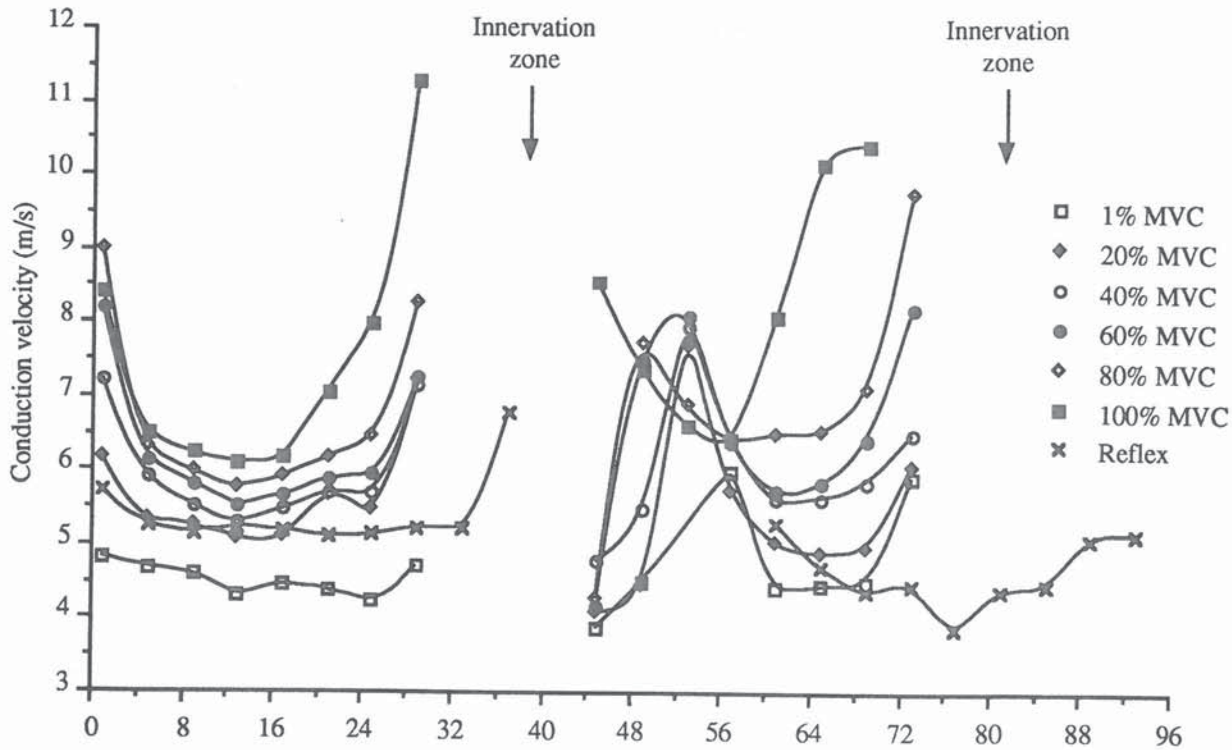


Figure 1. CV variations as a function of electrode location and contraction conditions (absolute values).

EMG signals were detected with 3 miniature Ag/AgCl surface electrodes (4 mm diameter) then amplified and digitized at 10 kHz. Interelectrode distance was held at 13 mm, center electrode common. The dominant leg was graduated every 4 mm from the distal to the proximal region. Three tests were performed for each location, the electrodes being randomly removed or not. After 2 or 3 rest days, the electrode bar was moved to the next location.

Time delay between signals, hence CV, was determined from the peak of the cross-correlation function (0.1 ms accuracy) on 1024 points-windows on which we assessed signal stationarity. 20 values were calculated for each second of the contraction (50% overlapping). CV estimates were accepted if the cross-correlation rate (R) was higher than 70% in voluntary contraction cases and 90% in reflex induced contractions. Within-location reproducibility was tested by ANOVA with repeated measurements. The gaussian nature of the CV distribution was evaluated using the Kolmogorov-Smirnov test.

RESULTS

Correlation rate variation with electrode location showed that the SEMG signal strongly fluctuates near its generation and extinction areas. This is due to the characteristics of these particular areas. This led to important variations in CV estimate as illustrated in Figure 1. The various contraction conditions outlined that CV estimate increased with contraction level. This could be explained not only by the usual justifications (like motor unit recruitment and firing rates) but also by modifications in volume conductor characteristics. The CV values and their variations with respect to time remained approximatively constant over the area of the distal region located between 4 and 16 mm. The width of this area decreased with increasing contraction level.

CV estimate showed a good reproducibility for any given electrode location and contraction modality. Except for very low contraction levels (1 and 20% MVC), CV estimate did not follow a gaussian distribution and had to be characterized by median and range rather than statistical mean and variance.

A cumulative CV distribution is given in Figure 2 for the location where CV was the least overestimated. The asymetric shape of the distribution is clearly demonstrated on the graph. We also verified that double differentiation lowered CV variability with respect to electrode location.

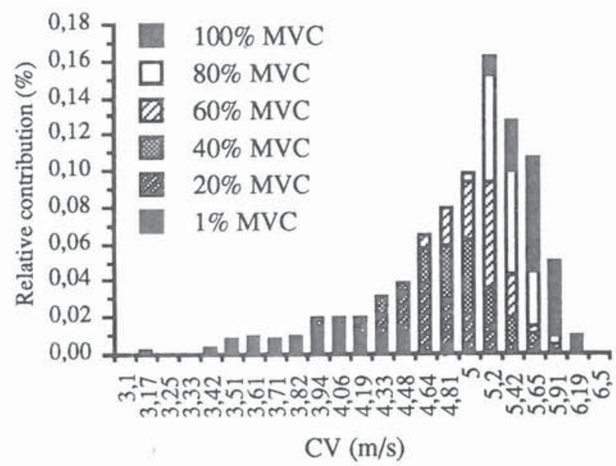


Figure 2: CV distribution for 12 mm location.

CONCLUSION

The CV estimate and its variation with respect to time are very sensitive to electrode location and contraction condition. This strong variability is due to the relative motion of myotendinous or active neuromuscular junctions and the electrode bar. Therefore, it seems difficult to adequately compare results obtained from various studies appearing through the litterature. There is a strong need for proper standardization methods based on morphological and methodological aspects.

REFERENCES

Andreassen, S., Arendt-Nielsen, L. (1987) J. Physiol., 391: 561-571.
Hägg, G.M. (1993) J. Electromyogr. Kinesiol., 3: 231-235.
Roy, S.H., De Luca, C.J., Schneider, J. (1986) J. Appl. Physiol., 61: 1510-1517.
Sadoyama, T., Masuda, T., Miyano, H. (1985) Med.&Biol. Eng.&Comput., 23: 339-342.

Jean-Yves Hogrel
Université de Technologie de Compiègne - Département de Génie Biologique, URA CNRS 858
BP 649 - 60206 COMPIEGNE Cedex - FRANCE

CHANGES IN THE EMG POWER SPECTRUM WITH INCREASING FORCE IN HAND MUSCLES

P. Noel, M. Bilodeau, A.B. Arsenault, E. Thibault, D. Gravel
School of Rehabilitation, Faculty of Medicine, University of Montreal and Research Center, Montreal Rehabilitation Institute, Montreal, QC, Canada, H3S 2J4.

SUMMARY

The aim of the present study was to characterize the behaviour of the EMG power spectrum, across force levels, of two hand muscles, the first dorsal interosseus (FDI) and the abductor digiti minimi (ADM). It was found that the median (MF) and mean power frequency (MPF) of the power spectrum presented a more pronounced increase, across force levels, for the FDI than for the ADM. This result could be explained by the greater difference between the diameter of type I and II fibers for the FDI.

INTRODUCTION

During an increasing muscular effort, the gradual recruitment of motor units with larger fibers (and higher conduction velocity (1)) has been associated with an increase in the MF and the MPF of the EMG power spectrum (2,3). Previous studies have shown that the FDI recruits motor units up to about 30-50% (4) of the maximal voluntary contraction (MVC), whereas the ADM has a larger recruitment range (i.e. up to 80% MVC) (5). If the increase in the average muscle conduction velocity associated with the additional recruitment of motor units (1) is the major mechanism explaining the increase in MF and MPF with increasing force levels then one could expect an increase in MF and MPF to about 30-50% MVC for the FDI and 80% MVC for the ADM. It was the aim of the present study to verify this hypothesis.

METHODS

Six male subjects (24.3 ± 4.1 years) volunteered for this study. Measures of skinfold thickness over the two muscles were taken with an adipometer. Abduction forces at the metacarpo-phalangeal joint of the second and fifth fingers were measured with a strain gauge transducer. Each subject produced three isometric ramp contractions, from 0 to 100% MVC in 5 sec. using visual feedback of the force. EMG signals were recorded with pairs of miniature surface electrodes (6-mm center to center), amplified (16-800Hz) and sampled at 2 KHz. The MF and MPF of power spectra were calculated from single 256-ms windows (Hamming window, 512 points, FFT) taken at every 10% MVC . The Root Mean Square (RMS) value of each window was also calculated to obtain the EMG/force relationship for both muscles.

RESULTS & DISCUSSION

Figure 1 (pooled data of all subjects) shows the EMG/force relationships for the FDI and ADM. It can be noted that the activity of both muscles increased linearly with force. Figure 2 illustrates behaviour of the MF across force levels for both muscles. It can be seen that: a) the increase in the MF (and MPF, not shown) of the FDI is more pronounced than that of the ADM, and b) a decrease in MF is present at higher force levels for both muscles (more pronounced for ADM). No significant difference was found between the skinfold thickness overlying the two muscles (Δ=0.78mm, $p>0.01$).

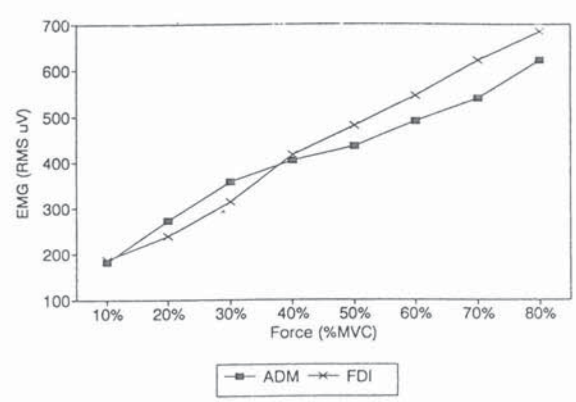


Fig. I : Average EMG/force relationship for ADM and FDI (n=6Ss)

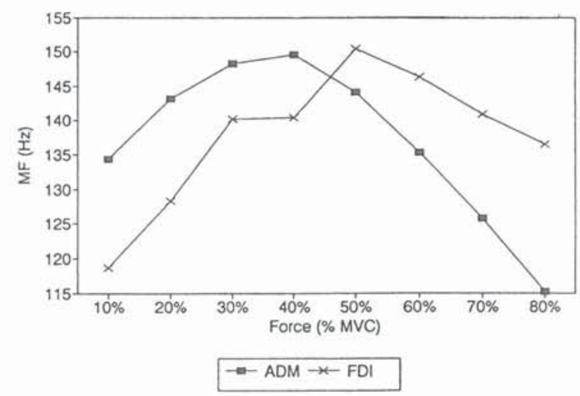


Fig. 2: Average Median (MD) frequency of the EMG power spectra for ADM and FDI (n=6Ss)

A more pronounced increase in the spectral statistics was expected for the ADM than for the FDI since this muscle possesses a larger recruitment range than the FDI. However, the more pronounced increase was observed for the FDI. These results suggest that other factors, besides recruitment (2), can determine the behaviour of the EMG power spectrum across force. One such factor might be the variation in fiber diameter of the different fiber types within a given muscle. For example, in the FDI the difference between type I and type II fiber diameter is three time larger than in the ADM (6). Consequently, the variation in average muscle conduction velocity would be larger in the ADM (1) along with greater changes in MF and MPF across force levels. The decrease in spectral statistics at higher force levels could possibly be explained by mechanisms such as synchronisation and fatigue.

REFERENCES

1. Masuda, T., De Luca, C. (1991) J. Electromyogr. Kinesiol. 1: 116-123.
2. Solomonow, M. et al. (1990) J. Appl. Physiol. 68: 1177-1185.
3. Bilodeau, M. et al. (1992) J. Electromyogr. Kinesiol. 2: 121-129.
4. Milner-Brown, H.S. et al. (1973) J. Physiol. 230: 371-390.
5. Tanji, J., Kato, M. (1973) Exper. Neurol. 40: 759-770.
6. Polgar, J. et al. (1973) J. Neurol. Sci. 19: 307-318.

AUTHOR'S ADDRESS

A. Bertrand Arsenault, Ph.D.
Research Center
Montreal Rehabilitation Institute
6300 Darlington Ave
Montreal, QC, Canada H3S 2J4
FAX: 514-340-2154

SURFACE AND WIRE EMG CROSS-TALK AMONG NEIGHBORING MUSCLES

M. Solomonow, R. Baratta, B. Zhou, M. Bernardi
Bioengineering Laboratory, Dept. of Orthopaedic Surgery, Louisiana State University Medical Center, New Orleans, LA, 70112, U.S.A.

SUMMARY

Surface and wire myoelectric activity of the medial gastrocnemius, lateral gastrocnemius and tibialis anterior of the cat were recorded during supramaximal stimulation of their nerves before and after the nerves of the LG and TA were cut. It was shown that the peak to peak (p-p) and mean absolute value (MAV) of cross-talk in the LG and TA did not exceed 5% of their maximal value for surface recording and 2.5% of maximal for wire recordings. Cross-talk values were appreciably higher, 20% and 16% in the LG and TA, in preparations where fatty tissue extended from the posterior of the knee to cover the MG and LG muscle. During increasing force contraction accomplished by orderly stimulation of motor units in the MG up to 100% of its maximal force the cross-talk in the LG and TA increased linearly up to the values indicated above. It was concluded that the cross-talk problem in surface recordings is negligible in most circumstances, and that EMG recording from muscles covered by adiposes tissue is not acceptable due to high cross-talk.

INTRODUCTION

The possibility that surface recordings of myoelectric activity from a muscle may be contaminated with signals from neighboring muscles is a controversial issue that has not been settled by recent studies. DeLuca and Merletti (1) as well as Koh and Grabiner (2) attempted to resolve this issue by recording EMG in normal adults subjected to surface electrical stimulation. But these studies are limited and deficient due to reflex activation of antagonists by stimulated tactile receptors as well as to inability to supramaximally stimulate the muscle due to pain sensation. High values of cross-talk (17%) were, expectedly, reported from neighboring muscles in the above reports. Wire recordings were shown to be relatively immune to cross-talk, as was described by Etnyre & Abraham (3). The possibility of fatty tissue, overlaying muscles in the subcutaneous space, as a major source of cross-talk was reported by Solomonow et al. (4), as a coincidental fact in a study of elbow flexors-extensors co-contraction study.

It is the objective of this report to determine, in definite values, the cross-talk in neighboring muscles, the behavior of cross-talk across the full range of force increase, and the effect of subcutaneous adipose tissue on cross-talk values.

METHODS AND RESULTS

Twelve adult cats were anesthetized with chloralose, and all leg muscles were denervated except the nerves of the MG, LG and TA. Bipolar cuff electrodes were placed on each of the muscle nerves. Surface recordings were made with bipolar electrodes (Ag-AgCl), 3 mm diameter and 7.5 mm center to center placed over the midline of each muscle from the motor point and distally. Wire electrodes were inserted into each muscle just underneath the surface electrodes.

Initially, all muscles were supramaximally stimulated, and the associated EMG in terms of peak to peak of M-waves and Mean Absolute Value were defined as 100%. The nerves to the LG and TA were then cut, and supramaximal stimulation repeated. EMG recordings of each muscle in that condition where normalized with respect to the maximal EMG recorded from them before their nerves were cut.

Finally, motor units of the MG were orderly stimulated (5) from 0 to 100% of the force, while EMG was recorded from each muscle. EMG cross-talk from the LG and TA was normalized, again, with respect to the maximal EMG of the respective muscle before the nerve was cut. Final dissection of the leg revealed if subcutaneous fatty tissue was present over the MG and LG muscles.

Data shows that during maximal activity of the MG, in the absence of fat, surface cross-talk values in the LG and TA are always less than 5% and wire cross-talk is less than 2%. In the presence of fat, cross-talk values are very high, in the LG, being up to 20%, but less than 2% in the TA which is not covered by fat.

During orderly stimulation of motor units in the MG, cross-talk values increase linearly in the TA and LG, reaching highest values of 5% in the LG and 2% in the TA. Wire electrodes cross-talk was also linearly increasing, but always less than surface EMG. Presence of fat did not introduce additional cross-talk in wire recordings.

CONCLUSION

The data and its analysis suggest that in the absence of fat, worst case cross-talk in surface recordings is always less than 5% of the maximal EMG of a muscle residing near another muscle which is fully active. In submaximal contractions the cross-talk is proportionally less. Cross-talk, therefore, could be neglected as a factor in most studies if the electrodes are of the appropriate size, interspacing and placed correctly over the muscle (6) with absence of fat. In the presence of fat, cross-talk in surface recordings is high, rendering any interpretation as unreliable. Wire recordings are nearly free from any cross-talk and do not suffer deterioration due to presence of fatty tissue.

ACKNOWLEDGEMENT

This work was supported by NSF Grant BCS 9207007.

REFERENCES

1. DeLuca & Merletti (1988) EEG & Clin Neurophys. 69:568-575.
2. Koh & Grabiner (1992) J. Orthopedic Research, 10:701-709.
3. Etnyre & Abraham (1985) 6th ISEK proceedings, page 8.
4. Solomonow et al. (1988) Exp. Neurology, 100:470-477.
5. Baratta et al. (1989) IRRR Ytsnd NMR 36:836-843.
6. Fuglevand et al. (1992) Biological Cybernetics 67:143-153.

AUTHOR'S ADDRESS

M. Solomonow, Ph.D.
Dept. of Orthopaedic Surgery
2025 Gravier Street, Suite 400
New Orleans, LA 70112
Phone: (504) 568-2251
Fax: (504) 568-4466

DESCRIPTION OF ELECTROMYOGRAPHIC LINEAR ENVELOPE FOR SYNERGY ANALYSIS IN ISOKINETIC TEST

M. Cesarelli, M. Pisaniello, F. Pizza
Electronic Engineering Department University Of Naples, Italy

(Keywords: electromyography, linear envelope, isokinetic test)

INTRODUCTION

During the last years the isokinetic methodology has become very significant to investigate the muscolar deficit and recovery. Isokinetic devices allow to study the motion of a joint only from a mechanical point of view, since their output variables typically are: the joint angular position; the angular velocity and the global torque output. In order to identify the contribution of single muscles the myoelectric signal can be used as a sign of muscle activation. Our aim is to decompose the linear envelope of EMG, using a modified version of the decomposition technique(developed by Shavi et alt.[4]), to get information about control and activation strategy of joint muscles. The correlation between this above mentioned information and the biomeccanic one allows us to determine correct and incorrect phasing of muscles. We have studied both healthy and pathological subjects in order to assess the differences between them.

METHODS AND RESULTS

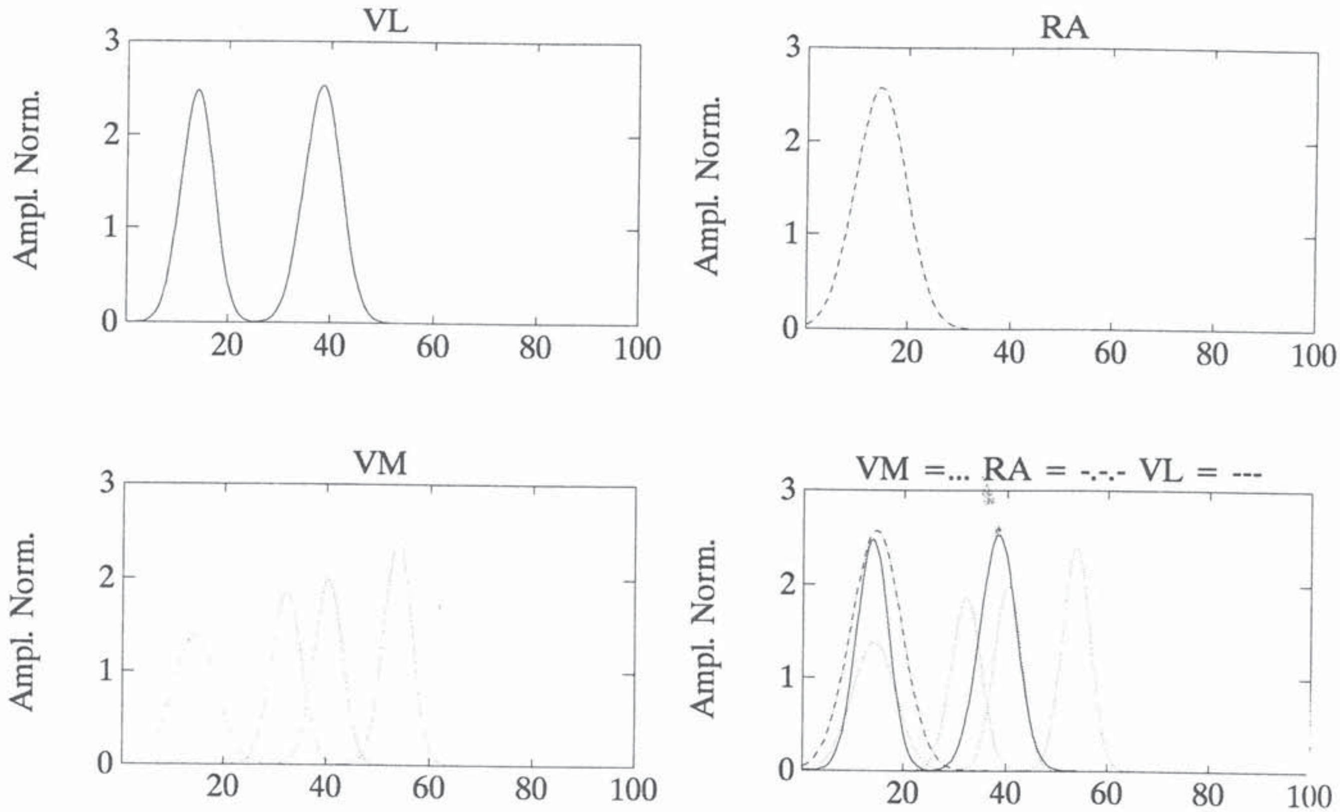
Our database consists of EMG signals of 20 healthy subjects and 10 pathological subjects. All subjects are men ranging between 18 and 45 years of age. Every man has executed a double isokinetic exercise, a cycle of 15 knee flexion-extensions, using the machine Lido Active in eccentric mode. In our database, EMG was sensed in a bipolar arrangement with surface electrodes and profiles have been developed from LE of three muscles of quadriceps: vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM). LE is generated through processes which are implemented in Matlab, a digital signal processing software. To remove motion artefacts and high frequency noise, a band pass filter FIR, using Hamming window cut-off frequencies of 40 and 400 Hz, was implemented. LE is obtained filtering low pass (10 Hz) the rectified EMG. To reduce the amplitude variability, due to variation of the skin resistance, the electrodes placements, etc., has been introduced an amplitude normalisation versus the maximum EMG level [5]. LE is modelled as a summation of unnormalized gaussian pulses of various lengths:

$$y(t) = \sum_{j=1}^{M} \alpha_i g_i(t-\mu_i) \qquad \mu_{i+1} - \mu_i > \mu_w \qquad g_i(t) = \exp\left(-\frac{t^2}{2\sigma_i^2}\right) \qquad |t| \le 2.4\sigma_i$$

where y(t) is the approximation of EMG LE over extension time; gi(t) is the gaussian basis function, μi is the time when activity occurs, μw indicates the minimum separation of the consecutive pulses, αi is the amplitude of the basis function and M is the number of the pulses. For extraction of temporal features we have implemented a modified version of the technique developed of Shiavi & oth. [4]. To reduce the complexity of the sinergic model, only the dominants phases of global activity from each LE are used. They are defined as the phases of activity which have a significant proportion of activity reflected in LE. In this work, the threshold of significativity is defined as the summation of total EMG activity (the area under the LE) divided by number of components minus one. In this way we have achieved two purposes: 1) to pay attention only to significant phases having strong weight in global activity of EMG; 2) to reduce sensibility to noise. The dominant phase having the highest amplitude is defined as peak activity of LE; the other dominant phases are defined major phases. Studying the normalisation method we have observed that VM is the muscle having the most regular course in the isokinetic test; instead, the RA is the muscle with great variability among subjects. In this work attention is focused particularly on VM. Healthy subject execute the first cycle of the test in many different ways; they don't show any preferential activation sequence. This is not significant because it could depend on the initial condition of the subject. Instead pathological subjects utilize mainly the most powerful muscle (RA), because of injuries of their articulation knee, which reduce the degree of freedom in movement control. In the central part of the test the differences between normal and pathological subjects are more evident. The pathological subjects prefer the VL, as first muscle in the activation sequence. The VL is, among the examined muscles, the richest of red fibres, so the most resistant to fatigue. We have observed that VM is never activated as first muscle. VM is the muscle showing the fastest and more consistent downfall of the muscular tone, when a traumatic event occurs on an anterior cruciate ligament. The extensions in final part of the test give prominence to muscular fatigue; the peak activities of all the three muscles tend to distribute on the whole extension interval. An important observation has been done about pathological subjects: on the whole test (15 extension) the VM amplitude component are very similar. We have observed also that pathological subjects tend not to alter the muscular control and activation strategy.

CONCLUSION

Analysis of LE EMG signals show that there doesn't exist an optimal activation strategy for all subjects, even if pathological subjects show a little range of solutions. The differences between the two populations make us think that research in this direction can be suggested to assess the pre and post-operative treatment of a trauma of the anterior cruciate ligament. The correlation of the force profile, recorded simultaneously with EMG signals, with the parametric variations of EMG phasic activity, can offer a way to investigate the origin of the different activation strategies in normal and pathological subjects. The origin may be seen in the evolution of movement control strategy, or in the modification of the muscle, considered as mechanical actuator, during isokinetic test.



REFERENCES

[1] M. J. CARLOTTO.: "Histogram analysis using a scale - space approach.", IEEE Trans. Pattern Anal. Mach. Intell., vol PAMI 9, pp. 121-129, Jan. 1987.

[2] M. CESARELLI, A. SPIEZIA, A. PEPINO, M. BRACALE, "The Electromyogram during isokinetic test", 9th International Congress of ISEK, Firenze, pp171.

[3] R. J. HATHAWAY : "A constrained EM algorithm for univariate normal mixtures ", J. Stat. Comput. Simulat., vol. 25, pp. 211-230, 1986.

[4] R. G. SHIAVI, J. J. CHEN, L. Q. ZHANG : " A Quantitative and Qualitative Description of Electromyographic Linear Envelopes for Synergy Analysis ", IEEE Trans. Biomed. Eng., pp. 9-18, vol. 39, n°. 1, Jan. 1992.

[5] J. F. YANG, D. A. WINTER : "Electromyographic amplitude normalisation methods: Improving their sensitivity as diagnostic tools in gait analysis ", Arch. Phys. Med. Rehabil., vol. 65, pp. 517-521, 1984.

AUTHOR'S ADDRESS

Mario Cesarelli
University of Naples
Electronic Department
via Claudio n.21
80125 Napoli
ITALY

A NEURAL NETWORK PATTERN RECOGNITION APPROACH TO SYNERGY ANALYSIS OF ELECTROMYOGRAPHIC DATA

Carole A. Tucker, H. John Yack, Scott C. White
Department of Physical Therapy and Exercise Science
State University of Buffalo, Buffalo NY.

INTRODUCTION

Pattern recognition techniques to analyze electromyographic (EMG) data have become increasingly sophisticated over the past decade. The majority of pattern recognition algorithms for movement analysis rely on statistical parameters of the LEEMG and consider a single muscle's activity. Synergy analysis requires the simultaneous study of multiple channels from muscles active during the movement. In addition to the statistical features, the underlying temporal relationships within and between channels are of importance. Pattern recognition in synergy analysis becomes increasingly complex as one attempts to define the underlying temporal parameters or features of the LEEMG, their interphasic relationships, and the classification scheme to be used. A neural network approach to pattern recognition requires no assumptions about data features nor the relative importance of different features, and the classification scheme need not be strictly defined. Network topology and training parameters are defined, and the network will settle to the appropriate classification algorithm by adjusting its internal parameters or weights. Neural networks can also process large inputs with relative ease in comparison to the more traditional approaches.The purpose of this study was to demonstrate the utility of a neural network approach for synergy analysis of electromyographic data.

METHODS

A single subject was studied during three different gait activities: walking at normal and fast speeds, and while simulating a pathology (ankle sprain). EMG activity, detected by surface electrodes, of five muscles: anterior tibialis (AT), extensor digitorum longus (EDL), medial gastrocnemius (MG), lateral gastrocnemius (LG), and soleus (SOL),and footswitch data were collected for 45 second periods during each activity. The raw EMG signal was processed and the resultant linear envelope was normalized in time to one gait cycle and in amplitude to the mean during the cycle. Footswitch data was used to determine the duration of each gait cycle. LEEMG's at 2% intervals for five strides from each of the 3 conditions were then chosen to be used in training and testing the network. The ensemble averages of each muscle's LEEMG were calculated to be used for comparison.The network was defined as a three layer, feedforward, backpropagation network, and was developed using Aspirin/MIGRAINES v6 Neural Network Software© (Russell R Leighton/Mitre Corporation) on a Sun4/SPARC computer system. The input patterns to the network consisted of the 5 muscle patterns for a single stride for a single condition and formed a 51*5 matrix. The hidden layer consisted of 25 sigmoidal nodes with outputs in the range of [0,1] fully connected to the input and output layers. The output layer consisted of three nodes whose values were in the range of 0 to 1.

An output pattern of [100] represented normal speed,[010] for fast speed, and [001] for the simulated pathology conditions. Training was accomplished by supervised learning using 9 of the 15 patterns. The network's learning rate was set to .35, and the error tolerance was 0.025.

RESULTS

The ensemble averages of the 5 strides for each muscle for a each condition had < 25% coefficient of variation (Fig 1). Network training took 600 iterations and less than 60 seconds to reach the specified error tolerance. The network was then tested on all 15 input patterns representing 5 strides from each of the 3 conditions. Six input patterns had not been used for training, and teh network demonstrated 93% correct assignment within the error tolerance for this test set, and 100% correct assignment of the patterns used for training. Only a single pattern, one from the normal condition, could not be defined within the error tolerance, but was correctly assigned to the normal pattern classification. Testing of the 15 patterns was accomplished in under 15 seconds.

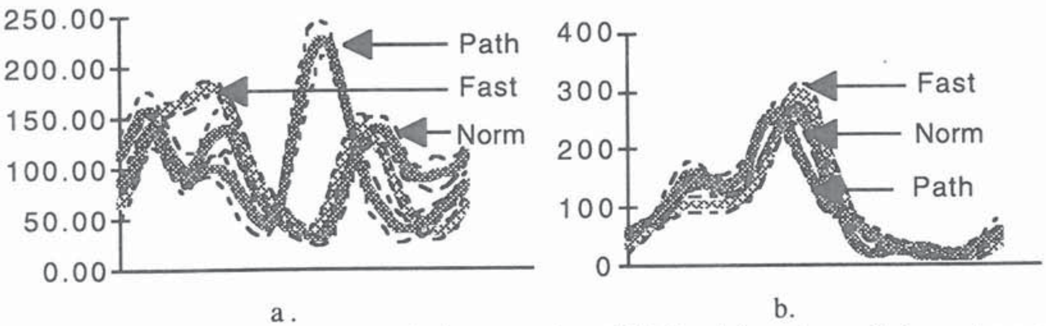


Figure 1: Ensemble averages of the anterior tibialis (a) and medial gastrocnemius (b) during gait demonstrating the variation of patterns for the three conditions

CONCLUSION

Synergy analysis has relied on more traditional methods for pattern classification (1), and the use of neural networks in the field of electromyography or kinesiology has been limited (2,3). This study demonstrates the use of a neural network pattern recognition approach for synergy analysis of EMG. This relatively simple network demonstrated excellent classification ability and was able to generalize its training to correctly recognize new patterns. Further study should include data from multiple subjects to test the robustness of the network's performance, as well as further optimization of the networks performance.

REFERENCES

1.Chen JJ, Shiavi R, Zhang LQ (1992)IEEE Trans BME39(1):9-18.
2.Sepulveda F, Wells D, Vaughan CL(1993) J Biomech 26(2):101-109.
3.Holzreiter S, Kohle, M (1993) J Biomech 26(6) 645-651.

AUTHORS ADDRESS

Carole A. Tucker, Department of Physical Therapy and Exercise Science, State University of New York, 410 Kimball Tower, Buffalo, NY 14214.Tel:716-829-2941x100/e-mail catucker@ubunix.buffalo.edu

QUANTIFICATION AND GRAPHICAL PRESENTATION OF MUSCLE COORDINATION PATTERN

Jiping He and Yang Wang
Dept. Neurosurgery, Thomas Jefferson University and Dept. EE, Temple University, Philadelphia

INTRODUCTION

What information could be extracted from surface electromyogram (EMG)? How to interpret the EMG information? And ultimately can EMG be quantified and provide concise and useful information for clinical diagnosis and evaluation of muscle coordination? These questions have been the core issue for study and application of EMG for many years. Using EMG for evaluation of muscle coordination has been applied for some stereotypical and cyclical movements such as gait. The data for that analysis, however, is so enormous that has been practically obstructing its own usefulness in clinical environment. The search for algorithms to quantify EMG for a consistent and reliable measure of muscle coordination has been extensive yet not very successful.

Surface EMG of individual muscles is notorious for its variability with many uncontrollable parameters. It was shown that EMG recorded simultaneously from all major muscles involved in a movement provided more information, and if processed appropriately might present the promising direction for development of a quantitative and objective measure of muscle coordination [1]. A computer program will automatically scale and analyze EMG from up to six muscles. The output from the program is a table of 24 numerical measures that will be presented in a graphic pattern for each movement. The major digression in that approach from the more traditional approach of quantifying EMG is that the relative measures with respect to a prime mover for the movement are used to minimize the fluctuation of EMG due to those uncontrollable parameters. The new progress in this direction will be discussed here.

METHOD

Ten healthy subjects and two patients with either multiple sclerosis or cerebral palsy participated in this study. Muscle coordination patterns of two functional movements using upper extremity were studied: feeding oneself and reaching for an object in front. Both movements started and ended in the same relaxed position with the hand laying on the lap. The subjects performed trials of the two activities at normal speed using arm on their dominant side. Surface EMG was recorded from six muscles of each subject: trapezius, anterior and medial heads of deltoid, biceps, triceps and brachioradialis. Bipolar electrode with preamplifier (gain=35, EMG-544, Therapeutic Unlimited, Iowa) was used. Data was recorded to computer disk using MIO-16 from National Instrument (16 bit resolution, 55K frequency for 16 channels).

Algorithms for scaling and analyzing EMG were developed using LabVIEW (National Instrument, TX). Four measures were calculated: intensity, duration, phase and centroid of muscle activity. The automatic detection of muscle activity ON and OFF times and calculation of intensity and centroid were developed and implemented. The ON and OFF times of a muscle's activity was detected by a modified moving window difference. The muscle activity intensity was calculated by taking the ratio between the total area under the EMG envelop and the duration of activity. The centroid was the time at which the muscle activity was equally divided for the movement. The phase was defined as the time difference between the movement onset and the muscle ON time.

Then the intensity and phase measures were scaled against the prime mover, anterior deltoid in our example, to obtain the relative intensity and phase measure for each muscle. The muscle activity duration and centroid were expressed as the percentage of total movement duration. Therefore a total of four numerical values was obtained as the relative activity measure of each muscle. This algorithm allowed us to generate a quantitative measure of each muscle's activity intensity, duration, centroid and phase relative to those of the prime mover of a movement.

RESULT

Even though the raw EMG data were reduced to four numerical scores for each muscle, a table contains scores for many muscles is still not a concise and intuitive presentation easily understandable. A graphical presentation for muscle coordination pattern is more desirable. Such presentation should be muscle based and contains all information about muscle activity for easy interpretation and comparison of both normal and abnormal patterns. We found that a spoked wheel type of graphical presentation satisfied most of our criteria. Each muscle could be represented by a spoke on the wheel. Four relative measures could be given by overlapping plots. Numerical scores could be shown by scales on the spoke.

With this new spoked wheel plot we present the summary muscle coordination pattern for the two movements we studied in Figs. 1 and 2. The average muscle coordination pattern from all ten subjects for the feeding motion is given in Fig.1. Six muscles are presented by six spokes. Each spoke is linearly scaled from -0.2 to 1.0 with zero indicated by the little circle in the center. Each score is measured from the zero circle inward or outward. The thick line without fill gives the relative intensity score for each muscle. The hatched pattern gives the duration. The crossed pattern gives the centroid. The dark dotted pattern in the center is the phase.

Figure 1. Muscle Coordination Pattern for Touch Movement

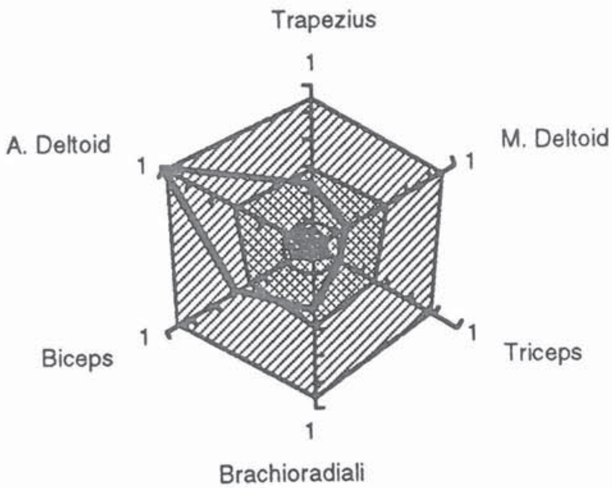
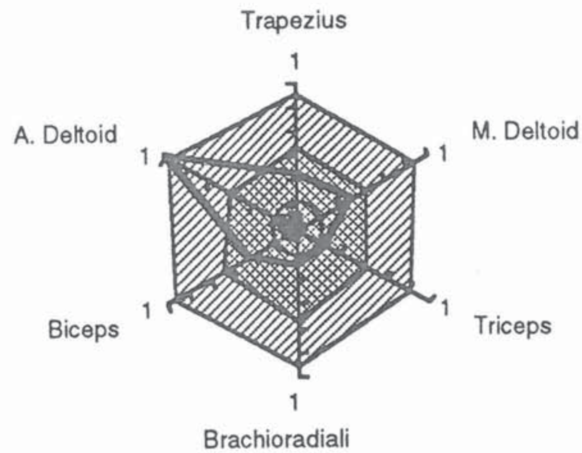


Figure 2. Muscle Coordination Pattern for Reaching Movement



The muscle coordination pattern for the reaching motion in our study is given in Fig.2. By comparing patterns in Figs. 1 and 2 we can clearly see that two different muscle coordination patterns are associated with the two movements studied. Both movements involve flexion of shoulder joint and occur in front of the body. The reaching motion requires more elevation of the arm and extension of the elbow joint. Therefore a larger activity intensity of deltoid muscle can be seen from Fig. 2 than that in Fig. 1. At the same time a smaller activity from biceps is presented in Fig. 2 with later occurrence of the centroid, presumably to brake the extension motion and to start flexing the elbow.

The muscle coordination pattern presented in this type of plot is easy to visualize and therefore can be readily used to diagnose and evaluate patients' muscle function. The new pattern plot for patients with motor disorder can be clearly shown to deviate from the normal pattern.

DISCUSSION

The novelty of the algorithm is two fold. The first is that by using relative measures the influence to EMG from many uncontrollable parameters, such as emotion, weather, skin surface, etc., is minimized in the final quantified scores. The absolute strength of each individual muscle is irrelevant in this measure. The second is the new graphical presentation of muscle coordination pattern. It is compact, concise yet comprehensive. The algorithm is automatically performed by computer program on a set of muscle EMG recordings to provide efficient and objective evaluation. The application of the algorithm and the graphical presentation in a clinical environment should be acceptable once the validity and consistency are established. This validity study is under way in a larger subject pool.

REFERENCE

[1] J. He, Y. Wang and M. Mesure. "A new representation for measure of muscle coordination pattern from EMG". Soc. Neurosci. Abstr., 1993

EMG AS A MEASURE OF MOTOR CONTROL IN MAN: TOWARD A BASIS FOR QUANTIFICATION

Arthur M. Sherwood, Ph.D., William J. Eaton, M.E.E., W. Barry McKay, B.S., and Noshir F. Kharas, M.S., Division of Restorative Neurology and Human Neurobiology, Baylor College of Medicine, Houston, Texas, USA

SUMMARY

While surface EMG recordings can provide important insights into patterns of motor control, there are many unresolved issues in their interpretation. Appropriate transformations are needed to maximize the information gained from their use. We present preliminary data regarding the relationship of surface EMGs and spasticity in spinal cord injured subjects and propose a theoretical framework for quantification and interpretation of surface EMGs as an index of motor control.

INTRODUCTION

We have used stripchart recordings of surface EMG to characterize abnormal motor control in subjects with upper motor neuron disorders (spinal cord injury, head injury, stroke) for the past twenty years in a protocol we have called *brain motor control assessment* (**BMCA**) [Sherwood and Dimitrijević, 1989]. These records provided important clues regarding the nature of abnormal motor control and were used to describe a new syndrome, "discomplete" spinal cord injury, i.e., subjects who are unable to move, but have neurophysiological evidence of residual brain control [Sherwood et al., 1992]. The observed patterns of motor unit activity have also been used to select appropriate intervention methods. We began continuously digitizing such EMG waveforms in 1989 to develop computer-assisted analysis. While this activity can be characterized by the site, duration, pattern and amplitude in single muscles and in its distribution across muscles, it can also be characterized by the total activity in response to each maneuver, which we have termed the motor control profile (MCP) of EMG activity. Although longitudinal studies in a single subject can be analyzed through self-normalization, there are no appropriate tools which permit quantitative analysis across subjects. This paper discusses an approach toward such analysis.

METHODS

Surface electrodes were placed over quadriceps, adductors, hamstrings, tibialis anterior and triceps surae muscles bilaterally, along with lower abdominal and lumbar paraspinal trunk muscles. EMG signals were amplified by 5000 over 50 Hz - 800 Hz and digitized at 1800 samples/s/channel. The results were continuously observed on the computer display and annotated by typing in protocol and extra-protocol events. In addition to the 12 EMG channels, two additional channels were allocated for position sensors, one channel was allocated for force transduction and one event channel.

Maneuvers were carried out with the patient supine on a comfortable, padded examination table. Following a 5 minute relaxation period, a standardized set of maneuvers was undertaken, each of which was repeated three times. The set consisted of: reinforcement maneuvers, attempted voluntary movements of the lower limbs, passive unilateral hip and knee and ankle movement, patellar and achilles tendon taps and clonus elicitation, vibration of quadriceps and triceps surae muscles and finally, stimulation of the plantar surface of each foot.

Envelope extraction and waveform compression to an effective sample rate of 20 s/s was accomplished by an RMS algorithm. The reduced data were analyzed by integrating each EMG channel over each subphase of each maneuver. Mean values for the three repetitions were computed. Activity was summed across all channels to form a single index of activity for each subphase, and plotted against the phase of the study to form the motor control profile.

RESULTS

The motor control profiles obtained for spinal cord injury (SCI) subjects were markedly different from those obtained for neurologically intact subjects. Even at such a coarse resolution, differences in motor control were distinctly evident, even in the absence of specific transformation factors.

Repeated studies in subjects yielded useful results. In a review of 9 SCI subjects who underwent an experimental surgical treatment of spasticity, the overall index of activity correlated with self-reported changes in spasticity better than did any of the clinical measures, matching both increases and decreases in eight of the nine subjects. We have also compared a profile of activity derived from the studies with clinical assessment of subjects and have found a correlation between clinical measures of spasticity and the EMG measures. The tibialis anterior EMG activity following plantar stimulation over 40 studies correlated with the clinical scoring of withdrawal reflexes ($r^2 = 0.69$).

DISCUSSION

The obtained results support the utility of surface EMG data as a means of representing motor control. While motor control can be defined on different levels [Loeb, 1987], in the absence of voluntary movement in humans, the choices for measurement are restricted. Motor control of the limbs first exists as a tangible entity at the spinal segmental level, where afferent and efferent paths converge. We propose that <u>the pattern of activity of the spinal motor centers should be considered as the "gold standard" for studies of motor control in individuals with impaired voluntary movements secondary to upper motor neuron dysfunction</u>. A transformed version of that information can be derived from the EMG data. The motor control profile proposed here is an attempt to quantitatively describe features of that motor control. Ultimately, it will be necessary to define the limits of accuracy imposed by individual variations in the transformation of motor neuron firings to surface EMG signals. The analysis done thus far assumes that an identity transformation matrix can be used to map surface EMG signals into pooled firing rates. Under sufficiently general conditions, the problem may be reduced to an estimation of the magnitude (between 0 and 1) of the elements of a matrix, and to the determination of how such elements might be dependent on factors such as which muscle, age weight, etc.

CONCLUSIONS

Interpretation of surface EMG data for studies of motor control should be based on transformation to a standardized reference system. We propose that the pooled firing rate is an appropriate measure of motor control, and that efforts should be devoted to the identification of the limits of accuracy which can be expected from such methods.

REFERENCES:

Loeb GE: Motor Control, in, Adelman G (ed), *Encyclopedia of Neuroscience*, Boston: Birkhäuser, 1987.

Sherwood AM, Dimitrijević MR: Brain Motor Control Assessment, *Proceedings*, 11th International Conference of the IEEE EMBS, 11:943-944, 1989.

Sherwood, A.M., Dimitrijević, M.R., and McKay, W.B., Evidence of subclinical brain influence in clinically complete spinal cord injury: discomplete SCI, *J. Neurol. Sci* 110:90-98, 1992.

ACKNOWLEDGMENTS:

The author gratefully acknowledges the support of the Vivian L. Smith Foundation for Restorative Neurology, and the Rehabilitation Research and Development, Department of Veteran's Affairs.

ADDRESS

Arthur M. Sherwood, P.E., Ph.D., Associate Professor and Associate Head for Academic Affairs
Division of Restorative Neurology and Human Neurobiology,
Baylor College of Medicine,
One Baylor Plaza, S809
Houston, TX 77030-3498
Telephone (713) 798-5153, Telefax (713) 798-3683
email: ams@bcm.tmc.edu

EFFECTS OF MUSCLE FIBER TYPE AND SIZE ON EMG MEDIAN FREQUENCY

S.H. Roy, E.J. Kupa, S.C. Kandarian*, C.J. De Luca
NeuroMuscular Research Center and *Department of Health Sciences, Boston University, Boston, MA, U.S.A.

SUMMARY

An *in vitro* experiment using isolated neuromuscular preparations of different fiber type composition was conducted in rat muscles to investigate the influence of muscle fiber-type characteristics on EMG median frequency (MDF). Muscles with higher percentages of fast fibers exhibited higher initial values and a greater rate of decrease of MDF during fatigue. Initial values of MDF (IMF) were significantly related to average muscle fiber area weighted according to the percentage of fast fibers. These findings demonstrate that EMG median frequency is influenced by the combined effects of muscle metabolic and morphologic properties.

INTRODUCTION

The specifics of the relationship between the EMG signal and the physiological and anatomical properties of muscle fibers remain unexplained. Muscle fiber-type characteristics are believed to influence the behavior of the EMG power spectrum during sustained contractions. Changes in EMG spectral parameters during a sustained contraction have been related to muscle fiber composition, however these were human *in vivo* studies that could not properly control for confounding physiological factors (Gerdle et al. 1988, Moritani et al. 1985). Also, *in vivo* studies on whole muscles in humans have not found a linearly increasing relationship between fiber size and EMG spectral parameters or conduction velocity (Wretling et. al. 1987), despite the fact that these relationships are well established in single muscle fibers (Hakansson 1956). These findings demonstrate the need for *in vitro* studies on whole muscle to precisely measure the interrelationships between EMG parameters and muscle fiber characteristics under isolated conditions.

METHODS

The soleus (SOL), extensor digitorum longus (EDL) and costal diaphragm (DIA) muscles (n=8 for each muscle) were excised with their motor nerves intact from female Wistar rats (wt. 150–175 g). The neuromuscular preparations were placed in an oxygenated bath of Krebs solution maintained at a constant temperature of 26°C and a constant pH of 7.2. The nerve was electrically stimulated at a frequency of 40 Hz and a pulse width of 0.2 ms. One 5 s contraction and two 20 s contractions were recorded for each muscle. Rest periods of 5 min were interposed between contractions. M–wave signals were detected using an electrode unit (interelectrode distance = 4.6 mm) positioned on the underside of the muscle preparation in contact with the muscle fascia and differentially amplified and bandpass filtered (20–2000 Hz). The MDF was calculated in software giving a frequency resolution of 2 Hz. The muscle was immediately removed from the bath and frozen (–20 °C) at optimal length. Fibers were typed as SO, FOG and FG based on mATPase, SDH and α–GPDH stains, and muscle fiber cross-sectional areas were calculated (fibers counted per muscle [±SD] was 421±115).

RESULTS AND DISCUSSION

Mean values (n=8) and standard error (SE) of normalized MDF are plotted with respect to the 20 s duration of the contraction in Fig. 1. Values were normalized to the first 0.50 s epoch. During the course of the contraction, the

MDF decreases at different rates according to the muscle's content of fast (glycolytic) fibers. Not shown is the finding that muscles with higher percentages of fast fibers exhibited higher IMF values.

Mean cross-sectional areas (μm2) were different across muscle groups for each of the fiber-types, with the SOL muscle having the largest cross-sectional areas for the SO and FOG fibers (3690±1232 and 2405±563) and the EDL muscle having the smallest cross-sectional areas (930±215 and 1033±386). The DIA muscle contained slightly larger FG fibers on average than the EDL muscle.

We did not find a linearly increasing relationship between average muscle fiber cross-sectional area and IMF. However, when a weighted average fast fiber area (wAREA) was calculated for each individual muscle (n=24) according to the following formula: wAREA = (% by area of fast fibers)(Average muscle fiber area), a highly correlated and linearly increasing relationship (r=0.92 and p<0.001) was present (Fig. 2).

The results of this study demonstrate that the behavior of the EMG median frequency during a sustained contraction is influenced by the combined effects of muscle fiber size and muscle fiber type.

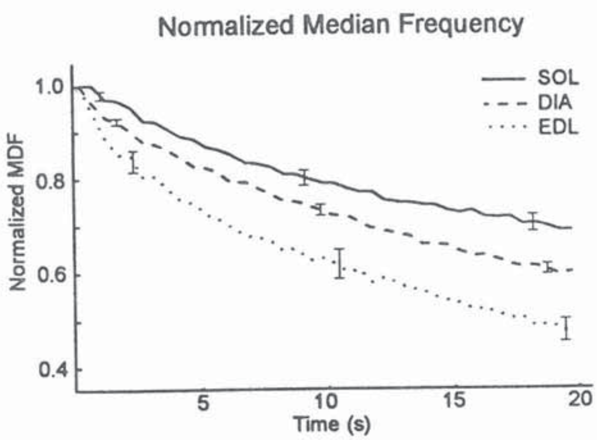


Figure 1. Normalized mean MDF (±SE) values are plotted with respect to contraction duration (n=8 for each muscle group).

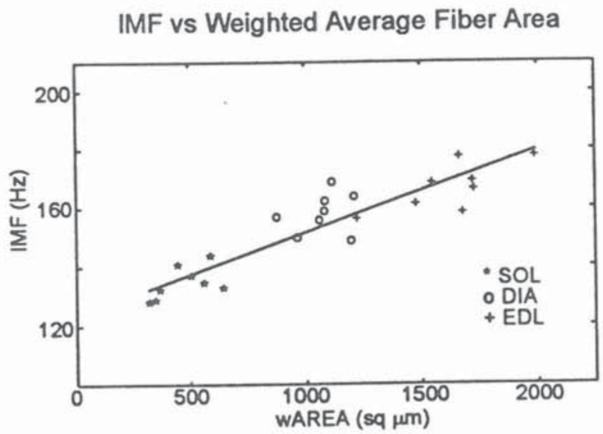


Figure 2. Whole muscle IMF values are plotted with respect to wAREA for each muscle specimen (n = 24, r = 0.92, p < 0.001).

REFERENCES

1. Gerdle, B., Henriksson-Larsen, K., Lorentzon, K., Wretling, M.L. (1991) Acta Physiol. Scand., 142:457-465.
2. Hakansson, C.H. (1956) Acta Physiol. Scand., 37:14-34.
3. Moritani, T., Gaffney, F.D., Carmichael, T., Hargis, J. (1985) In: Biomechanics IX-A, Vol 5a:287-292, Human Kinetics Publishers, Champaign, Il.
4. Wretling, M.L., Gerdle, B., Henriksson-Larsen, K. (1987) Acta Physiol. Scand., 131:627-628.

AUTHOR'S ADDRESS

Serge H. Roy, Sc.D.,P.T.
NeuroMuscular Research Center, Boston University
44 Cummington Street, Boston, MA, 02215, USA
tel: (617) 353-9633 fax: (617) 353-5737 e-mail: ROY@BUNMRG.BU.EDU

EMG LATENCIES IN SEVEN POSTURAL MUSCLES FOLLOWING POSTURAL PERTURBATION IN YOUNG ADULTS

G.V. Smith and P.A. Anderson
Department of Physical Therapy, School of Medicine,
University of Maryland, Baltimore, MD, U.S.A.

SUMMARY

Latency periods for perturbation-induced electromyographic (EMG) activity were determined for 7 muscles following postural challenges. Normal young adults (n=10) participated in the study. The results, subjected to a 3-way ANOVA and a post hoc statistical analysis, indicated a caudal to cephalic progression in the appearance of the EMG activity following anterior postural perturbations. Interestingly, the data revealed that the recruitment pattern changed following posterior postural perturbations, beginning, now, with the rectus abdominus and proceeding thereafter both caudally and cephalically. While the significance of this transformation remains unclear, it is not incompatible with the notion that postural reflexes may be modified by direction-specific afferent input. Based on the present data, it may also be suggested that individuals, particularly elderly individuals, with weak abdominals may be at increased risk of injury due to an unexpected posteriorly directed disruption of their base of support.

INTRODUCTION

Balance difficulties leading to falls are an all too common cause of injury, particularly among the elderly. Given the increasing number of senior citizens in this country and the attendant increases in health care costs for this segment of the population, the problem of falls clearly warrants further research into the mechanisms regulating equilibrium. Throughout the developed world, investigators in the field of balance have adopted a technique pioneered, among others, by Lewis Nashner (1, 2). This technique, which utilizes a moveable platform, has contributed much to our understanding of the biomechanics of human postural stability. For example, the function of the musculo-skeletal system in maintaining balance has been well studied using this technique. Unfortunately, the neural mechanisms involved in preserving equilibrium remain, at best, poorly understood. This is particularly true of the role of afferent input in integrating or modifying cortical and subcortical postural reflexes. The purpose of this investigation was to begin to amass the technology required to correlate movement-induced afferent neural input with subsequent motor output in humans. In the present study, we report on EMG latency periods obtained from 7 muscles following postural perturbation in both the anterior and posterior direction.

SUBJECTS AND METHODS

Ten healthy subjects, five females and five males (mean age 22.8 and 29.2 years respectively) participated in the study. The subjects were informed of the goals and purposes of the study and subsequently gave informed consent.

A BTE MotionSpec[R] Balance and Motor Control System (Baltimore Therapeutic Exercise Company, Baltimore, MD) was used to perturb the subject's balance in both the anterior and posterior direction. Platform movement parameters are computer controlled. In the present study, the movement kinematics were 10 cm displacement, 40 cm/sec velocity and 125 cm/sec^2 acceleration. These parameters have been demonstrated elsewhere to elicit a consistent postural response to perturbation (3). Three trials were recorded and analyzed later for each subject.

EMG activity was recorded using a MacLab[R] Data Recording System while the subject stood on the MotionSpec[R] platform. Surface EMG electrodes were fabricated for this study by Carlo DeLuca (Boston University). They had a built-in differential amplification of 10, a bandwidth of 713 Hertz and a common mode rejection ratio of 88 decibels. Electrodes were placed over the muscle bellies of the following muscles: 1.) tibialis anterior (TA), 2.) gastrocnemius (GN), 3.) medial hamstrings (MH), 4.) vastus lateralis (VL), 5.) rectus abdominus (RA), 6.) lumbar erector spinae (ES) and the 7.) sternocleidomastoid (SCM). Ground electrodes were placed on the fibular head and the clavicle. To determine background EMG activity, 50 msec of quiet standing was recorded prior to the onset of platform movement. The EMG recording was continued for a minimum of 500 msec after platform movement commenced. The raw EMG signal was integrated and rectified by the MacLab[R] system's software. The latency period for EMG activity was defined as the time from the onset of platform movement to the onset of EMG activity. The latency period for movement-elicited EMG activity was determined by means of a fixed threshold measure. The threshold was determined by a modification of the method of Studentski et al. (4). Time measures and EMG activity were recorded in 5 msec intervals.

The data was subjected to an ANOVA. In addition, a Student Newman-Kuels Post Hoc analysis was used to show specific between and within group differences. An alpha level of 0.05 was set for all statistical tests.

RESULTS

The following latencies were obtained after anterior perturbations (mean $\pm$ S.D.): 1.) TA 119.54 $\pm$ 13.31 ms, 2.) GN 166.87 $\pm$ 36.24 ms, 3.) MH 156.00 $\pm$ 44.33 ms, 4.) VL 172.22 $\pm$ 78.56 ms, 5.) RA 180.45 $\pm$ 64.04 ms, 6.) ES 213.33 $\pm$ 80.77 ms and 7.) SCM 224.28 $\pm$ 121.94 ms.

Following posterior perturbations, the following latencies were obtained (mean $\pm$ S.D.): 1.) TA 216.47 $\pm$ 97.27 ms, 2.) GN 151.11 $\pm$ 58.62 ms, 3.) MH 208.82 $\pm$ 91.21 ms, 4.) VL 185.00 $\pm$ 70.24 ms, 5.) RA 140.00 $\pm$ 53.61 ms, 6.) ES 289.70 $\pm$ 126.62 ms and 7.) SCM 305.00 $\pm$ 151.60 ms.

A 3-way ANOVA demonstrated both within and between subject differences, as well as, interaction differences. For example, latency periods as a function of the direction of perturbation were significantly different ($F=7.3346$, $p=.0075$). Similarly, there were statistical differences in individual muscle latencies ($F=5.1253$, $p=.00008$) and in the interactions between latencies times and directions ($F=2.4174$, $p=.0292$). A Student Newman-Kuels post hoc test showed no differences when the anterior perturbation were considered. There were, however, multiple latency period differences when the posterior perturbations were analyzed. There were also interaction differences.

DISCUSSION

Supporting the previous work of Nashner (1,2), the data recorded during the anterior postural perturbations demonstrated a caudal to cephalic progression in the activation of the targeted muscles. Unexpectedly, however, the results of the posterior perturbations revealed that EMG activity began in the rectus abdominus and progressed both caudally and cephalically thereafter. Given the heterogeneity and size of our subject population it was anticipated initially that establishing statistical differences in muscle latencies would be highly unlikely. Surprisingly, this was not the case. While there were no significant differences between muscle latencies following anterior perturbations there were following posterior perturbations. These results may indicate a major yet heretofore unrecognized role of the rectus abdominus in maintaining posture and balance. It may also be suggested, based on these results, that within the elderly population, those individuals with weak abdominals, may be at increased risk of injury due to falls. Further study of this issue appears warranted.

REFERENCES

1. Nashner, L.M. (1980) J. Neurophysiol. 44:650-664.

2. Nashner, L.M. (1982) Trends Neurosci. 2(5):536-544.

3. Anderson, P.A., Smith, G.V., Alon, G. (1994) J. EMG Kines. (Submitted)

4. Studenski, S., Duncan, P.W., Chandler, J. (1991) J. AM. Geri. Soc. 39:229-234.

EMG AND ANGULAR MOTION OF THE KNEE JOINT IN THE WALKING GAIT ANALYZED BY USE OF THE FAST FOURIER TRANSFORM

B.R. Brandell and C.L. Rice, Department of Anatomy, University of Saskatchewan, Saskatoon, SK Canada and School of Physical and Occupational Therapy and Department of Anatomy and Cell Biology, McGill University, Montreal, PQ Canada.

INTRODUCTION

The objectives were to: 1) compare level and upslope treadmill walking with respect to angular motions of the lower limb, and the simultaneously recorded EMG from the lower limb (antigravity) muscles; 2)explore the use of the Fast Fourier Transform (FFT) equation series as a method for mathematically defining curves plotted from recurring cyclic data as in the stride cycles of unimpaired walking gaits; 3) provide a model for testing the statistical significance of differences in walking gait parameters generally; and 4) explore the relationship between mathematically defined sinusoidal motion and lower limb angular motions during walking gait. Earlier studies which recorded EMG during upslope walking did not examine the concurrent lower limb motions (1,2). Brandell and Williams (3) used FFT to profile intersubject variability in lower limb angular motions, and as a smoothing technique to differentiate position into velocity and acceleration curves.

METHODS

Six (6) healthy male subjects walked on a treadmill at a constant speed of 3.1 mph (5 kmph) at 0° (level) , 3°, 6° and 9° of upward tilt, while motion was recorded at 50 fps (Bolex, Model #16 M camera) and while bipolar EMG recordings were made simultaneously with Beckman silver-silver chloride surface electrodes from the midbelly regions of the vastus, hamstring and gastrocnemius muscles. For the soleus muscle two electrodes were placed with a proximal distal orientation over the medial surface area of its belly between the medial head of gastrocnemius and the tibia. The EMG and footswitch data from the heel, ball and toe of the ipsilateral foot were stored on magtape and a Science Accessories Corp. sonic digitizer was used to digitize X - Y coordinate positions of points marked on the lower limbs, and a Dec 20-60 computer was then used to plot graphic curves of the EMG and position data versus time, and to make use of a FFT package from the Institute of Mathematical and Statistical Libraries (IMSL) to make visual and statistical comparisons between harmonic curves plotted in the time domain.

RESULTS

Only knee positions changes are reported in this paper. The main changes in the knee position data curve with an increased upward angle of treadmill tilt were in the amplitudes of its peaks: knee extension during swing phase before Heel Strike (HS) was decreased, but was increased in the stance phase before Toe Off (TO); and the angle of stance phase knee flexion was increased (deeper) because of the reduced swing phase knee extension.

These alterations are represented by the statistically significant later phasing of the first (fundamental) harmonic, and increased amplitude of the second harmonic of the knee position data curve (six strides for each of five (5) subjects).

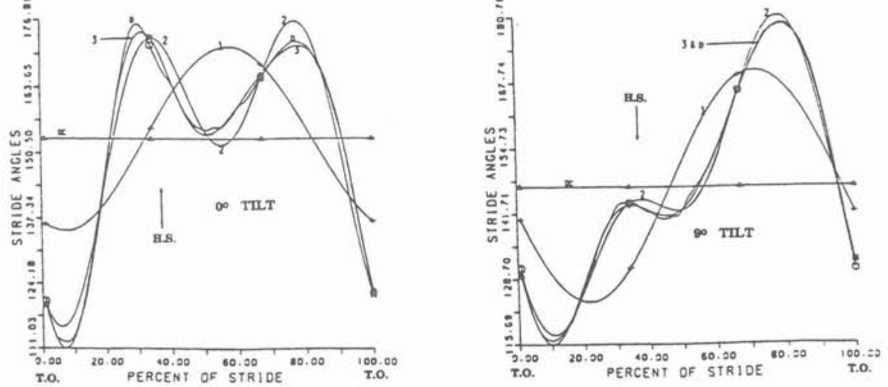
Upward tilt resulted in a statistically significant increase of mean EMG amplitudes for all four muscles, and statistically significant shifts to later phasing of the basic envelopes (first harmonics) of the vastus and hamstring EMG (8 strides for each of six (6) subjects).

CONCLUSIONS

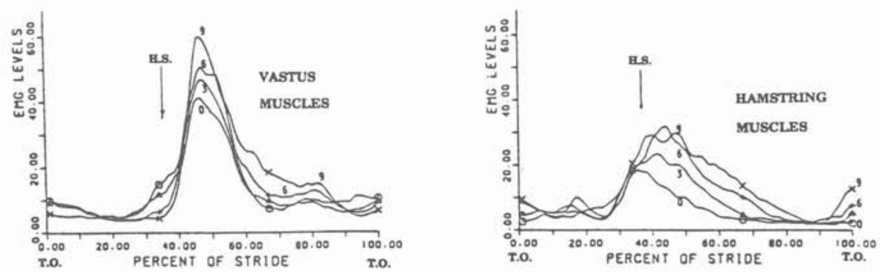
1. The Fast Fourier Transform is an effective means of statistically assessing amplitude and phasing changes in angular motions, as illustrated by the knee joint, and in the simultaneous EMG of related muscles.

2. The increased EMG amplitudes, and continuations of EMG into the later stance phase are related to greater ranges of flexion and extension of the knee, ankle and hip joints, during upslope walking.

3. Since the plotted position curves, which represent the sequential motions of the knee joint during the walking gait stride cycle, can be almost completely reconstructed by combining their first two harmonic sinusoid curves, it appears that the knee joint motion is itself basically sinusoidal, and that the muscle contractions are phased to maintain this motion characteristic.



Harmonic reconstructions of the complete plotted curve of **knee joint angles** during a typical walking stride cycle on a moving treadmill. 180° would represent complete knee extension 1 = first (fundamental) harmonic curve; DC = Direct Current voltage level, which is used as a zero baseline about which the harmonic sinusoidal curves fluctuate like an alternating (AC) current; 2 = composite of harmonics 1 + 2; 3 = composite of first 3 harmonics; D = complete curve of plotted data. HS = Heel strike; TO = Toe off. Each curve is an average of 6 strides from one representative subject.



EMG data curves at the **four conditions of treadmill tilt:** 0,3, 6 and 9 degrees; EMG levels are normalized to an arbitrary maximum voltage, represented as 100. Each curve is an average of 8 strides from one representative subject.

REFERENCES

1. Akihiro, T., Nagashima, H., Takechi, H. (1985) Arch. Phys. Med. Rehabil. 66:610-613.
2. Brandell, B.R. (1977) Am.. J. Phys. Med.. 56(2):59-74.
3. Brandell, B.R. and Williams K. (1974). Interntl. Series on Sport Sciences, 1, Biomechanics IV:91-97.

AUTHOR'S ADDRESS

Bruce R. Brandell, Ph.D., Department of Anatomy
University of Saskatchewan, Saskatoon, SK Canada
S7N OWO Fax: (306) 966-4298

AGE-RELATED CHANGES TO OPEN-LOOP AND CLOSED-LOOP POSTURAL CONTROL MECHANISMS

J.J. Collins*, C.J. De Luca*, A. Burrows+, L.A. Lipsitz+
*NeuroMuscular Research Ctr., Boston University, Boston, MA, U.S.A.; +Harvard Medical School and Hebrew Rehab. Ctr. for Aged, Boston, MA, U.S.A.

INTRODUCTION

Postural instability in the elderly is a poorly understood, multifactorial phenomenon. The results of earlier studies analyzing postural stability among individuals of different age groups have been ambiguous — some have reported age-related differences in postural sway (1), whereas others have found none (2). Additional work is needed to understand age-related changes to postural control mechanisms.

In an earlier study (3), we examined quiet-standing center-of-pressure (COP) trajectories as one-dimensional and two-dimensional random walks. These analyses revealed that over short-term intervals of time during undisturbed stance open-loop control schemes are utilized by the postural control system, whereas over long-term intervals of time closed-loop control mechanisms are called into play. Our approach, known as stabilogram-diffusion analysis, has the advantage that it leads to the extraction of repeatable COP parameters which can be directly related to the resultant steady-state behavior and functional interaction of the neuromuscular mechanisms underlying balance control.

In the present study, we utilized stabilogram-diffusion analysis to examine how the aging process affects the operational characteristics of the open-loop and closed-loop postural control mechanisms.

METHODS

COP time series were collected for 25 healthy young males (age: 19–30 yr, mean 22 yr) and 25 healthy elderly males (age: 71–80 yr, mean 75 yr). The subjects were of similar height and weight. They had no evidence or known history of a gait, postural or skeletal disorder. Each subject was instructed to stand in an upright posture on a force platform for a period of 30 s. Ten trials were conducted for each subject with his eyes open.

The COP trajectories were studied according to stabilogram-diffusion analysis. A plot of mean square COP displacement versus time interval is known as a stabilogram-diffusion plot. Stabilogram-diffusion plots were computed for each subject trial and then ten such curves were averaged to obtain a resultant stabilogram-diffusion plot for a particular subject. Stabilogram-diffusion analysis involves the extraction of three sets of posturographic parameters: diffusion coefficients, scaling exponents, and critical point coordinates (3). The short-term and long-term diffusion coefficients characterize the stochastic activity of the open-loop and closed-loop postural control mechanisms, respectively, whereas the corresponding scaling exponents quantify the level of correlated activity in such control schemes. The critical point coordinates approximate the transition region over which the postural control system switches from open-loop control to closed-loop control.

Standard statistical analyses were used to compare the results for the two populations.

RESULTS

Firstly, the short-term diffusion coefficients for the elderly subjects were significantly larger than those for the young subjects. Secondly, the short-term (long-term) scaling exponents for the aged subjects were significantly larger (smaller) than those for the young subjects, i.e., the increments in the COP trajectories for the aged subjects were more positively (negatively) correlated over the short-term (long-term) regions. Finally, the critical point coordinates—the critical time interval and critical mean square displacement, respectively—for the elderly population were significantly greater than those for the young population.

DISCUSSION

The aforementioned short-term diffusion coefficient calculations suggest that the overall stochastic activity of the open-loop control mechanisms increases in the elderly. Moreover, the finding that the short-term scaling exponents for the elderly subjects were larger than those for the young subjects suggests that the steady-state behavior of the open-loop control schemes is more positively correlated and therefore perhaps more unstable in the aged, i.e., the output of the overall system moves more quickly away from a relative equilibrium point. On the other hand, the finding that the long-term scaling exponents for the elderly subjects were smaller than those for the young subjects suggests that the steady-state behavior of the closed-loop feedback mechanisms is more negatively correlated and therefore perhaps more tightly regulated in the aged, i.e., following a perturbation, the output of the overall system is more quickly restored to a relative equilibrium point. Finally, the above critical point coordinate calculations suggest that the temporal and spatial characteristics of the interaction of the open-loop and closed-loop control mechanisms change as a function of age — in the elderly, for example, open-loop control schemes are utilized for longer time intervals and correspondingly larger COP displacements during periods of undisturbed stance.

REFERENCES

1. Overstall, P., Exton-Smith, A., Jimms, F., Johnson, A. (1977) Brit. Med. J., 1:261–264.
2. Fernie, G.R., Gryfe, C.I., Holliday, P.J., Llewelyn, A. (1982) Age Aging, 11:11–16.
3. Collins, J.J., De Luca, C.J. (1993) Exp. Brain Res, 95:308–318.

Supported by the Rehab R&D Service of the Dept. of Veterans Affairs. Elderly subjects were recruited with the assistance of the Claude D. Pepper Geriatric Research and Training Ctr.

AUTHOR'S ADDRESS

Dr. J.J. Collins, NeuroMuscular Research Center, Boston University, 44 Cummington Street, Boston, MA 02215, U.S.A., email: collins@buenga.bu.edu

The Effect of Pressure on Spinal Cord Excitability and Parameters of Gait Initiation

D. Brunt, J. A. Robichaud, R. Annan Department of Physical Therapy, University of Florida, Gainesville, FL, U.S.A.

SUMMARY

The effect of circumferential pressure on soleus motoneuron pool excitability during gait initiation was studied. The H-reflex was used as an indirect measure of reflex excitability. The amplitude of the H-reflex was found to diminish prior to tibialis anterior (TA) activity and onset of gait initiation. The same trend was noted with application of pressure but the H-reflex amplitude was always less then the no pressure data. Pressure also delayed the onset of initiation as well as reducing the peak forwards ground reaction force (Fx). Pressure as a modality to reduce muscle stiffness is discussed.

INTRODUCTION

The H-reflex may be used as an indirect measure of the excitability of the motoneuron pool. The application of pressure over the muscle belly decreases motoneuron reflex excitability and increases reaction time during isometric plantar flexion. However, executive components of the movement remained unchanged (1,2). Gait initiation is a task that requires the interaction of the ankle musculature to generate the ground reaction forces (Fx) needed to generate the forward motion of the center of mass. This is accompanied by the backwards movement of the center of pressure (3,4). This decoupling of the center of mass and center of gravity is directly related to TA activity (5). The purpose of this study was to determine if circumferential pressure applied around the leg during gait initiation would reduce spinal cord excitability. It was expected that this would bias TA activity and increase the Fx component of the ground reaction forces.

METHODS AND RESULTS

Subjects stood with their intended swing limb on a force platform. Muscle activity was recorded from the soleus and TA. A light signal cued the subjects to begin walking at a self paced speed. H-reflexes were elicited using an isolation unit attached to a Grass 44 simulator. Stimulations of one ms duration were applied to the skin over the tibial nerve in the popliteal fossa. All raw data was sampled on-line via an analog to digital converter at 1000 Hz for 1500 ms. Subjects completed four blocks of 15 trials which included tibial nerve stimulation at approximately 75, 90, 110, 125 ms following the light stimulus. This procedure was repeated with circumferential pressure applied to the calf of the swing limb. A 32.5 cm air-splint was used to apply pressure to the distal limb. Dependent measures included the amplitude of the H-reflex (% baseline) presented in blocks as determined from the onset TA activity; movement onset as indicated by a directional change in Fx; peak Fz, Fx and Fy ground reaction forces (% body weight); and time to toe-off and heel-strike of the swing limb. The independent variable was no pressure and pressure.

Figure 1 shows a no pressure trial where the M and H response precede onset of TA activity and movement onset. As expected the amplitude of the H-reflex diminished as stimulation became closer to TA onset. This decrease in amplitude was due to reciprocal inhibition. This same trend was noted for the pressure data except that the H-reflex amplitude was consistently less than the no pressure trials. Figure 2 shows second order polynomial correlations for individual subject data where the correlation between H-reflex amplitude and time prior to TA onset was .86 for no pressure trials and .80 for pressure trials. Movement onset occurred significantly earlier for the no pressure trials ($F=14.14$, $p=.02$). Onset of TA, toe-off and heel strike also occurred earlier but the difference failed to reach significance ($p=.06$). There was no difference between peak Fz and Fy but Fx was significantly less for the pressure trials ($F=27.51$, $p<.01$).

Figure 1

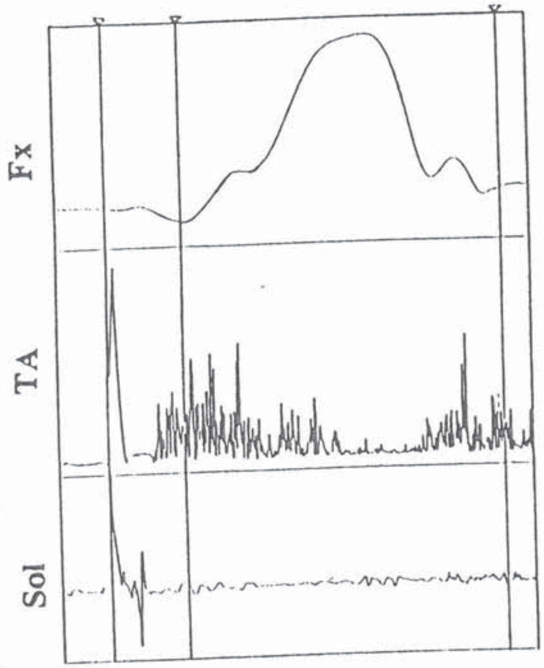


Figure 2

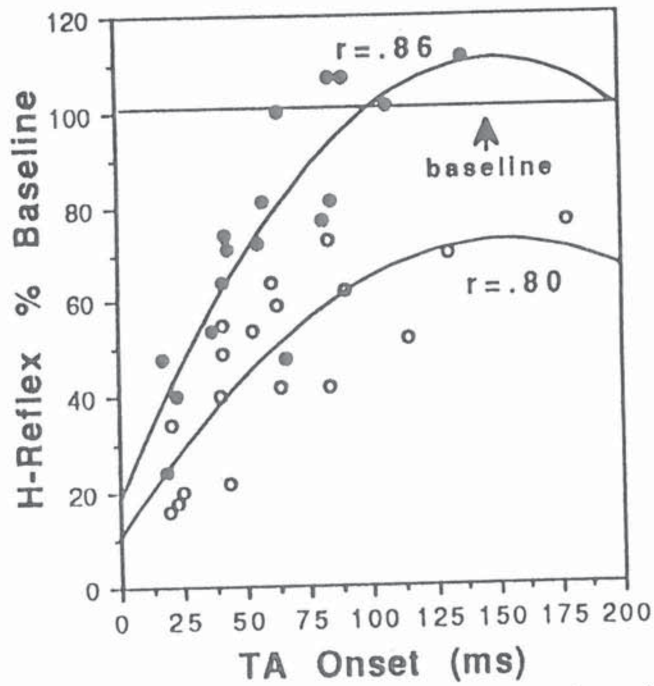


Figure 1. Data of a typical no pressure trial. The vertical lines from left to right represent time of stimulus, movement onset, and swing toe-off. Length of Figure is 700 ms. Peak Fx is 24 % BW. Data collection is triggered by the light stimulus.
Figure 2. Second order polynomials for reflex amplitude (% baseline) and stimulus time for individual subjects. The data points are blocked in 25 ms bins prior to the onset of TA activity.

DISCUSSION

Circumferential pressure reduced the excitability of the soleus motoneuron pool, and increased time to movement onset. However, Fx was found to be less with pressure while Fz and Fy remained unchanged. This is different from previous data if we consider Fx to be a component of movement output. It appears that the pressure also reduced the excitability of the TA motoneuron pool where the TA is primarily responsible for generating forces in the sagittal plane (5). The results of this study are relevant to the rehabilitation of stroke patients where it has been shown that acceleration forces are often absent. This was shown to be due to over activity of the posterior leg muscles and decreased activity of the TA (6).

REFERENCES

1. Robichaud, RA,. Brunt D. (1994) Percept Mot Skills, 78:427-434.
2. Brunt, D., Robichaud, RA. (1993) Neurol Rep (abs), 17:20.
3. Brunt, D. et al. (1991) Am J Phys Med Rehabil, 70:206-11.
4. Yuancheng, J. et al. (1993) Gait Post, 1:9-22.
5. Crenna, P., Frigo, C. (1991) J Physiol, 437:635-53.
6. Brunt, D., Vander Linden, D., Behrman, A. (1993) Neurol Rep (abs), 17:21.

AUTHOR'S ADDRESS

Denis Brunt, Ed.D., P.T.
Box 100154
Department of Physical Therapy
University of Florida
Gainesville, FL 32610

THE EFFECTS OF CRYOTHERAPY ON SELECTED BALANCE PARAMETERS

B. Pfleger, T. Whittaker, J.E. Lander, Life College, Marietta, GA

INTRODUCTION

Ankle injury is common in sports, accounting for almost 25% of all lost time.[1] Ice, compression, elevation, and stabilization (ICES) are commonly used as immediate treatment to injury. The effects of ice upon athletic performance are not well understood. Some research indicates that isokinetic torque, power, and work of the plantar flexor group are reduced immediately following cold water application[2], while other research concluded that ice immersion did not significantly affect the proprioceptors responsible for sensing position.[3] The purpose of this study was to determine if the application of ice to the foot adversely affects the athletes's ability to balance.

METHODS

Ten adult male athletes were recruited from the College Rugby team (ages: 23-35 years; mean height: 172.1 cm; mean weight: 83.0 kg). For each balance test, subjects were required to stand on a force platform for 20 consecutive seconds. Pre- and post-tests were recorded before and after three icing conditions: no icing with 30 minutes rest, 20 minutes icing with 10 minutes rest, and 30 minutes of icing. Icing consisted of immersing the right ankle of the subject in a 1-2 °C ice/water bath. Skin temperatures were recorded at two minute intervals during icing and testing. Four sets of data were collected in randomized order for each subject under the conditions of single foot (right)/eyes closed, single foot (right)/eyes open, two feet/eyes closed, and two feet/eyes open.

The force platform recorded forces and moments in three orthogonal planes each at a sampling frequency of 100 Hz. The middle 10 seconds of data collected was used for subsequent analysis. Software was written to calculate the minimum, maximum and mean (absolute) values for each of the six platform variables. Additional parameters calculated included the standard deviation, range, root mean square (RMS) and excursion of each of the six measurements. The center of pressure (COP) was computed so that the rectangular and circular areas and the path length of the COP could be computed. The mean radial distance of the COP was also calculated to find the average distance the subject travelled from their mean stance position.

Fourier analysis was employed to examine further the radial displacement of the COP and vertical torque parameters. Mean frequencies were analyzed in three windows: 0 to 6.25 Hz, 0 to 12.5 Hz, and 0 to 50 Hz.

RESULTS

The post-test values were subtracted from the pre-test values for each variable under the three icing conditions and four stance/eyes conditions. The pre/post-test differences were compared for the no ice and 30 minute icing conditions as these conditions were expected to produce the largest discrepancies. A factorial ANOVA was done with the four independent variables (subject by ice by stance by eyes) and the pre/post difference serving as the dependent variable. Of the 50 parameters derived from the force platform data, 6 showed a significant effect ($p < 0.05$) for the icing variable: anterior-posterior maximum force, anterior-posterior minimum force, medial-lateral mean absolute force, anterior-posterior excursion of COP, maximum frequency component of the fourier analysis of the radial distance of the COP, and mean frequency component of the fourier analysis of the radial distance of COP.

Factor analysis was also employed to reduce the data set. Three factors were determined to contribute at least 5% to the total variance. Factor one had 24 variables with a loading factor greater than 0.8, with factor two having 4 variables and factor three having 2 variables. Path length of COP was determined to be the principle parameter characterizing factor one, with factor two characterized by vertical torque and factor three characterized by mean frequency of radial displacement of COP (in 0 to 12.5 Hz window). None of these three parameters displayed a significant icing effect in ANOVA testing.

CONCLUSION

It is difficult to conclude that cryotherapy had a significant effect on the ability of athletes to maintain balance in this study. Although a significant icing effect was seen in 6 of the 50 parameters obtained from the force platform data, these could be due to Type I statistical errors. Using an alpha level of 0.05, one would expect to see at least 2-3 significant findings when testing 50 variables. The nature of the six significant parameters would indicate that the subject made corrections that were slower in velocity but more forceful under iced conditions than corrections made with no icing.

REFERENCES

1. Arnheim, D.D., Modern Principles of Athletic Training, 7th Edition, St. Louis, MO, Times Mirror Mosby College Publishing, 1989.

2. Michlovitz, S., Thermal Agents in Rehabilitation, 2nd Edition, Philadelphia, PA, F.A. Davis Co., 1986.

3. Lehman, J., and Delatuer, B., Therapeutic Heat and Cold, 4th Edition, Baltimore, MD, Williams and Wilkins, 1990.

AUTHOR'S ADDRESS

Bruce Pfleger, Ph.D.
Research Department
Life College
1269 Barclay Circle
Marietta, GA 30030

COMPARISON OF ACROSS-SUBJECT EMG PROFILES USING SURFACE AND MULTIPLE INDWELLING WIRE ELECTRODES DURING GAIT

Ross A. Bogey, DO, Jacquelin Perry, MD, Ernest L. Bontrager, MA and Joanne K. Gronley, MS. Department of Physical Medicine and Rehabilitation, University of California, Davis Medical Center, 4301 X Street, Suite 2030, Sacramento, CA 95817 (RAB), and Pathokinesiology Service, Rancho Los Amigos Medical Center, 7601 East Imperial Highway, Downey, CA 90242 (JP, ELB, JKG).

INTRODUCTION

Dynamic electromyography (EMG) is used with increasing frequency in the management of patients with neuromuscular disability, yet there is continuing controversy about the choice of recording electrodes. Most commonly, surface electrodes (affixed to the overlying skin) are selected to detect the muscle's electrical activity because they are non-invasive and allow recording over a large area. A common problem, however, is the inclusion of signals from adjacent or deep muscles ("crosstalk") by electric volume conduction [1]. Wire electrodes inserted directly into the designated muscle allow more precise definition of the onset and cessation of activity of specific muscles by markedly reducing intramuscular crosstalk. This is an important concern when the EMG is used to aid in surgical planning (tendon transfers or lenghthenings). Despite this, wire electrodes are often rejected because of the need for skin penetration and the presumption that the small electrode sampling area does not represent the electrical activity of the entire muscle. The purpose of this study was to test the hypothesis that there are no differences in the normalized wire electrode EMG obtained from multiple sites within the same muscle.

METHODS

Subjects consisted of 5 males with no history of gait pathology or neuromusculoskeletal disease. Electromyographic (EMG) activity of the proximal and distal soleus muscle was recorded with paired fine-wire electrodes (50 micron) using Basmajian's single needle technique [2]. In addition, paired surface electrodes separated 2 cm center-to-center were placed over the motor point of the soleus. Electrode placement was confirmed by electrical stimulation through the indwelling electrodes and by voluntary muscle contraction. The EMG system bandwidth was 40-1000 Hz with an overall gain of 1000. A maximum voluntary contraction (MVC) was performed to provide an EMG signal for normalization of the soleus gait data. The gait cycle interval was recorded with footswitches taped to the subject's bare feet. Footswitch and EMG data were collected at 2500 samples per second, rectified, integrated, and then corrected for baseline noise. The intensity of the soleus activity was calculated as the average %MVC at each 1% gait cycle interval over multiple gait strides (linear envelopes [3]). Individual subject and across-subject EMG profiles for soleus were obtained by linear interpolation [4]. To determine statistical significance of the duration of activity and maximum relative intensity, a repeated measures ANOVA was used to compare the across-subject EMG profiles in each electrode condition. Scheffe's test was used to identify significant differences between groups. A $p<.05$ level of significance was selected.

RESULTS

Soleus surface electrode EMG was continuously active throughout the gait cycle. The duration of activity of the surface EMG was significantly longer than either wire EMG signal. There was no significant difference in the duration of soleus activity between the proximal and distal wire electrodes and their onset and cessation times were essentially identical (Fig 1). Peak relative intensity showed no significant differences between the types or positions of electrodes.

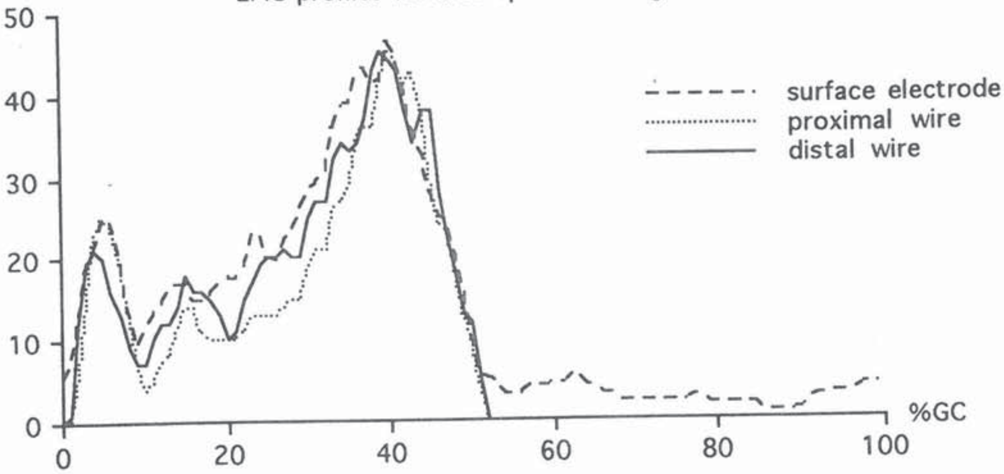


Figure 1. Across-subject soleus EMG profiles for free speed walking.

DISCUSSION

The ability of dynamic EMG to enlighten surgical planning depends on an accurate delineation of timing as well as relative intensity of the muscle contraction. As this study demonstrated, wire electrodes provide precision in phasing that was not available with surface EMG. Also confirmed by this study was the ability of the wire EMG recording to represent activity of the whole muscle despite the small contact area. This physiological event occurs as the result of the wide dispersion of each motor unit's muscle fibers [5-6]. Thus, the small sample size of the muscle from which a wire electrode recording is obtained does not appear to limit the validity of the wire EMG data.

REFERENCES

1. Koh, TJ and Grabiner, MD. Evaluation of methods to minimize crosstalk in surface electromyography. J Biomech 26: 151-157, 1993.
2. Basmajian, JV and Stedio, GA. A new bipolar indwelling electrode for electromyography. J Appl Physiol 17: 849-858, 1962.
3. Arsenault, AB, Winter, DA, and Marteniuk, RG. Is there a "normal" profile of EMG activity in gait? Med Biol Eng Comput 24: 337-343, 1986.
4. Bogey, RA, Barnes, LA, and Perry, J. A computer algorithm for determining the group profile from individual EMG profiles during gait. Arch Phys Med Rehabil 74: 286-291, 1993.
5. Burke, RE, Levine, DN, Salcman, M, and Tsairis, P. Motor units in cat soleus muscle: Physiological, histochemical and morphological characteristics. J Physiol 238: 503-514, 1974.
6. Bodine-Fowler, S, Garfinkel, A, and Roy, RR. Spatial distribution of muscle fibers within the territory of a muscle unit. Muscle Nerve 13: 1133-1145, 1990.

CORRESPONDENCE

Dr. Ross A. Bogey, Department of Physical Medicine and Rehabilitation, University of California, Davis Medical Center, 4301 X Street, Suite 2030, Sacramento, CA 95817

QUANTIFICATION OF HUMAN GAIT WITH THE PRESENCE OF BACK PAIN

L. Arendt-Nielsen, T. Graven Nielsen, P.E. Pedersen, M.D. Jepsen, H. Svarre[1], Dept. of Medical Informatics, Lab. for Motor Control, Aalborg University; and [1]Aalborg Hospital, Dept. of Rheumathology, Aalborg, Denmark.

SUMMARY

New strategies are needed for rehabilitation of chronic low back pain patients. Assessment of the muscle function (activity and co-ordination) is a way in which to gain more insight into the pain-related motor control strategies that are assumed involved when patients with low back pain (LBP) are walking. In the present study, we compare the back muscle function in patients and matched controls during walking on a treadmill.

INTRODUCTION

Low back pain (LBP) is a great clinical problem. According to Bradley (91), more than 70% of all adults experience a period with low back pain some time or another in their life. It is very difficult to diagnose the cause of the syndrome and more than 85% of the patients are left with the description "idiopathic low back pain". As a result of this, personal, medical, and social expenses in connection with LBP are very high.

More efficient methods for diagnosis and rehabilitation of LBP patients are therefore necessary. A few electrophysiological and kinematic studies have been performed on LBP patients but most of them under static conditions.

The aim of this project was to investigate the differences in muscle co-ordination during walking between a normal group and an LBP patient group.

METHODS

The co-ordination of the back muscles in the lumbar region was analysed, using surface EMG signals from eight paraspinal positions (Fig. 1).

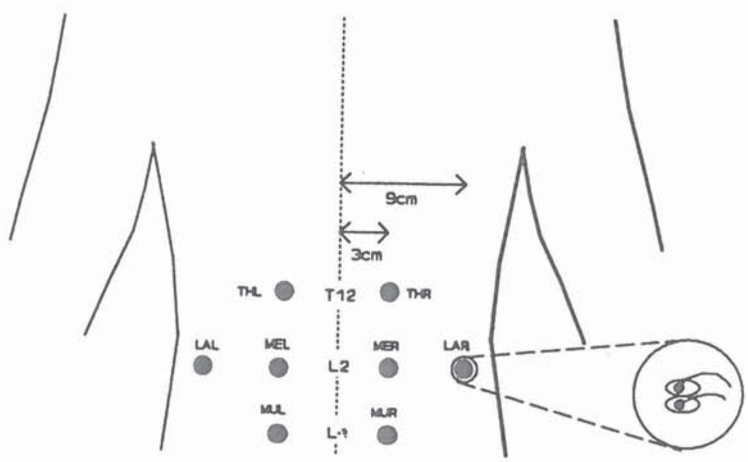


Fig.1 Positions of the electrodes used to assess co-ordination of the lumbar muscles.

EMG profiles constitute a way in which to monitor the strategies by which the human central nervous system controls movements, and, earlier, they have been used to describe pathologic gait (Winter, 84, Sinkjær and Arendt-Nielsen, 91).

The EMG signals are rectified, low-pass filtered and related to the ipsilateral heel strike. This signal processing results in EMG profiles from every stride period and EMG profiles from 40 strides are averaged.

EMG profiles form the starting point for an evaluation of the co-ordination patterns for the normal and the patient group. EMG recordings were made from 10 chronic LBP patients and 10 age- and sex-matched normal persons. All patients had unilateral back pain. The recordings were carried out during walking on a treadmill at a speed of 4km/h. The differences in muscular activity and co-ordination between the LBP patients and the normal group were calculated and the left/right side differences in activity and co-ordination were evaluated.

RESULTS

All patients were able to walk at 4 km/h. The main differences between patients and controls were increased activity during the late swing phase/early stance phase. The general EMG level was reduced in the patients. Interestingly, substantial differences were observed between the activity and co-ordination on the painful and painless side. When examining the muscle activity on the painful side, the agonist/antagonist periods seem to fit with the pain-adaptation model.

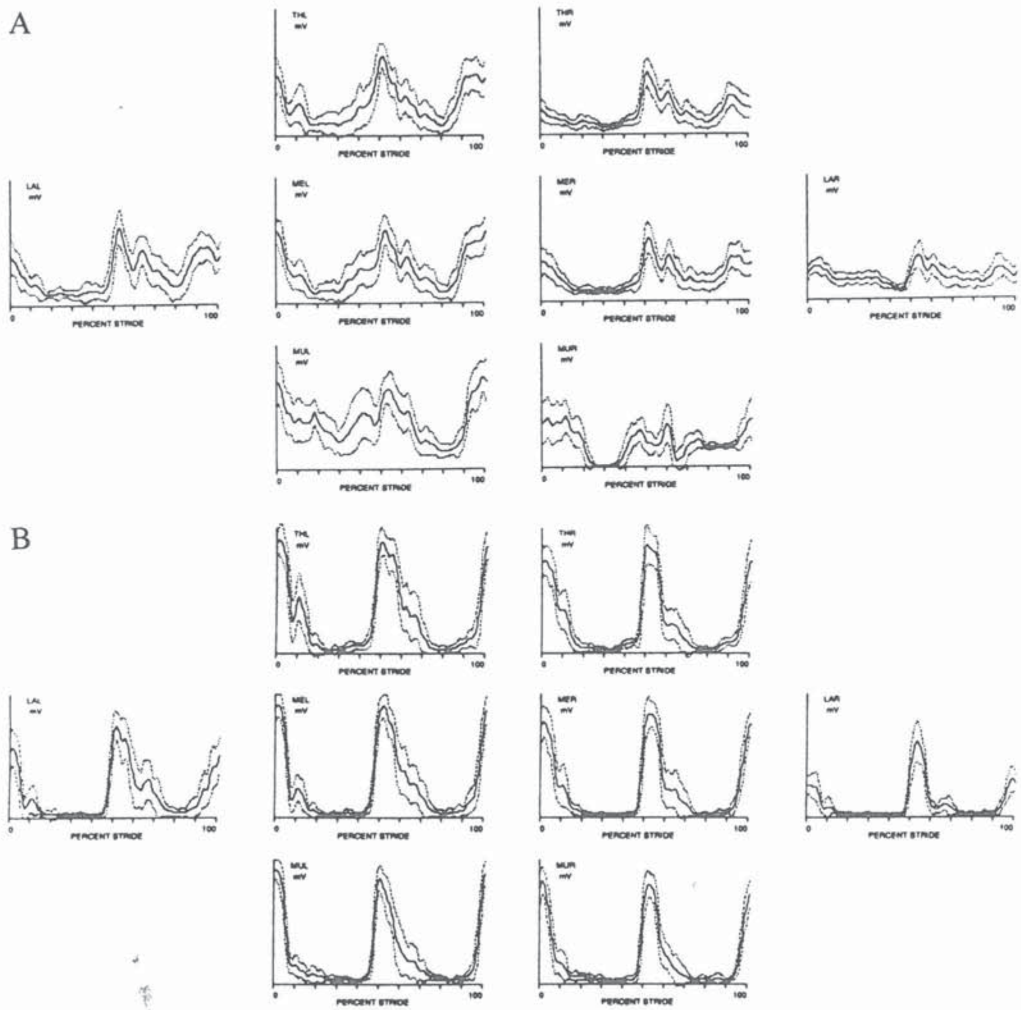


Fig. 2. A set of EMG profiles from one LBP patient (A) and one normal matched control (B) during walking. The recording positions are given in Fig. 1.

CONCLUSION

The present study presents the first investigation of the lumbar muscle activity and co-ordination patterns during walking with the presence of unilateral low back pain. The results suggest that there are possibilities for developing new rehabilitation strategies for patients with chronic low back pain.

REFERENCES

1. W G. Bradley (1991), Neurology in Clinical Practice, The Neurological Disorders, Butterworth-Heinemann, Boston, 405-414
2. T. Sinkjær, L. Arendt-Nielsen (1991), J. Electromyography and Kinesiology, 1: 209-217
3. D. A. Winter (1984) Arch. Phys. Med. Rehab., 65: 393-398

AUTHOR'S ADDRESS

Lars Arendt-Nielsen, Prof., Ph D., Dept. of Medical Informatics, Lab. for Motor Control, Fredrik Bajersvej 7D, Aalborg University, DK-9220 Aalborg, Denmark. Phone: +45-98-158522, ext. 4954; Fax +45-98-154008, Email: lan@miba.auc.dk.

A NOVEL APPROACH TO THE ESTIMATION OF MUSCLE ON-OFF TIMING DURING GAIT

P. Bonato *, T. D'Alessio °, M. Knaflitz •
*Dip. INFOCOM Università di Roma *La Sapienza*, Italy ° III Università di Roma, Italy
•Dip. Elettronica Politecnico di Torino, Italy

INTRODUCTION

In the last decades, surface myoelectric signal recording gained large popularity among researchers involved in gait analysis. In fact, by means of this methodology it is possible to investigate the causes of kinematic and kinetic modifications of gait with respect to the muscle activity. Although surface myoelectric signal recorded in dynamic conditions is not a reliable indicator of muscle contraction force [1], it may be effective to investigate muscle synergies by identifying the activation intervals of different muscles. Methods currently proposed in literature to detect the activation intervals require the arbitrary choice of a threshold, that discriminates background noise from the signal. These methods have a major drawback, since they do not allow the user to set (or to estimate) the probability of correct detection of a) the presence of noise only (specificity) and b) the presence of signal albeit corrupted by noise (sensitivity). The aim of this work is to present and characterize a technique for the detection of the activation patterns from surface myoelectric signals that contemporarily assures a fixed minimum specificity and optimal sensitivity.

METHODS

The algorithm herein proposed is implemented by a processing chain consisting of I) a whitening filter that relies on the Burg algorithm and II) a statistical detector based on a double threshold technique. The signal generated by muscles is considered as a Gaussian modulated process corrupted by Gaussian additive noise. The whitening filter, applied to the input series, guarantees the statistical independence of subsequent samples. A second series is derived from the output of the whitening filter by summing the squared values of couples of subsequent samples. This yields a χ^2 distributed series, thus allowing for calculating easily the probability that a group of r_o (first threshold) out of m subsequent samples passes a given threshold t_h (second threshold). This probability, if only noise is present, is related to the specificity P_s of the detector. When the myoelectric signal is superimposed to noise the sensitivity P_d may be readily estimated, since it is related to r_o, t_h, and to the signal to noise ratio (S/N).

A postprocessor dedicated to the extraction of the activation intervals in gait analysis has also been developed and tested. Basically, it rejects variations of the output state shorter than 30 ms. This is related to the generally accepted hypothesis that muscle activations shorter than 30 ms [2] have no functional significance in controlling joint motion.

RESULTS

To characterize the entire processing chain, the relations among P_s, P_d, and S/N were derived theoretically for different values of r_o and m, and then verified by means of computer simulations. Different choices of r_o and m were called *strategies*. The dependence of the bias of on and off instants of each muscle activation was investigated with respect to the strategy and the S/N ratio.

A comparison of different strategies shows that the specificity may be increased either by increasing the number of samples observed by the detector (thus reducing time resolution) or by decreasing the first threshold (r_o). In our case, the sampling rate was equal to 1 kHz and then m was set equal to 5, to obtain a time resolution of 10 ms. To reach the maximum sensitivity for a fixed specificity the first threshold (r_o) was set equal to 1.

The study of the bias that affects the estimates shows that high sensitivity results in early detection when the algorithm detects the onset of activation intervals (negative bias), and late detection of their ends (positive bias). High specificity has an opposite effect. If strategy $r_o = 1$ $m = 5$ and $0.88 \leq P_s \leq 0.95$, the absolute value of the bias is always lower than 5 ms ($S/N \geq 6$ dB).

Moreover, the standard deviation of the estimate was evaluated. Both the onset and the end of the activation intervals were considered. By posing $r_o = 1$ and $m = 5$, we showed that when $0.88 \leq P_s \leq 0.95$ the standard deviation of the estimate is always lower than 10 ms ($S/N \geq 6$ dB).

CONCLUSION

In this paper we present an original technique for detecting the activation patterns of muscles during gait. This technique allows the user to set the value of sensitivity and estimate the resulting value of specificity. Theoretical as well as experimental results demonstrate that the approach herein presented allows for an accurate detection of muscle activation intervals in practical situations. Moreover, the possibility of precisely characterizing detection errors makes this algorithm superior to single threshold methods usually presented in literature. It is concluded that this technique is innovative with respect to the state of the art and that its performances are adequate in clinics as well as in basic research.

REFERENCES

[1] J.V. Basmajian, C.J. De Luca, "Muscles Alive", *Williams & Wilkins*, 1985

[2] R. A. Bogey, L. A. Barnes, J. Perry, "Computer Algorithms to Characterize Individual Subject EMG Profiles During Gait", *Arch. Phys. Med. Rehabil.*, vol. 73, pp. 835-851, Sept. 1992

Acknowledgements

This work was supported by the Italian National Research Council under grant # 91.02901.CT07.

Dr. Marco Knaflitz
Dip. di Elettronica del Politecnico di Torino
C.so Duca degli Abruzzi 24
10129 - TORINO - ITALY
Tel. +39-11-5644135. FAX +39-11-5644134
E-MAIL: "KNAFLITZ@POLITO.IT"

CHANGES IN ANKLE STRETCH REFLEX WITH FATIGUE INDUCED BY SPONTANEOUS HOPPING

B.Maton, A.LePellec, E.Golomer. Laboratoire de Physiologie du Mouvement, Université de Paris Sud, France.

INTRODUCTION

In muscle stretch-shortening cycles which are common in various types of locomotion, stretch reflexes participate in muscle stiffness during eccentric contractions. They might also play a role in providing the upper nervous system with information about the state of the peripheral system. Changes in both short-term (M1) and long-term (M2-3) components of the stretch reflexes have been described for upper limb muscles, following both maximal and submaximal contractions (Duchateau & Hainaut(1), Komi et al.(2), LeBozec et al.(3)). However, these data can not be easily extrapolated to ankle muscles because the ratio of fatiguable (type IIB) to non fatiguable (type I & IIA) muscle fibres is quite different for example in soleus and in the upper limb muscles and also because descending influences from the upper structures of the central nervous system (CNS) may not be the same for upper and lower limb muscles, or for flexor and extensor muscles. Thus, the purpose of the present experiment was to analyse how the ankle stretch reflex is susceptible to be modified following submaximal spontaneous hopping.

METHODS AND RESULTS

6 healthy female students, 19-23 years old were examined twice. These subjects were instructed, first to hop at spontaneous frequency and amplitude (SH condition) during a few seconds, and then to hop with both knees extended (KEH condition), during the same amount of time. Then, depending on the experimental session, they were required to sustain one or the other of the hopping condition for as long as possible. The task was stopped when subjects were unable to keep frequency or amplitude at the initial value. The hopping tests were preceded and followed by a test of isometric contraction of the ankle muscles. It consisted in keeping an erect posture on tiptoe during 5 seconds.

Bipolar surface EMGs from soleus (SO), gastrocnemius medialis (GM), and tibialis anterior (TA) muscles were recorded simultaneously with the angular displacement of ankle and knee and the vertical component (Fz) of the reaction force from a force plate. All signals were computed, using a 1000 Hz sampling frequency.

Median frequency (MF) and mean frequency (MPF) of the DSP as well as RMS value of EMGs recorded during the isometric contractions were calculated. Comparison of these EMG parameters between the samples obtained before and those obtained after sustained hopping indicated that SO and GM were actually fatigued after the task. Indeed MF and MPF from these muscles were lower and RMS was higher after than before the hopping task. A less expected result was that TA-EMG also showed signs of fatigue in 10 of the experimental sessions.

Hopping was analyzed over 10 cycles at the beginning of the fatiguing task and 10 cycles just before the end of the task.

Each cycle consisted of a ground contact phase followed by a flight phase. The beginning of each ground contact phase was automatically detected as a brisk continuous increase of Fz following the nul values corresponding to the flight phase. EMGs were fully rectified and averaged over the 10 cycles synchronized with Fz onsets.

At the beginning of the hopping task M1 was often but not always evoked in the ankle muscles. It occured in 9 and in 6 of the 12 SH series and also in 9 and in 8 of the 12 KEH series, for SO and GM respectively. Fatigue enhanced M1 occurrence in SO: it occured in all the SH series and in 10 of the KEH series. In contrast, M1 occurence was not affected or even decreased in GM under fatigue: it occured in 6 SH series and only in 4 KEH series.

A late component was seldom distinguishable in initial and fatigued hopping series. M1 latency was measured as the time lag between the Fz onset and that of the reflex EMG burst, while M1 amplitude was defined as the peak amplitude of the EMG burst minus the EMG amplitude at the onset of the burst.
Kolmogorov-Smirnov two-sample statistical test showed no significant difference of M1 latency or amplitude between SH and KEH series, allowing them to be put together.
The same test revealed no significant influence of fatigue on M1 latency, nor on M1 amplitude. M1 soleus mean latency was 37 ms in both initial and fatigued series, while it slightly increased from 39 ms to 42 ms for the gastrocnemius.
As a matter of fact, when soleus M1 component was evoked in both initial and "fatigued" hopping series, its latency was either slightly shorter or slightly greater under fatigue, as compared to initial values, depending on the experiment. Similar observations were made with regard to M1 amplitude.
Since the occurence, latency and amplitude of stretch reflexes are known to depend on the initial level of EMG activity at the instant of the stretch (Matthews(4)), we also measured initial EMG activity. Changes with fatigue were not univocal and even if a linear correlation was shown between M1 peak amplitude and EMG initial activity, a great lot of points were out of a 99 % confidence interval.

DISCUSSION

On the basis of studies on submaximal high level cyclic contractions of the triceps brachii, Komi et al.(2) proposed that one of the mechanisms used by the CNS to compensate for the loss of contractile force due to fatigue, might be an enhancement of the stretch reflex. In the present experiment, the occurence of soleus M1 stretch component increased under fatigue. This result is in agreement with, and extends the above notion to submaximal cyclic contractions of the ankle joint, but with a specific effect on SO. This specificity of the reflex control of SO and GM has been described in the absence of fatigue, for example in walking and running by Duysens et al.(5). It does not seems to be a consequence of the biarticular function of GM as opposed to the monoarticular function of SO since there was no improvement of GM reflex when subjects were asked to keep the knees extended. The specificity of the reflex control rather suggests that besides changes in the common drive exerted by the CNS, there exists a premotoneuronal gating by peripheral afferents.
On the other hand, fatigue was never accompanied by univocal changes in M1 latency or amplitude. This might express a variability of the ankle angle at the instant of ground contact. Changes in M1 latency as a function of ankle angle have been described by Fellows & Thilmann (6). Another explanation might be that subjects used different strategies to compensate for muscle fatigue, as shown by Bonnard et al. (7) for the muscles excitation patterns from similar fatiguing hopping tasks.

REFERENCES

1 - Duchateau,J., Hainaut,K. (1993) Abst.XIV ISB Congress
2 - Komi,P.V., Gollhofer,A. (1987) Med.Sports Sc.,26:119-127.
3 - LeBozec,S., Rougier,P., Natta,F. (1993) Abst. XIV ISB Congress.
4 - Matthews,P.B.C. (1986) J.Physiol.(London),374:73-90.
5 - Duysens,J., Tax,A.,Van der Doelen,B., Trippel,M., Dietz,V. (1991) Exp.Brain Res.,87:193-214.
6 - Fellows,S.J., Thilmann,A.F. (1989) Exp.Brain Res.,77:135-139.
7 - Bonnard,M., Sirin,A.V., Oddsson,L. ,Thorstensson,A. (1994) Neurosciences Letters in press.

AUTHOR'S ADDRESS

Bernard Maton, Ph.D. Laboratoire de Physiologie du Mouvement. Université de Paris Sud. Bât.470, 91405 Orsay cedex, France. Fax: (33.1) 69.85.52.19

MUSCLE FATIGUE MANIFESTATIONS IN THE FOLLOW UP OF PATIENTS AFFECTED BY DUCHENNE MUSCULAR DYSTROPHY

M. Knaflitz•, G. Balestra°, C. Angelini*, M. Cadaldini*

•Dipartimento di Elettronica, °Dipartimento di Sistemi di Produzione ed Economia dell'Azienda, Politecnico di Torino, Torino, Italy, *Clinica delle Malattie Mentali e Nervose, Università di Padova, Padova, Italy

INTRODUCTION

It is well known and documented that the assessment of the electrical manifestations of muscle fatigue is an effective method for evaluating muscle function [1]. Moreover, a novel methodology of muscle fatigue evaluation that relies on the analysis of the myoelectric signal collected during electrically elicited contractions was presented and tested on a population of healthy adult subjects [2]. It was also demonstrated that results obtained by analyzing electrically elicited myoelectric signal are easier to comprehend and more repeatable than those obtained during voluntary contractions. The considerations above induced us to study the electrical manifestations of muscle fatigue in children affected by Duchenne Muscular Dystrophy (DMD). The goal of this work is twofold: a) to describe the electrical manifestations of muscle fatigue in DMD children, and b) to show how our methodology allows for quantification of changes of the muscle function due to either the progression of the disease or to a pharmacological treatment.

METHODS

Eighteen DMD patients and ten healthy children were enrolled in this study. Advantages and possible drawbacks of the study were explained to the subjects and to their parents, who then signed an informed consent form. The age of the children ranged from 5 to 10 years. All the patients underwent a controlled pharmacological trial for two years. Their performances, including electrical manifestations of muscle fatigue, were evaluated before starting the treatment and then every two months throughout the trial. Healthy children were examined since data relative to these subjects were not available in literature. Muscle fatigue was evaluated during a tetanic (35 Hz), isometric, and supramaximal stimulation of the Tibialis Anterior. The stimulated contraction lasted 30 s. The signal collected during the contraction was segmented into 1 s epochs, and, for each epoch, we computed the power spectrum mean frequency (MNF), the average rectified value (ARV) of the signal, and the muscle fiber conduction velocity (CV).

RESULTS AND CONCLUSIONS

The dark area of Fig. 1 comprises the time courses of MNF obtained from the entire normal population, while the lightly shadowed area includes the time courses of MNF in DMD patients. It may be observed that MNF percent decrement in DMD children is remarkably lower than that typical of the control group. This observation may be explained in part by the reduction of the muscle fiber density due to this pathology and in part by the compromission of the force generation capacity of residual muscle fibers. Moreover, by observing different time courses of MNF obtained from the same subject during the two-year

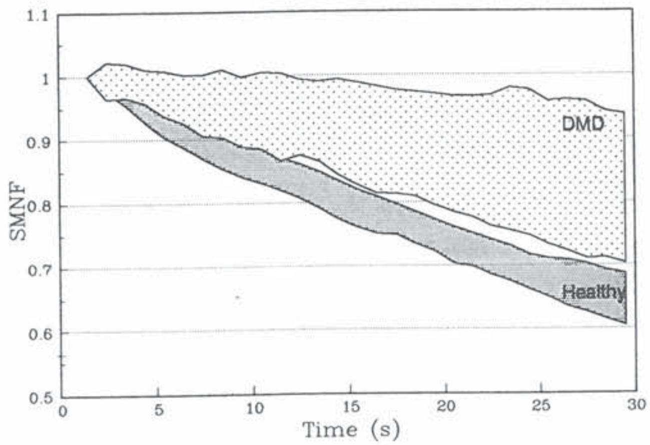


Figure 1: Comparison of MNF time courses in DMD patients and in healthy children

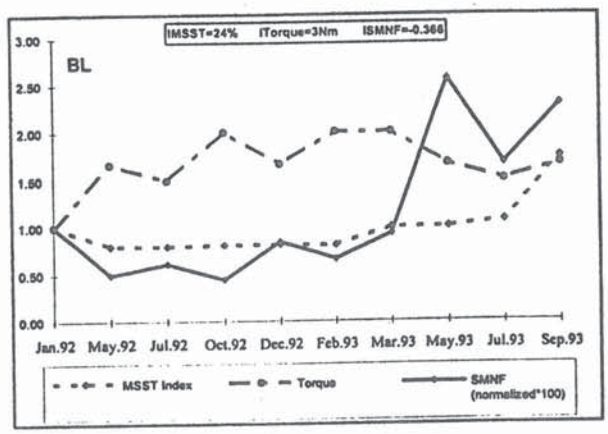


Figure 2: MNF % decrement, MVC torque, and MSST score in a specific patient

observation period, it is evident that the drug administration could either improve MNF percent decrement or keep it stable, while in untreated patients it decreased dramatically. By studying the data obtained from each child, we often noticed that myoelectric signal parameters are affected by changes of the muscle function long before motor performances. Fig. 2 presents a clear example of such a behavior; the solid line depicts the MNF percent decrement recorded over a 21-month period on a specific patient, the dash-dot line shows the behavior of the maximal voluntary contraction torque measured during dorsiflexion of the foot, and the dashed line reports the MSST (a functional index) score. It is evident that in March 1993 MNF percent decrement decreased remarkably, as well as the torque, while the functional index started worsening 4-5 months later. In conclusion, our data show clearly that electrical manifestations of muscle fatigue during electrically elicited contractions in DMD patients are significantly different from those typical of normal subjects. Moreover, our protocol allows for a reliable and accurate quantification of variations of muscle performances due either to the progression of the disease or to a pharmacological treatment.

REFERENCES

[1] C.J. De Luca, M. Knaflitz, *Surface Electromiography: what's new?*, Torino: CLUT, 1992.

[2] M. Knaflitz, R. Merletti, C.J. De Luca, "Inference of motor unit recruitment order in voluntary and electrically elicited contractions". *J. Appl. Physiol.* 68: 1657-1667, 1990.

ACKNOWLEDGMENTS

The financial support of Telethon-Italy to the project *Evaluation protocols for muscular dystrophies: a multicentric approach* is gratefully acknowledged.

AUTHOR'S ADDRESS

Marco Knaflitz, Ph.D.
Dipartimento di Elettronica - Politecnico di Torino
Corso Duca degli Abruzzi 24 - 10129 Torino - Italy
E-mail: KNAFLITZ@POLITO.IT

BRAKING PROCESS ADAPTATION TO MUSCULAR FATIGUE IN HUMAN RAPID AIMED FOREARM MOVEMENTS

S. Le Bozec, M. Gentil, J. Zouhri
Laboratoire de Physiologie du Mouvement, URA CNRS 631, Université de Paris-Sud, 91405 Orsay, France

SUMMARY

Normal human subjects made accurate, fast elbow extension movements of 20° amplitude to a 3° - wide target under normal and perturbed (antagonistic muscle fatigue) conditions. The effect of muscular fatigue on the temporal EMG patterns and kinematics parameters as a function of practice was analysed. There was no difference between the two conditions for the triceps brachii (TB) integrated EMG. The biceps brachii (BB) activity burst was the same for the first pre-fatigue trial and appeared larger as trial increased. Local fatigue led to a large shift in movement amplitude. Taking into account information induced by movement as trial increased, the adjustment of motor program is carried out.

INTRODUCTION

When subjects were trained to make rapid elbow extensions to a well defined target, by repetition of a task, a specific pattern of muscle activity is shown. The pattern consists of an initial burst of agonist activity which produces the force required to set the limb in motion, which is followed by an antagonist braking burst. How localized muscular fatigue affects the movement outcome accuracy and the pattern of aimed movements have not been a major focus of past research. In one of the few studies in this area, Le Bozec and Zouhri (1988) showed increased antagonist EMG burst following repetitive submaximal contractions. Due to local antagonist muscle fatigue, compensatory mechanisms affect specifically the antagonist muscle suggesting that antagonist is not linked to the agonist muscle. Therefore, one of the goals of the present study was to study the agonist-antagonist muscular activation of human subjects using a normally encountered perturbation of the neuromuscular system: antagonistic muscle fatigue. This work investigated changes in EMG and kinematics parameters for fast extension movements as a function of practice.

METHODS

Five normal adult subjects took part in the study. The first series of experiments involved elbow extension movements from a starting position of 110° to a visible 3° - wide target zone at 20°. The second series of experiments was carried out after local antagonist muscle fatigue and involved same extension movements. The subjects were instructed to: 1) accurately move from the starting position to the target zone, as quickly as possible, 2) and to stop within the target zone. Muscle fatigue was induced by a regime of repeated isometric contractions, according to a previously devised experimental paradigm (see Le Bozec and Rougier 1991). The surface EMGs of the agonist muscle (triceps brachii) and antagonist muscle (biceps brachii) were recorded. A cursor program was used to calculate the maximal velocity, the integrated EMG (IEMG) and the electromechanical delay of the agonist and antagonist bursts.

RESULTS

The timing of agonist and antagonist activities was analyzed for the first 10 pre and post-fatigue movements. The TB activity began at a fixed latency of about 30 ms prior to movement onset. The onset of TB activity did not change during fatigue. In contrast, the BB activity occurred significantly earlier under fatigued conditions than under unfatigued conditions. BB activity began shorter after the onset of movement as the post-fatigue trials increased. On the first trial following fatigue, the BB IEMG burst was similar to the last unfatigued BB IEMG burst. On and after the 5th post-fatigue movement, the BB IEMG burts were significantly larger. Fig. 1 shows the displacement and velocity for movements in unfatigued and fatigued subjects. On the first trial following fatigue, subjects overshoot the target by an increase of about 0.15 rad and progressively made accurate movements. Velocity was not significantly different between the post-fatigue trials and the last 5 pre-fatigue trials.

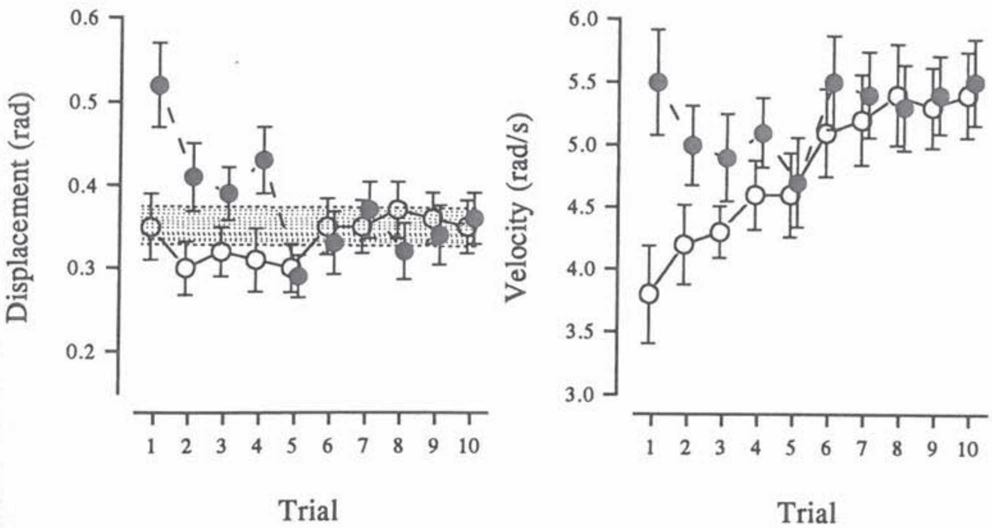


Fig. 1. Displacement and velocity for the pre-fatigue and post-fatigue trials. Dashed lines with filled circles show means ± SD of fatigued subjects. Solid lines with hollowed circles show means ± SD of unfatigued subjects. Dashed area corresponds to the target zone.

DISCUSSION

From these data, we concluded that antagonist muscle fatigue leads to amplitude perturbations during the first post fatigue trials. During these fast elbow extension movements, the ongoing corrections for movement amplitude occurred as trial increase. In the same way, compensatory mechanisms show a progressively increase in the fatigued antagonist muscle in order to maintain the same equilibrium point in fatigue and in unfatigued conditions. In our experimental conditions, for movements lasting less than 100 ms, nervous system should not be able to use visual and kinesthetic input to correct errors in the first post-fatigue trial (Cordo and Flanders, 1989). One can consider that adjustment of motor program is carried out, taking into account information induced by movement as trial increased. Finally, in the nervous system the desired trajectory of the moving limb might be compared with the real one The increase in motor activity of the antagonist muscle could represent the error correction signal resulting from this comparison.

REFERENCES

1. Cordo, F., Flanders, M., (1989) TINS, 12: 110-116
2. Le Bozec, S., Rougier, P., (1991) Eur. J. Appl. Physiol. 63: 293-299
3. Le Bozec, S., Zouhri, J., (1988) Arch. Int. Physiol. Bioch 96:42-43

AUTHOR'S ADDRESS

Serge Le Bozec
Laboratoire de Physiologie du Mouvement, URA CNRS 631
Université de Paris-Sud
Bât 470
91405 Orsay, France

LOCALIZED AND GENERAL FATIGUE INDUCED BY PROLONGED LIFTING TASKS

P. Capodaglio, M. Buonocore, EM. Capodaglio, G.Bazzini
"Clinica del Lavoro" Foundation, IRCCS, Medical Center of Rehabilitation, Montescano (Italy)
Service of Ergonomy, Service of Cl.Neurophysiol.

SUMMARY

Two trained subjects performed an incremental and an endurance lifting test, while cariorespiratory parameters (HR, VO_2) were monitored with a K2 apparatus and subjective perception of localized and general effort was obtained from Borg's CR10 scale. Before and 2 min after the completion of the two tests, the subjects were asked to maintain isometrically a 60% mean maximum voluntary contraction level for 30 sec (LidoLift dynamometer), while paraspinal EMG activity was recorded. Spectral analysis of EMG signals was performed.

INTRODUCTION

Quantitative spectral analysis of EMG signals collected from surface electrodes on the lower back during sustained static contractions monitors the development of muscle fatigue in terms of compression of the power density spectrum toward lower frequencies. Few studies in the literature combine well controlled EMG analysis with direct mechanical, physiological and psychophysical measures of fatigue to quantify dynamic lifting fatigue. The purpose of this study is to describe the relationship between observed quantified changes in myoelectrical activity due to localized fatigue and physiological, mechanical parameters of generalized fatigue, and subjective perception of effort caused by dynamic tasks over extended periods.

METHODS

Two trained subjects performed three maximal isometric contractions (MVC) with a LidoLift dynamometer. The isometric effort was performed with bent back and fully extended knees, according to Chaffin's protocol. The subjects were then asked to maintain isometrically a 60% mean MVC contraction level for 30 sec, while paraspinal EMG activity was recorded. A feedback display was provided. Myoelectric activity from longissimus thoracis muscle at the L1 spinal level and multifidus muscle at the L5 spinal level was recorded bilaterally by bipolar surface electrodes fixed to the skin over the muscle belly at the level of the lower margin of the spinal process. Subsequently, the subjects performed two prolonged dynamic lifting tasks in different sessions. The first test was incremental, with a starting load of 10 kg that increased 5 kg every 3 min. Subjective perception of effort, either localized to the back or general, was monitored with Borg's CR 10 scale. Physiological parameters (HR, VO_2) were sampled during the test with a K2 apparatus. The test was stopped at a CR 7 perception of effort or at 85% of the maximal HR level. 2 min after the completion of the incremental test, the subjects underwent the same EMG evaluation at 60% MVC. In a different session the subjects performed an endurance test with a load corresponding to a CR 3 perception of effort. The test was carried on 30 min and subjective and physiological responses were monitored every 5 min. Immediately before and 2 min after the completion of the endurance test, myoelectric activity during the 60% MVC isometric contraction was recorded. The median frequency (MF) was chosen as a reliable indicator of the spectral shift toward lower frequencies.

RESULTS

Load	*K Joules*	*CR10 general*	*CR10 back*	*HR*	*% max HR*	*VO_2*	*MET*	*Physical demand level*
10 kg	1.959	2	1.5	95	50	1.08	3.67	light-medium (2)
15 kg	2.938	3	3	105	55	1.29	4.38	medium (3)
20 kg	3.657	3.5	4	111	58	1.42	4.82	medium (3)
25 kg	4.571	5	5.5	123	64	1.67	5.68	medium (3)
30 kg	5.485	6	6.5	132	69	1.86	6.32	heavy (4)
35 kg	5.942	7	7	141	74	2.03	6.9	heavy (4)

Tab.1 - The table shows mechanical (K joules), physiological (HR, VO_2) and psychophysical parameters (CR10) at increasing weights lifted for 2 min each. The rate of lifting is constant (6 lifts/min). Physical demand levels are expressed accordingly to the U.S.E.S. classification and, in brackets, the Haskell scale (1-6). Mechanical work is calculated from the following parameters: weight lifted, average upward acceleration, height of lift.

The mean slopes of decay of the MF for both sides at L1 and L5 level, during a sustained effort at 60% MVC immediately before and 2 min after the completion of the incremental test, were respectively: - 0.0053 and - 0.0093 (fig.1).

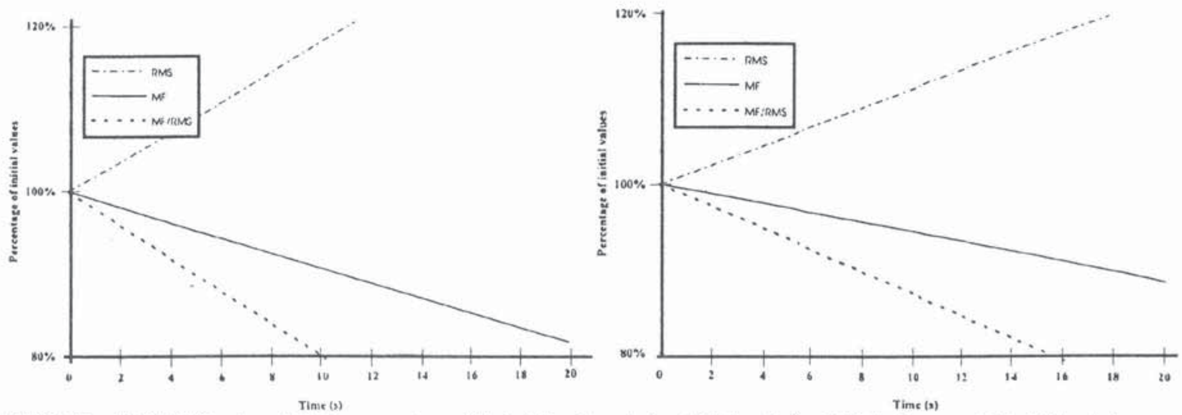


Fig.1 - MF, RMS and MF/RMS values plotted as percentage of the initial values, before (right) and after (left) the incremental test. The plots represent the mean values for both sides and both spinal levels.

In the steady state test the subjects lifted a 15 kg weight, corresponding to a CR3 subjective perception, at a frequency of 6/min for 30 min. Physiologic parameters remained constant during the endurance test, while spectral parameters slightly decreased (Tab.2).

min	*CR general*	*CR back*	*HR*	*% max HR*	*VO2*	*MET*	*slope MF*
10	3	3	120	62	1.34	4.5	- 0.005
20	3.5	3.5	120	62	1.34	4.5	
30	3.5	3.5	118	61	1.28	4.3	- 0.0065

Tab.2 - Physiologic and subjective parameters during a 30 min continous lifting task. EMG spectral parameters were obtained immediately before and 2 min after the endurance test.

CONCLUSION

Subjective and physiologic parameters of general and localized fatigue were highly correlated. Criticisms could be levelled at the method to quantify local fatigue, since the forces recorded during the isometric contraction are expression of the coactivation of different muscle groups. The observed changes in myoelectric activity could therefore be due either to physiological changes in the muscles or to a different recruitment of the muscles involved.

REFERENCES

1. Chaffin DB, Herrin GD, Keyserling WM. Pre-employment strength testing: an update position. JOM 20:403-8, '78
2. Jorgensen K, Jensen BR, Kato M. Fatigue development in the lumbar paravertebral muscles of bricklayers during the working day. Int J Ind Ergo 8:237-45, '91
3. Petrofsky JS, Lind AR. Metabolic, cardiovascular and respiratory factors in the development of fatigue in lifting tasks. J Appl Physiol 45:64-8, '78
4. Potvin JR, Norman RW. Quantification of erector spinae muscle fatigue during prolonged dynamic lifting tasks. Eur J Appl Physiol 67:554-62, '93
5. Roy SH, De Luca CJ, Casavant DA. Lumbar muscle fatigue and chronic lower back pain. Spine (14): 992-1001, '89

AUTHOR'S ADRESS

Paolo Capodaglio, MD
Servizio di Ergonomia
Centro Medico di Riabilitazione di Montescano
27040 Montescano (PV), Italy
Phone + (385)247 331   Fax + (385)61386

FATIGUE DURING LOW-INTENSITY CONTRACTIONS OF THE TRICEPS SURAE MUSCLE

J.H. van Dieën, P. Heijblom, M.H.B. Helmes, H.H.E. Oude Vrielink
Labour Management and Ergonomics Department, IMAG-DLO, Wageningen, The Netherlands

SUMMARY

Ten subjects performed sustained isometric plantar flexion at 5-8% of their maximum voluntary force (MVC) for 4.5 hours. The MVC and the force due to electrical stimulation decreased by on average about 30% and 40%, respectively. In only 2 subjects indications of central fatigue were found, in 4 subjects the force during electrical stimulation decreased more than the MVC. The electromyogram (EMG) amplitude indicated the soleus muscle to be more active than the medial and lateral gastrocnemius muscles. The decrease of the MVC parallelled the decrease of the EMG amplitude during MVC of the soleus muscle. The results can be explained by preferential recruitment and selective fatigue of low threshold motorunits.

INTRODUCTION

Sustained low-intensity contractions appear to play a role in the development of occupational musculoskeletal disorders such as RSI. It has been hypothesized that selective fatigue of low threshold (LT) motorunits is a key-factor in the etiology of these disorders (1). However, relatively little is known about fatigue development and fatigue mechanisms during low-level activity. The present study was designed to investigate fatigue development during sustained low-level contractions in a heterogeneous muscle group.

METHODS

Ten male volunteers (age: 21-29 years; body mass: 65-95 kg; length: 1.75-1.97 m) participated in the experiment. Two Ag/AgCl bipolar electrode pairs separated 0.01 m were applied over the lateral (GL), and medial gastrocnemius (GM), and soleus (SOL) muscles to detect the electromyogram (EMG). Subjects performed a sustained isometric contraction of the triceps surae muscle at 5-8% (target) of the maximum voluntary contraction force (MVC) for 4.5 hours. Before the onset of the task and every 15 minutes thereafter, the force during 20 Hz (ES20), and 100 Hz electrical stimulation (ES100), the MVC, and the EMG during MVC (EMGmvc) were recorded. At the onset of the task and subsequently before each electrical stimulation the EMG during the target force (EMGt) was recorded. The median frequency (MF) of the EMGt was estimated by means of fast Fourier transform. The EMG amplitude was expressed by the rectified and averaged EMG (RAEMG). Data from the two corresponding electrode pairs were averaged and all parameters were normalized to their initial values.

RESULTS AND DISCUSSION

The task resulted in fatigue as indicated by significant decrements of MVC in 7 subjects (overall mean final value: 73%, SD 15%), ES20 in 8 subjects (58%, SD 27%), and ES100 in 9 subjects (60%, SD 22%). The normalized MVC showed a stronger decrease as compared to normalized ES20 and ES100 in 2 subjects, indicating central fatigue. In the other subjects it was equal to or less than the ES100 and ES20 decrease, indicating fatigue of peripheral origin. However, the ratio of ES20 and ES100 changed in only 3 of these 8 subjects, implying that low frequency fatigue is not the major cause. ES20 and ES100 decreased more than the MVC in 4 subjects. This might be explained by selective fatigue of LT motorunits. These probably constitute a large fraction of the units activated during electrical stimulation, due to their lower activation threshold. The performance decrement of these units may, therefore, be more manifest during electrical stimulation as compared to MVC.

As expected from its high content of LT motorunits, SOL was the most active muscle. Initially the median RAEMGt was 10% of the initial RAEMGmvc, as compared to 2% for GL and 7% for GM. On average the differences throughout the 4.5 hr period were similar to those at the onset of the task. The RAEMGt increased in all muscles (final median values of 5, 27 and 14% of initial RAEMGmvc in GL, GM and SOL, respectively). As illustrated by the high final RAEMGt value for GM, in some subjects occassional large increases of activity of GM with concomittant decrements of the activity of SOL and GL were seen. The difference between the muscles at 4.5 hr was not significant.

No evidence was found for a hampered action potential conduction over the muscle, since the MF during target did not decrease. The RAEMGmvc decreased (final value: GL 69%, SD 24%; GM 65%, SD 24%; SOL 71%, SD 11%). The changes in all muscles were significantly correlated to the MVC decrement. The RAEMGmvc of SOL displayed a close correspondence with the MVC, with slopes of the relationship between the normalized parameters ranging from 0.76 to 1.28. Combination of this with the absence of central fatigue in most subjects suggests a hampered propagation possibly at the neuromuscular junction. This finding is in line with recent findings by other authors (2).

In conclusion, during low-level activity preferential recruitment of LT motorunits appears to result in selective fatigue of these units. The fatigue mechanism is predominantly of peripheral origin. However, fatigue is probably multi-causal at these low intensities and large interindividual variations occur.

REFERENCES

1. Vøllestad, N. (1993) In: Advances in Ergonomics and Safety V, London, Taylor and Francis, p. 125-129

2. Fuglevand, A.J., Zackowski, K.M., Huey, K.J., Enoka, R.M. (1993) J. Physiol. 460:549-572

AUTHOR'S ADDRESS

Jaap H. van Dieën, Ph.D.
Institute of Environmental and Agricultural Engineering (IMAG-DLO)
Labour Management and Ergonomics Department
P.O. Box 43
NL-6700 AA Wageningen
The Netherlands

INFLUENCE OF DIFFERENT CONTRACTION LEVELS ON SURFACE EMG SPECTRAL PARAMETERS OF BACK MUSCLES

L. Oddsson, S.H. Roy, J.E. Giphart, M. Emley, J.A. Levins & C.J. De Luca
NeuroMuscular Research Center, Boston University, Boston, MA, U.S.A.

SUMMARY

The behavior of spectral parameters of the surface electromyographic (EMG) signal were studied in a group of healthy males performing isometric trunk extensions (30 s) at different levels of their maximal voluntary contraction (MVC). EMG signals were recorded from six muscle sites of the lower back. The rate of decrease of the median frequency became significantly larger than zero at contraction levels higher than 30% of MVC. There was a non-significant decrease in the initial median frequency. Slopes were 3 times higher at 80% MVC as compared to 40% MVC. The amplitude of the EMG (RMS) increased linearly ($0.95<r<0.99$, $p<0.01$) with force at all muscle sites. A detailed knowledge of the behavior of these spectral parameters in healthy back muscles is necessary for an increase of our understanding of similar parameters in impaired back muscles.

INTRODUCTION

Surface electromyography (EMG) is a commonly used tool for the assessment of muscle function and fatigue (1, 2). Spectral parameters of the surface EMG have characteristics which make them attractive to use as indices of muscle function. They are influenced by metabolic processes in the muscle which the subject cannot perceive and voluntarily regulate (3). Usually, frequency domain properties of the EMG are studied during a fatiguing isometric contraction causing a compression of the power spectrum of the EMG signal towards lower frequencies. The spectral compression can be monitored as a decrease in the median frequency (MF) of the power spectrum (4). At low levels of contraction the rate of decrease of the MF is minimal suggesting a balance between build up and wash out of metabolites in the muscle. Thus, the contraction level at which MF starts to decrease would indicate when metabolites start accumulating in the muscle. This contraction level may be different in healthy as compared to impaired muscles. The aim of this study was to follow the behavior of spectral parameters including MF-slope intial MF (IMF) and amplitude (RMS) at different levels of contraction in healthy back muscles.

METHODS

Ten healthy male subjects performed sustained (30s) isometric trunk extensions at different levels of their maximum voluntary contraction (20, 30, 40, 50, 60, 70 and 80% of MVC) in a postural restraint and EMG aqusition device called the Back Analysis System (BAS) (1). Surface EMG was recorded using active bipolar surface electrodes (inter-electrode distance 1cm) placed bilaterally over the longissimus (L1-level), iliocostalis (L2-level) and multifidus muscles (L5-level) of the lower back. Spectral parameters including MF and RMS of the power spectrum were tracked in real time (10 Hz) and stored in digital form on a hard disk.

RESULTS AND DISCUSSION

Values of IMF and MF-slope for different contraction levels (%MVC) are shown in Fig. 1. Parameters were normalized to values obtained at 80% of MVC. There was a decreasing trend, however non-significant, in IMF values with increasing force levels (Fig. 1 left). Rate of decrease of the MF was not significantly different from zero at 20% and 30% of MVC. Between 40% and 80% of MVC the slope values increased by a factor of three (Fig. 1 right). The amplitude of the EMG (RMS) increased linearly ($0.95<r<0.99$, $p<0.01$) with force at all muscle sites.

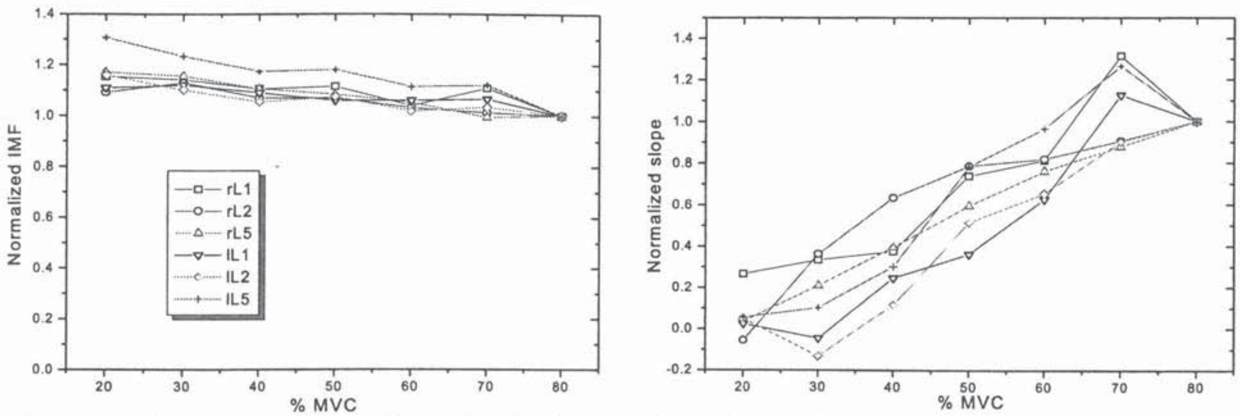


Figure 1. Average of normalized IMF values (left) and MF-slopes (right) for different levels of MVC for all subjects.

The results of this study suggest that the build up of metabolites in the muscle, as indicated by a significant decrease in the MF of the surface EMG signal during a 30s sustained contraction, appears at contraction levels somewhere between 30% and 40% of MVC. This is different from the behavior of MF-slope in muscles of the hand (5) which demonstrate slopes larger than zero already at 20% of MVC. The decreasing trend in IMF was not significant across subjects. However, certain subjects did show clear decreases in IMF with increasing contraction levels. We are currently investigating other parameters that may better quantify this behavior. Results of this study will be further compared to similar experiments in groups of subjects with impaired back muscle function.

REFERENCES

1. Roy, S.H., De Luca, C.J. & Casavant, D.A. (1989a), Lumbar muscle fatigue and chronic lower back pain, Spine, 14, 992-1001.
2. Mayer, T.G., Kondraske, G., Mooney, V., Carmichael, T., Butsch, R. (1989), Lumbar myoelectric spectral analysis for endurance assessement: A comparison of normals with deconditioned patients, Spine, 9, 986-991.
3. De Luca, C.J. (1985), Myoelectrical manifestation of fatigue in humans, CRC Critical Reviews in Biomedical Engineering, 11, 251-279.
4. De Luca, C.J. (1993), Use of surface EMG signal for performance evaluation of back muscles, Muscle & Nerve, 16, 210-216.
5. Roy, S.H. & De Luca C.J. (1989b), Evolving characterisitics of the median frequency of the EMG signal, In: Computer-Aided Electromyography and Expert Systems, ed Desmedt J.E., 205-221.

AUTHOR'S ADDRESS

Lars I.E. Oddsson, Dr Med Sc.
NeuroMuscular Research Center, Boston University
44 Cummington Street, Boston, MA, 02215, USA
tel: 617-353 9757 fax: 617-353 5737 e-mail: *ODDSSON@BUNMRG.BU.EDU*

EMG AMPLITUDE IN THE LUMBAR SPINE EXTENSOR AND FLEXOR MUSCULATURE DURING MAXIMAL AND SUBMAXIMAL CONSTANT FORCE CONTRACTIONS

M.B. Frazer and R.W. Norman
Occupational Biomechanics Laboratory, Department of Kinesiology, University of Waterloo, Waterloo, Ontario, Canada.

SUMMARY

Amplitude changes in sustained isometric contractions were studied for the flexor and extensor musculature of the lumbar spine. Sustained, 70% MVC isometric extension and flexion efforts produced EMG amplitudes equivalent to 88% and 115% of those observed during isometric MVCs. This approach would not appear to be a viable method for obtaining maximal EMG activation levels in individuals asymptomatic for low back pain.

INTRODUCTION

Tissue forces in the lumbar spine may be estimated using models that incorporate electromyography (EMG) as a signal to assist in the partitioning of the restorative moment (McGill (1) Marras (2)). For normalization purposes, these models require that maximal voluntary contractions (MVCs) be performed to illicit maximal EMG amplitudes. These contractions are problematic for individuals with low back pain, preventing the application of these models to this population. Moritani (3) and Petrofsky (4) report that for the biceps brachia, sustained, submaximal isometric contractions of 50% and 70% MVC result in EMG amplitudes equivalent to those produced during MVC efforts.

The purpose of this study was to determine if sustained, 70% MVC isometric contractions of the extensor and flexor musculature of the spine would illicit EMG amplitudes equivalent to those observed during MVC efforts.

METHODS AND RESULTS

EMG was recorded bilaterally from the rectus abdominis, external oblique, internal oblique, upper erector spinae (T9) and lower erector spinae (L3) for ten males (32.4 $\pm$ 13.0 yrs, 1.78 $\pm$ .08 m, 82.0 $\pm$ 9.5 kg), asymptomatic for low back pain. The connecting fastener of a shoulder harness was aligned with the torso centre of mass. For extension/flexion efforts, subjects lay prone/supine over a two-tiered bench, with their greater trochanter aligned with the edge of the higher tier and their upper body weight supported on the lower tier. The MVC moment-time history was measured via an LVDT connected to the harness and secured to the floor. The maximal extension moment was calculated from two, ten second isometric MVCs. Subjects also performed a series of ten second isometric extension/flexion efforts, ranging from body weight to 85% MVC, by raising loads attached to the harness, just clear of the floor.

Table 1 summarizes the results from the extension/flexion MVC and sustained contraction efforts. Linear regression of the submaximal contraction peak EMG amplitudes revealed the extension and flexion musculature were on average 21.5% MVC below, and 22.5% MVC above, the load moment respectively for a specific trial.

Table 1 Results from maximal and sustained contractions: Mean (± 1 SD)

	Maximal Contraction	70% Sustained Contraction		
		Peak EMG Amplitude (% MVC)		
	Moment (N.m)	Start	End	Duration (s)
Extension	328.7 ± 50.2	67.0 ± 14.8	87.8 ± 13.1 **	42.6 ±13.1
Flexion	307.2 ± 43.4	81.6 ± 27.5	115.6 ± 31.3 **	30.4 ±10.2

** $p < 0.01$ for significance of difference between means.

DISCUSSION

As expected, the extensor musculature produced a larger MVC moment than the flexor musculature. The flexion and extension moments are 2.0 and 1.3 times, respectively, those produced in standing postures in our lab (5).

The depressed peak extensor amplitude during the submaximal ten second and sustained exertions indicates that musculature not monitored (eg. multifidus) was contributing to the production of the required support moment.

Elevation of the peak flexor amplitude during the submaximal ten second and sustained exertions was very surprising. With the LVDT no longer affixed to the floor, subjects may have adopted a slightly more flexed posture than in the MVC condition. The resulting shorter muscle length would require increased EMG activation levels for each unit of force produced. Inbar (6) found that Median Power Frequency increased as muscle length decreased. Analysis of the Mean Power Frequency for the flexion trials showed significant increases for the right and left internal obliques, and rectus abdominis. Posture changes may also result in different motor units being monitored, altering the EMG-moment relationship produced during the MVC trials.

Although the EMG increased significantly during the 70% sustained trials, the respective undershoot and overshoot in the extension and flexion trials do not make this a viable method for obtaining maximal EMG activation levels in individuals asymptomatic for low back pain.

REFERENCES

1. McGill, S., Norman, R., (1986) Spine, 11:666-557.
2. Marras, W., Sommerich, C. (1991) Human Factors, 33:123-137.
3. Moritani, T., Muro, M., Nagata, A. (1986) J. of Appl. Phys., 60:1179-1185.
4. Petrofsky, J., Glaser, R., Phillips, C., et. al. (1992) Ergonomics, 25:213-223.
5. Frazer, M., Norman, R. (1993) ISB XIVth Congress Abstracts I, 422-423 Paris.
6. Inbar, G., Allin, J. and Kranz, H. (1987) Med. & Biol. Eng. & Comput., 25:683-689.

AUTHOR'S ADDRESS

Mardon B. Frazer, B.Sc.
Department of Kinesiology
University of Waterloo
Waterloo, Ontario N2L 3G1

POST FATIGUE FORCE AND EMG IN FES ACTIVATED PARALYZED MUSCLES

J. Mizrahi, D. Seelenfreund, A. Aviram and E. Isakov
Dept. of Biomedical Engineering, Technion–IIT, Haifa, Israel
Loewenstein Rehabilitation Center, Raanana, Israel.

INTRODUCTION

During fatigue, an exponential curve was reported to express the relationship between force and EMG peak-to-peak (PTP) amplitude of the M-wave signal in the muscles of paraplegic patients activated by FES [1]. The bioenergetics of paralyzed muscles under FES were studied *in vivo* using ^{31}P NMR spectroscopy during fatigue and recovery [2]. Profiles of the phosphorus metabolites indicated systematic variations in these parameters in the activation phase of the muscle and a gradual, slow process of recuperation of their rest-state values in the recovery phase. The force generation capacity in this latter phase is expected to increase, in accordance with the degree of metabolic restoration of the muscle. However, testing of post-fatigue force generation capacity requires stimulation of the muscle by as short as possible bursts of stimulation, if further fatigue has to be minimized. Both EMG and force transients were reported to last a few seconds from the onset of stimulation [3], implying that stimulation bursts should be at least of that duration, if muscle performance is to be reliably monitored. It is thus expected that post-fatigue burst stimulation, intended to test the force generation capacity, will induce further fatigue or delay recovery.

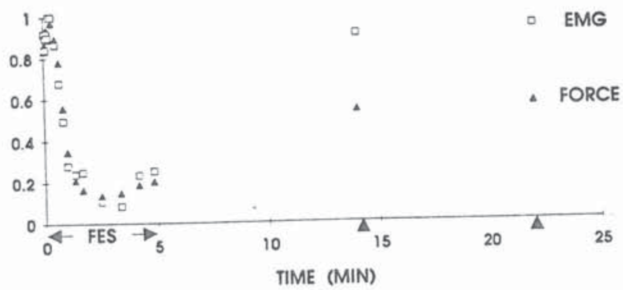
In the present study, we report on the post-fatigue force generation capacity as a function of the rest time of the muscle, i.e., the time between the beginning of fatigue and the first burst of stimulation. Of additional interest is to verify whether the force EMG relation reported in the fatigue process holds also in the post-fatigue phase.

METHODS

The right quadriceps muscle of a complete paraplegic patient, who was engaged in an FES training program were stimulated transcutaneously. The muscle was first isometrically fatigued by providing a continuous stimulation train of 70 mA intensity over the period of 5 min. Stimulation bursts of 70 mA and duration of 7 sec were thereafter given, following differing rest-times between the beginning of fatigue and the first burst of stimulation. Force measurements were made by means of a specially designed load cell, attached on an adjustable testing chair. From the measured torque, the quadriceps tendon force was calculated, using a previously described method [4]. Surface EMG was recorded by means of three 10 mm diameter gold cup-electrodes. It was important to suppress the stimulus artifact preceding the EMG signal and we, therefore, designed an amplifier capable of monitoring the artifact-free signal [5]. The EMG signal was represented by the amplitude of the M wave signal obtained [1] Both the force and the EMG signals were sampled on line at 1 kHz.

RESULTS AND DISCUSSION

Fatigue recovery results are shown in Fig. 1. The first burst of stimulation was given after 9 min from the beginning of recovery. A second burst followed after 8 min. The post-transient values of force and PTP, i.e., corresponding to the end part of the stimulation burst, are presented. The steep decay of force and EMG levels off at low force and PTP values after less than 2 minutes of continuous stimulation. A mild increase in both parameters is noticed after approximately 3 minutes of stimulation. A



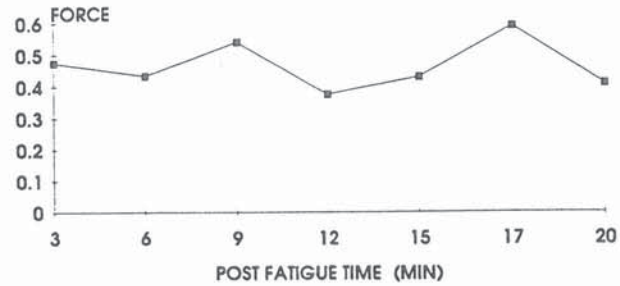


Fig.1: Force and EMG in Continuous FES and and post-fatigue bursts (vertical arrows)

Fig. 2: Force versus timing of first post-fatigue burst

possible explanation of these slight amplitude elevations is that some of the fibers may have recovered in the process of fatigue. Recent ^{31}P NMR spectroscopy studies have confirmed this possibility [2].

As seen from the figure, the PTP and force behave differently during recovery. The PTP is seen to gain almost full restoration from the very early phases of recovery. The force, however, was at 50% of the maximal pre-fatigue level. This indicates that the force-EMG relation reported during fatigue [1] is not correct during post-fatigue, stimulation bursts.

The effect of variation of the timing of the first burst of stimulation is shown in Fig. 2. It is noted that the force achieved was almost unaffected by the timing of the first burst of stimulation. In a previous paper, our group has shown that the muscle metabolites recover slowly but steadily in the process of recovery, and that full recovery after 3 min of stimulation is achieved after approximately 30 minutes [2]. These results, however, were obtained at full rest, i.e., with no stimulation during recovery. If muscle metabolites are indeed indicative of the force producing capability, as inferred from the relationship between force and metabolites during the fatigue process, it follows that the results obtained in the present study somewhat differ from the findings reported earlier, from which considerable recovery should be expected in the post-fatigue time studied. The difference should thus be attributed different durations of fatigue and to the mere introduction of the FES bursts, which would disrupt the recovery process.

ACKNOWLEDGEMENT

This study was supported by the Segal Foundation and by the Walter and Sandra Kaye Research Fund.

REFERENCES

1. Mizrahi J et al. *IEEE Tr. Rehab. Eng.* accepted 1994.
2. Levy M et al. *Magn. Reson. in Med.* 1993; 29:53-58.
3. Mizrahi J et al. *BAM*, in press, 1994.
4. Levy M, et al. *J. Biomed. Eng.* 1990; 12:150-156.
5. Minzly J et al. *Med. and Biol. Eng. and Comp.* 1993; 31:72-75.

Corresponding Author: Prof. J. Mizrahi, Dept. Biomed. Eng., Technion-IIT, Haifa, Israel 32000. Tel:972-4-294128; Fax:972-4-234131; e-mail:JM@BIOMED.TECHNION.AC.IL

COMPUTER AIDED ASSESSMENT OF RESPIRATORY SYSTEM FUNCTION

Aliverti A.[1], Carnevali P.[1], Ferrigno G.[1 2], Pedotti A.[1 2]

1. Centro di Bioingegneria - Politecnico di Milano - Fnd. Pro Juventute I.R.C.C.S.- Milano, Italy
2. Dipartimento di Bioingegneria - Politecnico di Milano, Italy

SUMMARY

Respiratory function reflects on the action of its passive and active components. The volume variation of the chest wall and its time course can characterise the status of the respiratory system in several pathologies. The present work introduces a new approach based on the integration of a new technique for the evaluation of trunk volume split in sectors (upper thorax (UT), lower thorax (LT) and abdomen (ABD) or right and left) with a special purpose software for the automatic extraction of several parameters (tidal volume, frequency, flow, Ti/Te, contribution of each sector to total volume) from the kinematics of the chest wall. Three examples of data output are shown in the results.

INTRODUCTION

The application to chest-wall motion analysis of an automatic motion analysis system, ELITE system (1), has allowed the computation of the volume variation of three sectors of the chest wall (UT, LT and ABD). The volume variations have been computed by fitting a geometric model to the 3D co-ordinates of several landmarks on the thoraco-abdominal wall. The co-ordinates of the landmarks, highlighted by small passive markers are computed in real time and automatically by the system. The quantitative reliability of this approach has already been evaluated by (2). The purpose of this work is to show the capability of a special software for the extraction of synthetic parameters useful for the assessment of respiratory system function, also in clinical environment and for statistics purposes, as intra and inter-individual comparisons and normality definition.

METHOD

The method proposed here is based on the use of an automatic motion analyser, the ELITE system, which measures the 3D co-ordinates of several passive, lightweight and small (6 mm diameter) markers applied to body landmarks at a sampling rate of 100 Hz or sub multiples. The measurement is carried out by using a set of four specially designed TV cameras which survey the region of space where the subject is located. The TV signal is sent to a hardware parallel processor which performs, in real time, a shape recognition of the markers. Their 3D co-ordinates are then computed by a stereo-photogrammetric technique with high accuracy (1).
In the present application three different experimental set-up have been used in order to analyse respiratory movements in different positions: sitting or standing, supine and on the side. These latter two positions are particularly useful for studying pathological subjects. Forty landmarks have been arranged on an anterior 4x5 grid and a symmetric posterior one. The anterior frame consists of 4 horizontal rows (2nd rib, xiphoid process, 10th rib and abdominal transversal line) and 5 vertical rows included between the 2 anterior axillary lines symmetrical with respect to the central line of the thoraco-abdominal wall. In the case of the supine set-up, the posterior markers are supposed to be fixed on the plane on which the subject lies. This arrangement allows the computation of the volume of 12 compartments which can be grouped into horizontal and vertical sectors. Either the contribution to volume variation due to UT, LT and ABD or the one due to left and right trunk can then be computed (2). In Fig. 1 the time course of the total volume of the

chest wall of a normal subject during supine quiet breathing is reported (bottom trace). The splitting of the volume into three parts corresponding to the contribution of UT, LT and ABD is reported as well.

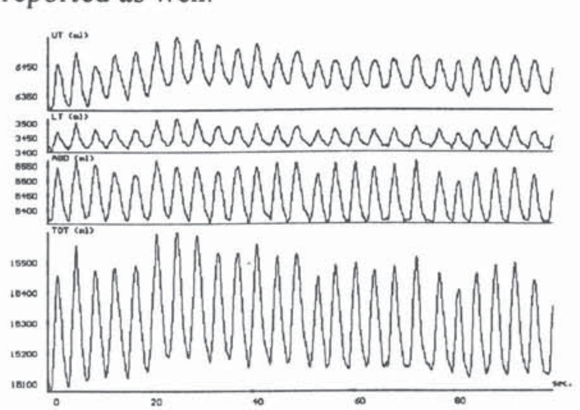


Fig. 1

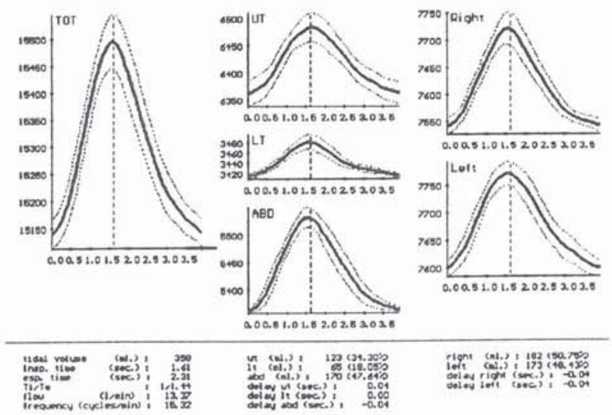


Fig. 2

In order to have a synthetic representation of the data, a dedicated software has been developed to recognise, starting from the time course of the total volume, each breathing act and its inspiratory and expiratory phases. The time base of the breathing period is normalised to a specified number of points. Then the mean and the standard deviation of these samples are computed and displayed. Several parameters can be computed on the normalised breathing (tidal volume as the difference between the end inspiration and the end expiration volume, inspiratory mean flow, UT, LT and ABD contributions as the corresponding volume variations in the same interval and timing data as in/expiratory time, frequency, delay of the maximum of each sector volume with respect to the maximum of the total one).

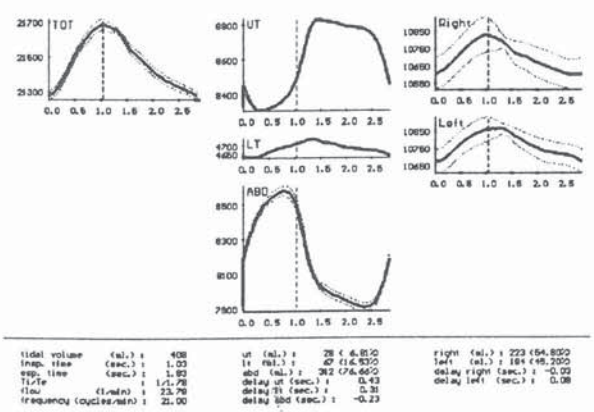


Fig. 3

Fig. 2 shows the result of this approach on the data of Fig. 1 and Fig. 3 refers to the breathing of a tetraplegic (C2 level) subject with phrenic stimulation. While in normal subject the three compartments expand simultaneously, in the tetraplegic one there is a paradoxical movement of the UT with respect to the other two compartments.

CONCLUSIONS

The results obtained with this approach allow both an immediate assessment of the breathing pattern and a quantitative evaluation of its main characteristics. Its simplicity, rapidity and reliability make it suitable both in research (statistic evaluations, studies of particular pathologies) and in clinical routine applications.

REFERENCES

1. Ferrigno, G., Borghese, N.A. and Pedotti, A. Pattern Recognition in 3-D human motion analysis. ISPRS J. Photogramm. Rem. Sens. 45: 227-246, 1990.
2. Ferrigno, G., Carnevali, P., Aliverti, A., Molteni, F., Beulke, G. and Pedotti, A. 3-D optical analysis of chest wall motion. J. Appl. Physiol. (in press)

Methodological aspects of kinematic analysis of pen movement during handwriting using the ELITE system

A.Campisi[1], M.Cincera[1,2], M.Rabuffetti[1] & A.Pedotti[1,2]

1) Centro di Bioingegneria, Fondazione Pro Juventute "Don Carlo Gnocchi" - Milano, Italy.
2) Politecnico di Milano - Dep. of Bioengineering, Milano, Italy

Abstract

The aim of the present study is to develope a method for kinematic analysis of handwriting using the ELITE system [Ref.1]. Since it is not possible to use a marker directly applied to the pen tip during writing, a procedure was developed to recostruct pen tip movement using markers projecting from the axis of the pen. This method assumes the rigidity of the pen and defines a reference system attached to the pen in which the pen tip coordinates are constant throughout the writing movement. In an attempt to improve precision, an optimization procedure was used to calculate the system of reference minimizing the effect of marker localization error. Precision obtained in this way allows for 2-D analysis of handwriting as well as the 3-D analyses of pen movement associated with handwriting.

Introduction

Traditionally, handwriting analysis has been carried out using ultrasonic devices [Ref.2], magnetic digitizing tables [Ref.2] or optical picture scanners [Ref.3]. The drawbacks of these methods include poor precision (ultrasonic devices), interference with the natural movement involved in writing (tables), and a lack of kinematic data (scanners) and a lack of 3-D information (tables and scanners).

The present study shows how 3-D kinematic data can be obtained with a high degree of precision and without the drawbacks mentioned above.

Methods

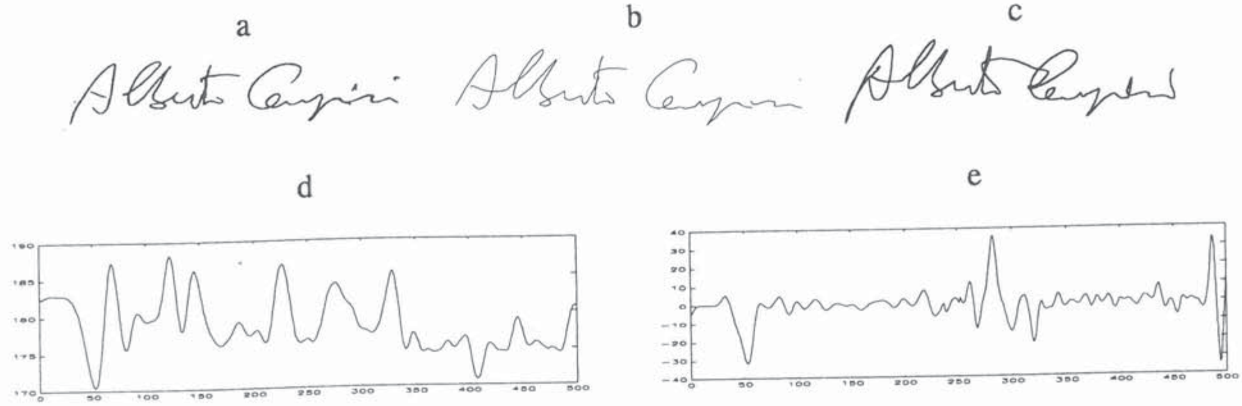
The ELITE motion analysis system [Ref.1] at a sampling frequency of 100 Hz was used for kinematic acquisition of technical markers. The system is able to reconstruct spherical passive marker centroid coordinates in real time.

The writing instrument (Staedler Microline 321s) is equipped with five light weight markers projecting at different angles and at a costant distance from the axis of the instrument.

Two preliminary acquisitions are carried out. The first consists of pen tip calibration: an additional marker is applied to the pen tip and a static acquisition is carried out. This acquisition allows for the determination of pen tip coordinates in a reference system attached to the pen. The second acquisition consists of writing plane calibration, during which a spiral is drawn while the pen tip is in constant contact with the writing surface. This acquisition allows for the estimation of the equation of the writing plane. Data acquisition during handwriting is then carried out using only the five primary markers. Data processing consists of a preliminary analysis of noise associated with marker positions. A non-linear simplex procedure for multi-objective function minimization [Ref. 4] was used to find the reference system attached to the pen which allows tip reconstruction with the least standard deviation about the tree axes. By means of subsequent rototraslations [Ref.5] 3D movements of the pen tip are reconstructed in an absolute XYZ system, where XY is the writing plane evaluated during the calibration phase. Finally a FIR filtering is performed on the reconstructed pen tip kinematics to smooth the signal.

Results

The suggested pen tip movement reconstruction procedure allows for handwriting analysis. The figure shows a comparison between the original pen trajectory (Fig.a), the reconstructed 2-D (Fig.b) and 3-D trajectory (Fig.c).



Original signature (a) and reconstructed ones (b and c). (b) has been obtained considering only points of the trajectory belonging to the writing plane. (c) has been obtained projecting on the writing plane the complete trajectory of the pen tip. Y displacement (d) (mm) and z velocity (e) (mm/ms) for 3-D movement are also plotted

Starting from a specific marker placement and a measurement error on marker localization with a standard deviation equal to ±0.1 mm, pen tip movement has been reconstructed with a mean error of ±0.01 mm. It was found that this error was equally distributed along the three axis of the reference system.

Discussion

Pen tip kinematics cannot be monitored directly because it is the impossible to place a marker on the pen tip during data acquisition without modifying the action of writing. The most important results of the suggested approach to handwriting analysis is that it permits acquisition of 3-D information during natural handwriting and the reconstruction of pen tip movements with a degree of precision comparable to or greater than that obtainable using other digitizing devices. Secondly, the method permits not only 2-D kinematic analysis of handwriting, but also a 3-D kinematic analysis of the pen tip movement associated with such activity.

In addition pen tip movement reconstruction precision can be increased by reducing the calibration field further, an important feature considering the small displacements of the pen during writing movement.

References

1. **Ferrigno G.C. & A. Pedotti.** *ELITE: a digital dedicated hardware system for movement analysis via real_time TV_signal processing*. IEEE Trans Biomed Eng 32:943-950, 1985.

2. **Lacquaniti F., C.Terzuolo & P.Viviani**. *The Law Relating the Kinematic and Figural Aspects of Drawing Movements.* Acta Psychologica 54 (1983) 115-130.

3. **Arazi, B.** *Automatic HandWriting Identification Based on the External Properties of The Samples.* IEEE Transaction on Systems, Man , amd Cybernetics, Vol. SMC-13, No.4, July/August, 1983.

4. **The Matworks, Inc.** MATLAB User's guide, 1991.

5. **John J.C**. Spatial description and transformation. In: *Introduction to Robotics*. Addison-Wesley Publishing Company, cap.II, pp. 19-67, 1989.

MULTIPARAMETRIC ANALYSIS OF LIPS AND JAW KINEMATICS IN NORMAL AND STUTTERING SUBJECTS

Redolfi M.[1], Ferrigno G.[1 2], Pedotti A.[1 2]

1 Centro di Bioingegneria - Politecnico di Milano - Fondazione Pro Juventute D.C. Gnocchi - Milano, Italy
2 Dipartimento di Bioingegneria - Politecnico di Milano, Italy

INTRODUCTION

This study focuses on the analysis of lips and jaw articulation involved in the production of repeated syllables /pa/ and /ba/ at two different velocities: normal and maximal.
Kinematic data of normal and rieducated stuttering subjects were recorded by means of the ELITE automatic motion analysis system and then compared.
The purpose of the study was to determine the differences between normal and stutterers by means of a data analysis realized both in time and frequency domain. Results show a significant difference between the two groups of subjects, especially when the sequences are pronounced at maximal velocity.

METHOD

Articulatory kinematics of three normal and two rieducated stuttering subjects were analysed. They had to repeat sequences /papapa.../ and /bababa.../ at normal and maximal velocity. Acoustic and kinematic data were recorded by means of the ELITE system (1), which assures a null interference in subject movements and the necessary precision to relieve very fast movements in a small field of view (30 cm x 30 cm). The system fits the requirements of the study in terms of small passive markers (1 mm. of radius) fixed in some particularly significant points of subject face (2). Articulatory parameters we want to analyse, lips and jaw opening, were geometrically calculated from 3D coordinates of these points (2). Parameters are completely unrelated to subject head movements because they are calculated as distances.
"Phase diagrams" were used to represent graphically kinematic spatial and temporal characteristics of the parameters. Moreover frequency contents of parameters, velocities and accelerations were calculated in order to perform a spectral data analysis. A statistical comparison between spectral parameters (mean and median) of the two groups of subjects considered (normal and stuttering) was carried out by means of the Student's t tests.

RESULTS

Phase diagrams show that normal subjects present an opening lips mean velocity clearly superior than opening jaw mean velocity while in stuttering subjects these mean velocities are similar. Moreover in normal subjects opening and closing movements seem to be repeated more regularly in time. These differences become more marked when /papapa.../ and /bababa.../ sequences are pronounced at maximal velocity.
From spectral data analysis, statistically significant differences between normal and stutterers were detected only when sequences are pronounced at maximal velocity. In normal subjects the values of spectral mean and median at maximal velocity are around 5 Hz both in /pa/ and /ba/ sequences in all the articulatory parameters we considered. In stuttering subjects these values fall down to 4 Hz. When /pa/ and /ba/ sequences are pronounced at normal velocity the values of spectral mean and median are around 3 Hz in both the two groups of subjects.

By means of Student's t test with level of significance of 5% was established that at maximal velocity spectral mean and median of normal and stutterers are statistically different.
Another interesting observation about spectral mean and median values is that their standard deviation is very low inside the two groups of subjects. This is important because generally there is a great inter and infra-subjective variability in articulatory kinematic merit figures commonly studied.

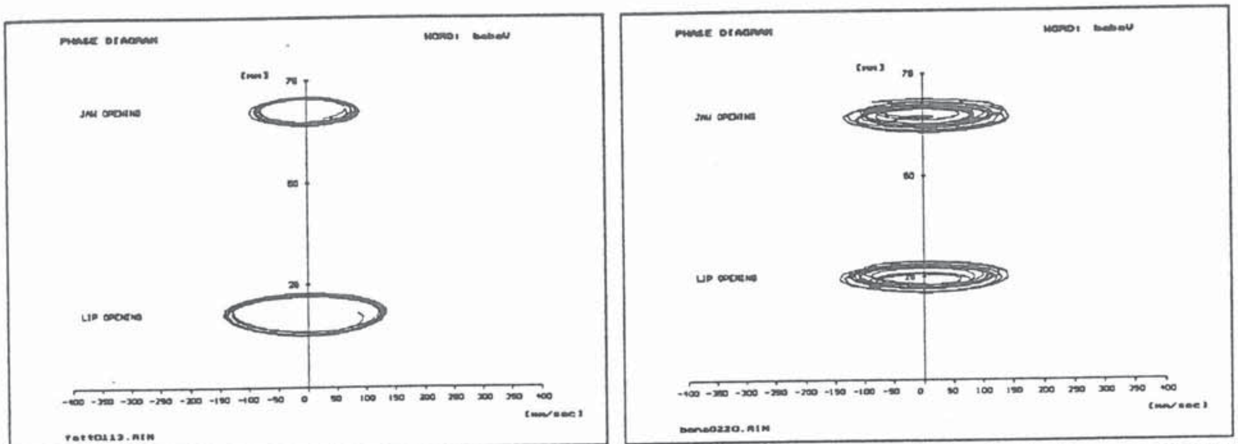


Fig.1 : Phase Diagrams of lips and jaw opening during the pronounce of /bababa.../ at maximal velocity of a normal subject (on the left) and of a stuttering subject (on the right)

Kinematics / Subjects	/PA.../ fast		/PA.../ normal		/BA.../ fast		/BA.../ normal	
	LO	JO	LO	JO	LO	JO	LO	JO
Normals MEAN	4.91	4.82	3.04	2.92	5.13	4.98	2.78	2.79
Normals MEDIAN	4.87	4.87	2.73	2.73	5.07	4.93	2.53	2.53
Stutterers MEAN	3.91	3.99	2.95	2.97	3.81	3.79	2.89	2.87
Stutterers MEDIAN	3.90	3.90	2.73	2.73	3.70	3.70	2.67	2.67

Tab. 1 : Spectral Mean and Median (in Hz) for Lips Opening (LO) and Jaw Opening (JO). Mean Values on three normal and two stuttering subjects.

CONCLUSION

Results show that rieducated stutterers are able to equal kinematic performances of normal subjects during the pronounce of reiterated sequences /papapa.../ and /bababa.../ when the velocity of production is normal while at maximal velocity significant differences are detectable by means of an analysis of articulatory kinematics in time and in frequency.
In particular spectral mean and median of lips and jaw opening, result significant merit figures to characterize the differences between normal and stuttering subjects.

REFERENCES

1. Ferrigno G., Borghese N.A., Pedotti A. (1990) ISPRS Journal of Photogramm. Remote Sensing, vol.45: 227-246.
2. Magno Caldognetto E., Vagges K., Ferrigno G., Zmarich C. (1993) Proc. of Eurospeech, vol.1: 409-412.

A THREE-DIMENSIONAL ANALYSIS OF RELEASE PARAMETERS IN ELITE NETBALL SHOOTING

Stuart Miller
Cardiff Institute of Higher Education, Cyncoed Road, Cardiff, Wales, CF2 6XD.

SUMMARY

The release characteristics of an elite netball shooter for a series of short distance shots was analysed using three dimensional video techniques (50 Hz). Considerable variation was found across trials in both the antero-posterior and medio-lateral foot positioning of the stance. This contributed to variation in the angular position of the hip and shoulder axes. Shoulder, elbow and wrist angles at release were found to be influenced by the timing of release, suggesting that had all shots been released at the same relative time, a greater degree of similarity would have been exhibited. It is possible that the technique adopted by the subject under investigation is repetitive, whilst the different ball release parameters required for shots of changing distance are achieved by alteration of the timing of release during the movement pattern.

INTRODUCTION

Netball, as an invasive team sport, has as its major focus, the objective of scoring more goals than the opposition in four 15 minute quarters of playing time. Thus it may be argued that the skill of shooting is the most important in the game.

The accuracy component of netball shooting is similar to that previously described in basketball (1). The conditions under which shots are taken in the two sports differ, however, in that netball rules dictate that the bodies of defensive players must be 0.9 m away from the shooter. Thus, netball shooting is more of a closed skill than for basketball, giving shooters the opportunity to develop technical consistency. It is therefore regarded as important that shooters should display consistencies in technique on an intra-individual basis.

On this basis, it was the objective of the study to perform a detailed kinematic analysis of the netball shot, as exhibited by an elite English performer.

METHOD AND RESULTS

Two Panasonic F-15 video cameras mounted approximately 1.5 metres above floor level were used to film the performance of an England international shooter during a practice match in the summer of 1993. The procedures recommended by BASS (2) were adhered to. Prior to filming, a volume (2 m x 2 m x 1 m) placed was close to the ring and symmetrical about a line joining the two goals, was calibrated.

Five sequences were subsequently analysed, beginning when either both feet made contact with the floor or on the first perceived initiation of the shooting movement, whichever was the sooner, and finishing 10 frames after ball release. A 14-segment, 18-point model of the human performer was employed to reconstruct a 3-D image from the related object space coordinates. The sampling rate was 50 Hz.

Ball release speed ranged between 3.16 and 4.22 $m.s^{-1}$, values similar to those reported for shots of similar distance in basketball (3). Values for shoulder and elbow angle at release were all above 135°, thus minimising the chance of the shot being blocked or deflected, and was exemplified in the mean release height of the ball above the head of 0.81 m. This was, however, compromised by a trunk angle of less than 90°. Shoulder, elbow and wrist angles at release would seem to be influenced by the timing of release, as correlations ranged between -0.72 (elbow) and -0.87 (shoulder), suggesting that had all shots been released at the same relative time, the shoulder, elbow and wrist angles at that time would have exhibited a greater degree of similarity. Ball release was found to occur prior to the centre of mass reaching its vertical peak, and the centre of mass velocity vector at release revealed a mean speed and direction of 0.61 $m.s^{-1}$ and 87°. Such movement must be compensated for in the selection of release parameters, variations in which rendered this selection increasingly difficult.

Both the shoulders and hip axes were found to be rotated leftwards between 60° and 82° at release. Similar phenomena have previously been reported for basketball shooting (3), and are regarded as beneficial to accuracy by allowing alignment of the eye, elbow, wrist and ball in a vertical plane with the basket. The antero-posterior (A/P) separation of the feet (X = 0.23 m, right foot forward) facilitates this alignment by tending the body to leftwards hip rotation. When combined with medio-lateral (M/L) separation, the stance position also increases the potential stability of the body by increasing the area of the base of support.

DISCUSSION

Netball rules dictate that a defending player must be 0.91 m away from the landing foot of the shooter, thus as all shots were characterised by the right foot being forward of the left, it is recommended that shooters should learn to land on their right foot and step away from the ring to maximise the distance from the defender. Ball release prior to the peak of trajectory has been reported as an aid to the provision of the required release speed, however, due to the use of a comparatively light ball and the close range nature of the shots under discussion, it is unlikely that such a strategy is necessary. It is possible that the technique adopted by the subject under investigation is repetitive, whilst the different ball release parameters required for shots of changing distance are achieved by alteration of the timing of release during the movement pattern.

In conclusion, it is recommended that consistency of technique for the subject investigated may be improved by; release of the ball at the same point with respect to the vertical peak of the centre of mass, preferably at the peak itself; constraining centre of mass movement during release to a vertical plane; establishing a consistent antero-posterior and medio-lateral base of support.

REFERENCES

1. Miller, S.A. (1993). Unpublished Masters thesis, University of Salford.
2. Bartlett, R.M. (ed.) (1993). *Biomechanical Analysis of Performance in Sport*. BASS.
3. Miller, S.A. and Bartlett, R.M. (1993). *Journal of Sports Sciences*, **11**, 285-293.

AUTHOR'S ADDRESS

Stuart Miller, Faculty of Education and Sport, Cardiff Institute of Higher Ed., Cyncoed Road, Cardiff, WALES, CF2 6XD. Tel: 011 44 222 551111 Fax: 011 44 222 747665

GAIT FEATURE CHARACTERIZATION THROUGH MOMENT/ANGLE RELATIONSHIP

C. Frigo* and L.M. Jensen**
* Centro di Bioingegneria, Fondazione Pro Juventute Don Carlo Gnocchi I.R.C.C.S., Politecnico di Milano, Via Capecelatro, 66, 20148 Milano, Italy.
** Dept. for Medical Informatics and Image Analysis, Inst. for Electronic Systems, University of Aalborg, Fredrik Bajersvej 7D, Aalborg, Denmark.

INTRODUCTION

Several biomechanical variables can be obtained from a well equipped gait analysis laboratory. Attempts to synthesize in few curves the main individual features of gait have been done in the past concerning the ground reaction forces (Vector Diagrams, Pedotti, 1977) and the lower limb joint angles (Grieve, 1969). Joint power time course (Winter, 1987), resulting from moment and angular velocity could be considered a way to summarize kinetic and kinematic aspects. Nevertheless the extent by which a joint can use its storage and restitution possibilities does not clearly appear on these curves. The present work has the purpose of demonstrating how the moment/angle cycles can synthetically give prominence to those phenomena connected to work exchange and how the elasticity aspects in locomotion can be depicted.

METHOD

Ground reaction forces and 3-D kinematic variables were obtained by a dynamometric force platform and an automatic motion analyzer respectively (ELITE System, BTS, Milan, Italy). Retroreflective markers were glued above the following anatomical locations: Posterior Superior Iliac Spines, prominence of the Sacrum bone, lateral femoral condyles, lateral malleoli and lateral aspect of the fifth metatarsal heads. An anthropometric model was used to predict the location of the hip, knee, and ankle joint centers. This was based on the following measurements directly taken on each subject: pelvis width and height, length of thigh, shank and foot, diameter of knee and ankle, and distance between first and fifth metatarsal heads. Flexion/extension angles and joint moments were computed for each joint in the sagittal plane. Stride temporal events were detected from ground reaction and kinematic data by an interactive procedure. To obtain comparable data joint moments were normalized to the body weight of each subject and joint angles were referred to the standing up-right position. All data were then normalized in time on the basis of the stride duration. Averages and standard deviations were obtained over ten strides by each subject. The mean SD referred to the peak to peak range of each variable was used as an Index of Variance (IV). Two kinds of barefoot locomotion were analyzed: walking at natural speed and jumping on one foot. All ten subjects analyzed were University students, males, and in good health.

RESULTS AND DISCUSSION

Moment/angle curves from a representative subject (25 years old, 78 kg of body mass, 1.77 m high) are reported in Fig. 1 for the two locomotor conditions. At natural speed a typical two loops cycle was observed at the hip joint. If one remembers that the area between the curve and the horizontal axis represents a work, one can realize that the area included in a loop correspond to the net work production (if the loop is counter-clock wise) or absorption (clock wise loop).

At the knee a spring-like behavior is observed during the yielding phase characterized by a steep rising of the moment quasi-linearly correlated with knee flexion. Return to extension is in large part superimposable to the previous phase. A second phase of linearity is observed during the extension phase which occurs after mid swing. At the ankle joint a loop with a positive net work (inner area) was always observed, characterized by two rather parallel curves in the rising and descending phases respectively. The enhancement of the elasticity phenomena during jumping is well depicted by the quasi-linear behavior of the ankle joint curve and by the reduction of the loop area. Thus, the possibility to exploit the mechanical elastic properties of the joints is well evident in the reported graphs. As to the interpretation of these data it would be tempting to say that modulation of muscles stiffness is a control strategy adopted by the Central Nervous System, at least in some phases, to efficiently fulfill the motor requirements. However the problem of how the pure rehological properties of muscles (passive stiffness and viscosity), active properties (force/length and force/velocity curves under varying activation levels) and reflex loops interact is still to be clarified and prevents drawing simplistic conclusions. Nevertheless the moment/angle representation looks to give an insight on the complex relationships among biomechanical variables and allow to characterize some basic features of a motor performance.

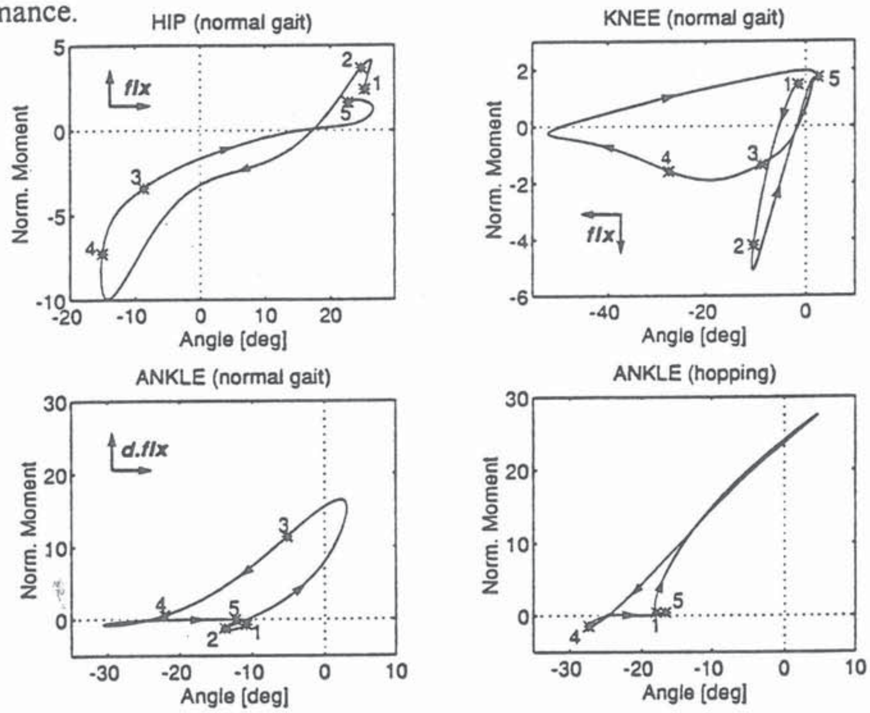


Fig. 1: Example of the moment/angle loop for the hip, knee, and ankle during normal gait and for the ankle when jumping. 1 & 5 corresponds to platform heel strike whereas 2, 3 & 4 indicates contralateral toe off, contralateral heel strike and platform toe off respectively.

REFERENCES

1. D.W. Grieve, "The assessment of gait", Physiotherapy, 55, pp. 452-460, 1969.
2. A. Pedotti, "Simple equipment used in clinical practice for evaluation of locomotion", IEEE Trans. on Biomedical Eng., vol. BME-24, n. 5, pp. 456-461, 1977.
3. D.A. Winter, "The biomechanics and motor control of human gait", Univ. of Waterloo Press, 1987.

THREE DIMENSIONAL CHARACTERIZATION OF MUSCLE LENGTH, LOAD AND VELOCITY

R.V. Baratta, M. Solomonow, G. Nguyen, R. D'Ambrosia
Bioengineering Laboratory, Department of Orthopaedic Surgery,
Louisiana State University Medical Center, New Orleans, Louisiana U.S.A.

SUMMARY

A method is developed to characterize the relationship among muscle load (force) length and velocity under conditions that approximate the free moving of loads. A weight-pulley system is used to apply loads of various weights to the passive muscle, stretching it until its passive elastic force is equal to the load weight. A supramaximal stimulation pulse train is then applied, resulting in the shortening of the muscle and a length-velocity trajectory. A set of trajectories is then assembled into a three-dimensional surface describing the interrelationship between muscle load, length and velocity. Traditional length-force and force-velocity relationships can be obtained by varying the projection of the three-dimensional figure. It was found that for different loads, the maximal velocity occurs at different lengths, and that at extremes of muscle elongation, the force-velocity relationships are not strictly monotonic. It is suggested that the simultaneous consideration of muscle force, length and velocity provides a more comprehensive view of muscle performance than separate length-force or force-velocity curves obtained under different conditions.

INTRODUCTION

Traditional methods of evaluating muscle contractile characteristics include length-force and force-velocity relationships. Length-force relations are generally obtained under isometric conditions, and force-velocity relations are constructed during isotonic or isokinetic contractions after a stretch-stimulate-release protocol. These conditions seldom occur during normal movement, and because of the experimental conditions being different, it is difficult to reconcile these two relationships. Abbott and Wilkie (1) attempted to reconcile these relations mathematically by setting the force variable in Hill's force velocity equation equal to force as a function of length. Bahler et al. (2) and Zierler (3) obtained a graphic representation similar to that envisioned by Wilkie. Experimental data in a three dimensional format were obtained by Fuglevand (4) and by Marshall et al. (5) during isokinetic knee extension. The experimental results were different from what was expected by previous researchers. The possibilities of sub-maximal subject effort, reflexive inhibition and antagonist coactivation must be considered when considering these studies, and as a result, it is still unknown whether the classical assumptions are valid during contractions performed under experimental conditions.

It is, therefore the objective of this study to obtain a three-dimensional characterization of isolated muscle contractile behavior assessing the relationship among load, length and velocity during electrically elicited maximal contractions, and to perform this protocol on several muscles of varied architecture in the cat's hind limb. The information resulting from this study is expected to be useful in ergonomics, human performance enhancement and in the design of Functional Electrical Stimulation systems. In fact, any endeavor necessitating a complete trajectory of muscle force, length and velocity is likely to benefit from the results of this study.

METHODS AND RESULTS

Cats anesthetized with chloralose were used. Their hind limbs were prepared by exposing the sciatic nerve, placing a stimulation electrode on it and denervating all muscles except those to be studied. The distal insertions were dissected free and a tendon holding device attached. A femoral pin was used to hold the proximal end of the limb, and the ankle was disarticulated to allow direct connection to anterior muscles. The muscles studied were Extensor Digitorum Longus, Flexor Digitorum Longus, Lateral Gastrocnemius, Medial Gastrocnemius, Peroneus Brevis, Peroneus Longus, Soleus, Tibialis Anterior and Tibialis Posterior.

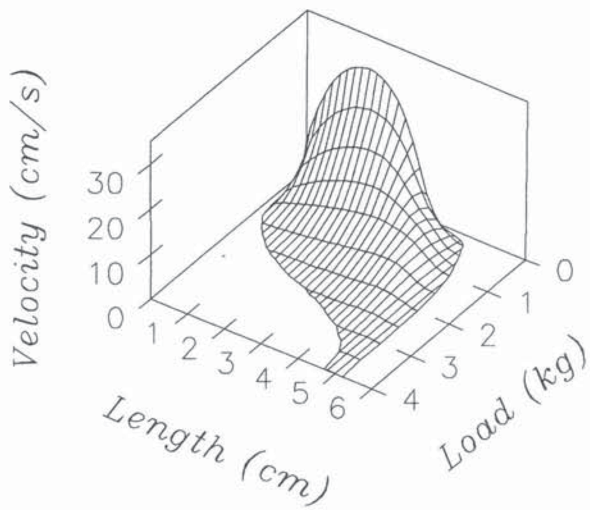


Figure 1

A weight-pulley system was used to apply loads of various weights to passive muscle, stretching it until its passive elastic force equaled the load weight. A supramaximal stimulation pulse train was then applied, resulting in the shortening of the muscle while its length was recorded and its velocity obtained by numerical differentiation, yielding a length-velocity trajectory. Four samples of each muscle were obtained, averaged and smoothed, creating a set of trajectories assembled into a three-dimensional surface such as Figure 1. The typical length-force and force-velocity curves can be obtained by viewing the projections of the figure on the zero-velocity and zero-length planes respectively. In addition, a representation of the maximal velocity as a function of length can be obtained by projecting the figure on the zero-load plane. In the figure, it can be observed that maximal velocity is not obtained at the same length at different loads.

DISCUSSION AND CONCLUSION

The surface obtained represents the upper limits of contractile performance available to muscle under load-moving conditions. An immediate utility to this approach is that any desired trajectory of movement must be contained within this boundary. Of direct relevance to the design of FES systems is that in a linear or quasilinear system, the desired trajectory may be approximated by scaling the stimulation parameters with respect to the surface boundary. From the physiological standpoint, it is interesting to note that the maximal velocity for increasing loads occur at increasing muscle length. This is a direct departure from the theoretical models (1,2,3), but was predicted by Fuglevand (4), and can be observed in Marshall et al.'s (5) graphic representations. In comparing the three dimensional characterizations of the nine muscles, it is apparent that the fiber arrangement which manifests itself in the length-force and force-velocity relations is also present. When compared to parallel fibered muscles, those with sharp pennation angles have generally "narrower" envelopes, with large load capacity at the expense of displacement and velocity. The results of the Soleus also indicate that fiber composition plays a role in determining the overall performance envelope by strongly affecting the maximal velocity.

REFERENCES

1. Abbott & Wilkie, *J. Physiol.,* 120:214-223, 1953.
2. Bahler et al., *J. Gen. Physiol.,* 51:369-384, 1968.
3. Zierler, in *Medical Physiology,* Mosby, 1974.
4. Fuglevand, in *Biomechanics X-A: Human Kinetics,* ISB, 1987.
5. Marshall et al., *Eur. J. Appl. Physiol.,* 61:263-270, 1990.

Supported by the National Science Foundation with Grants BCS 9006639 and BCS 9207007

AUTHOR'S ADDRESS

Richard V. Baratta, Ph.D., Louisiana State University Medical Center, Department of Orthopaedic Surgery, 2025 Gravier Street, Suite 400, New Orleans, Louisiana 70112. Tel. (504)-568-2251, Fax. (504)-568-4466, EMAIL : RBARAT@NOVMS.LSUMC.EDU

PROSTHETIC MASS AND INERTIAL MEASUREMENT PILOT STUDY

R.D. McAnelly, M.D., B. Rogers, M.S., & V. Faulkner, C.P.O.
University of Texas Health Science Center, San Antonio, TX, U.S.A.

SUMMARY

We have used Lephart's method to determine the mass, centers of mass and moments of inertia of 10 transfemoral (above-knee) prostheses and their components in order to develop regression equations for estimating prosthetic component mass and inertial characteristics

INTRODUCTION

In 1955, W. F. Dempster measured the mass, center of mass and moments of inertia of the body parts of 8 cadavers, measurements that could not be scientifically measured in an intact human body. From this and similar data, regression statistics have already been derived for estimating these body segment mechanical properties based on simple anthropomorphic measurements such as leg length, weight, etc. that combined with kinematic and force plate data are used to derive joint moments and powers.

When a transfemoral amputee is studied in gait analysis, however, a common practice is to assume that the parts of the prosthesis have the same mass, centers of mass and moments of inertia as the intact contralateral limb. The time and difficulty of mechanical measurement methods, including disassembly and reassembly of the prosthesis, often make this assumption necessary. Unfortunately, no regression equations exist to predict component mass and inertial characteristics of prostheses as they do for normals.

Using Lephart's Method from 1984, we have taken 10 disassembled, transfemoral prostheses with commonly used components, and have measured their mass, centers of mass and moments of inertia. From this prosthetics data, regression statistics will be derived to predict the mechanical properties of transfemoral prostheses based on simple external measurements, thereby allowing prosthetic kinetic studies to have similar accuracy as those of normal subject studies in human scientific gait analysis.

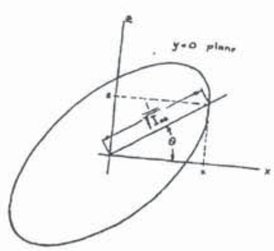


Fig. 1. MOMENTS OF INERTIA

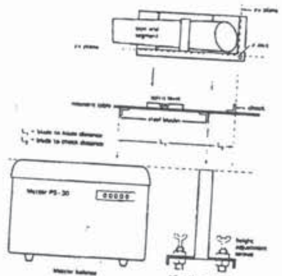


Fig. 2. MOMENT TABLE

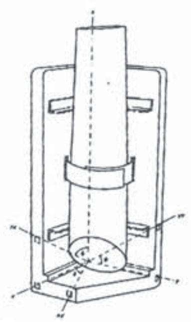


Fig. 3. HOLDING BOX

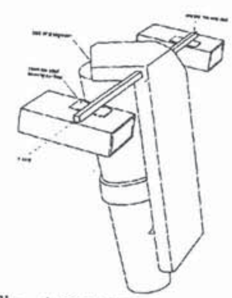


Fig. 4 OSCILLATION MECHANISM

METHODS

Three-sided prosthesis holding boxes were built to secure the prosthesis during testing and to provide a Cartesian coordinate system to reference the prosthesis in space. The boxes are constructed of clear plastic, with three walls at 90 degree angles to each other to correspond to the xy, yz and zx planes. Six pairs of threaded holes drilled into each box accept a pair of collinear rods to serve as a bearing surface for oscillation timing.

1. Mass of the intact prosthesis and each dissassembled component is determined on a Mettler balance.

2. The location of the center of mass is determined using an aluminum moment table.

3. Moments of inertia are determined using Lephart's method. The principal moments and axes of inertia form an ellipsoid in space, and measuring these moments and axes is done dynamically. The segment in question is suspended as a pendulum from an axis under the

influence of gravity, and then the moment of inertia about each of 6 axes is calculated by knowing the distance from the pendulum axis to the center of mass, and by measuring the duration of one pendulum cycle.

Moments of inertia are determined using the six paired holes for rods. The moment of inertia about each axis is determined, a tensor of inertial moments and products is directly calculated and the 3 principle axes of inertia are determined by rotation via the Jacobi method of rotating symmetric matrices.

RESULTS

We will evaluate how closely human cadaver regression equations for estimating human body segment mass, center of mass and moment of inertia fit the data we measure for prosthetic components. Statistically stated, what is the coefficient of determination (r^2 statistic) of traditional cadaver-based regression equation for the prosthetic component mass, center of mass and moments of inertia?

Our prosthetic data will give us our own regression formula for prostheses, and we will evaluate whether this prosthetic regression formula is a major improvement in estimation. Statistically stated, what is the coefficient of determination (r^2 statistic) of our own prosthesis-based regression equation for the same prosthetic component measurements?

CONCLUSION

A regression formula is calculated from prosthetic mass and inertia characteristics for transfemoral prostheses.

REFERENCES

1. Chandler, RF et. al., Investigation of Inertial Properties of the Human Body, Wright-Patterson AFB, Report No. DOT HS-801 430, March 1975
2. Contini, R, Body Segment Parameters, Part II, Artificial Limbs, Vol. 16, No. 1, pp. 1-19, Spring 1972
3. Dempster, WT, Space Requirements of the Seated Operator, Wright Air Development Center, Aero Medical Laboratory Contract No. AF 18(600)-43 Project No. 7214, July 1955
4. Hanavan, EF, A Mathematical Model of the Human Body, Wright-Patterson Air Force Base AMRL-TR-64-102, October 1964
5. Herriot, J. G. 1963 Methods of Mathematical Analysis and Computation. New York, John Wiley and Sons, Inc.
6. Lephart, S. A. (1984) "Measuring the Inertial Properties of Cadaver Segments". J. Biomechanics, vol. 17(7), pp. 537-543
7. Nazre, A.: M.S. Thesis "Design and Development of a New Prosthesis for Above Knee Amputees" Computer Simulation and Experimental Evaluation of Sagittal Plane Kinematic and Kinetic Responses of Prosthetic Limb." Michigan Technological University, 1988
8. Nazre, A.: Ph.D. Thesis "Optimization of a Prosthesis for Above-Knee Amputees", Institut fur Biomedizinische Technik und Krankenhaustechnik, 1991
9. Reynolds, Ph.D. Thesis: Measurement of the Inertial Properties of the Segmented Savannah Baboon, Dept. of Anthropology, Southern Methodist University, March 1974
10. Zatsiorsky, VM et. al. (1985), "Estimation of the Mass and Inertia Characteristics of the Human Body by Means of the Best Predictive Regression Equations" Biomechanics IX-B, Human Kinetics, Champaign, IL, pp. 233-239

AUTHOR'S ADDRESS

Robert D. McAnelly, M.D.
University of Texas Health Science Center
7703 Floyd Curl Drive
San Antonio, TX 78284
phone: (210) 567-5345
fax: (210) 567-5354

A PRELIMINARY KINEMATIC ANALYSIS OF NORMAL BALANCE

P.A. Anderson, G.V. Smith and G. Alon, Dep't. of Physical Therapy,
School of Medicine, University of Maryland, Baltimore, Md., USA

SUMMARY

The purpose of this study was to examine the kinematics of anterior and posterior postural response mechanisms maintaining balance in normal subjects. Two female subjects (mean age 21.5) with no history of disorders or injuries affecting balance were studied. Reflective joint markers were placed on footplate shoes, lateral malleoli, fibular heads, greater trochanters, and T12, C7 and occiput. Six trials were recorded for each subject; three trials per subject were randomly chosen for further study. Latency, magnitude of displacement, and duration of movement was recorded for each point digitized. This data included initial and secondary responses at each joint. The results revealed no trends in either magnitude, latency, or duration of joint excursion.

INTRODUCTION

It is widely recognized that the consequences of falls constitute a serious public health burden (2). In persons 65 years of age or older, falls are the number one cause of accidents (4).

Previous research has established there is a decline in standing balance with increasing age, and that poor balance contributes to a percentage of elderly that fall, (3). However, there is little information to explain this observation. Alterations to joint structures and consequently to alignment may impair mobility and balance and thus, possibly increase the risk of falling.

Balance, defined as the ability to maintain an individual's center of gravity within the base of support, requires the integration of information from cutaneous sensation, proprioception, vision, and the vestibular system, (1). The use of the balance platform test (3) in an anterior-posterior direction most directly simulates actual falling. The ultimate goal is to find the simplest balance test that can provide the highest level of success in predicting falling liability.

METHODS AND RESULTS

Subjects were two female volunteers (ages 21 and 22) with no history of balance disorders or injuries affecting balance. Informed consent was obtained. The study was approved by the Human Volunteers Research Committee (Document 505-58-4058-870003).

The equipment consisted of a MotionSpec balance platform (BTE, Baltimore Therapuetic Equipment). Filming was conducted with two Magnavox VHS HQ Digital cameras with a frame speed of 30 frames per second. The video tape was processed through the Ariel Performance Analysis System (APAS) (V.4.01, Ariel Life Systems, La Jolla, CA) for digitization and transformation into three-dimensional data. Reflective joint markers were attached with double-sided tape to bilaterally, see summary.

Initiation of joint movement was defined as the point at which angular displacement was at least one degree. Latency was determined by subtracting time of initial platform perturbation from time of initiation of joint movement. Analysis of the data revealed two distinct motion curves at each joint in response to the ant/post perturbation. The joint responded with an primary movement, followed by a secondary movement in the opposite direction. The results are summarized in the following table.

			R Ankle	L Ankle	R Knee	L Knee	R Hip	L Hip	T12 intver. Jt	C7 Intver. Jt
Primary Mvmt.	Ant. Perturbation	Lat. of Jt. Mvmt (mS)	0.11	0.06	0.1	0.13	0.1	0.2	n/a	0.12
		Mag. of Jt. Mvmt (deg)	0.1	5.4	12.2	9.8	2	5.2	n/a	6
	Post. Perturbation	Lat. of Jt. Mvmt (mS)	0.09	0.11	0.21	0.26	0.17	0.18	0.09	0.08
		Mag. of Jt. Mvmt (deg)	6.5	3.7	4.4	2.4	17.7	17.3	3	31.9
Secondary Mvmt.	Ant. Perturbation	Lat. of Jt. Mvmt (mS)	0.63	0.57	0.55	0.85	0.91	0.77	0.84	0.705
		Mag. of Jt. Mvmt (deg)	10.1	4.73	7.55	5.74	13.3	8.52	6.83	20.1
	Post. Perturbation	Lat. of Jt. Mvmt (mS)	0.76	0.79	0.71	0.88	1.22	n/a	1.06	0.88
		Mag. of Jt. Mvmt (deg)	0.39	1.11	0.69	0.66	0.99	1.44	0.84	1.2

DISCUSSION

The data about the ankles did not demonstrate any trend. This was for primary or secondary movements of plantarflexion or dorsiflexion direction of perturbation.

The knee's primary movement in both anterior and posterior perturbation was towards flexion. The large excursion produced by forward perturbation is consistent with the anatomy and biomechanics of the knee. The small magnitude of flexion produced by posterior perturbation is a natural reaction to a disruption of balance. The body attempts to lower the center of gravity to decrease the trauma of a fall. In the secondary knee motions, the posterior perturbation elicited greater movement into extension than did anterior perturbation. The data is consistent with the observation that the knee joints were not at full extension at the beginning of each trial.

With the hips, anterior perturbation produced no trends in primary movement. However, posterior perturbation consistently produced a primary movement of flexion. Secondary motions at the hips in response to backwards perturbation produced a greater extension than did anterior perturbation. Like the knee, the secondary reaction is greater in response to posterior perturbation. In both anterior and posterior perturbations, the magnitude of excursion at the hip was greater than that at the knee, which was greater than the range of motion at the ankle.

REFERENCES

1. Andres RO. IEEE/Ninth Annual Conference, Engineering in Medicine, Biology Society,CH 2513,1987,p757-758.
2. MacRae PG et. al. Measures ... Adults,JOSPT,1992,p123-128.
3. Maki BE et.al. J. of Am. Ger. Soc., 1988, p1-9.
4. Shanna JC et.al. Age and Aging, 1988, p94-102.
5. Tideiksaar R. Age, 1988, p111-114.

A FIR NEURAL NETWORK FOR MYOELECTRIC SIGNAL CLASSIFICATION

K.B. Englehart [1] , B.S. Hudgins, M. Stevenson and P.A. Parker
The University of New Brunswick, Fredericton, New Brunswick, Canada

SUMMARY

Recent work by Hudgins [1] has proposed a neural network-based approach to classifying the myoelectric signal (MES) elicited at the onset of movement of the upper limb. A standard feedforward artificial network was trained (using the backpropagation algorithm) to discriminate amongst four classes of upper-limb movements from the MES, acquired from the *biceps* and *triceps* muscles. The approach has demonstrated a powerful means of classifying limb function intent from the MES during natural muscular contraction, but the static nature of the network architecture fails to fully characterize the dynamic structure inherent in the MES. It has been demonstrated [2] that a finite-impulse response (FIR) network has the ability to incorporate the temporal structure of a signal, representing the relationships between events in time, and providing translation invariance of the relevant feature set. The application of this network architecture to limb function discrimination from the MES is described here.

INTRODUCTION

A dynamic network architecture (and an associated learning scheme) has been proposed by Waibel [3] that addresses these limitations of static networks. A time-delay neural network (TDNN) was applied to phoneme recognition. The phoneme recognition problem resembles the MES classification problem: the nonstationary origin of both signals challenges a classification scheme to incorporate the dynamic structure into its decision space. The scheme described by Waibel, however, neccesitates constraints amongst the network weights during training; a redundant architecture requiring a prohibitive training regimen results.

Wan [2] has described a network architecture - which he termed a "FIR network" - that yields the same feedforward response as the TDNN, but is configured such that a more elegant (and efficient) training scheme may be applied. Wan described the training scheme - *temporal backpropagation* - in the context of the problem of time series prediction. This work extends that of Wan to apply the FIR structure and temporal backpropagation to pattern recognition. The case of MES classification is presented as a specific application. The FIR network is very similar to a standard feedforward network, except that each synaptic connection is actually a *FIR filter*, rather than a simple scalar (Figure 1). The neural network no longer performs a simple static mapping from input to output: internal memory has now been added. Since there are no feedback loops, the network as a whole is still FIR.

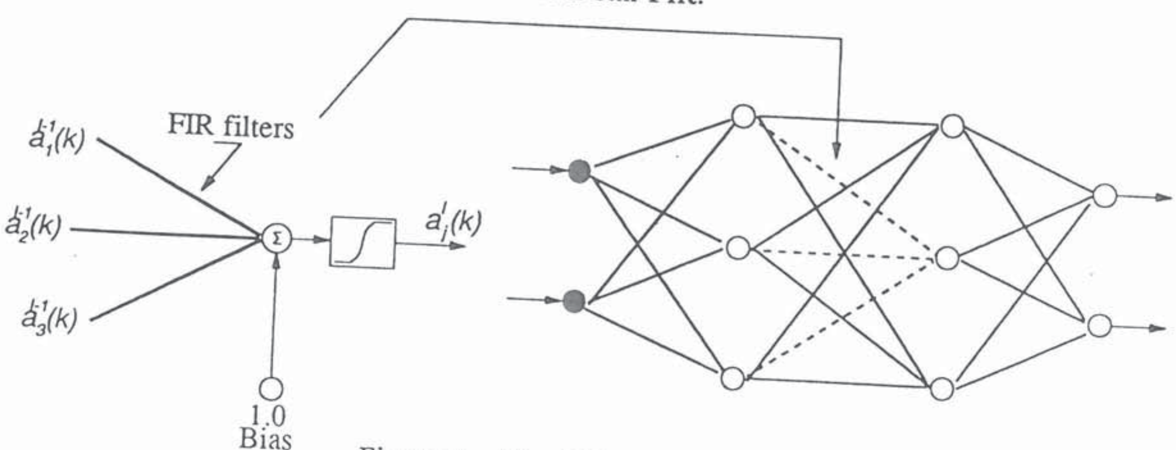


Figure 1 – The FIR neuron and network.

[1]Address correspondance to K.B. Englehart, Department of Electrical Engineering, University of New Brunswick, P.O. Box 4400, Fredericton, N.B. Canada, E3B 5A3
FAX: (506) 453-4827 email: C9LX@unb.ca

The coefficients for the synaptic filter connecting neuron i to neuron j in layer l is specified by the vector $\mathbf{w}_{ij}^{l} = [w_{ij}^{l}(0), w_{ij}^{l}(1), ...w_{ij}^{l}(M^{(l)})]$, where $M^{(l)}$ is the order of the filters in layer l. The output of neuron j in layer l at iteration k is given by the activation value $a_j^l(k)$.

The problem of pattern recognition imposes additional contraints upon the network architecture. Consider the task of classifying a signal observed over a time duration of T *frames*, with the primary information content of the signal occupying D frames ($D \leq T$). The $T - D$ time frames on either side of the signal account for misalignment within the observation window. The sum of the filter orders in the network is related to the observation period: $(M^{(1)} + M^{(2)} + ... + M^{(L)}) = T - 1$. The weights of the network are updated so as to minimize the squared error at the network output after the T frames of the data are clocked in.

The network was used to classify 240 *msec* bursts of MES, described above. Initially, the same feature set chosen by Hudgins was used (zero-crossings, trace length, mean absolute value and differential mean absolute value), acquired from each of six 40 *msec* time frames. Thus, $T = 6$ and the sum of the FIR network orders must be 5. A three-layer network with Nodes: 4x12x4 and Orders: 0 : 5 : 0 was specified. The classification performance of the FIR network and a standard backpropagation network are shown in Table 1; two data sets exhibiting different classification performance are included for illustration. For both the "good" and the "bad" data set, the classification performance of both networks is essentially identical. Reducing the frame size to 12 frames of 20 *msec* (such that $T = 11$) gave slightly better performance for both networks. The FIR has demonstrated that it yields *at least* the classification performance of a static network for this feature set. The strength of the FIR network however, is in capturing the temporal structure of a waveform, suggesting a signal representation that is as temporally resolved as possible – perhaps the raw signal itself. A static network would necessarily need to contain many time-delayed inputs to implement this approach – network training complexity grows as the square of the number of nodes. The FIR network implements the time delays longitudinally; additional time delays (and thus, data points per raw pattern presentation) add to training complexity linearly. This is the subject of current and future investigation. It is anticipated that the intrinsic representation and relation of temporal signal dynamics of this network architecture will provide a means of more resolved and meaningful analysis (and consequently, discrimination) of the myoelectric signal.

Subject	Time Frames	Static Network (correct/trials)	FIR Network (correct/trials)
SR	6	79/80	79/80
CP	6	64/80	64/80
CP	12	65/80	66/80

Table 1 – Classification performance.

REFERENCES

1. **Hudgins, B.S.**, "A Novel Approach to Multifunction Control", Ph.D. Dissertation, University of New Brunswick, Fredericton, New Brunswick, Canada, 1991.

2. **Wan, E.**, "Finite Impulse Response Neural Networks with Application in Time Series Prediction," Ph.D. Dissertation, Stanford University, Stanford, California, November, 1993.

3. **Waibel, A.**, *et al.*, "Phoneme Recognition using time-delay neural networks," *IEEE Transactions on Acoustics, Speech, and Signal Processing*, pp. 328 – 339, Vol. 37, No. 3, March, 1989.

This work was supported by Natural Sciences and Engineering Research Council grants (No.s OGP A000445 and WFA 0107353) and a Department of National Defense contract (No. W7707-3-2683/07-05C).

NEUROMUSCULAR PARTITIONING OF THE HUMAN LATERAL GASTROCNEMIUS: LEVELS OF ELECTROMYOGRAPHIC ACTIVITY DURING ISOMETRIC TASKS

[1,2]R.L. Segal, [1]L.-S. Cheng, [1]P.A. Catlin, [1,2]S.L. Wolf, [2]A.W. English
[1]Division of Physical Therapy, Department of Rehabilitation Medicine, [2]Department of Anatomy and Cell Biology, Emory University School of Medicine, Atlanta, GA, U.S.A.

INTRODUCTION

Muscles crossing more than one joint are usually described as having multiple functions.[1] An important, unanswered question is whether the CNS can control the different actions of multi-arthrodial muscles, and if so, which portion of the muscle form the elements controlled by the CNS to perform these actions. English and Letbetter[2] proposed that the fundamental peripheral elements are neuromuscular compartments. Neuromuscular compartments consist of the muscle fibers innervated by a primary branch of a muscle nerve.[3] Putative neuromuscular compartments (partitions) have been described in human subjects.[4]

The purpose of this study was to investigate the relationship of integrated electromyographic (IEMG) activity recorded from three architecturally defined heads (A, B & C) of the lateral gastrocnemius (LG) of healthy human subjects while performing quantified isometric contractions. These experiments extend the findings of Wolf et al.[5] by quantifying force levels produced while recording activity in the three heads of the human LG. In particular, we sought to answer two questions related to the concept that the architectural partitions of LG are functional partitions: 1) Are there task specific activation patterns within the heads? and 2) Are the heads selectively activated during specific tasks?

METHODS

Sixteen (7 male and 9 female) healthy volunteers (mean age 29.1$\pm$3.8 years) participated. Four pairs (2 in head A and 1 each in heads B & C) of fine-wire[6] EMG electrodes were inserted into sites determined from the results of Segal et al.[4] and measurements made in eleven cadaver legs, where the spatial relationships between adjacent bony landmarks and LG heads were determined. Because LG crosses both the knee and ankle joints, tasks were performed to examine isolated knee flexion and plantarflexion, and also combinations of knee flexion and plantarflexion. The five isometric tasks, sequenced in **random** order across subjects, were knee flexion, plantarflexion (ankle at 90°), plantarflexion (ankle fully plantarflexed), knee flexion with plantarflexion (ankle at 90°) and knee flexion with plantarflexion (ankle fully plantarflexed). Knee flexion and plantarflexion forces were recorded by force transducers. EMG activity was amplified, rectified, and integrated. The IEMG was normalized as a percentage of the middle 10 seconds of a 20 second maximal isometric plantarflexion contraction with the ankle at 90°, knee fully extended and the subject stabilized. Normalized IEMG values were used for all comparisons.

A one-way repeated measures analysis of variance (ANOVA) was used to test for significant differences in normalized IEMG activity recorded at different insertion sites for each task and in normalized IEMG activity recorded at individual insertion sites for different tasks. If the results of the ANOVA were significant, the Tukey's post-hoc test was used to find where the differences occurred. In addition, a random effects nested ANOVA was used to determine the major source of variability.

RESULTS

The anatomical partition designated Head A was more active when plantarflexion was part of the task (Table). Head B was relatively active across all tasks (Table). Head C tended to be more active when knee flexion was part of the task (Table). The largest amount of normalized IEMG activity recorded at all electrode sites during any task was during the combined action of knee flexion and ankle plantarflexion (ankle in fully plantarflexed position). Statistically significant differences in IEMG activity between heads was only found during plantarflexion {Both head A (proximal) and head B were more active than head C}.

Table: Ratios of Normalized Integrated Electromyographic Activity by Task. The ratios are in the parentheses and the LG heads are listed from left to right starting with the Head with largest amount of EMG activity. The ratios are obtained by using the mean activity from the location with the <u>lowest</u> amount of activity as the denominator. Abbreviations: A(P), head A proximal; A(D), head A distal.

Knee Flexion	C $\geq$ (4.7)	B > (4.31)	A(P) > (1.57)	A(D) (1)
Plantarflexion (Neutral)	A(P) $\geq$ (4.38)	B > (4.1)	A(D) > (3.1)	C (1)
Plantarflexion (Plantarflexed)	B $\geq$ (4.35)	A(P) > (3.92)	A(D) > (3.2)	C (1)
Knee Flexion, Plantarflexion (Neutral)	B > (1.87)	C $\geq$ (1.32)	A(P) $\geq$ (1.29)	A(D) (1)
Knee Flexion, Plantarflexion (Plantarflexed)	B $\geq$ (1.29)	A(P) $\geq$ (1.26)	A(D) $\geq$ (1.07)	C (1)

CONCLUSIONS

The first important feature of this study is related to differences in activation between single-jointed and two-jointed tasks. The motor unit activation patterns were clearly affected by whether the task required torques about one or two joints. Moreover, the patterns of activation during single-jointed torque generation depended on which joint was the target of the task. Second, the idea that partitions may form parts of different synergies for different tasks may radically change how we look at "whole" muscles. The organization of lower limb movement may not be centered around the traditional conception of muscles, as taught in gross anatomy classes, but may be centered around neuromuscular partitions.

REFERENCES

1. Williams PL, Warwick R, Dyson M, Bannister H (1989) Gray's Anatomy. Churchill Livingstone, New York
2. English AW, Letbetter WD (1982) Anatomy and innervation patterns of cat lateral gastrocnemius and plantaris muscles. Am J Anat 164: 67-77
3. English AW, Weeks OI (1984) Compartmentalization of single motor unit in cat lateral gastrocnemius. Exp Brain Res 56: 361-368
4. Segal RL, Wolf SL, DeCamp MJ, Chopp TM, English AW (1991) Anatomical partitioning of three multiarticular human muscles. Acta Anat 142: 261-266
5. Wolf SL, Segal RL, English AW (1993) Task oriented electromyographic activity from the human lateral gastrocnemius muscle. J Electromyogr Kinesiol 3:87-94
6. Basmajian JV, Stecko GA (1962) A new bipolar indwelling electrode for electromyography. J Appl Physiol 17: 849

STRUCTURE AND FUNCTION OF THE ABDOMINAL MUSCLES DURING PREGNANCY AND THE IMMEDIATE POST-BIRTH PERIOD

W.L. Gilleard#, J.M.M. Brown##
#Dept. of Biological Science, University of Sydney, Lidcombe, NSW, Australia.
##Dept. of Biomedical Science, University of Wollongong, Wollongong, NSW, Australia.

SUMMARY

The gross structural adaptations of Rectus Abdominis (RA), the level of EMG signal activity produced by the abdominal muscles during an isometric task, and the muscles' functional capabilities during pregnancy and the immediate post-birth period were investigated in six primigravida subjects. Increases were seen in RA length, angles of insertion, and separation as pregnancy progressed. Separation of RA showed significant reversal by week four post-birth but was not complete by week eight. There was no significant variation in the level of EMG signal activity and mean frequency of the signal from test to test. The muscles' functional capabilities decreased as pregnancy progressed, and remained low post-birth. The observed functional decrement of the abdominal muscles as pregnancy progressed and continued decrement post-birth was thought to be related to an altered biomechanical environment consequent to structural adaptations rather than a reduction in the ability to produce force.

INTRODUCTION

The functional capability of a muscle is related to its ability to produce torque, which may be affected by a change in either force or moment arm length. Anecdotally, the abdominal muscles are thought to thin as pregnancy progresses, and therefore are able to generate less force and consequently will have reduced functional capability. However, Lalatta Costerbosa et al., (1) have reported hypertrophy in animal Rectus Abdominis (RA) as pregnancy progressed. Therefore functional decrement in human abdominal muscles may be the result of structural adaptations that affect moment arm length.

Limited data is currently available on the structure, ability to produce force, and functional ability of the abdominal muscles during pregnancy and the immediate post-birth period. Therefore, the aims of this study were to determine the gross structural adaptations of a representative abdominal muscle RA, to investigate the level of EMG signal activity produced by the abdominal muscles during an isometric task, and to determine the muscles' functional capabilities during pregnancy and the immediate post-birth period.

METHODS AND RESULTS

Six primigravidas participated in nine test sessions from 14 weeks gestation until eight weeks post-birth. Three dimensional structural data was collected using Direct Linear Transformation with two laterally placed still cameras, to enable the calculation of RA separation above, at and below umbilicus, length of RA, and RA proximal and distal angle of insertion in sagittal and coronal planes. Surface EMG from upper and lower RA, and External Oblique were recorded during a supine maximal isometric contraction. The integrated EMG (IEMG) and mean frequency (MF) were calculated. The IEMG for each electrode was then summed to give total IEMG (tot-IEMG). Abdominal muscle functional ability was assessed by their ability to stabilise the pelvis against resistance. Ten nullipara participated in four EMG tests at intervals of four weeks.

Significant increases ($p<0.05$) were seen in primiparas' RA length, angles of insertion, and separation between 18 to 30 weeks. A further significant change was also seen between weeks 26 to 38 for all structural measures except RA length and RA separation below umbilicus. RA separation showed significant reversal at all sites by week four post-birth but reversal was not complete by week eight. There was no significant variation in nulliparas' and primiparas' tot-IEMG and MF from test to test. The ability to stabilise the pelvis against resistance was found to decrease as pregnancy progressed, and for the majority of subjects remained low post-birth.

DISCUSSION

The length of RA was increased as pregnancy progressed. However, as MF is related to muscle fibre diameter (2), the results indicate that muscle fibre diameters were not altered. The level of muscle activity, as indicated by tot-IEMG was also unchanged. Therefore the ability of the abdominal muscles to produce tension may be maintained throughout pregnancy. The muscles' biomechanical environment however was significantly altered. For RA, significantly increased angles of insertion in the sagittal plane by 30 weeks gestation may decrease moment arm length, as shown in Figure 1. Therefore the ability to produce torque may be reduced. A reduced ability to produce torque was reflected in the reduced ability to stabilise the pelvis against a resistance as pregnancy progressed. Post-birth, angles of insertion were not able to be calculated, however, continued functional decrement and incomplete reversal of RA separation by week eight post-birth indicates the abdominal muscles may remain biomechanically disadvantaged up to eight weeks post-birth.

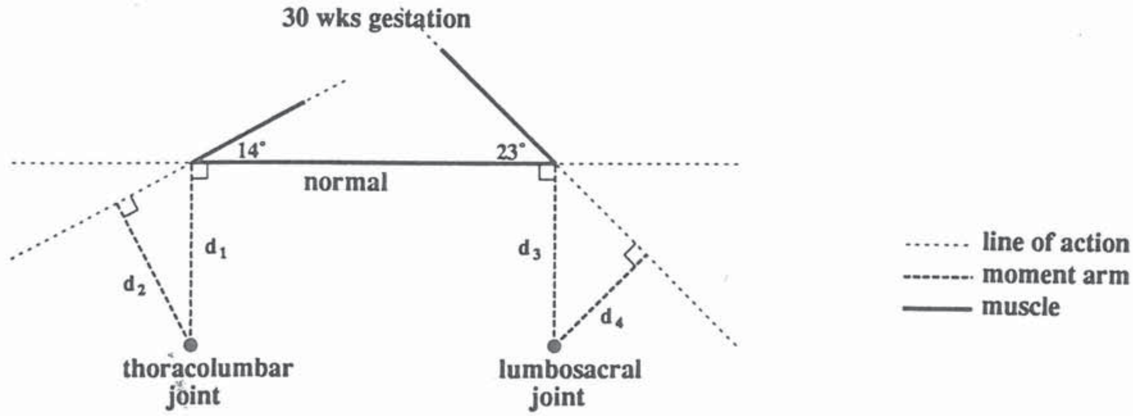


Figure 1. The Normal and 30 Weeks Gestation Line of Action for RA in the Sagittal Plane.

REFERENCES

1. Lalatta Costerbosa, G., Barazzoni, A., Lucchi, M., et al. (1987) Anat. Rec. 217: 23-29.
2. Basmajian, J., DeLuca, C. (1985) Muscles Alive, Williams and Wilkins.

AUTHOR'S ADDRESS

Wendy L. Gilleard, MSc(Hon)
Department of Biological Science, Faculty of Health Science
University of Sydney
PO Box 170, Lidcombe, NSW, 2141, Australia.
Phone: 02 646 6528; Fax: 02 6464853; E-mail: bi_gilleard@cchs.su.edu.au

MOMEMT ARM AND MUSCLE LENGTH EFFECTS ON EMG PRODUCTION IN HUMAN GASTROCNEMII MUSCLES

MOHAMMAD R. NOURBAKHSH AND CARL G. KUKULKA

Tehran University, Physical Therapy Dept, Tehran, Iran and Ithaca College, Physical Therapy Dept., River Road Laboratories, 300 East River Rd, Rochester, NY 14623

INTRODUCTION

The effect of muscle length on the force generating capability of muscle is well established in animal preparations where the manipulation of muscle length can be well controlled. Although experiments in humans have in general confirmed many of the findings from animal experiments, the isolated manipulation of muscle length in human experiments is not possible. That is, in most experiments on humans, muscle length changes are produced by manipulation of a joint angle, a procedure which changes not only the muscle length, but also the muscle moment arm. Previous reports of changes in motor neuron activity attributed to muscle length changes could to some extent be due to contributions of moment arm changes subsequent to joint angle alterations. In an effort to better understand the contributions of both moment arm and muscle length changes to EMG output, experiments were designed to attempt to separately manipulate moment arm and muscle length changes of the human gastrocnemii muscles.

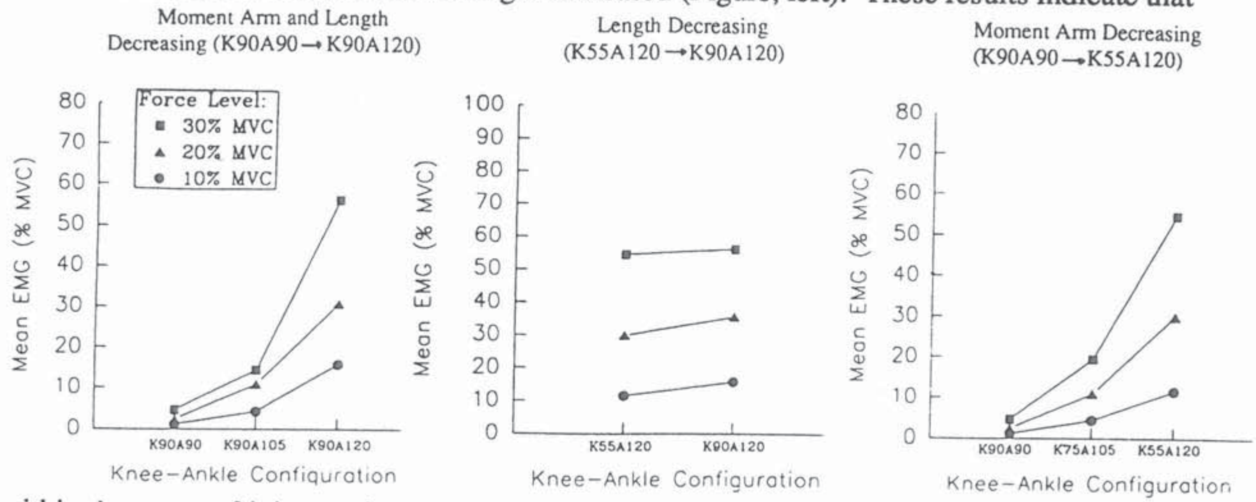
MATERIALS AND METHODS

This study was conducted in two phases. The purpose of the first group of experiments was to establish a set of knee-ankle angles inwhich the triceps surae moment arm at the ankle would vary but the overall muscle length of the medial gastrocnemius muscle remained relatively constant. Eight unembalmed, frozen human lower leg specimens from 6 different cadavers were thawed, the triceps surae muscle complex was exposed, and the muscles about the distal femur were dissected away to expose the distal shaft of the femur and knee joint. The leg was secured to a stabilization board which allowed manipulation of both the knee and ankle joints. The muscle length of MG was measured in a designated neutral position (ankle 90^{o}-knee 90^{o}), and from there, the ankle angle was positioned at either 105^{o} or 120^{o}. At each new ankle angle, the knee joint was manipulated so as to restore the MG muscle length to its original resting length, thereby allowing us to establish 3 pairs of knee-ankle angles for which MG muscle length remained relatively constant.

In phase two of the study, EMG changes in triceps surae were measured during isometric ramp up-hold-ramp down contractions for 5 different knee-ankle angle configurations. Three of these configurations were those established in phase one of this study and were designated as conditions under which moment arm changed, but muscle length remained constant. Two other knee-ankle angle configurations were also chosen in which both the moment arm and muscle length changed. Ten subjects performed constant velocity (5% MVC/s) isometric plantarflexion ramp contractions to 10, 20 and 30 % MVC under 5 different knee-ankle angle configurations . Surface EMG was recorded from MG, LG and SOL muscles although results only from MG muscle during the holding phase of the contractions will be reviewed here.

RESULTS and DISCUSSION

The left graph of the Figure depicts the pooled results for SOL EMG changes as both the moment arm and muscle length were changed concurrently and shows that as both ankle moment arm and MG muscle length decreased (K90A90→K90A120) EMG increased. When the ankle moment arm was held constant and the muscle was shortened (K55A120→K90A120) the EMG displayed a modest, statistically insignificant increase (center Figure). Finally, when the moment arm was decreased, assuming a constant muscle length (K90A90→K55A120) the EMG again was seen to systematically increase (Figure, right) in a manner similar to that for when both combined moment arm and muscle length decreased (Figure, left). These results indicate that



within the range of joint motions tested, changes in triceps surae EMG activity were determined predominately by changes in Achilles tendon moment arm. Although it was anticipated that moment arm effects should contribute to EMG output, the magnitude and predominance of this effect were unanticipated. An explanation for this finding may reside in possible nonlinear changes in both moment arm and muscle fiber lengths occurring throughout the arc of joint motion. Achilles tendon moment arm has been shown to be greatest in the ankle neutral (90^o) position, decreases slightly when the ankle is dorsiflexed to 70^o and reaches a minimum at 120^o plantarflexion (Spoor et al. 1990). MG muscle has been estimated to undergo approximately a 43 mm change in length (Woittiez et al. 1983) although the precise relationship between joint angle and muscle length over the entire arc of motion has yet to be determined. We hypothesize that muscle length changes might predominate when the ankle is dorsiflexed beyond 90^o and display only minimal changes for plantarflexion angle changes as tested here. Such a scenario would result in moment arm effects predominating between 90^o and 120^o plantarflexion, which were examined in this study and muscle length effects predominating between 90^o and 70^o of ankle dorsiflexion. Further experiments are required to elucidate upon this possibility.

References

Spoor CW, van Leeuwen JL, Meskers CGM, Titulaer AF, Huson: Estimation of instantaneous moment arms of lower-leg muscles. J Biomechanics 12:1247-1259, 1990.

Woittiez RD, Rozendal RH, Huijing PA: The functional significance of architecture of the human triceps surae muscle. In Biomechanics IX-A. Winter et al (eds) Human Kinetics Pub. Champaign, IL 1983, pp 21-26.

WRIST JOINT RESTRICTION: EFFECT ON FUNCTIONAL UPPER LIMB MOTION DURING PERFORMANCE OF THREE FEEDING ACTIVITIES

J.E. Cooper[1], E. Shwedyk[2], A. Quanbury[1], J. Miller[2], K. Nemeth[1]
Division of Occupational Therapy[1], Department of Electrical and Computer Engineering[2], University of Manitoba, Winnipeg, Manitoba, Canada.

INTRODUCTION

Ability to carry out essential activities of daily living (ADL) such as feeding is dependent upon upper limb function and successful task performance requires a functional range of motion (ROM) at the shoulder, elbow, forearm and wrist joints as well as normal sensation and prehension of the hand.[1] Any disruption of upper limb mobility can interfere with normal execution of ADL and may result in disability.[2] Syndromes that give rise to pain and subsequent loss of function in the upper limb are a major cause of morbidity in all age groups; it has been estimated that 9.6 million people in the U.S. between the ages of 18 and 64 have limitations in performance of ADL.[3,4] Treatment of upper limb joint pain commonly involves joint protection techniques, including provision of an orthosis to restrict motion during performance of ADL.[5] Evaluation of the effectiveness of an orthosis requires baseline knowledge of normal functional ROM and the impact of joint restriction on total upper limb motion. Grip strength and a standardized hand function tests have been used to quantify the effect of wrist orthoses, however little is known about their impact during performance of actual functional tasks.[6,7] Because a previous investigation[8] showed that elbow joint restriction led to increased upper limb ROM, primarily at the glenohumeral joint, a study was conducted to determine if similar changes would result from wrist joint restriction.

METHODS

Two groups (10 male, 10 female) of volunteer subjects age 18-49 years participated in the study. All were right dominant and in good health. Because pain levels and preexisting joint range restriction could not be held constant across subjects, individuals with upper limb and/or shoulder girdle pathology were excluded. A commercially available elastic wrist orthosis adjusted to 20° extension was fitted to the wrist of the dominant limb to provide the restricted condition. Each subject performed three standardized feeding tasks (eating with a fork and spoon, drinking from a cup) under both restricted and unrestricted conditions. The sequence of tasks and conditions was randomly assigned; six repetitions of each task under both conditions were performed. A kinematic model of the upper limb was used which provides for eight rotations in four upper limb joints (glenohumeral, elbow, forearm and wrist); an Euler angle system was used to define the joint motions. Motion was recorded simultaneously on three videocassette recorders; three of the four middle repetitions were analyzed using the software of the UM^{2}AS system.[9] The mean of the minimum and maximum wrist joint positions and resultant arc of motion during each repetition were calculated for each task and condition and compared using a randomized block design ANOVA followed by a post hoc Newman-Keul's Multiple Range Test. Significance was set *a priori* at .05.

RESULTS

The orthosis had no significant effect on the maximum and minimum positions and ROM at the glenohumeral, elbow and radioulnar joints. In both male and female subjects wrist motion was confined to extension, however only the males had significantly less ROM in the restricted condition. There was a significant difference between males and females in wrist deviation in the unrestricted condition - motion in males was confined to the radial deviation range while in females range extended into both radial and ulnar deviation. Application of the orthosis resulted in a significant decrease in ROM in both groups and in females the motion shifted entirely to the ulnar deviation range.

CONCLUSION

The rationale for prescribing an orthosis for joint protection is to restrict unwanted motion. A commercially available elastic orthosis was used in this study to determine if it had an effect on motion at the wrist joint. It was found that ulnar deviation was restricted in both males and females, however there was no difference in joint position or ROM in flexion/extension in females. Wrist motion in females was therefore performed against the resistance offered by the orthosis; this would require increased muscle force which in turn would generate a compression force across the joint. This suggests that over time the wrist joint might be at risk for articular and soft tissue damage. The investigation should now be extended to subjects with wrist pathology and electromyography should be used to compare levels of forearm muscle activity in both unrestricted and restricted conditions.

REFERENCES

1. Jonsson, E., Brattstrom, M., Lidgren, L. (1988) Scand J Rheumatol, 17:17-26.
2. Bowers, W.H. (1992) Occupational Disorders of the Upper Extremity. London, Churchill Livingstone: 291-299.
3. Hadler, N. (1977) Arth Rheum, 20:1019-1025.
4. Pope, A., Tarlov, A. (1991) Disability in America:Toward a National Agenda for Prevention. Washington DC, National Academy Press.
5. Nordenskiöld, U. (1990). Arth Care Res, 3:158-162.
6. Stern, E. (1991). Am J Occup Ther, 45:42-49.
7. Carlson, J., Trombly, C. (1983). Am J Occup Ther, 37:167-175.
8. Cooper, J., Shwedyk, E., Quanbury, A., Miller, J., Hildebrand, D. (1993) Arch Phys Med Rehabil, 74:805-809.
9. Safaee-Rad, R., Shwedyk, E., Quanbury, A. (1990) Med Biol Eng Comput, 28:569-573.

Supported by the Rehabilitation Centre for Children Research Foundation, Winnipeg MB

AUTHOR'S ADDRESS

Juliette E. Cooper, Ph.D.
Division of Occupational Therapy
University of Manitoba
770 Bannatyne Avenue, Room T263
Winnipeg, MB
R3E 0W3

Telephone: (204)789-3413
Fax: (204)775-7104
e-mail: cooper@bldghsc.lan1.umanitoba.ca

FINGER REACTION ANALYSIS WITH JAPANESE LANGUAGE ENTRY KEYBOARD

Yoshitomo Okada
Department of Industrial Engineering, School of Engineering, Nippon Bunri University

Katsumi Suzuki
Department of Orthopedic Surgery, School of Medicine, University of Occupational and Environmental Health, Japan

INTRODUCTION

In order to obtain a truly ergonomic Japanese Language Entry Keyboard, it costs much time and care to cope with the multi-variable parameters of the keyboards which causes the problems of both the appratus and the operator.

This research has determined each learning curve of various types of entry keyboards, i.e., JIS keyboard, new type JIS keyboards, and M type keyboard, so as to totally estimate their respective efficiency.

METHODS AND RESULT

Three persons for each type keyborad were selected from among the sixty students of the nursing course of a junior college. They had no experience in operating any entry keyboards, and their main body dimensions are 5%ile, 50%ile, or 95%ile. The experiments were carried out to determine the 90 hours' learning curve of each type keyboard.

As a result, the Romaji entry operation with JIS keyboard shows comparatively high entry speed at the earlier stage of learning, but in the course of practicing, 30% comes to suffer slowdown, so consequently this type is considered to bring demerit in a long-term use. The new type JIS keyboard gives entry speed almost equal to that given by the Sholes Keyboard which is used for English writing. The initial entry speed of the M type keyboard varies widely, through the M type gives almost higher entry speed than the Sholes keyboard.

DISCUSSION

On the ergonomic aspect, the M type is the best of the three types, and reduces the working stress in the keybord entry operation because of the low EMG load on the hands and arms.

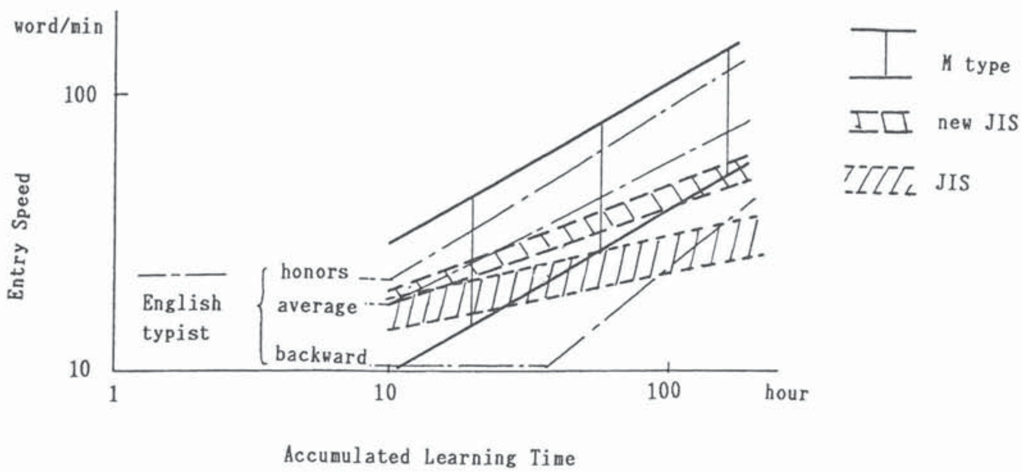


Fig. 1 Learning Curve (Total)

REFERENCES

1. Gen Ooiwa, (1984) Touch Type Entry Method of Japanese, Proc. of Bio-mechanism , Japan
2. Masanori Morita, New Entry Method of Japanese , Japanese Language Entry Methods, 13-2

AUTHOR'S ADDRESS

Yoshitomo Okada, Professor
Nippon Bunri Unversity
Department of Industrial Engineering
1-1 Itigi, Ooita-si, Ooita, JAPAN

PERTURBATION OF HUMAN MASTICATORY EMG-ACTIVITY BY NOCICEPTIVE LASER STIMULI

Peter Svensson[1] & Lars Arendt-Nielsen[2].

[1]Department of Prosthetic Dentistry & Stomatognathic Physiology, Royal Dental College, University of Aarhus, [2]Department of Medical Informatics, Experimental Pain Laboratory, University Center of Aalborg, Denmark.

INTRODUCTION

Jaw-muscle reflexes have mainly been elicited with innocuous mechanical or electrical stimuli and during static conditions. Little is known of muscle responses evoked by stimuli activating predominately nociceptive afferents and during dynamic conditions. The aim of this study was to apply painful laser stimuli and describe the EMG-response of human jaw-closer and jaw-opener muscles and the movement of the mandible during masticatory activity.

METHODS

An argon laser was used for brief (0.2 sec) nociceptive stimulation of the upper lip. A pain threshold (PT) was defined as an instant sharp and painful pin-prick perception. Bipolar surface electrodes were used to record the EMG-response of the right masseter (jaw-closer) and the suprahyoid muscles (jaw-opener) during voluntary opening and closing of the jaws. In the closing phase the subjects contracted the jaw-closers with a constant percentage of their maximal voluntary contraction (MVC). This level was controlled by visual EMG-feedback from the masseter muscle. The vertical movement of the mandible was monitored by a mechanical tracking device attached to the lower incisors. An auditory feedback was provided by headphones to keep a constant duration of the masticatory cycle (1 sec). The onset of the opening phase marked the start of the masticatory cycle and the laser could be triggered by a computer at various percentages of the cycle. For averaging, 8 nociceptive laser stimuli were applied and the EMG-activity and the movements were recorded 3 sec pre-stimulus and 5 sec post-stimulus.

The study comprised of 3 dynamic conditions each including ten young men.

Phase dependence: Laser stimuli were presented at 20%, 50% and 70% of the masticatory cycle in random order. In each cycle, the maximal EMG-activity during closing was attempted to match 20% MVC; stimulus intensity was constant (1.5 times PT).

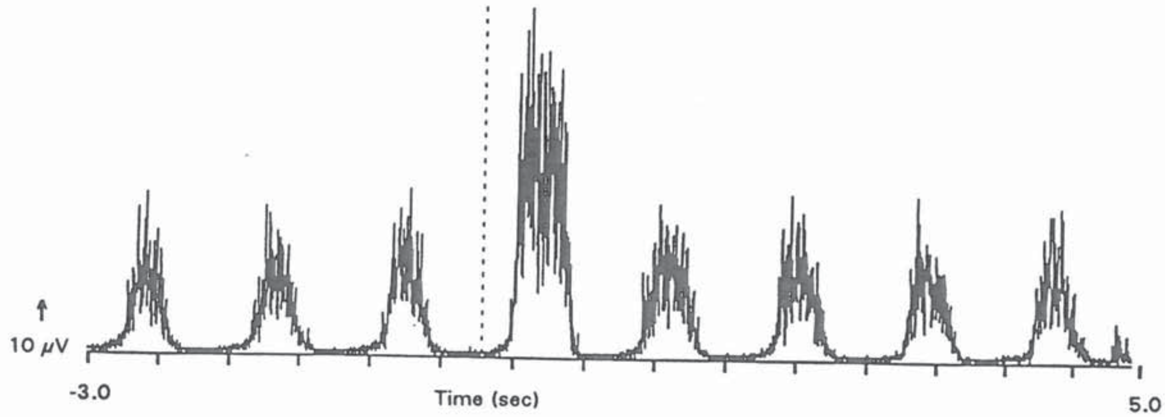
Force dependence: Laser stimuli were applied to subjects with maximal closing activity matching about 10%, 30%, and 50% MVC in random order. Each stimulus was applied at 20% of the cycle; stimulus intensity was constant (1.5 times PT).

Stimulus intensity dependence: Laser intensities of 0.4, 1.1, and 1.8 times PT were applied in random order. Each stimulus was applied at 20% of the cycle and with maximal closing activity near 30% MVC.

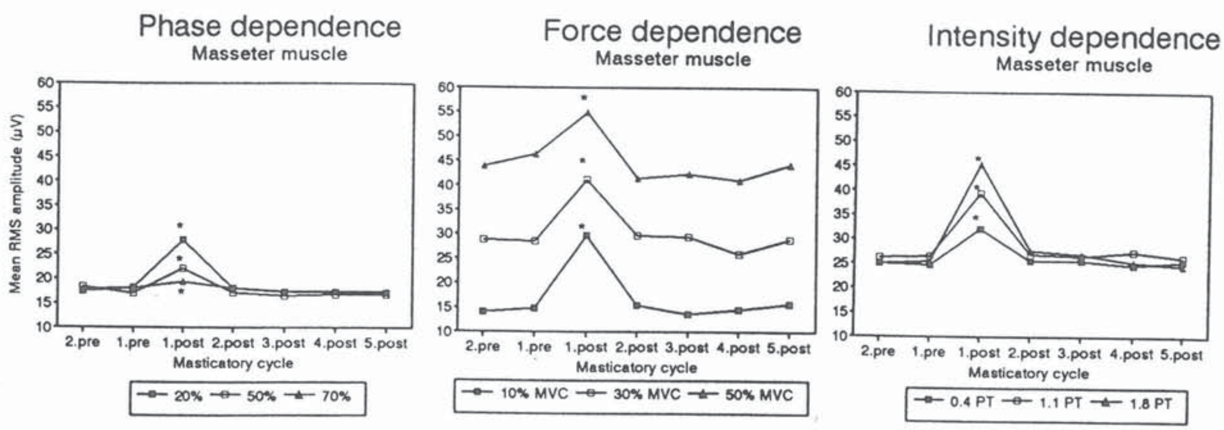
The root-mean-square (RMS) values of the recorded 8-sec EMG-signals were calculated consecutively in 1-sec time windows. Pre-stimulus activity was used as control. Significance was accepted at $P < 0.05$ using Wilcoxon's signed rank test for paired samples.

RESULTS

Overall, this study demonstrated a consistent and highly significant increase in jaw-closing EMG-activity in the first post-stimulus masticatory cycle (Figures). The excitatory reflexes of the jaw-closers were phase- and stimulus dependent but were not influenced by the level of MVC. The jaw-openers showed a similar but weaker trend.



Example of averaged EMG-recordings from the masseter muscle with 3 pre- and 5 post-stimulus masticatory cycles. Vertical line indicates onset of laser stimuli.



Mean RMS-values (N=10) of EMG-activity in 2 pre- and 5 post-stimulus masticatory cycles. * $P < 0.05$.

CONCLUSIONS

Nociceptive activity in cutaneous afferents perturbed masticatory activity of both jaw-openers and closers. This may represent a functional modulation of the motor strategy designed to continue the movement and preserve the function.

AUTHOR'S ADDRESS

Peter Svensson, DDS, Ph.D.
Dept. of Prosthetic Dentistry and Stomatognathic Physiology, Royal Dental College, Faculty of Health Sciences, University of Aarhus, Vennelyst Boulevard, DK-8000 Aarhus C, Denmark. Phone: + 45 86 13 25 33 - ext. 3291
Fax: + 45 86 19 56 65
E-mail: PSVENSSON@AADENT.DK

INFLUENCE OF FORCE OUTPUT AND MUSCLE LENGTH ON ERECTOR SPINAE EMG MEDIAN FREQUENCY.

A.F. Mannion, P. Dolan. Comparative Orthopaedic Research Unit, Department of Anatomy, University of Bristol, UK.

INTRODUCTION

In fresh, unfatigued muscle the median frequency of the EMG power spectrum is proportional to the average muscle fibre conduction velocity (Stulen and De Luca 1981). It has been suggested that conduction velocity can be considered - along with twitch torque, rise time and half-relaxation time - as one of the "size principle" parameters on which the orderly recruitment of muscle fibres is based (Andreassen and Arendt-Nielsen, 1987). Accordingly, the increase in median frequency with increasing force output, observed in many skeletal muscles, is attributed to the recruitment of progressively faster conducting motor units containing muscle fibres with correspondingly higher conduction velocities. However, since the muscle fibre conduction velocity is also known to be related to the diameter of the active fibres, it may simply be the typically larger size of the later recruited muscle fibres that is responsible for the observed increase in conduction velocity with increasing force output in most muscles. The erector spinae are unusual muscles in that the mean cross-sectional area of the Type II (fast twitch) fibres is reputedly smaller than that of the Type I (slow twitch). If the critical factor controlling motor unit recruitment is indeed motor unit type (Sypert and Munson, 1981) then, in this muscle group, the relationship between force output and EMG median frequency may be quite different. The aim of the present study was to examine this relationship in the erector spinae during force generation at differing muscle lengths.

METHODS

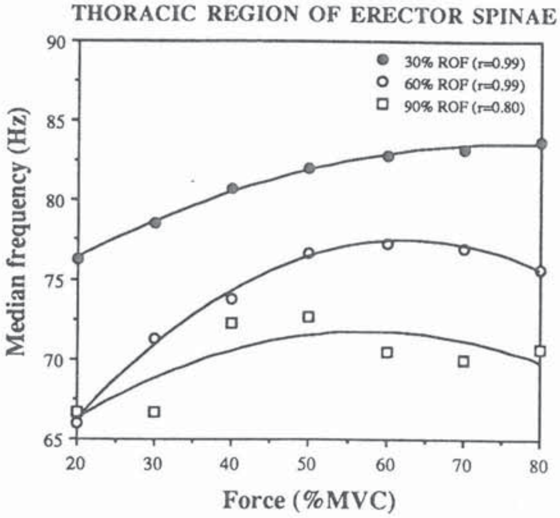
Eight young healthy male volunteers took part in the study. Range of flexion (ROF) of the lumbar spine was determined as the difference between the lumbar curvature (as measured by the 3-Space Isotrak) in full flexion and in erect standing. The percent ROF was used to indicate the relative length of the erector spinae, which was expected to increase with increasing lumbar flexion. For assessment of maximum voluntary contraction (MVC) of the back extensors, subjects stood with the lumbar spine positioned in 60 % ROF and pulled up on a floor-mounted load cell with a maximal effort for 3 seconds. Forces corresponding to 20-80% MVC (in 10% MVC increments, presented randomly, and with 2 min rest between each) were then held for 4-6s whilst surface EMG signals were recorded from the erector spinae muscles at the levels of the 10th thoracic and 3rd lumbar vertebrae. The experiment was repeated with subjects standing in 30% ROF and 90% ROF (an MVC was not performed at 90% ROF; the value obtained at 60% ROF was used to set the submaximal forces for this position). The raw EMG signal was band pass filtered (5-300 Hz), amplified, and A to D converted at a sampling rate of 1024 Hz. The EMG power spectral density was computed for 1 s sampling periods using a Fast Fourier Transform algorithm. The median frequency (MF) was defined as that which divided the spectrum into two regions containing equal power. The intercept of the regression of MF on time was used as the representative value for each level of force; where no fatigue occurred, the slope would be close to zero and thus equivalent to the mean value over the 4-6 s period. Repeated measures ANOVA was used to examine the influence of muscle length (given by %ROF) and load on MF for each region of the erector spinae.

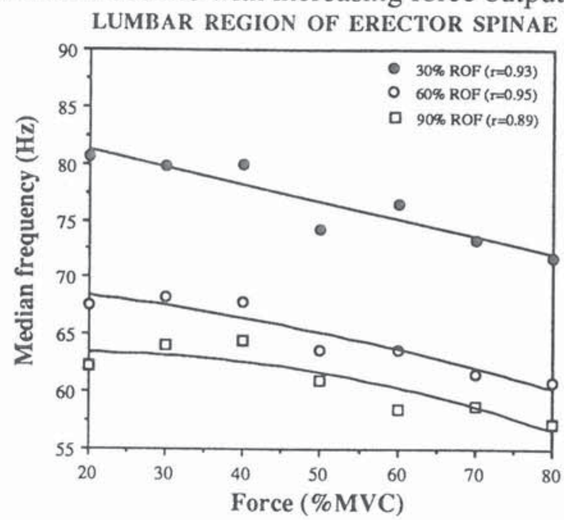
RESULTS

Analysis of variance revealed a significant effect of muscle length on median frequency in both thoracic and lumbar regions of the erector spinae ($P=0.003$ and $P=0.004$, respectively). In both regions, MF was significantly lower in the more flexed postures of 60% and 90% ROF than at 30% ROF ($P<0.05$). In each region, MF was lower at 90% ROF than at 60% ROF, but the difference failed to reach significance.

A significant effect of load on MF was evident, in both thoracic ($P=0.015$) and lumbar ($P=0.009$) regions of the erector spinae. No significant interaction between load and muscle length was observed, at either recording level, indicating that the shape of the relationship between force and load remained fairly constant regardless of muscle length.

The relationship between MF and force differed between the two EMG recording sites (see regression analyses below). In the thoracic region, MF steadily increased with force up to 50-60% MVC and then tended to level off or slightly decline. In the lumbar region, MF was relatively stable up to 30-40% MVC, and then showed a consistent decline with increasing force output.





DISCUSSION

The reduction in EMG median frequency with increasing erector spinae muscle length accords with observations on other skeletal muscles. If the average conduction velocity (and hence EMG median frequency) of a muscle depends on the effective cross-sectional area of its constituent fibres, then the decrease in MF in lengthened muscle may be the result of a reduced diameter of the stretched muscle fibres.

The initial increase in MF with force at the thoracic recording site most likely reflects the gradual recruitment of motor units containing progressively larger muscle fibres. The levelling off of MF could indicate either that recruitment of all motor units is complete by 50-60% MVC or that the muscle fibre size of the later recruited motor units does not change dramatically. To our knowledge, only one study has investigated the fibre type areas in this region of the erector spinae, and this reports that the Type I and Type II fibres are of comparable size (Sirca and Kostevc, 1985). No data on the size of the sub-types of the type II fibre are currently available.

Compared with most skeletal muscles the steady decline in MF with increasing force, displayed in the lumbar region of the erector spinae, is quite atypical. It suggests that in this region of the muscle group there is a progressive reduction in fibre size from the early recruited type I fibres through to the later recruited Type IIb fibres. This agrees with the reports of a smaller type II than type I muscle fibre cross-sectional area in the lumbar region (e.g Sirca and Kostevc, 1985).

With respect to motor unit recruitment, if the median frequency does indeed reflect the average muscle fibre conduction velocity, then these results challenge the notion that the conduction velocity can be considered a "size principle" parameter (Andreassen and Arendt-Nielsen, 1987).

REFERENCES

Andreassen S and Arendt-Nielsen L (1987) J Physiol 391: 561-571.
Sirca A and Kostevc V (1985) J Anat 141: 131-137.
Stulen F and De Luca CJ (1981). IEEE Trans Biomed Eng 28: 512-523.
Sypert GW and Munson JB (1981) Neurosurgery 8:608-621.

AUTHOR'S ADDRESS

Anne F. Mannion, PhD, Comparative Orthopaedic Research Unit, Department of Anatomy
University of Bristol, Southwell St, Bristol BS2 8EJ, UK

The support of the Medical Research Council of Great Britain is gratefully acknowledged

NONINVASIVE IDENTIFICATION AND CHARACTERIZATION OF SINGLE MOTOR UNITS BY EMG: PART I: PRINCIPLE AND BASIC PHYSIOLOGICAL APPLICATIONS

G. Rau, C. Disselhorst-Klug, J. Silny
Helmholtz-Institute for Biomedical Engineering,
Aachen, University of Technology,
Pauwelsstr.20, 52074 Aachen, Germany

SUMMARY

In contrast to the standard noninvasive EMG-techniques the high spatial resolution EMG-procedure (HSR-EMG) allows the detection of the single motor unit (MU) activity in a noninvasive way. The higher spatial resolution of this EMG-procedure is reached by using a multi-electrode array in combination with a spatial filter processing. In this way the HSR-EMG methodology allows the noninvasive measurement of the muscle conduction velocity (CV) in single MU and the localization of the endplate region.

INTRODUCTION

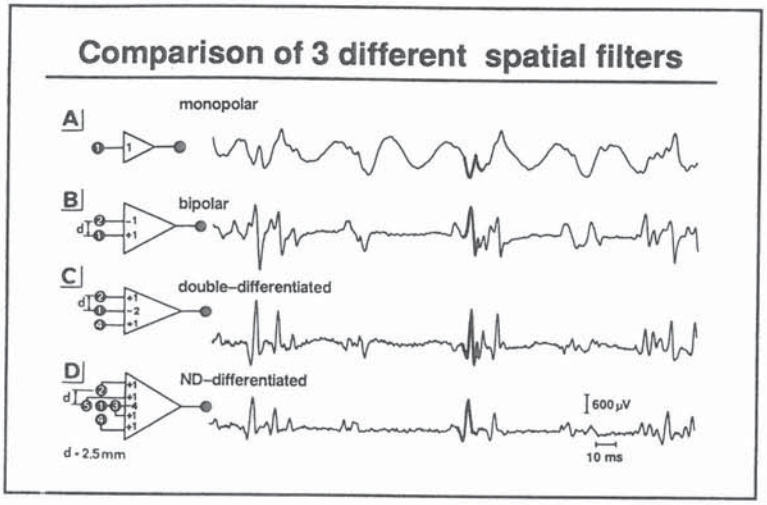
The standard surface-EMG detects, due to its low spatial resolution, only the integrated activity of a high number of MUs but does not yield any information about the single MU activity. The needle-EMG commonly used for investigations of single MUs is invasive and painful. We have developed a new EMG-procedure with a high spatial resolution which allows the acquisition of the single MU activity in a noninvasive way.

METHODS

By the excitation of all active MU a variable potential distribution is generated on the skin surface. MUs which are located close to skin surface contribute to the resulting potential distribution with a spatial steep potential gradient. This is in contrast to MU located more distantly which yield a flat potential course. A bipolar lead with an interelectrode distance in the mm-range, the simplest form of a spatial filter, performs the first spatial derivative of the potential distribution. In this way, the slope of the spatially steep part of the potential distribution contributes more to the resulting difference signal than the spatially flat potential part of distantly located MUs. However the spatial filtering of a bipolar lead is not sufficient to distinguish the activity of single MUs in the signal course (Fig.1).

A weighted summation of three EMG-leads arranged in a consecutive row performs the second spatial derivative of the potential distribution. The spatial selectivity of this double differentiating filter is compared to the bipolar lead essentially improved (Fig.1). The highest spatial selectivity has the Normal-Double-Differentiating filter (NDD-Filter), which consists of five electrodes arranged crosswise. The NDD-Filter amplifies only the activities of those MUs located directly under the center of the filter and reduces the signals of more distantly located sources. In this way the activity of single MUs becomes clearly distinguishable the signal course (Fig.1).

Fig. 1: Comparison between different spatial filter. The excitation of one MU has been emphasized.



A series connection of more than one NDD-Filter allows the detection of the excitation spread along the muscle fibers of single MUs. Therefore, multi-electrode arrays are used for HSR-EMG recordings which allow the calculation of several NDD-filtered channels. They consist of up to 32 gold covered pin electrodes (diameter 0.5 mm) in a two dimensional arrangement. The interelectrode distances are adapted to the muscle size and depth location and can vary between 2.5 mm and 5 mm.

APPLICATIONS

The HSR-EMG allows the noninvasive determination of the single MU activities as well as a statement about the excitation spread. From the time delay between the responses in two defined NDD-filtered channels and their spatial distance the CV in single MU can be calculated. Using the HSR-EMG methodology the effects of the muscle temperature, of the muscle force level and of inhomogeneities in the volume conductor on the noninvasive measured CV have been investigated and will be discussed. The results show that especially the muscle temperature as well as the presence of tissue inhomogeneities have to be taken into account when measuring the CV noninvasively. A change in the signal shape as a change in the conduction direction characterizes the location of the endplate region. Thus, the HSR-EMG can be also used to localize the site of the endplate region with an accuracy lower than the interelectrode distance of the electrode array.

CONCLUSIONS

The HSR-EMG seems to be well suited for a noninvasive detection of the activity of single MUs in superficial muscles. It allows the calculation of the CV in single MUs as well as a localization of the endplate region even at maximal voluntary contraction of the muscle.

REFERENCES

Reucher, H.; Rau, G.; Silny, J. (1987) IEEE Trans. on Biomed. Eng. BME-34: 98-113
Schneider, J.; Silny, J.; Rau, G. (1991) IEEE Trans. on Biomed. Eng. 38:9: 851-860
Schneider, J.; Rau, G.; Silny, J. (1989) Electromyogr. clin Neurophysiol. 29: 273-280

NONINVASIVE IDENTIFICATION AND CHARACTERIZATION OF SINGLE MOTOR UNITS BY EMG: PART II: FIRST CLINICAL APPLICATIONS IN DIAGNOSIS OF NEUROMUSCULAR DISORDERS

C. Disselhorst-Klug, G. Rau, J. Silny
Helmholtz-Institute for Biomedical Engineering,
Aachen, University of Technology,
Pauwelsstr.20, 52074 Aachen, Germany

SUMMARY

The noninvasive EMG procedure with high spatial resolution (HSR-EMG) was validated in children and young patients with different neuromuscular disorders. The HSR-EMG signal shows for each disorder typical pattern. These pattern can be quantified by different physiological- and signal-parameters. In this way, the specifity and the sensitifity of this noninvasive diagnostic technique can be determined.

INTRODUCTION

The diagnosis of neuromuscular disorders is essentially based on the analysis of the single MU activities. The conventional surface EMG is, due to its insufficient spatial resolution, not suitable for this diagnostic purpose. That is why regardless to it's painfulness the invasive needle-EMG methodology is used in clinical practice. However, a new EMG-methodology with a high spatial resolution (HSR-EMG) allows the detection of the single MU activity in a noninvasive way.

METHODS

The HSR-EMG procedure is based on the use of a multi-electrode array in combination with a spatial filter processing. The multi-electrode array consists of 32 gold covered pin-electrodes (diameter 0.5 mm) in a two dimensional arrangement (interelectrode distance 2.5 mm). The used spatial filter is build by a weighted summation of five adjacent EMG-leads which are arranged crosswise. This filter amplifies the EMG-signals of MUs located directly beneath the center electrode of the filter and reduces the signals of more distantly located sources. A series set-up of such spatial filter allows the determination of the excitation spread. In this way the determination of the conduction velocity (CV) in single MUs and the localization of the endplate region becomes possible.

The HSR-EMG has been recorded in 62 volunteers aged between the infancy and 24 years during maximal voluntary contraction of the m. abductor pollicis brevis. Additionally 21 patients with neuronal disorders (mainly spinal muscle atrophy) and 35 patients with muscular disorders (mainly Duchenne muscle dystrophy) in the same range of age were investigated with the HSR-EMG.

RESULTS

The results show a distinct difference between the HSR-EMG patterns from volunteers, patients with muscular and such with neuronal disorders (Fig.1). The HSR-EMG pattern of volunteers contains equally distributed peak amplitudes (Fig.1). This characterizes a high activity of a large number of MUs. The typical HSR-EMG pattern of patients with muscular disorders shows low and not clearly distinguishable peaks in the signal course; the HSR-EMG pattern of patients with neuronal disorders only high and isolated peaks with no activity between them.

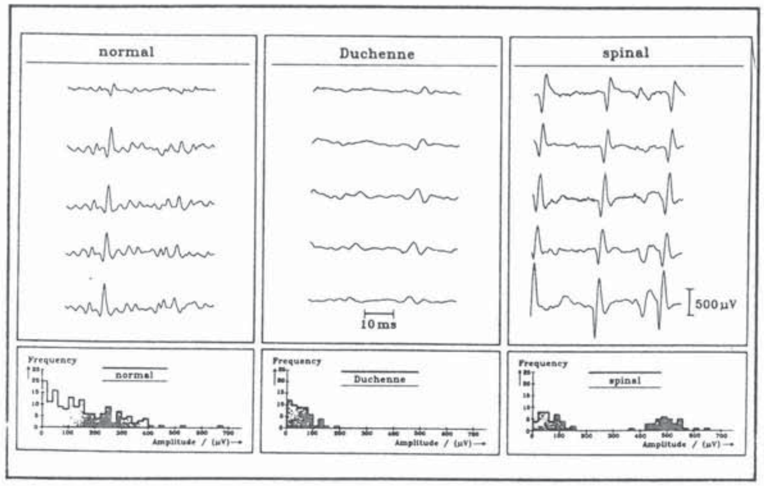


Fig.1: HSR-EMG recordings in patients and volunteers. Represented are five spatial filtered channels and the peak-amplitude frequency distribution of one channel.

This typical differences in the HSR-EMG pattern are evaluated by different parameters regarding the excitation spread, the duration of signal segments, the shape of isolated peaks and the peak-amplitude frequency distribution (Fig.1). Parameters like the CV in single motor units, the auto correlation function of the signal, the maximal peak amplitude, the gradient of the peak edge and the peak-amplitude frequency distribution separate clearly patients with muscular disorders. The dwell time over the RMS, the peak-energy to gap-energy ratio and the peak-amplitude frequency distribution distinguish patients with neuronal disorders. Using this parameter for a quantification of the HSR-EMG pattern the diagnostic sensitivity of the HSR-EMG methodology in children with different neuromuscular disorders has been determined to 82% and the specifity to 96%.

CONCLUSIONS

The noninvasive HSR-EMG detects for each disorder typical pattern in the electrical activity of the muscle. Using the HSR-EMG technique a quantitative distinction between healthy volunteers, patients with muscular disorders and patients with neuronal disorders becomes noninvasively possible. Even the type of the neuromuscular disorder can be detected.

REFERENCES

Reucher, H.; Rau, G.; Silny, J.; (1987) IEEE Trans. on Biomed. Eng. BME-34: 98-113

Ramaekers, V.; Disselhorst-Klug, C.;Schneider, J.; Silny, J.;Forst, J.; Forst, R.; Kottlerek, F,; Rau, G. (1993) Neuropaediatrics 24: 134-138

COMPUTER-CONTROLLED INSTRUMENTATION FOR DYNAMIC MEASUREMENTS OF SPASTICITY IN THE LOWER LIMBS OF MS-PATIENTS

L. De Wolf[1], L. Janssens[1], G. Nuyens[2,3], W. De Weerdt[2], A. Spaepen[2], P. Ketelaer[3], H. Carton[4]

[1]GROEP T Institute of Engineering, Leuven, BELGIUM

[2]Catholic University of Leuven, Faculty of Physical Education and Physiotherapy, Leuven, BELGIUM

[3]National Center for Multiple Sclerosis, Melsbroek, BELGIUM

[4]Catholic University of Leuven, Department of Neurology, Leuven, BELGIUM

This project is supported by the Nationaal Fonds voor Wetenschappelijk Onderzoek (Levenslijn Project) and by the National MS-Society of Luxembourg

INTRODUCTION

Various diseases of the central nervous system are at the origin of movement disorders. They are associated with changes in passive and active muscle properties. Many multiple sclerosis patients suffer from spasticity in the lower limbs. This causes problems in gait and transfers and impairs the functional activity in daily life. Spasticity is characterized by exaggerated tendon reflexes and muscle hypertonia. Quantification of spasticity has been recognized to be important but seems to be a difficult problem. Spasticity has many aspects and can be influenced by several factors, such as body posture, pain, medication, patient's emotional state, fatigue. It is manifested in different ways in rest and in voluntary movements. In the clinical evaluation of spasticity the therapist measures the velocity-dependent resistance of a muscle group against a passive movement and classifies it on a 5-point scale (Ashworth scale). This clinical evaluation has some disadvantages: the muscles are tested at rest and the accuracy is limited. Other assessment methods, such as neurophysiological and biomechanical tests have been proposed but no universal measurement tool has been accepted yet.

EXPERIMENTAL METHOD

In order to study spasticity in multiple sclerosis patients during active and passive movements we have designed two computer-controlled instruments. As spasticity is a velocity-dependent phenomen, the movement can be imposed at different speeds. The movements are velocity-controlled at a constant rate. However, the feedback system decreases the velocity when the resistance becomes too large.

An experimental set-up has been built to perform analytical tests of the ankle and knee joint. A rotation at constant angular velocity around the joint is imposed. A sequence of angular velocities between 20 and 180 degrees per second during flexion and extension for knee and ankle movements can be chosen by the therapist. During the movement, the angular position and the torque are continuously measured and displayed in real time on the screen. EMG-activity is measured by surface electrodes from different muscle groups: M. quadriceps, M. hamstrings, M. tibialis anterior, M. triceps surae. An isolated 8-channel EMG-amplifier developed at our institute with active electrodes will be used. The instrument consists of a direct drive motor mounted on a tube frame. The motor shaft is connected to a rotating arm to which the foot or the lower limb is attached. An adapted patient table allows tests to be performed with the trunk in horizontal and vertical position in order to investigate the influence of body posture on spasticity. Measurements are recorded on hard disk for later processing and interpretation. A safety button allows manual interruption in any position of the motor. Before each test, the therapist sets the maximum amplitude of the movement by positioning the motor arm to one end-point and moves it to the other limit. During the test, movements beyond those limits are not possible. When the torque, which is a measure of the resistance against the movement, exceeds a certain level, the velocity is reduced. In addition, a safety coupling mounted on the shaft uncouples the motor when the torque becomes too large.

A second instrument is designed to investigate spasticity during global flexion-extension movements of the lower limbs. Patients are sitting on a saddle and perform a cycling movement. As in the first instrument, a direct drive motor is connected to the shaft which moves with an angular velocity, controlled by the computer. The pedals of this set-up are specially designed with integrated load cells. This allows to measure the pedal force in two perpendicular directions. This force results from muscle action in the ankle, the knee and the hip. The positions of the crank and the pedals are continously measured. The instrument allows cycling at angular velocities up to 360 degrees per second. The EMG-activity will be measured in the same way as in the first set-up. The measurements will be performed during 5 revolutions in active and passive movement. An emergency stop button allows interruption at any moment by the operator. The software for the two instruments was developed on a 486 compatible PC with DYNASERV motor control unit and MICROSTAR data-acquisition card.

RESULTS

The first experimental set-up has been completely constructed and subjected to initial tests. Some problems with vibrations of the frame could mainly be resolved by choosing the right parameters for the motor control servo system. Tests on healthy persons have been performed to determine the experimental conditions for the test battery. A number of measurements on healthy persons for reference will be performed. After that the instrument will be used for measurements on multiple sclerosis patients.

The second set-up will be operational in a few months. The pedal structure with the force transducers has already been constructed.

CONCLUSION

Two experimental set-ups have been designed to investigate spasticity in an objective and quantitative way. The instruments allow us to assess spasticity in analytical and functional movements, and to investigate the influence of external factors, such as body posture, pain, medication, emotional state, fatigue. The instruments are velocity-controlled and measure force, torque and electromyographic signals. They are provided with different safety precautions and will be used to collect data from healthy persons and from multiple sclerosis patients with spasticity in the lower limbs. The measurement results will be compared with those of other tests (gait analysis). They will also be used for the evaluation of therapeutic methods.

Author's Address: Luc De Wolf, dr.ir.
GROEP T Institute of Engineering
Vuurkruisenlaan 4, B3000 Leuven, BELGIUM
Telephone: 32 16 23 08 50 Fax: 32 16 22 83 43

A COMPARATIVE STUDY OF PHYSIOLOGICAL TREMOR IN NORMAL AND RHEUMATOID ARTHRITIC SUBJECTS.

D.M. Halliday, B.A. Conway[1], W.L. Ng[2], J.R. Rosenberg
Institute of Physiology, University of Glasgow , [1]Bioengineering Unit, University of Strathclyde, Glasgow, and [2]Dept. Medicine, Royal Infirmary, Glasgow, Scotland, U.K.

INTRODUCTION

Rheumatoid Arthritis (RA) results in joint deformity, stiffness, pain and muscle wasting. In this report we present some preliminary results which examine the influence of RA on physiological tremor. Physiological tremor results from the interactions between the neural drive to muscles and the mechanical properties of musculoskeletal system. Any pathology which may influence a contributing factor, such as a joint disease, will therefore alter the resultant tremor. Time and frequency domain measures of association were used in this study to assess the dependency of finger tremor on single motor-unit discharges and the surface EMG.

METHODS

Tremor measurements were made in 24 normal and 8 RA (4 stage 2 and 4 stage 3) subjects. Patients with excessive hand deformity and whose tremor may have originated from neurological disorders were excluded. The tremor signal was derived from an accelerometer (Entran EGAX-5) which was attached to the underside of the distal phalanx of the unsupported middle finger. The subject's other fingers, wrist and forearm were supported by a rigid polypropylene cast. Single motor unit recordings were derived from concentric needle electrodes inserted into the extensor digitorum communis muscle (EDC). Two such electrodes were used to simultaneously record activity in pairs of motor units. Surface EMG activity was recorded by a pair of electrodes placed over the muscle. During data collection the subject was asked to extend and maintain the middle finger in an approximately horizontal position. Recordings (1-3min) were made with the finger unloaded and with added masses up to 25g, in 5g increments. Tremor and surface EMG signals were sampled at 1 kHz while single motor-unit data was stored as spike times with a 1ms resolution. For analysis, the tremor and surface EMG signals were assumed to be realisations of stationary zero-mean time series, and the motor unit firing times were assumed to be realisations of stochastic point processes. The frequency domain measure of association used was the coherence function. This function can be defined and estimated for both time series data and hybrid data (where one signal is a point process and the other a time series). The time domain measure of association, called the 2nd order cumulant, was estimated as the inverse Fourier transform of the cross-spectrum between the same two processes used in the estimation of the coherence. For time series cumulants can be interpreted as impulse response functions, whereas for hybrid data they have a form similar to spike triggered averages.

RESULTS

In figure 1 the hybrid cumulant density estimates between single motor unit discharges and finger tremor for a normal subject (Fig. 1a) and a RA subject (Fig. 1b) are shown. In both cases the middle finger was unloaded. Clear differences exist between these estimates. Firstly the strength of coupling between the single motor unit and the finger tremor is greatly reduced in the RA patient. This is more rigorously demonstrated by the corresponding coherence estimates (not shown) where direct comparison is possible using a statistical test (ref: 1). Secondly, for the normal subject the cumulant density exhibits oscillations after the initial acceleration peak following the motor unit discharge. These oscillations reflect a resonance of the finger which is induced by each motor-unit event. This ringing behaviour, is modified by adding mass to the finger. In normal subjects additional mass reduces the resonant frequency. However, in all RA subjects no equivalent behaviour can be seen illustrating a major difference exists in the behaviour of the musculoskeletal system in RA and normal subjects performing this simple task. The results suggest that coupling between motor-units and finger acceleration is reduced in RA. It is likely that this reflects joint pathology. Similar observations can be seen when time and frequency domain estimates are made between surface EMG and finger tremor records.

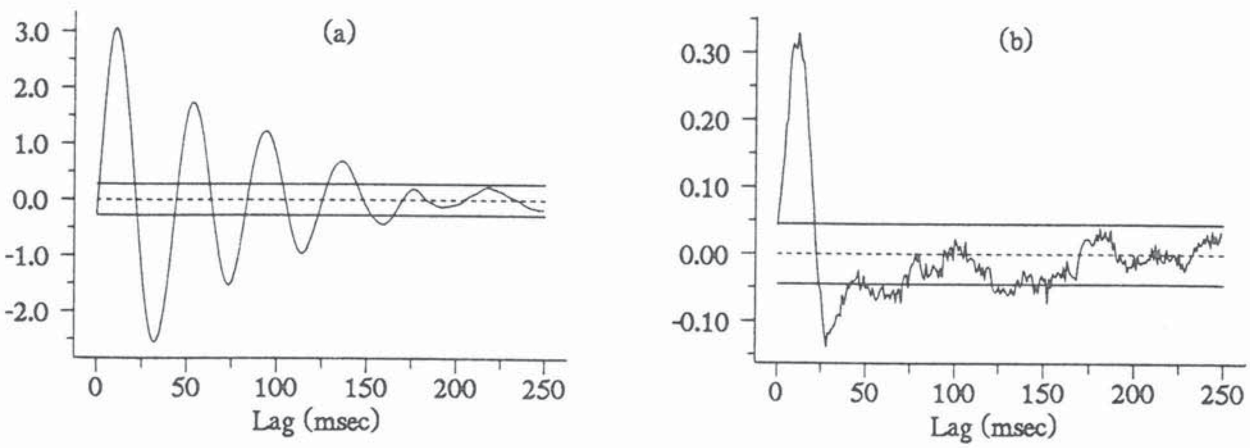


Figure 1. Cumulant density estimate between single motor unit spike and finger acceleration (a) normal subject (b) rheumatoid arthritic subject.

CONCLUSIONS

In this study we have demonstrated that differences exist in the correlations between EMG (single-unit and surface) and finger accelerations measured in normal and RA subjects. The resonant behaviour of the finger to a mechanical perturbation resulting from a motor-unit discharge appears highly damped in RA subjects suggesting that joint stiffness has increased. This becomes apparent as a reduced physiological tremor and appears to suggest that the disease acts to uncouple muscle and finger. Further studies will determine if this approach to the study of joint behaviour has any diagnostic potential.

REFERENCES

1. Rosenberg et al. Prog. Biophys. molec. Biol. (1989). 53. 1-31

This work was supported by The Wellcome Trust

AUTHOR'S ADDRESS

Dr. D.M. Halliday
Institute of Physiology
University of Glasgow
Glasgow, Scotland
Tel: 41 339 8855, Fax 41 330 4100

ADVANCES IN HYBRID SYSTEMS FOR PARAPLEGIC WALKING

Hermie J. Hermens, Gert Baardman, Henry M. Franken*, Peter H. Veltink*, Herman B.K. Boom*, Gerrit Zilvold
Roessingh Research and Development, *University of Twente Enschede, The Netherlands

INTRODUCTION

In Twente, The Netherlands, restoration of locomotion for paraplegic patients is adressed in an extensive joint research program. The main theme of this program is the development of the so-called MOSES system (Modular Orthosis with multichannel Surface Electrical Stimulation).

Clinical evaluation shows that hybrid systems currently prescribed in our rehabilitation centre, being 4-channel open loop surface electrical stimulation combined with a reciprocating gait orthosis[6] fail in terms of functionality.[1] Generally the result of this is that the hybrid system is used at home for therapeutic walking only.

We believe that two major shortcomings are responsible for the low functional value of current hybrid systems. Firstly, user friendliness is poor: the system is cumbersome and hinders the patient during almost all ADL functions. Taking this into consideration, the effort required to don and doff the system outweighs by far the functional merits. Secondly, the high energy consumption required during gait (though perhaps favourable in terms of therapeutic use) contradicts extended functional use.

Since none of the currently available orthoses was designed as such, it can not be expected that their properties are attuned to combined use with (surface) electrical stimulation. It is our opinion that integration of mechanical and electrical designs is essential for the eventual functional performance of the system. Furthermore, we believe that both user friendliness and system performance will be enhanced by incorporation of (automated) closed loop control of electrical stimulation. This paper describes our research activities directed towards the development of a new truly hybrid system, based on the above considerations, that should offer more functionality to the user in the near future.

ORTHOSIS

Current HKAFO's have two major drawbacks with respect to functional hybrid application: the rigidity during wheelchair use and transfers (thus contradicting user friendliness), and the fact that the knee joints are locked during the swing phase of gait. The latter property reflects on both energy consumption and cosmesis. A removable trunk brace is currently being developed to overcome the first drawback, and the second is encountered with a knee mechanism that provides automatic unlocking of the knee joint during the swing phase of gait. During the stance phase the knee is locked in extension. Actuation of the mechanism, conditional on a knee extension moment, is provided from the hip. A first prototype, developed for external mounting to the ARGO, was evaluated with respect to gait kinematics and energy consumption. The results of the study indicated that the energy consumption may indeed be reduced, especially if the mechanism is incorporated in the orthosis. The knee mechanism has to be combined with electrical stimulation of knee extensor muscles in order to ascertain knee extension at the start of the stance phase.[5]

STIMULATION METHODS

Comparative studies have been carried out towards the controllability and propulsive properties of different hip muscle stimulation methods. The results of these studies show that control performance of efferent stimulation methods is better than that of afferent methods (i.c. flexion withdrawal response).[2] Furthermore, propulsion provided by stimulation of hip extensor muscles during the stance phase was shown to outweigh that of the flexion withdrawal response evoked during the swing phase.[3]

CONTROL

Current control schemes for FES assisted locomotion are based on a hierarchical decomposition of the control into 3 levels. In the conscious level the subject controls the initiation of the system. The high level of control interfaces the subject and it coordinates the desired motor task by supervising a set of low level controllers. The low level controllers actually realize the muscle actions by manipulating the stimulation parameters. A number of projects are carried out on these aspects.

Finite state representation of paraplegic gait

As a first step towards the automated control of walking, a finite state representation was developed for the control of the electrical stimulation patterns during gait. Based on sensor information reflecting hip angles and crutch ground reaction forces, an algorithm was defined that splits the gait pattern into 9 different states. Depending on the state (transfer), low level controllers are activated.[4] The algorithm was checked using video registrations and tested succesfully in three patients. Recently it has been implemented in hardware. The crutch force sensor consists of a mechanical switch which is preloaded by a spring. It is currently tested in a more clinical setting.

Low level control

One of the major problems in the application of electrical stimulation to muscles is fatigue. In order to overcome this problem in hip flexion a control algorithm has been developed that compensates for a fatigue-induced decrease in hip flexion angle by adjusting the stimulation for a next cycle.[5] Each low level controller contains a discrete-time PID controller to adjust the duration of the hip flexors from cycle to cycle. It has been shown in several trials that the controllers succesfully achieved symmetry in gait by accounting for the different mechanical output of the stimulated muscles. The controllers also incremented the duration of the stimulation to minimize the effect of the fatigue induced decrease of muscle force.

REFERENCES

1. Baardman G, Lissone AGM, Hermens HJ (1990): Evaluation of the functionality of walking systems. Proc. Xth ECHE, Belgrade, pp. 381-386.
2. Baardman G, van Toor MW, Veltink PH, Franken HM (1993): Sensitivity analysis of hip muscle stimulation methods with respect to leg cyclical swing response. Proc. Ljubljana FES Conference, pp. 76-78.
3. Baardman G, Vastenholt JM, Hermens HJ (1992): Assessment of propulsive properties of hip muscle stimulation methods in hybrid systems. Proc. 4th Vienna Int. Workshop on FES, pp. 69-72.
4. Franken HM, Vries W, Veltink PH, Baardman G, Boom HBK (1993): State detection during paraplegic gait as a part of a finite state controller, Proc. IEEE 15th ann. Conf. Eng. Med. Biol., pp. 1322-1323.
5. Franken HM, Veltink PH, Baardman G, Redmeijer H, Boom HBK (accepted): Cycle-to-cycle control of the swing phase of paraplegic gait induced by surface electrical stimulation, Med. & Biol. Eng. & Comp.
6. Zilvold G, Hermens HJ (1992): Goal of FES and clinical aspects of patient management, Resoration of walking for paraplegics, ISBN 90-5199-094-4 IOS press, 355-358.

DIFFERENT THERAPEUTIC ELECTRICAL APPROACHES TO DIFFERENT KINDS OF SPINAL CORD LESIONS (SCL) SPASTICITY

R. Prati, S. Visconti and V. Alfieri
Rehabilitation Center, Lonato, Brescia, Italy

ABSTRACT

Inasmuch as spasticity shows different features, 15 spinal cord injured patients were studied with the aim to find which approaches with electrical stimulation are effective. Three kinds of spasticity are described, one prevalently clonic, one prevalently tonic extensor, one characterized by severe intensiti of spasms, prevalently tonic flexor at the lower limbs. The first two kinds are sensitive to electrical stimulation, provided different approaches are used; the third kind isn't sensitive to electrotherapy.

INTRODUCTION

An antispastic effect of therapeutic electrical stimulation (TES) is reported by many Authors. The SCL of traumatic and viral origin can show different clinical features of spasticity. Kawamura et al. (1), studing 13 quadriplegic patients, found 3 types of spasms in upper and lower limbs and 2 types in the trunk; they classified also 4 patterns of spasms in the limbs. It seems reasonable to suppose that some features of spasticity are connected with different physiopathological mechanisms. This assumption seems confirmed, from a clinical point of view, by the different effects of electrical stimulation on some different kinds of spinal cord spasticity, as referred by Alfieri and Marchetti (2). The aim of this presentation is to differentiate some clinical kinds of spinal cord spastdicity on the basis of the therapeutic effectiveness of electrical stimulation.

MATERIAL AND METHOD

We considered 15 SCL patients treated in the last few years with TES. The levels of lesions are: 6 cervical, 8 thoracic, 1 lumbar. Three patients were not affected with spasticity. Twelve were spastic: 9 with clonic contractions at lower limbs, prevalently in extension, 4 of these with low intensity (1 cervical, 3 thoracic), 5 with medium intensity (2 cervical, 3 thoracic); 1 cervical had tonic extensor spasticity of high intensity in his lower limbs and trunk, 2 patients, both cervical, had tonic intense flexor spasms at lower limbs and extensor in the trunk. All of these patients were treated also with TES for preparation to orthosis (RGO) or for strengthening muscles or for reducing spasticity or for other beneficial side-effects of TES. The parameters used were: trains of impulses of 600 usec at 30 Hz, lasting 5 sec with a duty cycle 1/4 for 20 min per muscle group, 3 times a day, for months, at home after training. The muscle groups stimulated were mainly quadriceps, hamstrings and extensors and eversors of foot; gluteal, abdominal and lumbar muscles when needed.

RESULTS

Out of the 12 spastic patients, 2, affected with high intensity tonic spasticity mainly in flexion had no improvement of spasticity by TES. The other 10 patients had an evident, considerable reduction of spasticity (fig.s 1 and 2).

DISCUSSION

We have identified 3 forms of spinal cord spasticity in relation with electrical stimulation. The first form is the most common and shows rapid clonic contractions at lower limbs, mostly in extension usually of low or medium intensity (9 pat.); this form is the most sensitive to TES. The second form is located prevalently in the antigravitational musculature as a tonic and persistent contraction more or less severe; this form can benefit from TES only if we avoid stimulation to the quadriceps and

and triceps surae muscles, but apply TES on their antagonists (1 patient). The third form of spasticity is characterized by a very strong, uncontrollable, untamable tonic spasm of trunk and legs, involving agonist and antagonist muscles; the two patients we observed in these conditions during this trial did not get any benefit from TES, as well as from any drug.

We conclude that the patients included in the first category above described can be treated with TES without worry, especially for preparation to standing up and walking; the same is for the patients included in the second category, but with the caution to stimulate the antagonist muscles of the tonically spastic-ones only.

References on request .

AUTHOR'S ADDRESS:
Dr. Vittorio Alfieri, M.D.,
Via Robecco 30, tel. ****.2.66049448,
20092 Cinisello B., Milano, Italy.

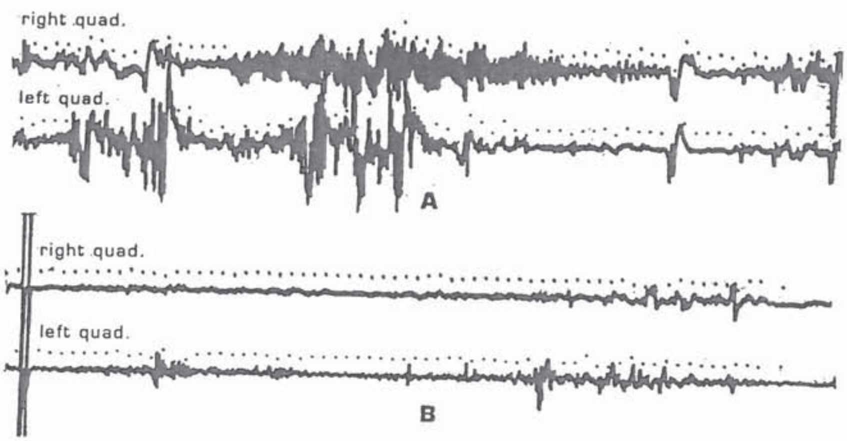


Fig. 1.- Quadriceps muscle's reflex activity in response to turning posture: A, before TES; B, 5 min after 20 min of TES.

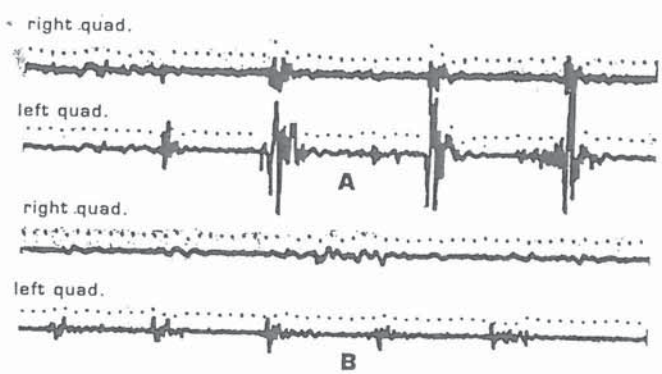


Fig. 2.- Quadriceps muscle's reflex response to cough: A, before TES; B, 5 min after 20 min of TES.

COMPENSATION OF LOSS OF POSTURAL MUSCLE ACTIVITY IN SPINAL CORD INJURED PEOPLE.

H.A.M. Seelen, Y.J.M. Potten
Institute for Rehabilitation Research, Dept. of Posture and Movement Research, Hoensbroek, the Netherlands.

INTRODUCTION

In people with spinal cord injury (SCI) at thoracic level postural balance during sitting is impaired. This can most prominently be seen in dynamic conditions i.e. during manual task execution. During rehabilitation SCI subjects are trained to use alternative strategies to partially compensate the loss of postural muscle activity including increased use of "non-postural" muscles which have both a favorable anatomical/biomechanical position relative to the trunk, spine and shoulder girdle and an innervation level cranial to the spinal cord lesion. The reorganization of postural control in seated SCI subjects takes place at both central information processing level and peripheral level resulting in different muscle activation patterns in SCI persons. One aim of the present study is to investigate which muscles are used by seated SCI subjects to compensate the loss of postural muscle activity of e.g. the erector spinae muscle (ES).

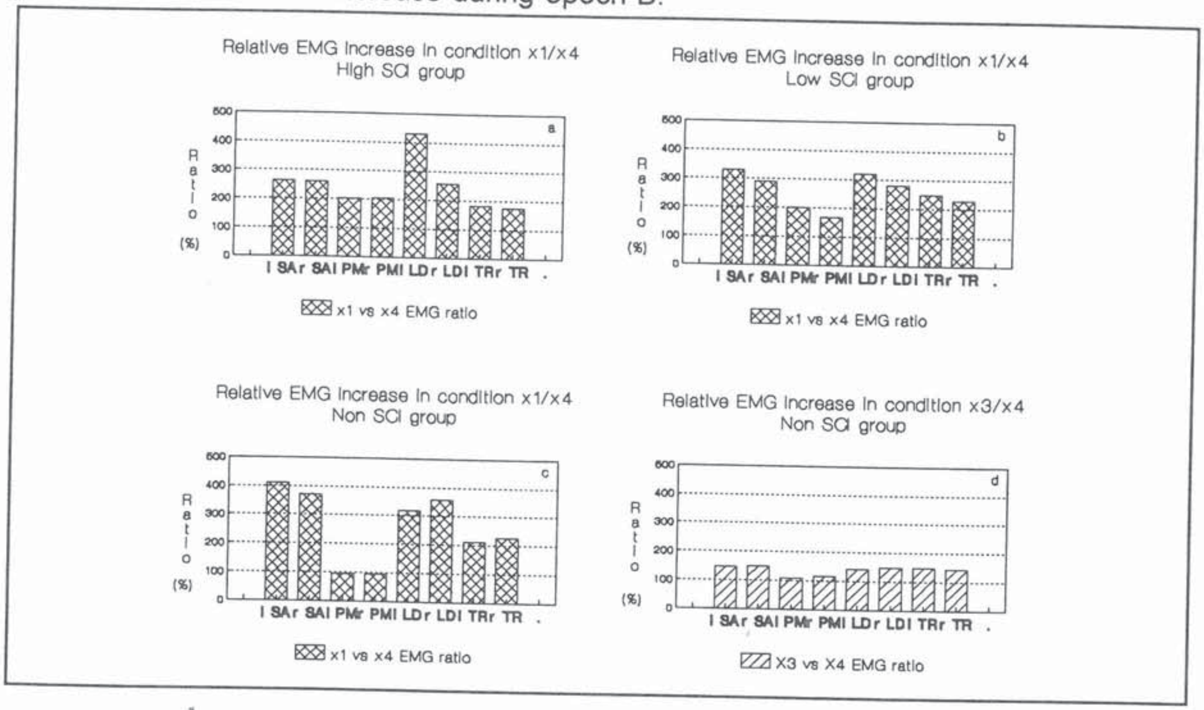
METHODS

15 SCI subjects with a complete lesion at level T2-T9 (H-group), 15 SCI subjects with a complete lesion at level T10-T12 (L-group) and 15 non-SCI subjects (N-group) participated in the experiment. Groups were matched for age, sex, dexterity, height, weight and educational level. All SCI subjects had completed their rehabilitation at least one year. Subjects were seated in a multi-adaptable chair placed on a force platform (Biovec-1000, AMTI, Newton, Mass.) behind a table. Sitting balance was perturbed by using a task in which bimanual forward reaching movements were to be made towards buttons placed at 90%, 75%, 30% and 15% of the individual maximum reaching distance in the medio-sagittal plane (position X1, X2, X3 and X4). The task was presented as a visual choice reaction time task in which the imperative stimulus indicated the reaching distance (i.e. X1, X2, X3 or X4) to be covered. Every trial was started by operating a centrally placed button proximal to the subject. Reaction time (RT), movement time (MT) and reaching distance were recorded. Simultaneously to task execution bilateral muscle activity of the erector spinae (ES) at level L3, T9 and T3, the trapezius pars ascendens (TR), the latissimus dorsi (LD), the serratus anterior (SA), the pectoralis major pars sternocostalis (PM) and the external oblique abdominal (OA) was recorded using surface electromyography (K-Lab MF-118/SPA-12, K-Lab, Amsterdam). Sitting balance i.e. changes in the position of the centre of pressure (COP) was computed from the force platform signals recorded synchronously. Button status was recorded on a separate event channel. To record all signals the reaching task series was carried out twice in a counter-balanced design. Per task series subjects had to reach 16 times over each of the 4 different distances in a random order. EMG data analysis included full wave rectification and ensemble averaging after signal synchronization to movement events linked to RT and MT. For each muscle mean EMG activity was calculated over the epoch between the onset of the imperative stimulus and the touching of the peripherally placed buttons (epoch A). Secondly an epoch representing 90% to 100% of relative sitting balance perturbation per trial was identified from the COP recording (epoch B). Within this epoch B mean EMG activity was computed for all trials for all muscles. Mean EMG of epoch for all conditions (X1, X2, X3 and X4) was standardized using the mean EMG of epoch A in condition X1. Differences both in single muscle EMG and in multiple muscle EMG patterns during epoch B were evaluated between all conditions. Statistical analysis included Kruskal-Wallis tests.

RESULTS

Relative increase in mean EMG activity within the three groups at increasing submaximal reaching distances was found to be significant ($p \leq 0.02$) except for both TR in the H-group and both PM in the N-group in all four conditions. No significant differences were found in both TR between condition X1 and X2 in the N-group. Percentual increase in EMG activity in X1 conditions relative to X4 conditions for all groups for muscles whose function was not primarily impaired due to the SCI i.e. the SA, LD, TR and PM muscles are given in figure 1a, 1b and 1c. Significant differences in multiple muscle EMG profiles were found between groups. When comparing conditions in which all participants showed approximately the same absolute COP. changes (i.e. X3 condition for the N-group and X1 for the SCI groups) more prominent differences in EMG activity in the SA, LD, TR, and PM were found between the SCI groups and the non-SCI group, as is depicted in figure 1a, 1b and 1d.

Figure 1. Relative EMG increase during epoch B.



DISCUSSION

In order to stabilize the spine and the shoulder girdle during manual task execution and subsequent sitting balance perturbation alternative use of muscles not primarily affected by the lesion is necessary in SCI. Our data indicate that activity patterns of non-postural muscles change in SCI subjects during standardized functional balance perturbation. Further assessment of these kind of multiple muscle profiles may give insight in the adaptation of SCI subjects to the lack of balance control resulting in new patterns of postural control. More insight in the evolution of these EMG patterns may prove to be valuable in the clinical assessment of balance control in SCI patients during rehabilitation.

AUTHOR'S ADDRESS

Henk A.M. Seelen
Institute for Rehabilitation Research
Department of Posture and Movement Research
Zandbergsweg 111
6432 CC Hoensbroek, the Netherlands

COMPARISON OF VOLUNTARY MOVEMENTS AND LOCOMOTION IN SPASTIC CEREBRAL PALSY

M. Elise Johanson, MS, RPT, Stephen R. Skinner, MD.
Shriners Hospital, Orthopaedic Biomechanics Laboratory, 1701-19th Avenue, San Francisco, California 94122, U.S.A.

INTRODUCTION

Investigations of motor dysfunction in patients with spasticity often focus on the clinical signs of spasticity including increased resistance to passive movement, clonus, and hyper-reflexia. In addition, patients may demonstrate abnormal motor control characterized by a lack of coordination, inadequate postural control, and limited or ineffective movement due to incorrect muscle timing. Patterns of muscle activation during passive or voluntary movements may be different when the same movements are compared during functional activities, such as walking.

The purpose of this study was to compare motions and patterns of activation in antagonist lower extremity muscle groups during passive movements, voluntary movements, and locomotion in children with cerebral palsy. Comparisons of similar movements under different neural control were expected to provide information about the motor control strategies used by the patient.

METHODS

Ten children with spastic cerebral palsy (8 diplegic, 2 hemiplegic) and dynamic equinus during ambulation were recruited from the out-patient clinic. Two girls and eight boys with a mean age of 7.7 (range 3-12) years were included. None of the children had any surgery on the lower extremities within five years of testing. All of the children had passive dorsiflexion to neutral or greater, but walked in equinus.

Each subject was asked to perform ankle dorsi-and plantarflexion and knee flexion and extension in sitting and standing positions. The confusion test (resisted hip flexion in the sitting position) was also included in the testing protocol. Subjects were asked to walk at a free walking speed.

Muscle timing of the quadriceps, hamstrings, pretibial and calf groups was recorded with surface electrodes. Motions at the knee and ankle were recorded with strain gauge electrogoniometers.

RESULTS

The point biserial correlation coefficient was used to measure the association between co-contraction and joint motion. Co-contraction occurred during only 14% of the tests. There was no correlation between co-contraction and the ability to perform any of the movements ($r_{pb} = -.02$). Simultaneous activation of flexor or extensor pattern muscles was observed when the subjects attempted to perform isolated movements in sitting and standing positions.

The maximum degree of motion achieved during passive (PASS) motion and voluntary motions in sitting (SIT), and standing (ST), the confusion test (CONF), at initial contact (IC) and swing (SW) phases of gait were ranked to compare motions in each condition. The mean ranked scores for the dorsiflexion and the knee extension are listed below. The highest scores represent the greatest degree of motion. Standard deviations are in parentheses.

	PASS	SIT	ST	CONF	IC	SW
DORSI	5.5 (1.0)	2.7 (1.3)	2.0 (1.7)	4.2 (1.7)	2.9 (1.2)	3.7 (1.0)
EXTEN	3.8 (0.4)	2.5 (0.7)	2.3 (1.1)	-	1.4 (0.7)	-

DISCUSSION

The active range of motion during the voluntary movements or locomotion was always less than the passive range. This finding could not be explained by contraction of an antagonist muscle that limited movement. The motion data suggest that more dorsiflexion could be achieved during the confusion test and the swing phase of the gait cycle, when the knee and the hip were also in flexion. Knee extension was limited at initial contact, when the hip and ankle were in flexion. Isolating movement to one joint was difficult in the sitting and standing positions.

Co-contraction was not a limiting factor in the performance of movements or during walking for this subject population. Rather, dependence on flexor or extensor synergy patterns and lack of selective control were more important determinants of motion. These findings suggest that reducing the spasticity will not necessarily result in improved motor control and emphasize the importance of assessing voluntary movement in patients with cerebral palsy.

BASIC FEATURES OF GAIT IN PATIENTS AFFECTED BY HIP DYSPLASIA: PRE AND POST OPERATIVE MULTIFACTOR ANALYSIS

C.Frigo* and C. Romanò**

* Centro di Bioingegneria, Fondazione Pro Juventute Don Carlo Gnocchi I.R.C.C.S., Politecnico di Milano, Via Capecelatro, 66, 20148 Milano, Italy

** Centro dell'anca displasica, Istituto Ortopedico G. Pini, Piazza Cardinal Ferrari, 1, 20122 Milano, Italy

INTRODUCTION

Patients affected by congenital hip dysplasia progressively show changes in the basic locomotor features which reflect the progressive joint degeneration as well as the adaptation of the neural control system to the primary alteration. Quantification of gait parameters can be a method for better understanding the role of different mechanisms and to predict the evolution of the functional impairment (Murray et al., 1971).

METHOD

Twenty-one adult subjects (both sexes) affected by congenital hip dysplasia were analysed quantitatively by a gait analysis system. Clinical and radiographic data were considered as well. In particular the Harris Hip Score was adopted as a general indicator of the patient's conditions. Subjects were analysed bilaterally, and the asymptomatic limb always was in the maximum range (Hip Score 98-100) while the affected one was is the range from 96 (slightly symptomatic) to 27.1 (maximum impairment).

Gait analysis was performed by means of the ELITE System (BTS, Milan, Italy). Four TV cameras were located in the gait laboratory S.A.F.Lo. in such a way that the two sides and the back of the patient could be detected within a working volume of 3 x 2 x 1.5 m (Pedotti and Frigo, 1992). Retro-reflective markers (8 mm in diameter) were glued on the skin at the following locations: C7 and T8 spinal processes, Posterior Superior Iliac Spines, bone prominence of the Sacrum, lateral femoral epicondyles, lateral malleoli, head of the fifth metatarsi. The 3-D coordinates of each marker as computed by the ELITE System together with some anthropometric parameters directly measured on the patients and on the X-Ray pictures, allowed an estimation of the location of the following internal points: center of C7 and T8 vertebral bodies, lumbo-sacral joint, hip, knee and ankle joint centers, mid forefoot points. Ground reaction forces and myoelectric signals from eight relevant muscles were analysed as well by means of a dynamometric platform (KISTLER) and a multichannel portable device respectively. By a special purpose program it was possible to process all the kinematical, dynamometrical and electromyographical variables and to compute the joint moments and joint powers in the sagittal and frontal planes. Stick diagrams in different perspective view and time course of all the variables obtained were presented in a way that comparison with a homogeneous control population and comparison between subjects was easily feasible. Spatio-temporal parameters were analysed as well, based on kinematics and dynamometric data. Patients were analysed barefoot during quite standing and during walking at their natural cadence. The more severely affected patients who underwent a surgical intervention of hip joint replacement were re-analysed by the same protocol after six month and one year.

RESULTS AND DISCUSSION

A number of typical features appeared in all the patients analysed, even in the nearly asymptomatic ones. In synthesis they are: 1) pelvis orientation in space: all the subjects showed a tendency to maintain the hemipelvis of the affected side higher and more advanced than the contralateral one. This means enhancement of hip adduction and extrarotation; 2) reduction of hip extension. In the most severe cases the extension movement was very abruptly stopped at a given angle (close to the standing up-right value); 3) reduction of the extensory moment of forces in the second half of stance phase. In several cases the initial, flexory moment was very prolonged along the stride cycle; 4) joint power very reduced at all the joints and particularly in the absorption phases.

Most of the above parameters showed a relatively good statistical correlation with the Harris Hip Score (r in the range 0.61 - 0.82). Other gait parameters of biomechanical interest, showed a more complicated correlation with the clinical degree of severity. For example the moments in the frontal plane (adductory) were enhanced in patients with high HHS and again in those with very low HHS. Patients in the range from 70 to 40 showed in general lower than normal adductory moments at the affected hip joint. Concerning the spatio-temporal parameters decreased velocity and stride length were observed. Stance phase duration in percentage of the stride cycle was increased on the un-affected side and reduced on the affected one. Despite rather large asymmetrical features in all the variables analysed, double support time and anterior step length were very similar on left and right side, even if they differed from normal.

Patients analysed six months after hip joint replacement showed some interesting changes in the spatio-temporal parameters, but preserved the same individual peculiarities in most of the biomechanical variables analysed. In particular the angular limitations and the joint moments looked to be very similar to those obtained before the surgical intervention.

Patients analysed one year after the endoprosthesis application showed interesting signs of recovery, either in terms of angular excursions and in terms of moments, powers and spatio-temporal parameters. Thus it seems that a longer period than usually assumed is required for a complete, documented recovery. Pain relievance which is consequent to the surgical intervention seems not to be sufficient for the physiological patterns to re-emerge after long term adaptation to joint disease. A more focused physical treatment based on analytical data would in all likelihood help an efficient speed out of the recovery process.

REFERENCES

1. M.P. Murray, D.R. Gore and B.S. Clarkson, "Walking patterns of patients with unilateral hip pain due to osteo-arthritis and avascular necrosis", J. Bone and Joint Surgery, 53-A, pp. 259-274, 1971.
2. A. Pedotti and C. Frigo, "Quantitative analysis of locomotion for basic research and clinical applications", in: Functional Neurology, vol. 7 Suppl. 4, pp. 47-56, 1992.

TREMOR AND EMG MEASUREMENTS OF LOCALIZED MUSCLE FATIGUE IN VDU WORK

Gunnevi Sundelin (1), Lage Burström (2), Mats Hagberg (3)
(1) Department of Occupational Medicine, University Hospital, S-901 85 Umeå Sweden, National Institute of Occupational Health, (2) Division of Technical Industrial Hygiene, S-907 13 Umeå, (3) Division of Work and Environmental Physiology, S-171 84 Solna, Sweden

INTRODUCTION

Evaluation of localized muscle fatigue in industry has often been assessed by using EMG measurements of elevation of signal amplitude and concomitant shift in power spectrum toward lower frequencies. Muscle tremor measurements can also be used to detect posturally induced muscle fatigue to augment EMG metrics of fatigue in low exertion and variable duty cycle tasks.

The aims of this study were to determine development of localized muscle fatigue in shoulder and neck by analysing EMG signal characteristics and postural tremor response.

METHODS

Twelve secretaries typed from a written manuscript continuously for 2 hours on a computer. The work place was initially adjusted for each person and thereafter the elbow-rest of the chair was raised to further induce static load on shoulder and neck muscles. The EMG activity was recorded from the lateral and the cervical portions of the descending trapezius muscle and from the levator scapulae with surface electrodes on the right side during the two working hours. Small Medinik amplifiers (Örbyhus Sweden) were used for linear amplification with a bandwidth of 5-500 Hz. The amplified signals were recorded on a FM tape recorder with a bandwidth of DC to 1250 Hz. The analogue electrical signals were converted to digital representation at a sampling frequency of 4 kHz per channel with 12 bit accuracy over the signal range ± 5V. FFT technique was used to obtain the power spectral density function for an average time of 250 ms with a bandwidth of 10-500 Hz.

Postural tremor of the right arm was measured using a accelerometer (Brüel & Kjaer 4384) mounted on top of a small cylinder which the secretaries held in their right hand for one minute, with 90 degrees shoulder flexion, before and after the VDU work. The signals from the accelerometer were amplified by a charge amplifier (Brüel & Kjaer 2635) before it was fed to a real time analyser (Brüel & Kjaer 2032) with a bandwidth of 1-2000 Hz.

Discomfort ratings were assessed every five minute during the work from the neck, shoulder, elbow and hand.

RESULTS

There was a significant increase ($p<0.05$) in the mean tremor activity (m/s^2) between measurements before and after the VDU work. The relative change in tremor activity can be seen in figure 1. The greatest changes can be seen between 8 to 20 Hz and minor changes up 100 Hz.

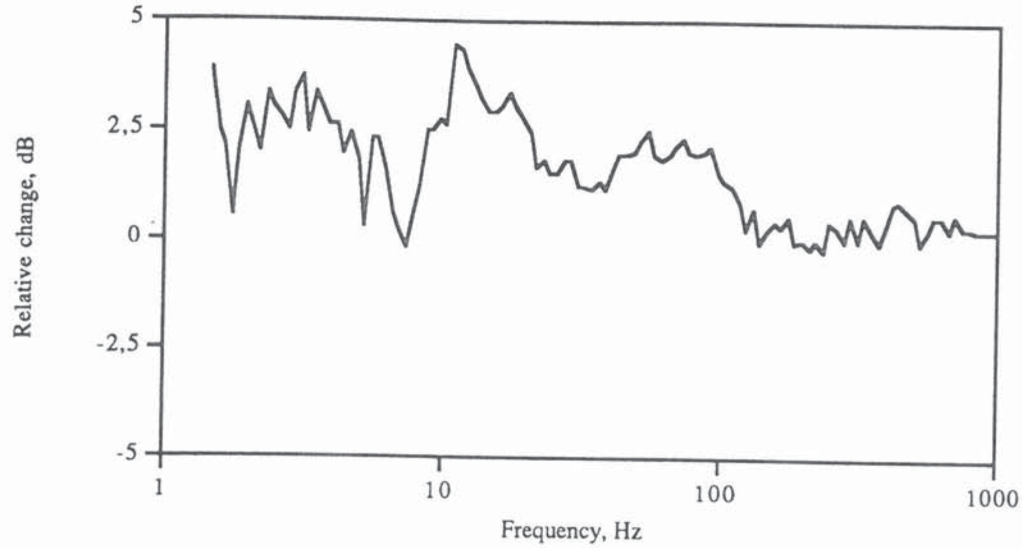


Fig 1. The relative change of tremor activity as a function of frequency for all subjects.

Slope coefficients for the regression line for EMG, MPF and RMS amplitudes, were calculated for the whole work period. The correlation coefficients between the slopes for the regression lines and the difference in tremor were, in RMS 0.42 for the levator scapulae muscle, 0.42 for the cervical part and 0.20 for the lateral part of the descending trapezius muscle. The corresponding correlations for MPF were -0.53, -0.22 and -0.48.

The correlation between tremor measurements and ratings of perceived discomfort in different body regions was low.

CONCLUSION

In agreement with other studies (1,2) the present results showed that physical effort in shoulder and neck muscles (fatigue) during VDU work influences tremor amplitude. However, the relationship between EMG metrics and tremor measurements in real work situations needs further investigation.

REFERENCES

1. Wiker, S.F., Chaffin, D.B., Langolf, G.D., (1989) Ergonomics 32:211-237
2. Viitasalo, J.T., Gajewski, J., (1993) ISB, abstract 14145-1415

AUTHOR'S ADDRESS

Gunnevi Sundelin, Ph.D.
Department of Occupational Medicine
University Hospital
S-901 85 Umeå, Sweden
FAX 009-46-90-102456

NEUROMUSCULAR KNEE JOINT CONTROL DURING DOWNHILL WALKING

M. Kuster[1], G.A. Wood[2] and S. Sakurai[3]

[1]Department of Orthopaedic Surgery, Kantonsspital, St. Gallen, Switzerland.
[2]Department of Human Movement, The University of Western Australia, Perth, Australia.
[3]Research Centre of Health, Fitness and Sports, Nagoya University, Nagoya, Japan.

INTRODUCTION

This investigation sought to identify neuromuscular mechanisms by which we cope with the potentially destabilising joint stresses during downhill walking. Theoretical models of the human knee joint have indicated that coactivation of the hamstring and quadriceps muscle groups progressively unloads the anterior cruciate ligament (ACL) between full extension and 22 degrees of knee flexion (O'Connor, 1993). Moreover, the gastrocnemius muscle is purported to be an important knee joint stabiliser for knee flexion angles beyond 22 degrees. Insofar as downhill walking on a 20% slope necessitates knee flexion angles of 40 degrees during stance (Wall *et al*, 1981) it was predicted that electromyographic (EMG) coactivity of the above mentioned muscle groups would be evident during this locomotor task, and that such coactivity would be of particular benefit to the ACL-deficient person.

METHODS

Surface electromyograms were recorded from the rectus femoris, biceps femoris and gastrocnemius muscles of 12 normal healthy subjects during both level and downhill walking at a standardised cadence of 120 steps/sec. Downhill walking was undertaken on a ramp of 6m length and 19% gradient. EMG signals were sampled at 500 Hz, full-wave rectified and lowpass filtered (6 Hz) and then averaged over 10 - 19 step episodes from 0.1 sec before heelstrike (HS) to 0.6 sec after heelstrike (i.e. end of stance). These ensemble averages were then normalised to a percentage of each subject's linear envelope peak value and finally averaged over all subjects.

RESULTS

Similar patterns of activity of the quadriceps and hamstring muscles were observed during both level and downhill walking with a high degree of coactivity around heelstrike. However, significant reciprocity was observed between these two muscles during late swing and late stance in downhill walking. The major burst of quadriceps activity occurred later and was more tightly tuned to early support during downhill walking, and hamstring activity increased in late stance while quadriceps activity decreased during this period.

The most pronounced neuromuscular adjustment to walking downhill was displayed by the gastrocnemius muscle. The lack of propulsive force needed at toe off produced the expected reduction in muscle activity during late stance. However, a dramatic increase in muscle activity was observed in this muscle around heel contact. Insofar as a net dorsiflexor moment of force prevails at heel contact during both downhill and level walking (Kuster, *et al*, 1993) this increased activity of the biarticular gastrocnemius muscle is evidently necessary to effect knee joint control. This finding is consistent with the knee joint model predictions mentioned above and suggests that the gastrocnemius muscle becomes an essential adjunct to quadriceps and hamstring coactivity in order to prevent excessive tibial anterior shear during knee flexion angles greater than 22 degrees. This knee joint control mechanism would be particularly important for the ACL-deficient subject given the greatly increased mechanical demands of downhill walking (Kuster *et al*, 1993).

CONCLUSIONS

During downhill walking the much higher knee joint shear forces necessitated greater reciprocity of quadriceps and hamstring activity, together with a large burst of gastrocnemius activity at heelstrike in order to provide adequate stability for the knee.

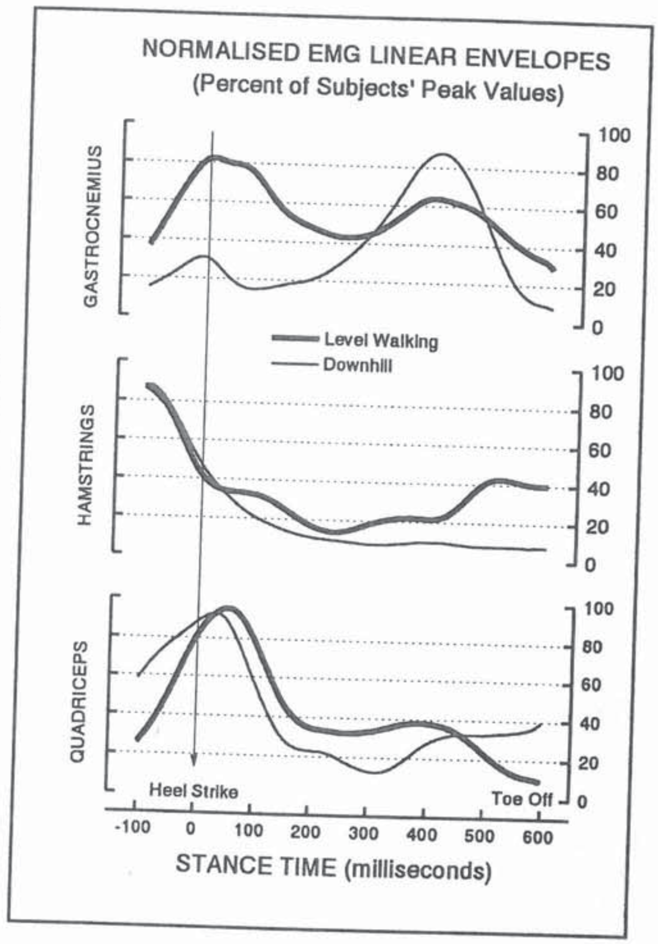


Figure: Averaged linear envelopes of EMG activity recorded from rectus femoris, biceps femoris and gastrocnemius during downhill and level walking. Individual envelopes were normalised to each subject's peak level of EMG activity before averaging.

REFERENCES

O'Connor, J.J. Can muscle co-contraction protect knee ligaments after injury or repair. **Journal of Bone and Joint Surgery**, 75-B(1),41-48,1993.

Wall, J.C., J.W. Nottrodt, and J. Charteris. The effects of uphill and downhill walking on pelvic oscillation in the transverse plane. **Ergonomics**, 24(10),807-816, 1981.

Kuster, M., S. Sakurai and G.A. Wood. A Kinematic and kinetic comparison of downhill and level walking. Submitted to **Clinical Biomechanics**, 1993.

Corresponding Author: Dr Markus Kuster, Department of Orthopaedic Surgery, Kantonsspital, 9000 St. Gallen, Switzerland.

DYNAMIC BALANCE CHANGES FOLLOWING ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION

M.A. Finley, P.A. Anderson
Department of Physical Therapy, University of Maryland School of Medicine, Baltimore, MD, U.S.A.

SUMMARY

The balance responses of post-operative anterior cruciate ligament (ACL) reconstruction subjects and healthy individuals were studied. It was found that no difference existed in the contraction timing , amplitude or reaction and recovery times between these groups. It was also discovered that with greater acceleration of perturbation, the activation patterns in the subjects as one group resembled a more of a hip strategy, or combination of strategies.

INTRODUCTION

Balance studies have indicated the need for proper sensory, particularly proprioceptive, inputs in order for normal balance responses to occur. Dynamic balance control is the ability to maintain an upright posture, during motion. This is accomplished through mechanical balance responses. Postural adjustments, or balance responses refer to the changes in postures that occur in order to maintain the COM within the base of support, or to return the COM within the limits of the BOS after a perturbation has occurred (1).

The goal of this study was to document dynamic balance reactions in post-operative ACL reconstruction subjects as compared with healthy individuals. This study proposed that with anterior cruciate (ACL) reconstruction, changes in muscular contraction timing, amplitude, and general reaction and recovery times may be seen during balance disturbances.

METHODS AND RESULTS

Male subjects (eight normal/eight ACL) were exposed to four anterior and four posterior platform perturbations at two accelerations(125cm/s^2, 150cm/s^2). During these perturbations, EMG data from rectus femoris(RF), medial hamstrings (MH), medial gastrocnemius (MG), and tibialis anterior(TA) and data on the reaction and recovery times from foot switches were collected. Analysis of variance statistical procedures were utilized to determine if significant differences existed between groups, muscles, platform direction or acceleration in terms of the dependent measures.

No significant differences were found between the groups in EMG onset time, EMG amplitude, reaction or recovery time. Collapsed across groups, significant interactions between muscles and platform direction were found for both EMG variables (Fig 1 & 2). MG and MH muscle activity preceded the onset of the RF and TA during posterior perturbation. No muscle onset differences occurred during anterior translation, but the amplitude of TA was significantly greater than the other muscles. During posterior translation the amplitude of the MG was found to be significantly greater than TA and RF, but not the MH.

FIGURE 1
ONSET TIME
MUSCLE X DIRECTION

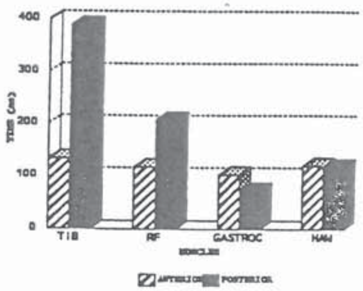
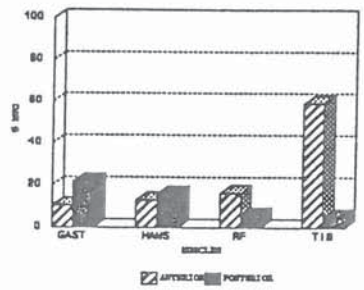


FIGURE 2
AMPLITUDE
MUSCLE X DIRECTION



DISCUSSION

A possible explanation for the findings of no significant difference between the groups may be that these experimental subjects had sufficient proprioception, and balance responses. This proprioception may have been developed through the retraining of muscles and tendons, or through the mechanisms adapted from higher centers as in the Konradsen (2) study.

Additionally, the higher acceleration of platform perturbation resulted in a combination of hip and ankle strategies, unlike the findings by Nashner (3,4). Other investigations studied the responses at lower velocities and accelerations. The choice of the higher acceleration may have resulted in the different recruitment patterns and onset times.

REFERENCES

1. Duncan,P., Studenski,S., Chandler,J., Bloomfield,R., LaPointe, L. (1990). Physical Therapy, 70:88-96.
2. Konradsen L, Ravn J, Sorensen A. (1993) Proprioception at the ankle: The effect of anaesthetic blockade of ligament receptors. Br J Bone and Joint Surgery, 75B:433-436.
3. Nashner L. (1976) Adapting reflexes controlling human posture. Exp Brain Res, 26:59-72.
4. Nashner L. (1977) Fixed patterns of rapid postural responses among leg muscles during stance. Exp Brain Res, 30:13-24.
5. Schutte M, Dabezius E, Zimny,L, Happel L. (1987) Neural anatomy of the human anterior cruciate ligament. J Bone and Joint Surgery, 69A:243-247.
6. Anderson P, Smith G, Alon G.(1992) Dynamic performance parameters in young adult and elderly human subjects. Paper presented at the APTA Annual Conference, Denver, Colorado.

ESTIMATION OF ACTIVATION PATTERNS FROM SURFACE MYOELECTRIC SIGNAL IN TKR PATIENTS DURING GAIT

M.G. Benedetti°, P. Bonato•, F. Catani°, T. D'Alessio †, M. Knaflitz ‡, A. Battistini°, A. Leardini°, M. Marcacci°, S. Giannini°,

°Biomechanics Laboratory, Istituto Ortopedico Rizzoli, Bologna, Italy • Dipartimento INFOCOM, Università di Roma La Sapienza, Roma, Italy †III Università di Roma, Roma, Italy ‡Dipartimento di Elettronica, Politecnico di Torino, Torino, Italy

INTRODUCTION

In motion studies, surface myoelectric signal analysis is commonly used to estimate muscle on-off timing, in order to correlate muscle activity to the kinematics and kinetics of motion. Early researchers used to collect the myoelectric signal by means of indwelling electrodes, but this technique is not easily applied to moving subjects and also causes obvious discomfort. More recently, the advent of active probes made it possible to utilize surface detected myoelectric signal. Surface electrodes are comfortable even during dynamic contractions, easy to use, and safe. Unfortunately, this technique often leads to poor signal to noise ratios (S/N), it is exposed to crosstalk, and relative movements among active fibers and detection surfaces cause large variations of the signal amplitude. These variations are not related to changes of the muscle force output, and may be misleading [1]. The information content of muscle on-off timing is not as rich as that of the signal envelope, but its estimation and interpretation are much more robust and rational. Typically, on-off timing is obtained by first extracting the signal envelope and then by observing in which time intervals the envelope crosses a threshold arbitrary chosen, that discriminates background noise from the signal. This procedure does not allow the user to choose (or to estimate) the sensitivity (true positives) and the specificity (true negatives) of the detection. In this work we apply a novel algorithm based on the statistical properties of the signal, previously characterized both theoretically and experimentally on a group of normal subjects. The aim of this study is to objectively estimate the on-off timing of eight muscles of the trunk and lower limbs during walking, in a group of pathological subjects operated on total knee replacement (TKR). Moreover, the correlation between muscle activation and kinematics and kinetics of locomotion in these subjects is discussed. Previous authors [2] reported that TKR patients improve gradually their walking ability after surgery, but some abnormalities persist even after long periods of time. In the past, the abnormal knee pattern was attributed to different possible causes, as a) the weakness of the extensor muscles, b) the proprioceptive deficit after replacement, c) the multijoint degenerative involvement, and d) the implant geometry.

METHODS

Five patients operated on TKR (International by Howmedica) for primary osteoarthrosis were analyzed six months after surgery. Clinical assessment was performed in all the patients. Mean age was 64.8 years. Gait analysis was performed by means of an ELITE System (by BTS, Milan, Italy) for kinematics and a KISTLER platform for kinetics. Surface myoelectric signals were recorded through a TELEMG (by BTS, Milan, Italy) system. Raw signals were recorded with a sampling frequency of $500\,Hz$, bandpass filtered ($20-200\,Hz$), and processed by the statistical detection algorithm previously cited. Eight muscles were examined: the

Homolateral Longissimus Dorsii, the Contralateral Longissimus Dorsii, the Gluteus Medius, the Rectus Femoris, the Biceps Femoris, the Hamstrings, the Gastrocnemius, and the Tibialis Anterior.

RESULTS

The muscle activation pattern of TKR patients resulted remarkably different from the normal one, on both the operated and the sound side. Particularly, we noted I) a prolongation of the activity of the Rectus Femoris and Hamstrings during the stance phase, II) a premature activation of the Gastrocnemius at the heel strike, and III) a protracted activity of the Tibialis Anterior until the mid-late stance phase. These findings may partially explain the abnormalities found in the kinematics and kinetics of the locomotion. The knee pattern was characterized bilaterally by a decreased flexion during the loading response and the mid-stance phases. Three patients had also a decreased maximum knee flexion moment during mid-stance. The external knee extension moment at the loading response phase was reduced, as well as the power at the knee throughout the stance. The muscle on-off timing in TKR patients resulted generally repeatable in the intra-subject evaluation. Clinical assessment pointed out the absence of pain at the operated knee, a mean flexion-extension range of motion of 0-110 degrees and a mild quadriceps weakness.

CONCLUSION

Surface myoelectric signals processed by the statistical algorithm provide interesting and reliable information about the muscular activity during gait. In TKR patients this methodology showed an abnormal gait pattern characterized by a continuous prolonged activity of the Rectus Femoris and Hamstrings, an early activation of the Gastrocnemious at heel strike, and cocontraction of the Gastrocnemious and Tibialis Anterior during early and mid-stance. This abnormal, stiff-legged, pattern is functional to stabilize the body weight of subjects that underwent arthroplasty, thus reducing external moment shear forces at the knee joint. Knee power is reduced suggesting muscle contraction can be considered as isometric during loading phase. The abnormal contralateral muscular and joint angular patterns probably improve gait symmetry.

REFERENCES

[1] J.V. Basmajan, C.J. DeLuca, "Muscles Alive", William & Wilkins, 1985

[2] M.E. Steiner, S.R. Simon, J.C. Pisciotta, "Early Changes in Gait and Maximum Knee Torque following Knee Arthroplasty", Clin. Orthop., no. 238, pp. 174-182, January 1989

Acknowledgements

This work was partially supported by the Italian National Research Council under grant # 91.02901. CT07 and partially carried out in the frame of the CEC programme AIM, project a-2002CAMARCII.

Dr. Fabio Catani
Biomechanics Laboratory
Istituto Ortopedico Rizzoli
Via di Barbiano 1/10
40136 - BOLOGNA - ITALY
Tel. +39-51-6366529. FAX +39-51-583789
E-MAIL: "VI6BOQ71@ICINECA.CINECA.IT"

EMG ACTIVITY EXHIBITED BY ACL DEFICIENT KNEE OF HIGH LEVEL SKIERS USING A CUSTOM-MADE KNEE BRACE DURING DOWNHILL SKIING

M. Lamontagne*, G. Németh, K.S. Tho#, E. Eriksson
Dept. of Orthopaedics, Karolinska Hospital and Dept. of Sports Orthopaedics Surgery, S-171 76 Stockholm, Sweden. * Guest researcher from the School of Human Kinetics and Dept. of Anatomy and Neurobiology, University of Ottawa, Ottawa, Canada; # Fellow from Dept. of Orthopaedics Surgery, Alexandra Hospital, Singapore.

INTRODUCTION

In downhill skiing, large incidence of knee injuries has been reported in literature (1). Most of these injuries occur mainly to the anterior cruciate ligament (ACL)(2,3). The injury prevention of this anatomical structure becomes the utmost importance that can be done by improving ski equipment. After an ACL injury, athletes or simple participants who want to get back to their favourite activity, can use a knee brace to secure ACL deficient knee joint. What is the effectiveness of a knee brace during this kind of activity?
The purpose of this study is to compare the electromyography (EMG) activity of vastus medialis (VM), biceps femoris (BF), semi-menbranosus (SM), and gastrocnemius medialis (GM) of both legs while using and not using a custom-made knee brace on ACL deficient knee (ACL-DK) during downhill skiing in a slalom course.

METHODOLOGY

Seven high level skiers with ACL-DK (4 females and 3 males) between 23 and 46 years of age participated in the study. All subjects were tested according to the Lachman, valgus-varus stability, and pivot shift tests to evaluate the level of deficiency of the injured knee. EMG surface electrodes were placed on the VM, BF, SM, and GM of both lower limbs. The custom-made knee brace designed for each subject (Defiance, Smith & Nephew Donjoy Inc.) was adjusted by a specialist before performing their four runs (two runs with and two runs without brace). A slalom course was set up of 10 turns (gates) in a slope rated double black diamond. The skiers went down the slalom course with and without knee brace on the injured knee while raw EMG and videography were recorded. A telemetric EMG system (TELEMYO 16; Noraxon OY) recorded eight EMG channels of the above mentioned muscles at 1000 Hz per channel. The video camera (PANASONIC; model NV-MS4 Super VHS) signal was used along with VGA-to-Video conversion board (VGALink) which allows to superimpose the computer screen and video image on the video monitor and then recorded on a video cassette recorder (PANASONIC; NV-FS200EC). This allowed us to synchronize video image and EMG recordings.
Each run consisted of two cycles; one cycle was defined by one successive left-right-left turn. Usually, the six middle turns in a run were used for the data analysis. The EMG signal of all muscles during a cycle was rectified and low-pass filtered (Butterworth; critically damped 4th order dual passes) and normalized by peak amplitude and by time. Each cycle for the same condition is averaged and the integral of the LE EMG was calculated and saved for statistical analyses. An analysis of variance was performed to predict any statistical difference in the EMG signal between injured and normal knee with or without knee brace during downhill skiing. One subject was discarded because of the bad EMG signal recordings.

RESULTS AND DISCUSSION

The ensemble average and peak normalized EMG activity of all muscles on the ACL deficient leg was higher with the custom-made brace then with no brace during downhill skiing in a slalom course (Figure 1). Moreover, the subjects exhibited higher EMG activity on all muscles with the normal leg than the injured leg. However, only the biceps femoris EMG activity on the injured leg showed a significant difference with the knee secured with the brace and not secured. These findings do not support the general belief that the ACL deficient knee is stabilized by active contraction of hamstring musculature to compensate its joint laxity (4). Again, the present results, in conjunction with other investigations, did not show the automatic hamstrings excitation when the ACL is injured (5). The automatic hamstring contraction in response to the ACL deficiency was obviously not solicited in this activity.

From the figure 1, the results suggest that the subjects used more of their muscles with the injured leg fastened with a knee brace. Also from other findings of this study (not shown), the subjects used more of their normal leg then the injured leg which was not secured with a knee brace. It seemed that the investigated subjects were more confident in using their normal leg while the injured leg was not secured with a knee brace and they were more confident in using the injured leg while secured. Therefore, the EMG activity of the injured leg while secured with a knee brace increases and the injured leg exhibites a similar EMG pattern as the normal leg.

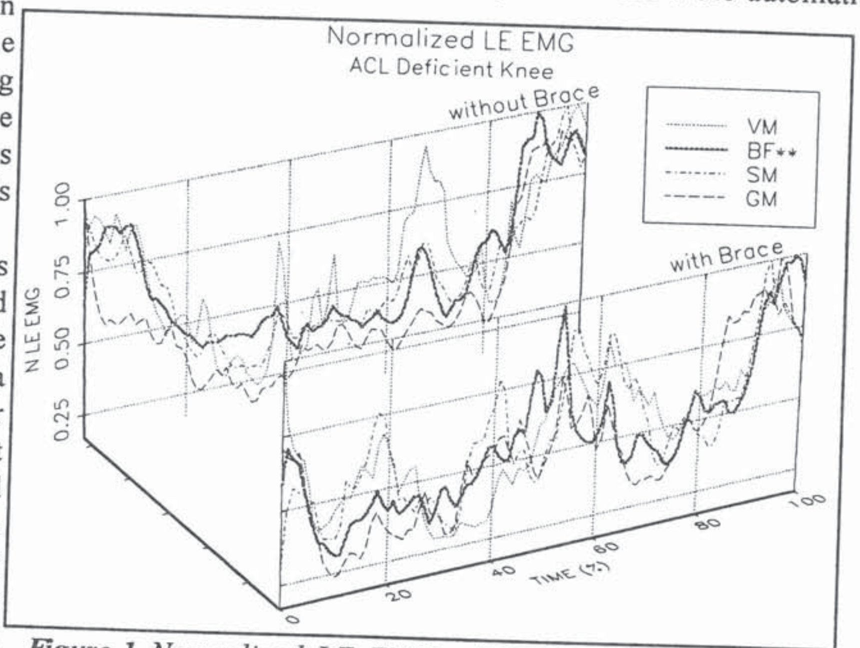


Figure 1 Normalized LE EMG of ACL deficient Knee with and without brace during downhill skiing

REFERENCES

1. Weaver JK and al. (1985) Clin. Orthop. (199): 185-191
2. Feagin JA and al. (1987) Clin. Orthop. (216): 13-18
3. Higgings RW & Steadman JR (1987) Am. J. Sports Med., 15(5): 439-447
4. Tibone JE & Antich TJ (1993) Clin Orthop. (288): 35-39
5. Gradiner MD, Koh TJ & Miller GF (1993) Med Sc Sports Exerc., 24 (10):1075-89

ACKNOWLEDGEMENTS

This project has been funded by Smith &Nephew Donjoy Inc and Karolinska Institute. The authors want to thank Lieve van den Berghe for the knee brace adjustment on subjects.

A NEW TECHNIQUE TO IMPROVE REHABILITATION AFTER ACL-RECONSTRUCTION USING EMG OF THE VASTUS MEDIALIS MUSCLES :

Wißmeyer, Th.; Sterk, J.; Boos, O.; Kinzl, L.
From the Departments of Orthopedic Surgery and Computer Science,
University Hospital, Ulm, FRG

INTRODUCTION

Selecting the appropriate program of muscle training is an important factor to improve the results of knee surgery. Every effort is made to avoid the atrophy of the vastus medialis muscle (VM).After reconstruction of the anterior cruciate ligament (ACL) active extension of the knee should be done very cautiously.

For many years the widley held opinion about the function of the VM was, that he had a specific action during the last degrees of knee extension. And so the last 20 or 30 degrees of knee extension are considered to be of great importance. But despite many efforts no electrophysiologic research could prove this opinion. From the anatomical point of view nowadays is emphazised the possibility to separate the VM into two different parts concerning direction of pull and innervation, the vastus medialis obliquus and longus (VMO and VML)(1,2). But it could not be demonstrated a clear difference in function of the two heads of this muscle, especially no differences occuring over the range of knee extension (recently in 4).We attempted to address this "vastus medialis controversy" using the following method.By means of a highly developed electromyographic equipment we show the participation of the individual parts of the quadriceps muscle during isometric knee extension. We are able to differenciate the action of the vastus medialis obliquus muscle (VMO) from that of the vastus medialis longus (VML) at precisely defined muscle forces and during different angles of knee flexion/extension. To our knowlege this is the first isometric study using different exactly defined forces, different knee angles and integrated and normalized raw-EMG derived from intramuscular electrodes.

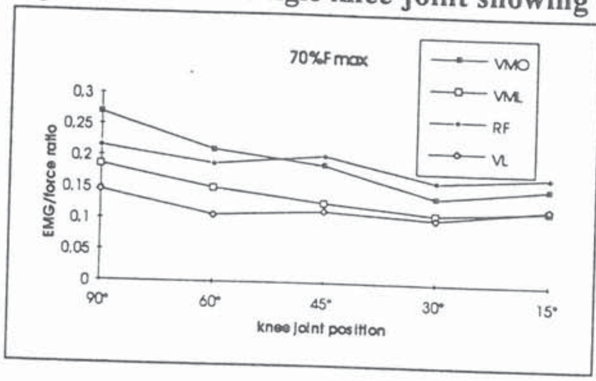
METHODS

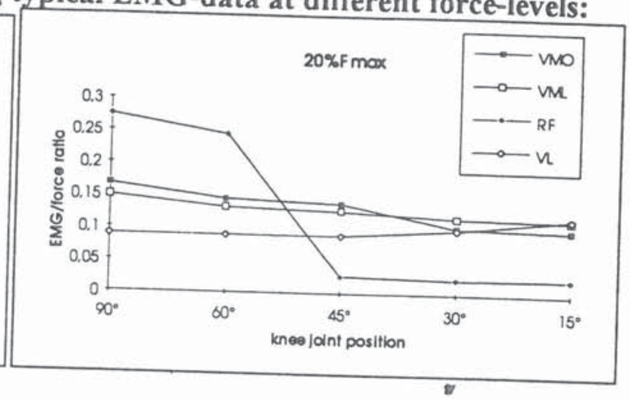
23 healthy volunteers (10f, 13m; 16 to 47 years) without prior history of knee trauma or symptoms were used as probands. Using a pair of intramuscular wire electrodes (Pl/Ir; 70μ; Teflon coated) per channel inserted in a modified manner according to Basmajian(3) with a inter-electrode-distance of 2cm we recorded a four channel EMG with a band width from 20Hz to 7,5KHz. The raw EMG-signal was AD-processed with a sample frequency of 20 KHz per channel and registrated on optical disk. The exactly defined anatomic points of needle insertion avoiding the motor-endplate-regions and the standardized and controlled position of the proband avoiding rotational movement of the leg made the measurements very reliable. After determination of the maximal muscle power (Fmax) in each of five knee-positions (15°, 30°, 45°, 60°, 90°) the probands have to perform an isometric knee extension with defined muscle force (20%, 40%, 60%, 70% Fmax). An optical/acoustic feed-back instrument made a constant active power adjustment possible by the proband with an accuracy of 1%. The integrated EMG of a four second period of isometric knee extension in defined positions is calculated and normalized on the force value in each trial, which was registrated as a fifth channel. The values of four muscles are compared: VMO, VML, m.rectus femoris (RF) and m.vastus lateralis (VL). No absolute EMG-data were compared to avoid the problem of variable EMG-signals depending on the location in the muscle. We always made up the relation between two different muscles in one position and force, or between different positions of one single muscle.

RESULTS

In 80-90% of cases the activity of the VMO at 90° knee flexion was the highest of the 4 muscles when the test was carried out at 70% of Fmax respectively. The VMO and VML differ in their behaviour, the EMG-activity of the VMO was greater in 80% of our subjects. In contrast the RF at 90° shows higher values concerning 40% and 20% Fmax respectively, his activity decreases however to almost zero during the last 45 degrees of knee extension at low forces (compared to VMO $p<0,00001$), see Fig.4. The VMO showed higher activity in 90° compared to 15° normalized to force ($p<0,00001$). In contrast to widely held opinions there is no dominance of the vastus medialis muscles during the final degrees of knee extension.

Fig.1 and 2: One single knee-joint showing typical EMG-data at different force-levels:





Comparing the "Activation-Ratio: VMO/VL ;90°;70%" (relation of the normalized EMG-value of VMO to that of VL in 90° at 70% Fmax) to the "Activation Ratio: VMO/VL;15°;70%" there were significantly higher values to see at 90° ($p<0,05$; Wilcoxon-Rank-Sum-Test). See Fig.3. This clear prevalence in activation was not to be seen in the VML.

Fig.3: VL / VMO at 70% Fmax
N=46; normalized EMG

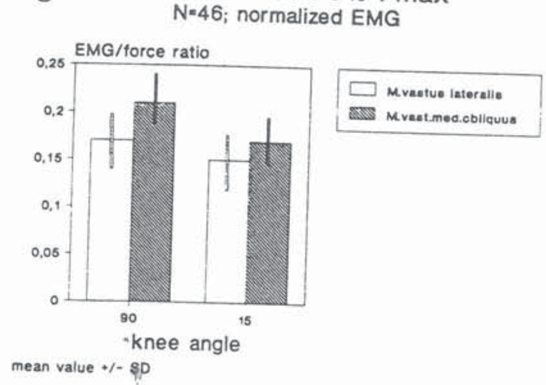
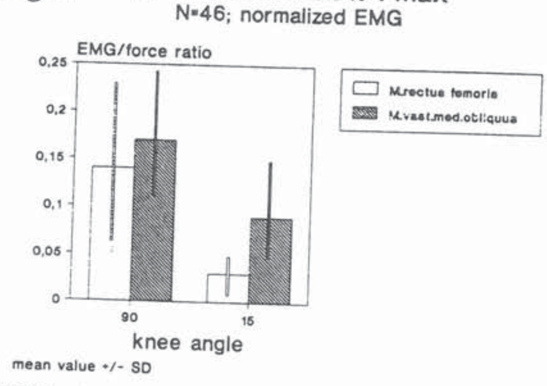


Fig.4: RF / VMO in 20% Fmax
N=46; normalized EMG



CONCLUSION

The above shows that the main action of the VMO is registrated at 90° knee extension at high isometric forces. Quadriceps training does not require active final knee extension. In isometric testing the maximum strain within the ACL occurs at a knee flexion of 45° to full extension (5), therefore exercise of the VMO should take place at 50% to 70% Fmax between 90° to 50° of knee extension, so this optimized training avoids undue stress to an ACL repair after surgery.

REFERENCES

1) Weinstabl R, Scharf W, Firbas W: The extensor apparatus of the knee joint and its peripheral vasti: anatomic and clinical relevance. Surg Radiol Anat, 11 (1989) 17-22.
2) Thiranagama R: Nerve supply of the human vastus medialis. J Anat, 170 (1990) 193-198.
3) Basmajian, JV and DeLuca CJ: Muscles alive, Williams&Wilkins, Baltimore, 1985.
4) Karst GM, Jewett PD: Electromyographic analysis of exercises proposed for differential activation of medial a.lateral quadriceps femoris components. Phys Therapy, 73 (1993) 286-299.
5)Beynnon B, Howe JG, Johnson RJ, Fleming BC: The measurement of anterior cruciate ligament strain in vivo. Int Orthop (SICOT), 16 (1992) 1-12.

ANALYSIS OF SIX DEGREES OF FREEDOM KNEE KINEMATICS OF ANTERIOR CRUCIATE LIGAMENT INJURED AND UNINJURED KNEES DURING LOCOMOTION

Li-Qun Zhang and Richard Shiavi
Department of Biomedical Engineering, Vanderbilt University
Sensory Motor Performance Program, Rehabilitation Institute of Chicago

INTRODUCTION

The anterior cruciate ligament (ACL) is a very important structure for controlling knee joint stability and movement. Because of the complex and interdependent ways the ligaments, capsule, muscles, and bones work together to control knee joint stability and motions in the three-dimensional space, the biomechanical alterations caused by ACL injury, the underlying physiological processes following ACL injury, and the neuromuscular compensatory responses adopted by the patients are still not well understood. The objectives of the research were to compare the knee kinematics during locomotion between the ACL injured and uninjured populations, to study the alterations of knee joint kinematics caused by ACL injury, and to study how the patients compensate for the ACL injury.

METHODS

Forty six subjects were studied. Sixteen of them had a completely ruptured ACL in one of his/her knees. The rupture occurred at least a year before the experiment. A six DOF goniometer was mounted on the subject's lower limb (Shiavi *et al.* 1987). At the beginning of the experiment, the subject was asked to stand still with their second metatarsals pointing forward. The posterior aspect of the greater trochanter, fibular head, and lateral malleolus were aligned vertically, and the goniometer angles were recorded as the "standing position" (Shiavi *et al.* 1987). The subjects were then asked to walk straight on a 12m long walkway at their natural speed. Data were sampled in the middle 5m by a computer, and the same trial were repeated four times.

The six channel goniometer measurement was low-pass filtered, transformed into the following six DOF anatomical descriptions to characterize the tibial movement relative to the femur: three rotations in the sequence of the knee flexion-extension, adduction-abduction, and internal-external rotation, and three translations along the tibial x (lateral-medial), y (posterior-anterior), and z (upward-downward) axes. The data were then normalized along the time axis so that every stride was represented by 55 data points. The data were visually inspected and those trials with large noise, unstable measurement over the trial were then excluded from further analysis. For each of the forty-six subjects studied, five to twenty-four strides of data were left after the editing.

An unbalanced ANOVA statistical analysis was used to test whether the two group means were equal or not. To get a fair weight for each of the subjects, five strides were selected for every subject. In total, 150 strides from 30 normal subjects and 80 strides from 16 ACL injured subjects were used in the analysis. The ANOVA analysis was performed for each of the 55 points across a stride and for each of the six anatomical descriptions. The null hypothesis was that the two group means are equal.

RESULTS

Fig. 1 shows the comparison of mean knee kinematics patterns between the normal and ACL injured subjects. Fig. 2 shows the corresponding p-values given by the ANOVA test. Several major tendencies shown by the ACL injured subjects are summarized below:

- Compared to that of uninjured subjects, ACL injured subjects externally rotated their tibia shortly after heel-strike, and such an external rotation lasted until near the end of swing phase. This may reflect the subject's tendency to avoid stretching the injured ACL (whether it was completely torn or not) because tibial internal rotation elicits the most ACL elongation (Kennedy *et al.* 1977). This is also consistent with a previous result that the ACL injured subject delayed the major semitendinosus muscle activity from the end of swing phase to the beginning of stance phase (Zhang and Shiavi 1992). The explanation is that if activities of other muscles remain the same, the delay of ST activity may cause a significant increase in external rotation moment on the tibia at the late swing phase.

- The ACL injured subjects tended to pull their tibia laterally and this may be because after ACL injury the vastus lateralis and biceps femoris took over part of ACL's stabilizing function, especially in the direction of preventing the medial translation of the tibia relative to the femur.
- The tibia of ACL injured subjects was in a more anterior position in comparison to that of uninjured knees during locomotion. Restraining tibial anterior displacement relative to the femur is probably the most important and obvious function of the ACL. A rupture of the ACL reduced such restraining force greatly, and the posterior pulling generated by the hamstrings muscles was not strong enough to pull the tibia *fully* back. Notice that the excessive anterior tibial displacement of the ACL injured knee was mostly shown during the swing phase when the knee joint was not heavily loaded.

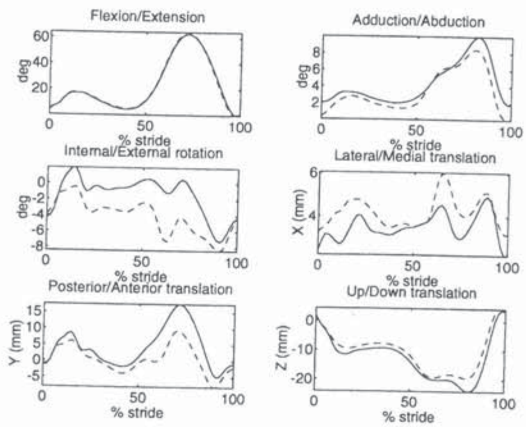


Fig. 1. Comparison of kinematics of normal and ACL injured knees during locomotion. The solid curves correspond to the uninjured knees, and the dashed curves represent the ACL injured knees. The positive directions for the six DOF are: flexion, adduction, internal rotation, and lateral, posterior, and upward translations, respectively. The abscissa represents the percentage of stride during locomotion.

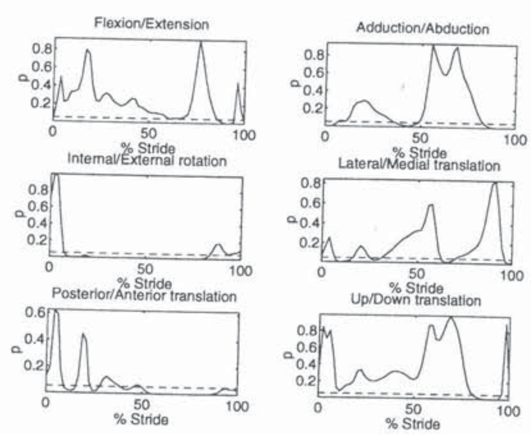


Fig. 2. ANOVA statistical analysis results on knee kinematics of normal and ACL injured knees. The solid curves represent the p-values and the dashed lines correspond to the level of 0.05 (5%). The abscissa represents the percentage of stride during locomotion. The ordinate is the p-value.

REFERENCES

J. C. Kennedy *et al.*, "Strain Gauge Analysis of Knee Ligaments", *Clin. Orthop. & Rel. Res.*, **129**:225-229, 1977.

R. G. Shiavi *et al.*, "Helical Motion Analysis of the Knee — I. Methodology for Studying Knee Kinematics During Locomotion", *J. Biomech.*, **20**:459-469, 1987.

L.-Q. Zhang & R. G. Shiavi, "A Study of EMG and Kinematic Patterns of Normal and ACL Deficient Subjects During Locomotion", *Proc. 2nd N. Amer. Congr. Biomechanics*, pp. 135-136, Chicago, 1992.

ELECTROMYOGRAPHIC ANALYSIS OF HABITUATION PROCESSES IN TREADMILL WALKING

Jia-Jin Chen, Ing-Shiou Hwang, Jiann-Jhy Liou, Tse-Chien Huseh*
Institute of Biomedical Engineering, *Department of Rehabilitation Medicine, NCKUH
National Cheng Kung University, Tainan, Taiwan, R.O.C.

INTRODUCTION: Treadmill is widely accepted in clinical study such as gait analysis and retraining. However, on the account of validity, the difference of treadmill walking and ground locomotion has long been a debatable issue. There is a need for caution when interpreting EMG data and other biomechanics parameters from studies on pathological gait analysis or biofeedback gait training on treadmill [2]. The aim of this research is to study the characteristics of muscle activity pattern during habituation process in treadmill walking in comparison to floor walking.

METHODS: Six healthy subjects, all males, without history of neuromuscular disorders, participated in this study. Ankle antagonists in both limbs, pretibial group and calf group were examined. At first, all subjects were requested to walk at their comfortable walking cadence on treadmill for one minute. As preliminary walking finished, EMG signals of pretibial and calf muscle groups were sampled at a rate of 1 KHz. In each trial, there were 8 data collection section which are at 30 sec, 1 min, 2 min, 4 min, 6 min, 8 min, 10 min, and 12 min after walking initialized. The interval of each collection was 10 seconds. Then, all the subjects were requested to walk on the floor with the same data collection scheme.

In this research, EMG activity is presented in a form of linear envelope (LE) which is commonly used in gait analysis. The EMG LE is generated from the raw EMG signal through band-pass filtering, rectifying, integration, normalization, ensemble averaging procedures. The variance ratio (VR) of EMG LE is applied to analyze the repeatability of EMG activity and thus to quantify the habituation process of normal adults walking on treadmill. The VR value, ranging from 0 to 1, is an index of similarity of signals. If EMG signals of different gait cycle within 10 seconds measuring period are totally identical, the VR value is zero. The VR trend of four muscle groups, measuring at 8 different walking stages were analyzed to find out the steady performance of each muscle group during the treadmill walking. The minimum time of reaching an acceptable VR was defined as habituated state. The basic assumption of habituation process of motor control is that proper practicing makes voluntary movement more accurate and less variant [1].

RESULTS: Fig. 1 shows representative examples of the VR trend during treadmill walking of two subjects. D-TA and D-GS denote the pretibial group and calf group of the dominant leg respectively. Similarly, ND-TA and ND-GS denote the pretibial group and calf group of the non-dominant leg. The pretibial group of dominant leg and non-dominant leg have higher VR trend than those of the calf group of both sides. The VRs of the pretibial group for subjects did not subside to a steady state within 12 minutes' exposure to treadmill walking. However, most of the calf group of the subjects will finally reach steady state during 12 minutes' treadmill walking.

The habituation process, especially in treadmill walking, is supposed to affect the muscle performance which can be observed from the EMG LE. Fig. 2 shows the EMG LEs of the pretibial group during treadmill walking in comparison with pooled EMG LEs of floor walking. We can observe that the phasic activity of pretibial group during treadmill walking is low phasic activity during loading phase and rather intense activation during swing phase.

DISCUSSIONS: By using EMG LE and VR repeatability measure, this research presents an approach for describing EMG phasic activities generated during human habituation on treadmill walking. Our results indicate that pretibial group of all of our six subjects would never habituate over 12 minutes' treadmill walking. In contrast, calf groups of four subjects reached habituated state after 2 minutes' exposure to treadmill walking, and the calf groups of the other two subjects demonstrate prolong habituation process. The cause of habituation difference is possibly related to the forward momentum imposed by the constantly mobilized treadmill belt. Further development of this approach is toward the assessment the prognosis of coordination before and after rehabilitation in helping clinician in gait evaluation, especially on the pathological gait measured on treadmill.

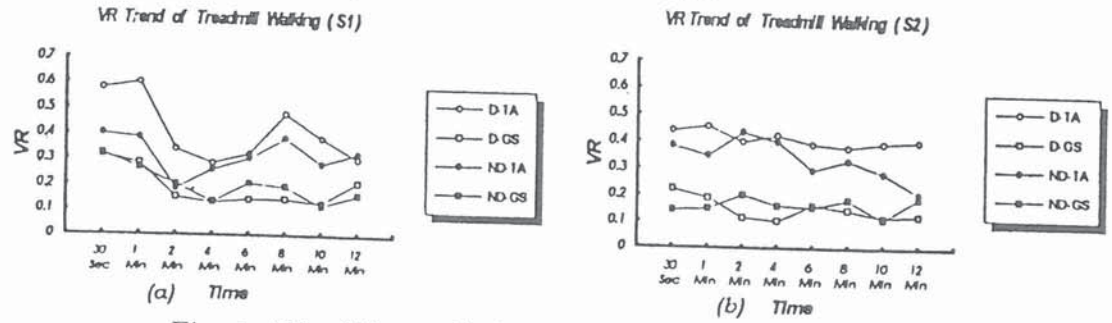


Fig. 1. The VR trend of treadmill walking in both legs.

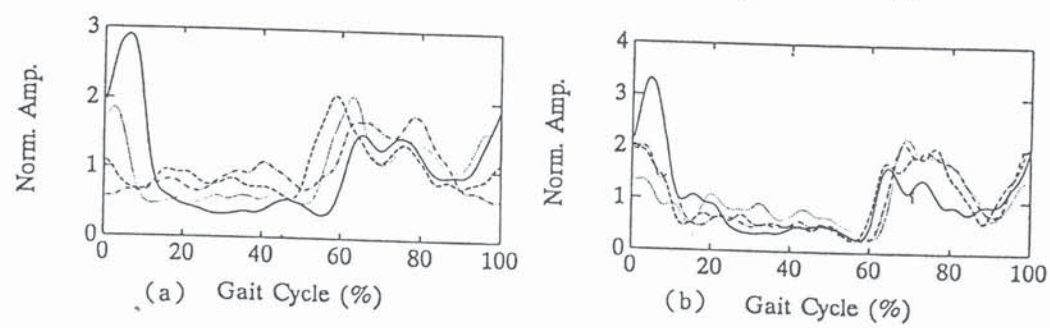


Fig. 2. Three linear envelopes of pretibial group during treadmill walking (dashed lines) and floor walking (solid line) for (a) Subject 1 (dominant), (b) Subject 2 (dominant).

REFERENCES

1. Arsennault, A.B., Winter, D.A., and Marteniuk, R.G. (1986) Treadmill versus walkway locomotion in humans: an EMG study. *Ergonomics,* 29:665-676.

2. Murray, M.P., Spurr, G.B., and Gardner, G.M. (1985) Treadmill vs. floor walking: kinematics, electromyogram, and heart rate. *Journal of Applied Physiology,* 59:87-91.

3. Wagenaar, R.C. and Beek, W.J. (1992) Hemiplegic gait: A kinematic analysis using walking speed as a basis. *Journal of Biomechanics,* 25:1007-1015.

Supported by grants in part from the National Science Council, R.O.C. (NSC 80-0404-E-006-53) and in part from the Department of Health (DOH83-HR-321).

PRECISE ESTIMATION OF FIRING RATES OF MULTICHANNEL SURFACE ELECTROMYOGRAPHY: A BISPECTRAL APPROACH

Jia-Jin J. Chen, Yue-Ren Lin, Thy-Sheng Lin*, Yue-Nan Jenp**

Inst. of Biomed. Eng., * Dept. of Neurology, **Dept. of Rehab. Med., NCKUH
National Cheng Kung University, Tainan, Taiwan, R.O.C.

INTRODUCTION: The spectrum of surface EMG can be considered as the multiplication of the spectrum of single motor unit (MU) and the spectra of MU firing rates [1]. The peaks below 40 Hz of the surface EMG spectrum are correlated to the firing rates. However, these peaks tend to be over-estimated because the phase coupling of harmonics. Thus, in previous studies only the first peak of the surface EMG spectrum was considered as the dominant firing rate of MU's. In this study, we develop a precise firing rates estimation scheme which utilize autoregressive (AR) spectrum analysis to detect the all the peaks and bispectrum analysis to eliminate the over-estimates. The bispectrum, a class of higher order spectral analysis, calculates the third moment of the signals, instead of second moment in traditional power spectrum in which the phase information is suppressed [2][3].

METHODS: In our experiment, fourteen pairs of EMG electrodes with differential preamplifiers are applied to a ring region of two major muscles, biceps and triceps, about elbow joint movement. Multichannel surface EMG signals and torque generated are measured during isometric extension and flexion contraction in step and ramp movements. This multichannel design facilitates the bispectral analysis on the study of nonlinear interaction between EMG signals of different channels.

The bispectrum for three zero-mean stationary processes, $\{X_j(n), j = 1, 2, 3\}$, is defined as the Fourier transform of the third order cross-cumulate sequence [3]. For a real pth order autoregressive (AR) process $X(n)$ can be described by

$$X(n) + \sum_{i=1}^{p} a_i X(n-i) = W(n) \tag{1}$$

where $E\{W^3(n)\} = \beta \neq 0$. The corresponding AR filter transfer function is

$$H(z) = 1/[1 + \sum_{i=1}^{p} a_i z^{-i}] \tag{2}$$

Finally, the cross-bispectrum is defined as

$$B(\omega_1, \omega_2) = \beta H(\omega_1) H(\omega_2) H(\omega_1 + \omega_2) \quad |\omega_1|, |\omega_2| \leq \pi \tag{3}$$

A peak in the cross-bispectrum indicates that the energy of $X_3(n)$ is the result of phase coupling between processes $X_1(n)$ and $X_2(n)$. If processes $X_1(n)$, $X_2(n)$ and $X_3(n)$ are replaced by the same process, $B_{XXX}(\omega_1, \omega_2)$ becomes the bispectrum.

The autoregressive (AR) spectrum analysis method is utilized to detecting all the peaks, i.e. the poles at AR model, below 40 Hz of the surface EMG spectrum. The order of AR model is determined by Akaike's information criterion (AIC). The over-estimates caused by the phase coupling of harmonics can be eliminated by detecting peaks in cross-bispectrum of EMG signals in adjacent channels. These peak spectra after eliminating over-estimates eliminating the over-estimates, these peaks in EMG spectra are represented in a time-varying form for analyzing the time-dependent firing patterns.

RESULTS: From the power spectrum of channel 5, shown in Fig. 1(a), only two peaks can be observed; however, in Fig. 1(b) more precise 4 poles of AR model can be detected at 12.2, 20.0, 31.3, 37.75 Hz. To observe the phase coupling phenomenon, a noticeable peak at (20.2, 10.68) Hz, marked with '*', can be seen from cross-bispectrum diagram of channel 5-4-5 in Fig. 1(c). This indicates that the peak at 31.3 Hz in channel 5 could result from the phase coupling of harmonics at 20.2 Hz of ch. 5 and 10.68 Hz of an adjacent EMG channel, ch. 4. Similarly, peaks in cross-bispectrum of channel 5-6-5, Fig. 1(d), harmonics of (20.0, 16.96) Hz of ch. 5 and ch. 6 could result in the peak at 37.75 Hz in channel 5. After eliminating the over-estimates from phase coupling, poles at 12.2 and 20.0 Hz in channel 5 are the detected firing rates. Following the same procedure, the time-varying firing rates of EMG from step or ramp isometrical contractions of normal, parkinsonism and stroke patients can be compared.

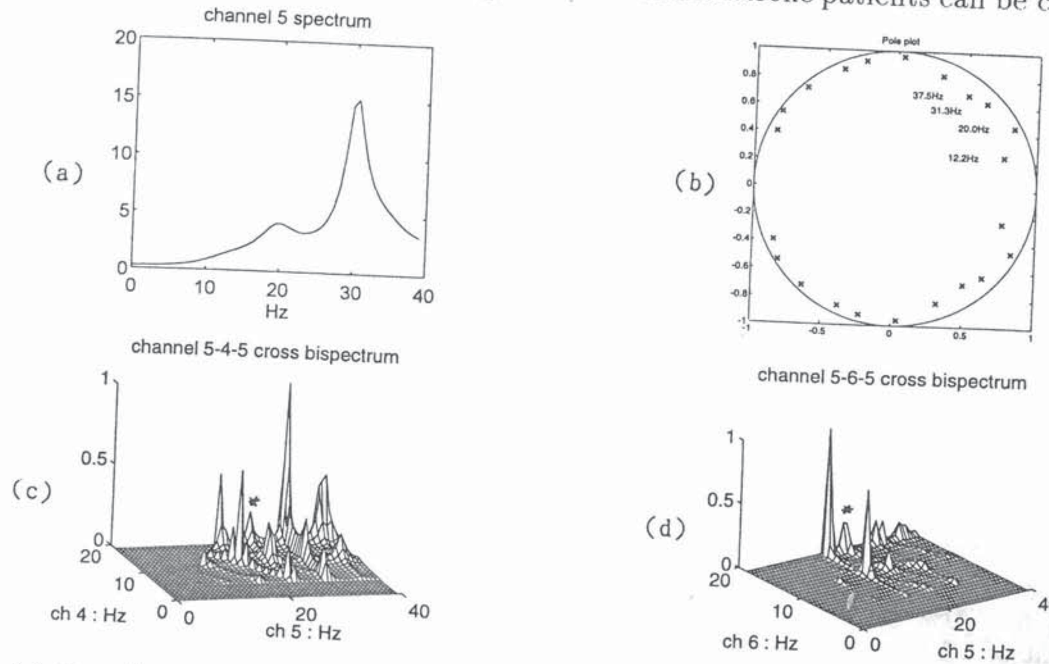


Fig. 1: (a) Power spectrum and (b) poles of EMG of ch. 5. The cross-bispectrum of (c) ch. 5-4-5 and of (d) ch. 5-6-5.

REFERENCES

1. A. Van Boxtel and R. B. Schomaker, "Motor unit firing rate during static contraction indicated by the surface EMG power spectrum," *IEEE Trans. Biomed. End.*, vol. 30, no. 9, pp. 601-608, 1983.

2. M. R. Raghuveer and C. L. Nikias, "Bispectrum estimation: A parametric approach," *IEEE Transactions on Acoustics, Speech, and Signal Processing,* vol. 33, no. 4, pp. 1213-1230, Oct. 1985.

3. C. L. Nikias and A. P. Petropuou, Higher-Order Spectral Analysis: A Nonlinear Signal Processing Framework, Prentice-Hall Inc., 1993.

RANK-ORDERED REGULATION OF MOTOR UNITS

Zeynep Erim*, Carlo J. De Luca*†, and Kiyoshi Mineo‡

*NeuroMuscular Research Center and †Departments of Biomedical Engineering and Neurology, Boston University, ‡ Department of Rehabilitation Medicine, Shizuoka City Hospital, Shizuoka, Japan

INTRODUCTION

This investigation was undertaken in order to explore the recruitment and firing rate modulation strategies that are used in force generation by the muscle. The recruitment and firing rate properties of the muscle were studied in the whole force range of the muscle. The conclusions of this study, augmented by earlier findings lead to the development of a model for the rank-ordered regulation of motor units.

METHODS

Myoelectric signals were detected from the Tibialis Anterior muscle of five healthy subjects with a special quadrifilar needle electrode during voluntary isometric contractions. Details of the experimental set-up are discussed in [1]. The subjects generated isometric linearly increasing (10% of maximal voluntary contraction / s) forces up to the maximal voluntary level. The acquired signals were decomposed into the constituent motor unit action potential trains and an accurate record of the interfiring intervals was obtained using Precision Decomposition [2].

The continuous *mean firing rate* signal for a motor unit was calculated by passing a 400 ms Hanning window over an impulse train corresponding to the firing times of that motor unit. Motor unit firing rates were investigated as a function of muscle force by plotting the average values (in a 1 s window) of the mean firing rate signals against corresponding averages of the force recording at 1 s intervals. The recruitment threshold for each motor unit was also calculated for the purposes of establishing any dependence between the control properties of recruitment and firing rate modulation.

RESULTS

The relationship between the force output of the muscle and the firing rates of individual motor units was found to contain three distinct contiguous regions. The first region, which began at the recruitment threshold of the motor unit lasted during an increase of 10 to 20 % of maximal voluntary contraction level, had a rapid increase in firing rates. The second region, which continued to 70% of the maximal voluntary contraction level marking the completion of recruitment of all the available motor units, displayed a slower increase in firing rates with increasing force. The last region, which covered the range where recruitment of new motor units was not possible and extended to maximal voluntary level, showed a sharp increase in firing rates.

The rate of change of firing rate was correlated to the recruitment threshold. Motor units with higher recruitment thresholds displayed greater rates of change . The dependence of gain on recruitment threshold was less pronounced in the second region of the activation profiles.

The firing rates of motor units at any instant were found to be ordered according to the recruitment order: at any given time in the contraction motor units that had lower recruitment thresholds had higher firing rates than units that had higher recruitment thresholds. This phenomenon provides the appearance of an "onion skin" to the plots of the firing rates. This rank-ordered organization is apparently not designed to maximize the force output of the muscle, because the higher-threshold motor units generally have faster force twitches and would require greater firing rates to fully tetanize. Instead, the control scheme may be organized to balance the force generating potential with the fatigue properties of the motor units.

The firing rates of all motor units identified in a contraction were observed to converge to the same value as the force approached maximal levels. This convergence property is consistent with the observations of a negative correlation between the recruitment threshold and firing rates at submaximal levels, and the positive correlation between slopes of the firing rate-force curves and recruitment thresholds.

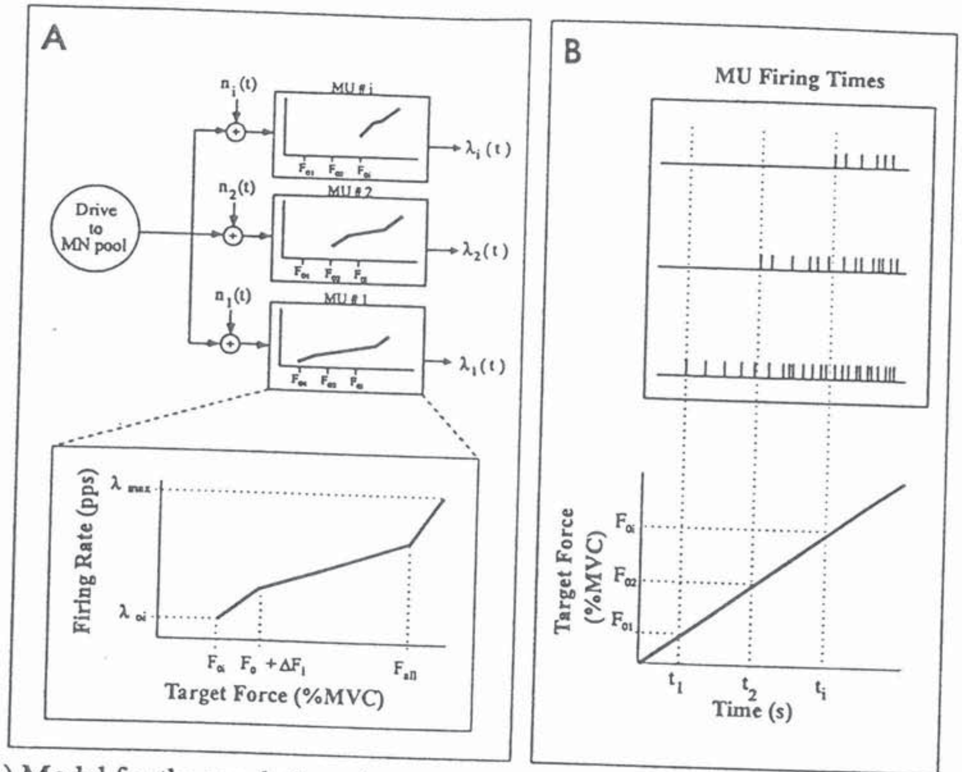


Figure 1: (A) Model for the regulation of recruitment and firing rates of motor units. Bottom panel displays a blow-up of the input-output curve describing the behavior of a motor unit as a function of the targeted force level, as revealed by the results of this study. (B) The response of three motor units with different recruitment thresholds when driven by a linearly increasing input.

A model is proposed to explain the mechanisms underlying the recruitment and firing rate modulation of motor units in the regulation of muscle force. The model employs the formerly advanced concept of *common drive* to the motoneuron pool [3], along with present findings. In this model, the firing activity of individual motor units are effected, not through separate command inputs, but through inherent differences in the way individual motor units respond to a common drive. Furthermore, the response of a given motor unit to this drive is determined by its recruitment rank, or equivalently its recruitment threshold. Such a systematic organization would relieve the central nervous system from the burden of keeping track of and controlling each motor unit separately, and provide a strategy whereby a single command could be employed to regulate the activity of all the motor units in a given motoneuron pool.

REFERENCES

1. Broman, H., De Luca, C. J., and Mambrito, B. "Motor Unit Recruitment and Firing Rates Interaction in the Control of Human Muscles." *Brain Research* 337:311-319, 1985.
2. De Luca, C. J. "Precision Decomposition of EMG Signals." *Methods in Clinical Neurophysiology* 4:1-28, 1993.
3. De Luca, C. J. "Control Properties of Motor Units. Evolving Concepts." In Computer-Aided Electromyography and Expert Systems, ed. J. E. Desmedt. 103-109. 2. Amsterdam: Elsevier, 1989.

AUTHOR'S ADDRESS

Zeynep Erim, Ph.D.
NeuroMuscular Research Center, 44 Cummington Street, Boston, Massachusetts 02215
Fax: (617) 353 5737; e-mail: erim@bunmrg.bu.edu

ACCURATE AND ROBUST ESTIMATION OF AVERAGE FIRING RATE OF SURFACE EMG INTERFERENCE SIGNAL.

Chris T.M. Baten, Hendrik J. Hamberg, Hermie J. Hermens,
Roessingh Research & Development BV, Enschede, Netherlands.

SUMMARY

A robust and accurate estimator for the average firing rate in voluntary contractions is introduced, based on surface EMG interference signal deconvolution. Testdata from EMG simulations and from normal subjects are presented to assess validity, reproducibility and sensitivity of the estimator.

INTRODUCTION

The two basic mechanisms for force regulation in skeletal muscle are recruitment of motor units and regulation of the individual motor unit firing rates [e.g. 5]. Recent papers report about clinically useful application of estimation of average firing rate (AFR) to assess disturbance of these force generation mechanisms [e.g. 3]. However the methods employed are not suitable for practical clinical use. More over their application is limited to contraction levels up to 30% MVC. This paper presents a robust and reliable estimation method for AFR based on spectral deconvolution of the surface EMG interference signal, which is not limited to low contraction levels and can be implemented on any regular PC system.

METHODS

The algorithm used for our AFR-estimator is based on the model representing the surface EMG signal as a summation of independent MUAP-trains, which result from convolution of a dirac impulse firing pattern and the surface Motor Unit Action Potential (MUAP) [6].
Basically estimation of AFR is performed by spectral deconvolution of the EMG interference signal with the estimated average MUAP power spectrum. Subsequently the resulting estimated firing pattern power spectrum were fitted with theoretically derived firing rate pattern power spectra (Summated Ten Hoopen functions, [4]).
Validity, reproducibility and sensitivity of the estimator are tested both with simulated and in vivo recorded EMG signals.

RESULTS

The estimator's quality is illustrated in Fig. 1 were the estimated AFR of 30 sets of 30 simulated EMG signals is shown as function of the exactly known AFR of the simulated EMG signal [2]. For the range of 6 to 20 Hz the standard deviation was less than 2 Hz. For AFR > 20 Hz a slight underestimation was observed (<5%).

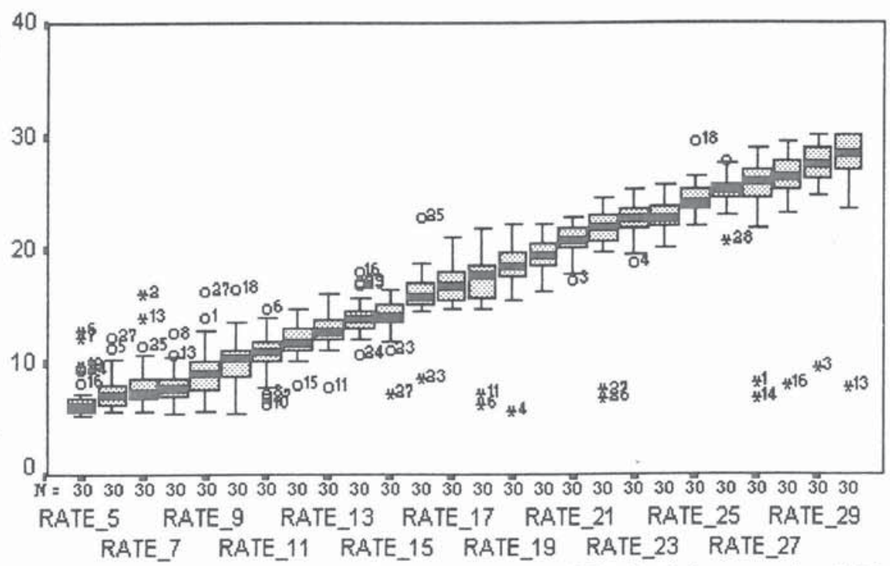


Figure 1: Estimated AFR as function of simulated AFR. Each box contains 50% of the 30 datapoints and the whiskers denote the last datapoint not being a outlier. * and o denote outliers.

In additional simulation runs the insensitivity of the AFR-estimator was found for MUAP duration, amount of variation in FR among the motor units and synchronization.
Application of the estimation method to surface EMG recordings of 10 normal subjects in 6 different muscles showed a weak positive dependency on contraction level in most recordings (Fig. 2) and a small but systematic decrease in all recordings during sustained contractions at 80% MVC (Fig. 3).

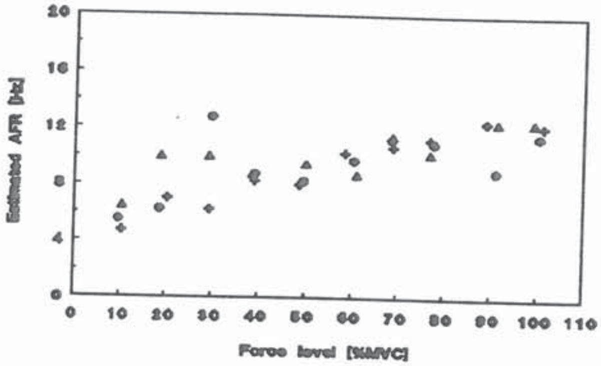


Figure 2: A typical estimated AFR against force level trace for the Rectus Femoris muscle.

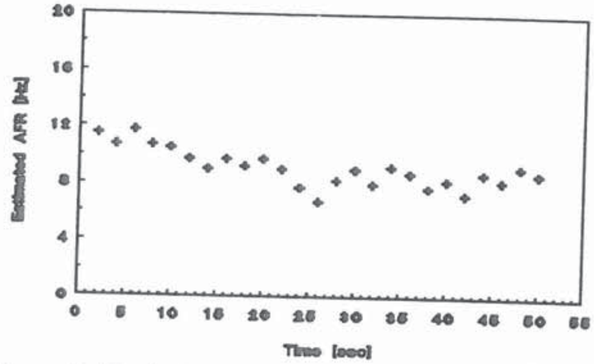


Figure 3: Typical estimated AFR against time for a sustained at 80% MVC in the Rectus Femoris muscle.

CONCLUSIONS

-In 10 normals in 6 muscles AFR appears to be only weakly sensitive to the contraction level. It should be noted that AFR is the weighted sum of the active motor units FR values and that it therefore reflects not only firing rate changes in already active motor units but also changes in recruited motor unit pool.

-In 10 normals in 6 muscles in 80% MVC level sustained contraction there is a small but marked decrease in AFR with time. Assuming no changes in recruited motor unit pool [1] this can entirely be attributed to changes in firing rates of active motor units.

REFERENCES

1. Baten C.T.M., Iren A.C.U.E., Hermens H.J., Rutten W.L.C., Boom H.B.K., 'The role of synchronization in localized muscle fatigue.', proc. 9Th Congres of the ISEK, 1992.
2. Hermens H.J, van Bruggen T.A.M., Baten C.T.M., Rutten W.L.C., Boom H.B.K., 'The median frequency of the surface EMG power spectrum in relation to motor unit firing and action potential properties.', J. Electrophysiol. Kinesiol., 1991.
3. Leslie J. Dorfman, Jane E. Howard, Kevin G. McGill, 'Motor unit firing rate variability in the detection of neuromuscular disorders.', Electroencephal. Clin, Neurophysiol.73, 215-224, 1989.
4. Ten Hoopen M., 'Examples of power spectra of univariate point processes', Kybernetik, 16, 145-154, 1974.
5. Solomonow M., Baten, C.T.M., Smit J., Baratta, R., Hermens H.J., D'Ambrosia R., Shoji H., 'Electromyogram power spectrum frequencies associated with motor unit recruitment strategies.', J. Appl. Physiol., 68 (3), 1177-1185, 1990.
6. De Luca C.J., 'Physiology and mathematics of myoelectric signals', IEEE Trans Biomed Eng, 26, 313-325, 1979.

AUTHOR'S ADDRESS

Chris T.M. Baten
Roessingh Research & Development BV,
Roessinghsbleekweg 33, 7522 AH Enschede, Netherlands
Phone: +31 53 875 734; Fax: +31 53 340849

KOHONEN MAPS FOR CLUSTERING MOTOR UNIT ACTION POTENTIALS

G. Balestra°, A. Capra•, E. Pasero•

°Dipartimento di Sistemi di Produzione ed Economia dell'Azienda, •Dipartimento di Elettronica, Politecnico di Torino, Torino, Italy

INTRODUCTION

The motor unit action potential (MUAP) clustering is an important phase of a decomposition algorithm. Several methods for identifying and clustering MUAPs are presented in literature; for a comprehensive review, the interested reader may refer to the papers referenced in [2]. This work presents a method based on self-organizing neural networks and describes its characteristics. This approach is appealing because it permits identification of the number of different MUAPs that may be convincingly recognized throughout a contraction.

METHODS

Self-organizing Kohonen maps [3] are neural networks based on the competitive learning module. In this category of nets, the neurons are organized like the elements of a square matrix. Each neuron is linked with all the others. Learning follows the Hebbian rule.

The feasibility of the proposed approach was evaluated by using a general purpose neural net simulator, and, since preliminary results were encouraging, we developed a specific Kohonen map simulator running on MS-DOS personal computers.

The clustering procedure works as follow: (1) the MUAPs are extracted from the signal by means of a threshold algorithm, then records are time windowed and padded with zeros to obtain a fixed number of samples; (2) a typical action potential is taken as a reference, and extracted MUAPs are aligned with it by means of a high resolution algorithm; (3) the aligned MUAPs are then used to train a specific map; (4) the nearest neighbor algorithm is applied to the trained map to select the principal templates.

This procedure was tested on simulated and real signals. The simulated signals were obtained by summing the contribution of up to 15 MUs by means of a myoelectric signal simulator described elsewhere [1]. The real signals were collected by means of quadripolar needle electrodes by our colleagues at the NeuroMuscular Research Center of Boston University.

RESULTS and CONCLUSIONS

After a large number of trials with maps of different dimensions, we concluded that 6x6 and 12x12 maps yield best results with acceptable learning time. By increasing the map dimension, the learning time increases and results worsen. Parameters of the clustering procedure were tuned by means of simulated signals in which the number of active motor units and their characteristics (firing rate, shape, conduction velocity variation modality) were chosen to reproduce the combinations most likely encountered when decomposing real signals.

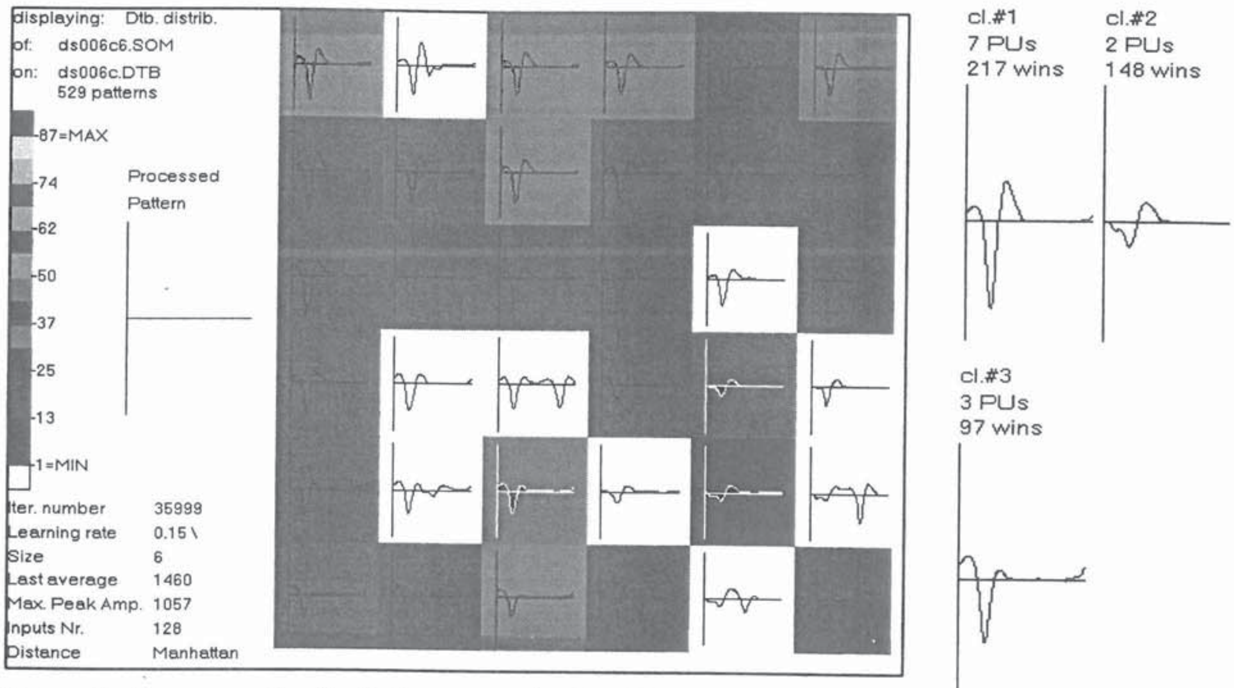


Figure 1: An example of a 6x6 maps trained with a real signal and the result of the clustering algorithm

REFERENCES

1. G. Balestra, M. Knaflitz, C.J. De Luca, "EMGGEN: a software package for myo-electric signal simulation designed for research and computer aided instruction", in *Electrophysiological kinesiology*, Eds: P. A. Anderson, D. J. Hobart and J. Danoff, Amsterdam: Elsevier Science Publisher, pp. 87-90, 1991.

2. J.E. Desmedt, *Computer-Aided Electromiography and Expert Systems*, Amsterdam: EElsevier Science Publishers, 1989.

3. T. Kohonen, "Self-organized formation of topologically correct feature maps", *Biological Cybernetics*, vol. 43, pp. 59-69, 1982

ACKNOWLEDGMENTS

The authors gratefully acknowledge the cooperation of Dr. C.J. De Luca, director of the NeuroMuscular Research Center of Boston University, who provided the real signals used to test the algorithm.

AUTHOR'S ADDRESS

Gabriella Balestra, Ph.D.
Dipartimento di Sistemi di Produzione ed Economia dell'Azienda
Politecnico di Torino
Corso Duca degli Abruzzi 29
10129 Torino -Italy
E-mail: BALESTRA@POLITO.IT

REPRESENTATION OF MOTOR UNITS IN THE SURFACE ELECTROMYOGRAM

K. Roeleveld, D.F. Stegeman, H.M. Vingerhoets, A. v. Oosterom
University Hospital Nijmegen, Inst. of Neurology, Dept. of Clinical Neurophysiology,
Nijmegen, The Netherlands.

INTRODUCTION

Surface electromyographical (SEMG) techniques are mostly used to quantify muscle activity in isometric situations, to investigate the timing of muscles under dynamic conditions and to detect muscle fatigue. When information is needed about the morphology or structure of the muscle or parts of the muscle, intramuscular needle emg (NEMG) is used, since surface electrodes are expected to be insufficiently selective. However, the propagation of motor unit action potentials (MUAPs) can easily be detected with arrays of SEMG electrodes (2,3). The observed change in shape of the MUAPs during propagation can be explained in terms of dispersion in motor endplate position and excitation, and a variation in conduction velocities. The present study further explores the representation of motor units in SEMG.

METHOD

The potential distribution of 30 motor units from the m. biceps brachii (BB) in nine healthy subjects, aged 20-50 years were collected. Also 20 motor units from the m. vastus medialis (VM) in five healthy subjects, aged 20-30 years were obtained. The potential field distribution of the active motor units at the skin surface was obtained with triggered averaging of the MUAPs, recorded at 30 monopolar skin surface electrodes. The surface electrodes were placed over the muscle with electrode arrays both in the direction of and perpendicular to the muscle fibres. Inter-electrode distance was 6 mm. The EMG signals were collected during voluntary, low force, isometric contractions. To explain the experimentally obtained MUAPs an electrophysiological model (1) was used .

RESULTS

Fig.1 shows simulated SFAPs and typical MUAPs measured from the VM and the BB, respectively. Each plot contains recordings of 6 electrodes, placed at sequential sites in the direction of the muscle fibres. In all measured MUAPs a mainly negative propagating part and non-propagating parts could be distinguished. In the model simulations of SFAPs these components were brought about by an intracellular action potential originating at the motor endplate, propagating along a muscle fibre and colliding at the muscle-tendon-transition. Except for the duration of the components, the MUAPs from the VM shared the main elements of the SFAP model simulations. However, most BB MUAPs were more complicated. To clarify the waveshapes, the propagating parts were separated from the non-propagating parts by taking the spatial derivative of the MUAPs (bipolar reconstruction) and to subsequently integrate the signals. The results of this manipulation are shown in fig.2. This view of the BB's MUAPs makes clear that the propagating parts of that motor unit exist of two fibre populations. Fig.3 shows the differences between the measured MUAPs and the propagating parts of fig.2 as an estimation for the non-propagating parts. These parts include the initiation of the potentials and their collision at the muscle-tendon-transition.

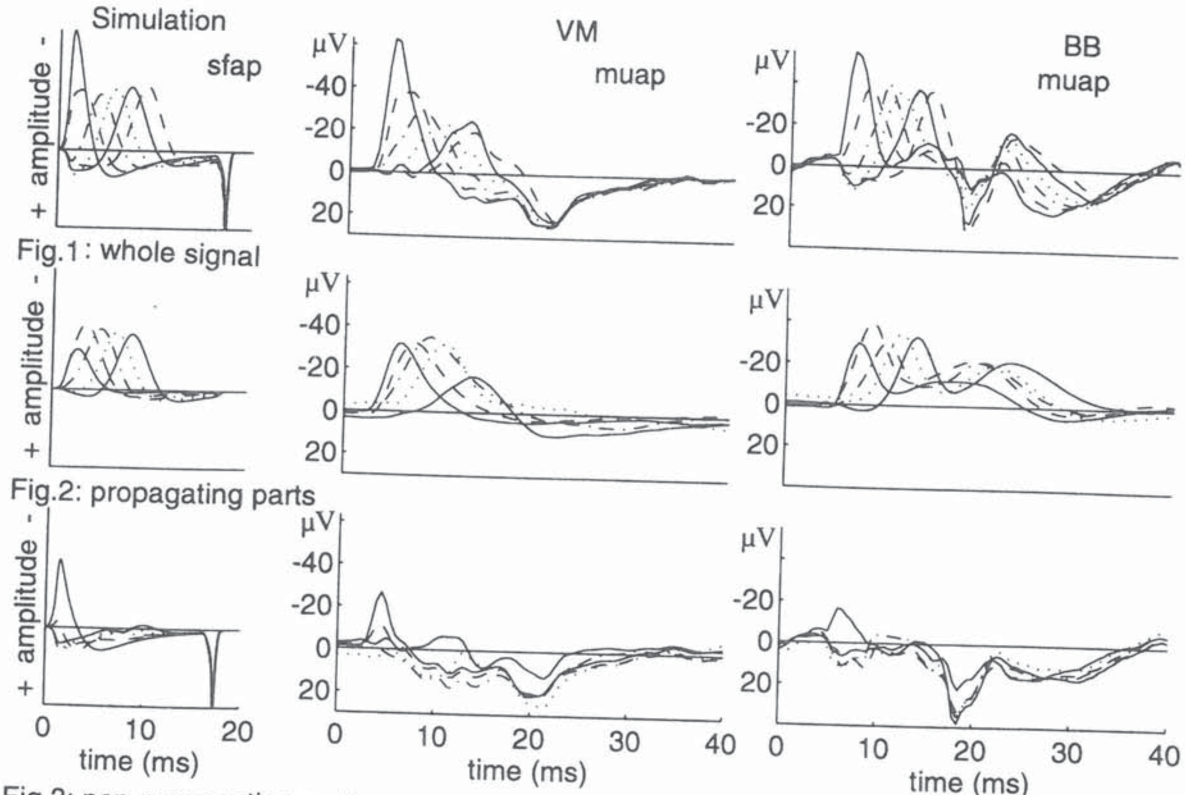


Fig.1 : whole signal

Fig.2: propagating parts

Fig.3: non-propagating parts

DISCUSSION

The SFAP simulations illustrate the basic components of the measured MUAPs. MUAPs are expected to be composed of such SFAPs with a certain temporal difference in the individual fibres belonging to the same motor unit. For the propagating potential, these temporal differences can be caused by a wide spreaded motor endplate region, a dispersion in motor endplate excitation, and a variation in fibre diameter. When electrodes are placed on both sides of the motor endplate region, the motor unit characteristics predominantly responsible for these temporal dispersions can be estimated. The variation of moments of collision of the action potentials at the muscle-tendon-transition is caused by these same parameters and by a variation in the spread out of this transition region. Combined with the conduction velocity, the occurrences of the end-effects relative to the initiation at the motor endplate give an estimation of the variable fibre lengths. Future work will be addressed to the validation and quantification of these concepts.

REFERENCES

1.Gootzen,T., Stegeman,D., v.Oosterom, A.(1991) Electroenc.clin.Neurophysiol.,81:152-162.
2.Gydikov,A., Kosarov,D. (1972) Electromyograph., 12(2):127-147.
3.Masuda,T., Sadoyama,T. (1986) Electr.clin.Neurophysiol., 63:590-598.

AUTHOR'S ADDRESS

K. Roeleveld, University Hospital Nijmegen, Inst. of Neurology, Dept of Clinical Neurophysiology, Postbus 9101, 6500 HB Nijmegen, The Netherlands.
Phone: 080-615283, Fax: 080-541122, Email: NEURO_KR@AZNVX1.AZN.NL

MOTOR UNIT RECRUITMENT STRATEGY OF THIGH ANTAGONIST MUSCLES IN STEP-WISE AND RAMP INCREASING ISOMETRIC CONTRACTIONS

M. Bernardi[1,2], M. Solomonow[1], R.V. Baratta[1]
[1] Bioengineering Laboratory, Dept. of Orthopaedic Surgery, Louisiana State University Medical Center, New Orleans, LA, U.S.A. [2] Istituto di Fisiologia Umana, Universita' "La Sapienza", Rome, Italy.

INTRODUCTION

The study of motor unit recruitment during muscular coactivation is of relevant importance in human physiology, clinical work and in rehabilitation. While it is commonly accepted (Petajan, 1991) that, when a muscle acts as agonist, motor units are recruited according to Henneman's "size principle" (1957), the literature regarding motor unit recruitment when a muscle acts as antagonist is very scarce. Based on a mathematical model which relates the surface electromyographic power density spectrum (PDS) with the conduction velocity of the action potentials in the muscle (Lindstrom et al., 1970), Solomonow et al. (1990) studied the contributions of motor unit recruitment and firing rate to the median frequency (MF), the frequency that divides the area under the PDS into two regions of equal power. The authors found that the increasing average muscle fiber conduction velocity during recruitment is the major contributor to variations in the MF, whereas changes in the firing rate have only negligible effects. The MF will be used in this study as a tracking parameter to describe the motor unit recruitment strategy of quadriceps and hamstrings during knee flexion and extension. The purpose is to highlight possible differences in muscle recruitment according to the role (agonist/antagonist), the motor task executed (flexion/extension) and the kind of isometric contraction performed (stepwise/linearly increasing contractions).

METHODS and RESULTS

Quadriceps and hamstrings surface myoelectric signals (MES) were collected at the same time during knee flexion and extension as performed by 18 volunteers, both in step-wise increasing isometric contractions (steps) and linearly increasing isometric contractions (ramps). Each step consisted of a trial of 2 second duration, at each 10% force level increment from 10% to 100% maximal voluntary contraction (MVC). In the ramps the force increased from 0% to 100% MVC in 3 seconds. The Fast Fourier Transformation was applied on the stored MES, the PDSs were obtained and the MFs were calculated. In both experiments mean MF values and standard deviations were obtained pooling together the data of the subjects. A first order polynomial was used to fit the data up to the force level in which the MF trend changed direction. The F test was utilized to evaluate the linearity of the data trend. MF mean values, standard deviations and regressions lines of all the data are shown in Figure 1. In all the conditions studied the MF showed an increment up to a certain value. The increment was demonstrated to be linear by the regression evaluation. In the stepwise experiments the mean MF trend was observed to decrease in all the conditions studied after the MF peak value. In the ramp experiment, both in extension and in flexion, when the muscles acted as agonist after the linear increment the MF stayed relatively stable with slight increments at the highest force levels. When the muscles acted as antagonist, the MF trend showed, in the quadriceps (flexion), a linear increment up to 40% MVC and then a drastic decrement. In comparison the MF trend for the hamstrings (extension) showed the peak value at higher force level and then a slight decrease.

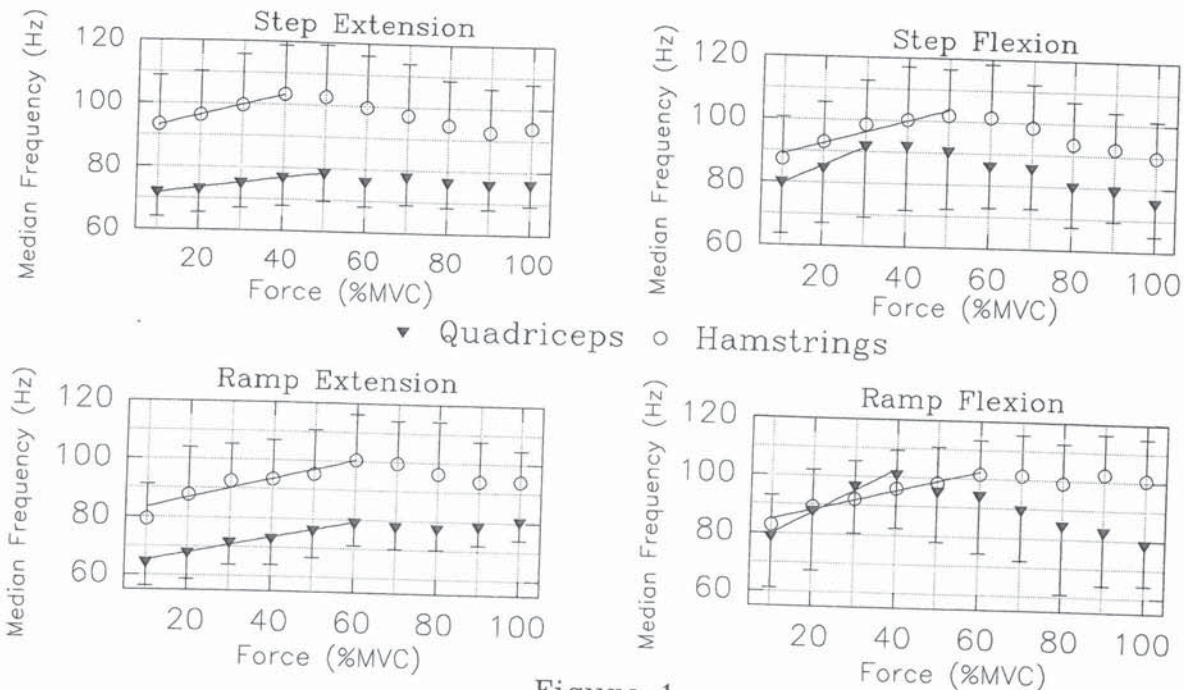


Figure 1

DISCUSSION and CONCLUSIONS

The force range in which the MF increases linearly is the range in which the most significant MF increment occurs. This suggests that at these force levels motor unit recruitment is the major mechanism for the muscle to increase its force. When the MF stops increasing, the firing rate becomes the most relevant factor in obtaining higher force. Differences were demonstrated in the motor unit recruitment control strategy in the coactivation of the thigh muscles depending on the role acted, the motor task and the kind of contraction performed. The motor units are recruited up to higher force levels in the muscle acting as agonist, but the "size principle" is still valid in the antagonist motor unit recruitment. The role of the hamstrings as antagonist (extension) is more relevant than the quadriceps acting the same role both in the steps and in the ramps. The recruitment process seem to be slower in the ramp contractions: the full recruitment for the agonist and the motor unit activation for the antagonist is obtained at higher force levels than in the steps. In conclusion, if a strict quality control of the data acquisition process is applied and the data within subject and among subjects are pooled together, the MF is a useful parameter to study the relationship between force and recruitment control strategy in whole muscle.

REFERENCES

Henneman E: Science 126: 1345-1346, 1957.

Solomonow M, Baten C, Smit J, Baratta R, Hermens H, D'Ambrosia R, Shoji H: J. Appl. Physiol. 68 (3): 1177-1185, 1990.

Lindstrom L.R., Magnusson R., Petersen I.: Electromyography 4: 341-353, 1970.

Petajan J.H.: Muscle and Nerve 14: 489-502, 1991

AUTHOR'S ADDRESS

M. Bernardi, M.D. Dept. of Orthopaedic Surgery, Louisiana State University Medical Center, 2025 Gravier Street, Suite 400, New Orleans, LA, U.S.A. Phone: (504) 5682251. Fax: 5684466

APPLICATION OF THE METHOD OF SPECTRAL-STATISTICAL ANALYSIS EMG IN THE OBJECTIVE DIAGNOSIS AND FOR EVALUATING THE INFLUENCE OF MEDICAL PREPARATIONS IN THE TREATMENT OF PARKINSON'S DISEASE

O. E. Khutorskaya

Institute of Control Sciences of the Russian Academy of Sciences Moscow , Russia.

SUMMARY

The method of Spectral-Statistical analysis of the electrical activity of muscles (EMG), has been developed in the Institute of Control Sciences, the Russian Academy of Sciences, used for the study of physiological and pathological tremors, was described [1]. The application of the method in neurological clinics allowed to objectively demonstrate the diagnostic symptoms of certain forms of Parkinson's Disease. The results of using the method in the evaluation of the influence of single pharmacological tests for Parkinson's Disease are described in this article.

INTRODUCTION

In this article, new symptoms are considered, which are obtained with the aid of this method and which allow the authors to conduct objective diagnosis of certain forms of Parkinson's Disease. The range values of each of the diagnostical indicators were computed for three symptoms of Parkinson's Disease: trembling, rigidity and akinesia. Analysis of the influence of the medicine is carried out by varying the diagnostic indicators, obtained through the Spectral -Statistical method for each muscle in both the "on" and "off" periods.

METHOD

With four muscles (extensor carpi radialis longus and tibialis anterior of both right and left sides) integrated EMGs are introduced into the computer. EMG are detected and filtered in the 0-15 Hz. range, thus forming the rounded EMG signals (i. e. EEMG) Spectral Analyses of 20 one-minute realizations of EEMG signals are conducted for each muscle. Diagnostic symptoms were obtained on the basis of a statistical treatment of the spectral parameters, from all the different spectra is conducted. The indicators applied for studying 120 patients diagnosed with Parkinson's Disease of different etiologies, forms and stages. The ages of the subjects vary from 35 to 80 years old with the continuation of the illness stretches from 2 to 15 years

The study sample, during the period of the pharmacologic test, consisted of 38 patients in the early stages of Parkinson's Disease. The study is carried out before taking the medicine (in the "off" mode) and 40-60 minutes after taking the medicine (in the "on" mode), when the effect of the medicine is at its highest. Patients are prescribed a one-time, habitual, individualized doses of anti-Parkinson's medicine anticholinergic and dopaminergic drugs.

RESULTS

In a researched group, in 113 out 120 cases, clinical diagnosis coincided with computer one. In six cases, doctors corrected the disgnosis of the illness with diagnostic indicators. In one case, computer and clinical diagnosis did not coincide.

Analysis of the materials show that in the "on" period, in 23% of the cases, for the group of muscles under study, change in the tremor frequency in the direction of normality was observed. In 2% of the cases, reduction in the magnitude of tremor was observed. In 10% of the cases, both frequency and magnitude of tremors were registered. In 65% of the cases, no change whatsoever was indicated. For those patients for which the effectiveness of the medicine was manifested independently in certain groups of muscles, corrective medical therapy was recommended.

CONCLUSIONS

The method developed here offers the possibility to objectively characterize in some way the symptoms for each muscles under study. Consequently, this allows us to objective the disgnosis of forms of Parkinson's Disease. Preliminary studies of patients, by conducting single pharmacological tests with anti-Parkinson's medications, showed that it is possible to use the Spectral-Statistical analysis EMG in the following applications: 1. For individualized selection of medication for mono-anti-Parkinson's or combined therapy.
2. For the evaluating the effectiveness of anti-Parkinson's medication.

REFERENCES

1. Andreeva Ye. A., Ivanova-Smolenskaya I. A., Kandel E. I., Khutorskaya O. Ye. Envelope EMG spectral analysis in the studies of fhysiological and pathological tremor. J. Eiectromyography and clinical neurophysiology. 25, 5, 273-293, 1985

ELECTROMYOGRAPHIC POWER SPECTRUM ANALYSES FROM DYNAMIC AND STATIC CONTRACTIONS

H. Christensen, K. Søgaard, B.R. Jensen, L. Finsen, G. Sjøgaard
Dept. of Physiology, National Institute of Occupational Health, Copenhagen, Denmark

INTRODUCTION

An increase in the RMS-amplitude and a decrease in the mean power frequency (MPF), or median power frequency (MF) of the EMG signal have been used as indicators of muscle fatigue during sustained contractions. However, the amplitude and the power spectrum are not only modulated during development of fatigue but have been shown to change with changes in the force level and muscle fibres length occurring during dynamic contractions. Consequently, development of muscle fatigue during occupational work has been studied during standardized submaximal static test contractions, performed before and after occupational work tasks. The purpose of this study was to evaluate if EMG recordings during occupational work tasks which usually include dynamic contractions can be used to register muscle fatigue caused by the work. Therefore the aim was to evaluate changes in the EMG variables recorded from a non fatigued muscle during a slow dynamic contraction compared to a static contraction. Further the aim was to elucidate the findings from the surface EMG by analyses of the recruitment pattern during dynamic and static submaximal contractions.

METHODS

Three females participated (age 37-41 years). The subjects were sitting with a 90° shoulder flexion, the left upper arm resting horizontally, the elbow flexed and the wrist supinated. The subject pulled against a custombuild strain-gauge ring attached in one side to a cuff which encircled the wrist and in the other side to a 2 kg load. Two types of contractions were performed 3 times each in randomised order: (I): A 7°/s elbow flexion/extension starting from 80° up to 100° and back to 80°. (II): A static isotonic contraction holding the load against gravity with a 90° flexion in the elbow. Bipolar surface electrodes were placed on the skin above the long head of the left brachial biceps muscle. For intramuscular recordings from the long head of the biceps muscle a quadripolar needle was used. In the dynamic contraction (I) the EMG signal was analyzed for 3 s during the concentric phase and for 3 s in the eccentric phase, corresponding to isotonic contractions. During the static isotonic contraction (II) 10 s of recording was analyzed. The intramuscular registrations were decomposed into trains of single motor unit action potentials (MUAPs) using the computer algorithm called Precision Decomposition Program (Stashuk, De Luca, 1989). For each train of MUAP the interspike intervals (IPIs) were registered and mean firing rate (MFR) were calculated. Only trains with more than 6 firings were included in the results.

RESULTS

The 2 kg load corresponded to as a mean 9.4 (8.1 - 10.1) % MVC. The RMS amplitudes, the MPF and the MF from the power spectrum, during the concentric phase, eccentric phase and static isotonic contraction are presented in the table. During the dynamic contraction a total of 45 motor units fulfilled the criteria of 6 or more firings and the mean firing rate, MFR, for these motor units was 14.2 (5.8-29.6) Hz. Activity in both the concentric phase and the eccentric phase were recorded for 27 motor units (60%), while 10 motor units (22%) were active in the concentric phase without being active in the eccentric phase. MFR for these 10 motor units was significantly higher compared to MFR for the 8 motor units (18%) active only during the eccentric phase. During the static isotonic contractions a total of 53 motor units were analyzed. Among these 72% showed a continuous firing pattern.

	RMS(μV)	MPF(Hz)	MF(Hz)	MFR(Hz)	MFR(Hz)
Concentric (3 s)	50±21.1	61±13.4	48±4.8	12.5(27)	15.4(10)
Eccentric (3 s)	46±21.0	61±14.5	45±4.7	11.0(27)	10.0(8)
Static isotonic (10 s)	32±17.3	69±7.9	51±5.2	11.7(53)	-

DISCUSSION AND CONCLUSION

A higher MFR was seen for the motor units which were only active during the concentric phase, compared to the MFR from the motor units only active during the eccentric phase for approximately the same number of active motor units. However changes in firing rate have no or only minor influence on the power spectrum, while changes in shape and duration of the active motor units influence the power spectrum significantly. Since, the main part of the identified motor units (60 %) were active during both periods, shape and duration of these action potentials were the same. Further, those motor units recruited during only the concentric or eccentric phase may also be recruited from the same low-threshold motor unit pool. This may explain that there were no differences in MPF, MF or RMS amplitude between the two phase in the dynamic contraction, or between the dynamic and static contraction at the same low force level. Many occupational work tasks are composed of slightly dynamic movements, often with a high degree of repetitiveness performed in a stereotype pattern of movement. We suggest that in occupational studies concerning evaluation of muscle activity during repetitive monotonous work at low force including slow dynamic contractions, a possible muscle fatigue with time can be tested using EMG recordings from the workcycles per se.

REFERENCES

Stashuk D, De Luca CJ (1989) Update on the decomposition and analysis of EMG signals. In: Desmedt JE (ed) Computer-aided electromyography and expert systems. Elsevier, Amsterdam, New York, Oxford, pp 39-54

AUTHOR'S ADDRESS

Hanne Christensen,Ph.D.
National Institute of Occupational Health
Department of Physiology
Lersø Parkallé 105
DK-2100 Copenhagen
Denmark
Tel: +45 39 29 97 11
Fax: +45 39 27 01 07
E-mail: amilpost@uts.uni-c.dk

THE EFFECT OF ANKLE JOINT VISCOELASTIC PROPERTIES ON THE DYNAMIC RESPONSE OF THE CAT'S MEDIAL GASTROCNEMIUS MUSCLE

B-H. Zhou, R.V. Baratta, M.Solomonow, R. D'Ambrosia
Bioengineering Laboratory, Department of Orthopaedic Surgery,
Louisiana State University Medical Center, New Orleans, Louisiana, U.S.A.

SUMMARY

The effect of the cat ankle joint passive viscoelastic properties on the dynamic response of the Medial Gastrocnemius (MG) was examined under isometric and load-moving conditions. Frequency response models were constructed of the muscle-joint system during sinusoidal isometric force oscillations and constant-mass lifting movements. These responses were compared against those obtained under the same loading condition after separating the muscle from its distal joint insertion. It was found that in isometric conditions, the muscle could be modeled as a critically damped second order system with a pure time delay, with the joint adding a pole-zero combination which reduced the system bandwidth. In load-moving conditions, two sets of double poles, two zeroes and a pure time delay were used to model the muscle, with the joint adding a pole-zero combination affecting the phase with no significant influence in the gain. It was concluded that the joint introduces dynamic effects to the muscle's response, which vary depending on the type of loading experienced by the joint.

INTRODUCTION

The dynamic behavior of the neuromusculoskeletal system is a manifestation of the dynamic properties of all of its components. Muscle, tendon and reflex behavior have been extensively studied as subsystems (1,2,3), but critical information is lacking as to the effect of the joint on the overall musculoskeletal system response to electrical stimulation. The joint plays an important role in transducing changes in muscle length to angle and muscle force to torque. Its components of non-linear and viscoelastc nature such as ligaments, articular cartilage, distal limb inertia and fluid filled joint capsule give rise to the possibility of significant modifications to the muscle's dynamic output which may have to be addressed in the design of systems intended to restore limb function via Functional Electrical Stimulation. In addition, because of the displacement of fluid within the collagen matrix of articular cartilage, it can be expected that these properties may have different effects under different loading conditions.

It is therefore the intent of this study to examine the influence of the ankle joint's properties on the dynamic response of the muscle-joint system. Isometric conditions are used to cyclically load a fixed area of the joint, and mass-moving conditions are used to provide a constant load that moves cyclically across the joint surface, simulating the movement of a loaded limb.

METHODS AND RESULTS

Eight adult cats, anesthetized with chloralose were used. Their hind limbs were prepared exposing the sciatic nerve, denervating all muscles except the MG, applying a tripolar stimulation electrode and placing a pin across the femoral condyles for rigid fixation. The foot was placed on a rotating armature designed to record isometric force at 90° of plantar flexion or joint angular displacement against a 0.2 Nm plantar flexion moment provided by a 1 kg weight suspended from a 2 cm pulley. Frequency response studies were performed under sinusoidal recruitment-derecruitment of motor units concurrent with increase-decrease in firing rate over the frequency range of 0.4 to 5.0 Hz. The calcaneal tendon was then severed and the frequency response study repeated with direct connection of the tendon to the isometric transducer or to the weight-pulley system. Gain and phase data were calculated using Fast Fourier Transforms, pooled and plotted in conventional Bode plots. ANOVA with repeated measures was used to detect significant differences between the gain and phase data corresponding to with or without tendon under similar loading conditions, and linear models consisting of pole-zero combinations and a pure time delay were fitted to each

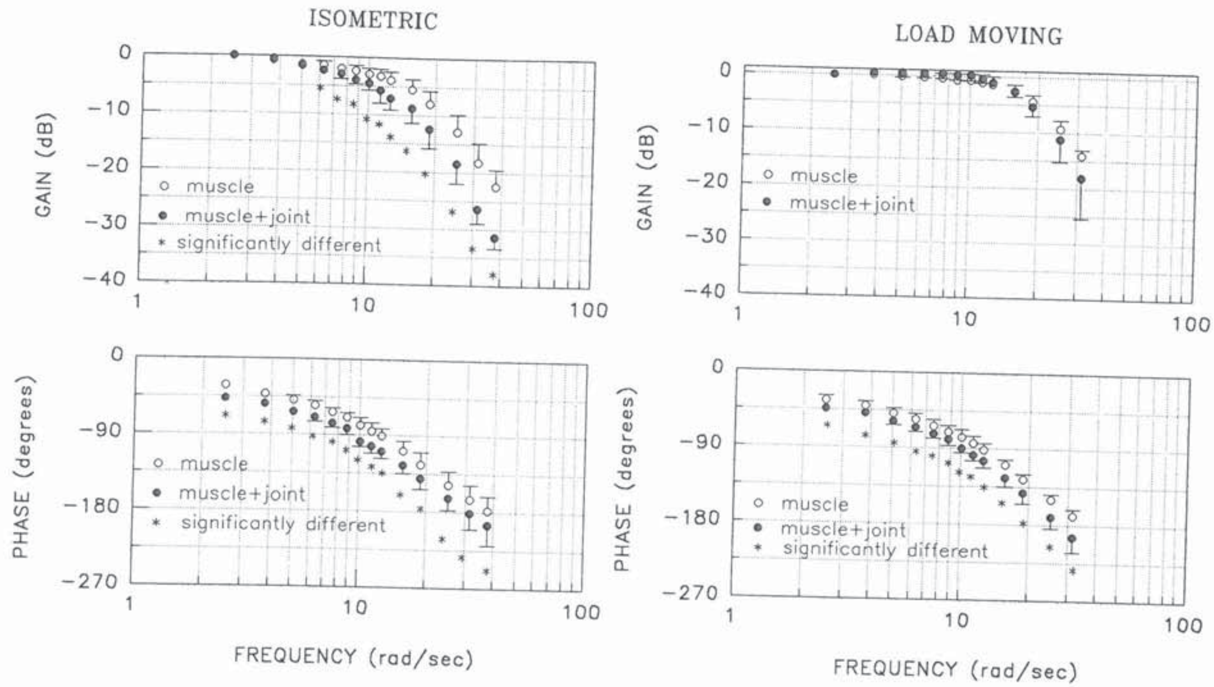


Figure 1

condition. Figure 1 shows the pooled results of gain and phase in isometric and load-moving conditions with the best fit models obtained through least squares. Asterisks indicate data pairs with significant difference ($p < 0.05$) between with-joint and without-joint conditions.

DISCUSSION AND CONCLUSIONS

The results of this study show that the joint introduces significant modifications to the dynamic response of muscle under both isometric and load moving conditions. In isometric conditions, the joint results in increased phase lag and decreased amplitudes at higher frequencies, while in load moving conditions, the results is increased phase lag without a concomitant decrease in amplitude bandwidth, thus behaving as a lag-lead network. These differences are probably caused by the different loading of the articular cartilage. In isometric conditions, a constant area is loaded cyclically, whereas in load-moving conditions, the area of loading varies throughout each oscillation cycle, resulting in different fluid flow patterns within the collagen matrix. This dynamic effects must be considered in the design of FES system, for they reduce the feedback loop stability margins.

REFERENCES

1. Baratta et al., *IEEE Trans-BME,* 37:243-251, 1990.
2. Roeleveld et al., *J. Appl. Physiol.*, 74:232-241, 1993
3. Rosenthal et al., *J. Neurophysiol*, 33:713-749, 1970.

Supported by the National Science Foundation with Grant BCS-9207007

AUTHOR'S ADDRESS

Richard V. Baratta, Ph.D., Louisiana State University Medical Center, Department of Orthopaedic Surgery, 2025 Gravier Street, Suite 400, New Orleans, Louisiana 70112. Tel. (504)-568-2251, Fax. (504)-568-4466, EMAIL : RBARAT@NOVMS.LSUMC.EDU

LIGAMENTO-MUSCULAR REFLEX ARC IN THE SHOULDER

M. Solomonow, C. Guanche, T. Knatt & R. Baratta
Bioengineering Laboratory, Dept. of Orthopaedic Surgery, Louisiana State University Medical Center, New Orleans, LA, 70112

SUMMARY

A reflex arc originating from mechano-receptors in the shoulder capsule and mediated via sensory branches of the axillary and musculocutaneous nerves to the spine and then to the biceps, subscapularis, supraspinatus, infraspinatus and acromiodeltoid muscles in the feline model was demonstrated. The reflex was elicited by electrical stimulation of anterior, inferior and posterior axillary articular nerves (AAAN, IAAN and PAAN, respectively) and the musculocutaneous articular nerve (MAN), while recording M-waves in the muscles via wire electrodes. Dissection of the articular nerves proximal to the stimulating electrodes verified the sensory origin of the reflex. It was concluded the strains and stress of ligaments within the shoulder capsule relatively elicit muscular activity with the objective of preserving joint stability.

INTRODUCTION

While the ligaments are the major structures responsible for joint stability, recent work (1), confirms that a reflex arc exists from afferents in the knee ligaments to muscles crossing the joint. The reflex was shown to be active upon stress in the ligaments. To date, it was not verified whether such reflex activity is unique to the knee or is a general phenomena existing in other joints.

The objective of this study, is to determine if such ligamento-muscular protective reflex exists in the shoulder.

METHODS AND RESULTS

Six adult cats were anesthetized with chloralose, and brachial plexus was exposed with anterior incision. Three axillary articular nerves terminating in the anterior, posterior and inferior shoulder capsule were identified (2). One musculocutaneous articular nerve was similarly identified.

Bipolar hook electrodes were placed under the articular nerves, one at a time, and supramaximal stimulation at 10 Hz was applied. M-wave activity was recorded from the biceps, supraspinatus, infraspinatus, subscapularis and acromiodeltoid with two fine wire electrodes inserted in each muscle. M-wave activity was amplified by differential amplifiers and recorded via data acquisition card on an IBM-PC computer.

To confirm the sensory origin of the M-waves, a fine string was tied around the nerve proximal to the electrodes. The nerve was then cut, and stimulation repeated. Lack of M-wave activity indicated that stimuli pulses from the articular nerves did not excite motor units in the muscles.

M-wave activity was recorded in the five muscles mentioned above before the articular nerves were dissected. M-wave activity did not exist after dissecting the nerves, confirming the existence of a ligamento-muscular reflex via several nerves.

DISCUSSION

The existence of a ligamento-muscular reflex arc in the shoulder joint confirms and generalizes the concept that ligaments and muscles are jointly and synergistically responsible for maintaining the integrity of a joint in the face of forces that by virtue of their magnitude and direction may disrupt and damage the anatomical structures. The fact that muscles can reduce the displacement of the bones of a joint relative to each other and bring them closer to their normal anatomical resting position was already confirmed for the knee and the shoulder (3,4).

It is also suggested that the objective of muscular co-contraction, often observed in many agonist-antagonist pairs, is in part to maintain joint stability.

REFERENCES

1. Solomonow, et al. (1987) Am. J. Sports Med., 15:207-213.
2. Wrete (1949) Acta Anatomica,7:173-190.
3. Rodosky, et al. (1990) Trans. ORS, 15:276.
4. Hirokawa, et al. (1991) J. EMG & Kinesiology, 1:199-208.

AUTHOR'S ADDRESS

Moshe Solomonow, Ph.D.
Louisiana State University Medical Center
Department of Orthopaedic Surgery
2025 Gravier Street, Suite 400
New Orleans, LA 70112
U.S.A.
Phone: (504) 568-2251
Fax: (504) 568-4466

MECHANICAL AND ANATOMICAL EQUILIBRIUM IN CYCLISTS

J.P. Clarys, E. Zinzen, P. Van Roy
Dep. Experimental Anatomy, Vrije Universiteit Brussel - Laarbeeklaan 103, 1090 Brussel, Belgium

SUMMARY

A cyclist's leg makes a simple 360° circular and rhythmic movement, activated by a simple flexion-extension function in a sagittal plane. However, because of the simultaneous rotation in the hip, knee and ankle joints with translation of the upper body, the general motion becomes quite complex (Hay, 1985).

The muscular equilibrium of the knee and hip flexors and extensors is therefore a conditio sine qua non for a good performance.

How are these flexors and extensors activated during cycling and how are the muscular equilibria distributed in the lower limb ?

METHODS AND RESULTS

11 semi-professional Belgian cyclists are tested on a isokinetic device (Kin/Com). The knee-extensors and flexors were tested bilateral on three different velocities (60°/s, 120°/s and 180°/s) in a concentric and eccentric mode with a range of motion of 90° (90° flexion > 180° extension).

The cyclists were fixed on the machine with a strap around the trunk, the hip and the upper-part of the leg. The inclination between trunk and femur was 120°.

Muscular intensity (IEMG) data were used from Clarys et al. 1988 and Antonis et al. 1989.

The activity of six lower limb muscles was measured under field circumstances using a portable EMG data acquisition system and active surface electrodes allowing remote (non-telemetric) monitoring of the cyclists' muscle activity patterns. Since the propulsion does not continue up to 180°, the flexor muscles have to be activated before the extension activity ends in order to generate the continuation of the circular motion until (and beyond) the bottom dead centre (180°) (Clarys et al. 1988).

Therefore EMG patterns of track cycling at submaximal (tempo-) and maximal (sprint-) velocity and/of road cycling both at sea level (0%) and steep up hill (20%) allowed for verfication of the equilibrium model.

1) comparison of the developed maximal torque values at 60°/s with a similar study in Australia (Telford, 1990) showed approximately the same results (table 1). We assume that the Belgian semi-professional cyclists are not different (as to force development) from other cyclists.

Table 1 : Maximal Torque comparison between Australian and Belgian road cyclists

Angular velocity	Maximum Torque in 17 Australian national training roas cyclists*	Maximum Torque in 11 Belgian semi-professional road cyclists
60°/s	191.0 Nm ± 26.7	204.1 Nm ± 20.4

2) Left/right comparison of both muscle groups indicated an almost perfect equilibrium between both body sites.
3) The flexion/extension (H/Q) ratio showed mean values of ± 80% for the eccentric mode. In comparison e.g. to football players (Davies, 1985), our values are about 20% higher, which means that the hamstrings produce more strength in cyclists than in football-players. This confirms the high IEMG activity of the hamstrings during cycling in all possible circumstances.

REFERENCES

1. Antonis, J., Clarys, J.P.,Cabri, J., Gregor, R.J., Clijsen, L., Welbergen, E.(1989) Congress Proceedings of the XIIth International Congress of Biomechanics, Los Angeles, UCLA Press, 445-446.
2. Clarys, J.P., Cabri, J., Gregor, R.J. (1988) J. of Sports Sciences 6: 229-237.
3. Davies, G.J. (1985) A Compendium of Isokinetics in Clinical usage, sec. ed. S&S publishers, La Crosse USA, 184-186 and 414-422.
4. Gregor, R.J., Cavanagh, P.R., Lafortune, M.A. (1985) J. of Biomech.18:307-316.
5. Hay, J.G. (1985) The Biomehcanics of Sport Techniques, 3rd. Ed., Englewood cliffs: Prentice Hall.
6. Telford, R.D., Hahn, A.G., Pyne, D.B., Tumilty, D.M. (1990) Excel 6, 4: 20-22.

AUTHOR'S ADDRESS

Jan Pieter Clarys, Ph.D.
Vrije Universiteit Brussel
Dept. Experimental Anatomy
Laarbeeklaan 103
1090 Brussels, Belgium

MOVEMENT-RELATED CORTICAL POTENTIALS AND THE SELECTION OF MOVEMENT

P. Praamstra[1], D.F. Stegeman[1*], Th. Oostendorp[2], M.W.I.M. Horstink[1], A.R. Cools[3]
[1]Inst. of Neurology and Depts. of [2]Medical Physics and [3]Psycho/Neuropharmacology, University of Nijmegen, The Netherlands. *presenting author

INTRODUCTION

Movement related cortical potentials form a class of event-related potentials that occur before and during the execution of voluntary movements. This report deals with the well-known Readiness Potential (RP). This is a slowly rising negative shift preceding movement at about 1500 ms and peaking at about movement onset. The highest over-all amplitudes are measured at the frontocentral midline and there is an asymmetry with higher amplitudes contralateral than ipsilateral on the side of movement (Figure 1).

PET studies have shown that regional cerebral blood flow (rCBF) in frontal areas is influenced by the mode of movement selection. The supplementary motor area (SMA) is more active when subjects freely select the direction of a joystick movement than when performing repetitive movements in a fixed direction (Deiber et al., 1991).

The present study deals with the hypothesis that the mode of movement selection, apparently influencing the SMA activity, also modulates the RP.

METHODS

Sixteen subjects (9 M, 7 F; age 21-47, mean 26) participated in the experiment. Two types of joystick movements were compared: (1) Single movements in a fixed direction (lateral) and (2) Single movements in four freely selected directions (either forward, backward, left or right). The movements were performed in a self-paced way at a rate of approximately once every 5-10 seconds. EEG recordings were made from 29 sites over the scalp and averaged time-locked to a switch closure in the joysticks.

RESULTS

In Figure 1 a grand-average over all 16 subjects for left hand movements is presented. The thin curves apply to the fixed direction condition, the bold curves are measured in connection with the freely selected movement. The averaged rectified EMG response from the dorsolateral surface of the forearm is shown in the lower right of the figure. Analyses of variance on individual data confirmed a main effect for the Mode of selection in the time window between -600 and 0 ms before movement. An interaction of Mode of selection by Electrode indicated the largest effects at electrode sites near the midline. The over-all wave shape did not show important differences between the two conditions.

DISCUSSION

Given the poor resolution of PET, it is difficult to obtain rCBF measures that selectively index the *planning* as opposed to the *execution* of movements. The present electrophysiological results unambiguously relate to the selection of movement. An important question that arises is whether the RP amplitude differences between movements in fixed and in freely selected directions can be attributed to the SMA. The distribution of the effect as well as the referenced PET results suggest indeed that the SMA is involved. Recent results from intracranial recordings of the RP (Ikeda et al., 1993), however, have challenged the traditional model of RP generation, which involves sequential activation of SMA and primary motor cortex. In addition, attempts to use source localization methods (e.g. Scherg 1990, own observations) in separating the RP signal in different contributions from separate locations in the brain have not yet unambiguously supported an SMA contribution to the RP. Further improvements of such techniques, e.g. by using more realistic head and source models are currently developed.

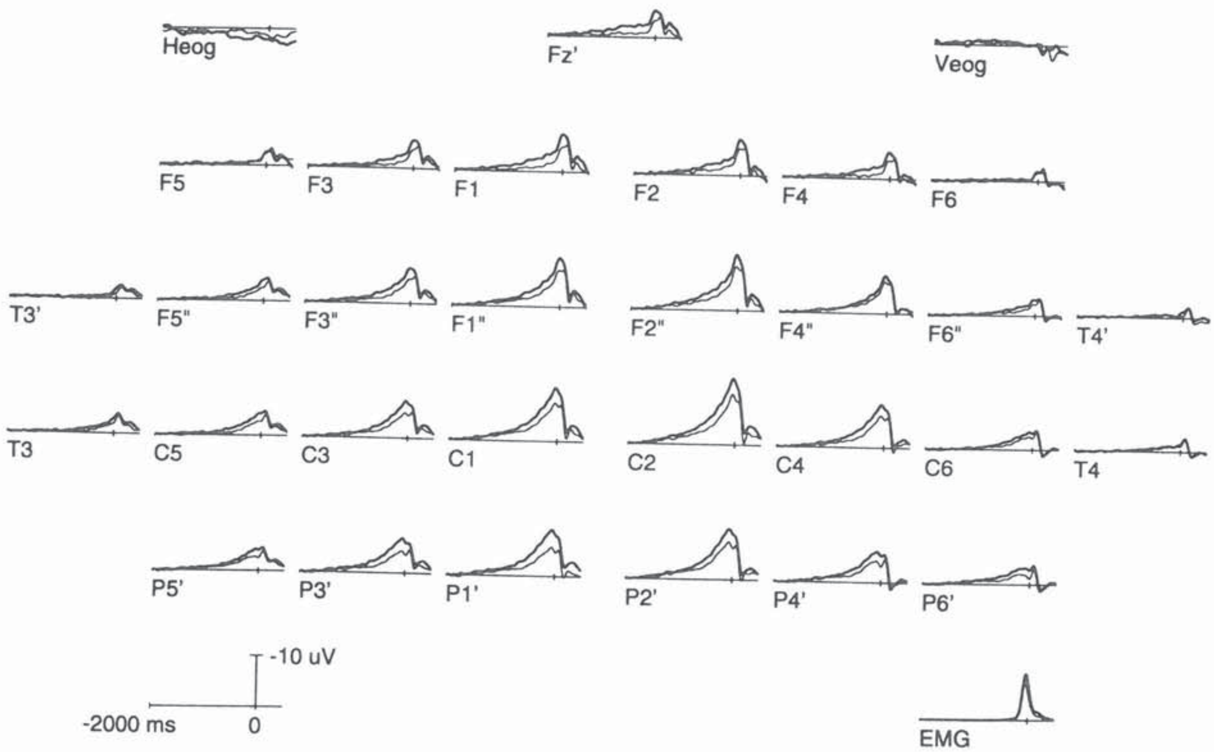


Figure 1: *Grand average readiness potential distribution for prescribed (thin) and free selected (bold) movements. Movement at time zero.*

REFERENCES

1. Deiber,M.P.,Passingham,R.E.,Colebatch,J.G.,Friston,K.J.,Nixon,P.D., Frackowiak,R.S.J. (1991) Exp. Brain Res. 84:393-402.
2. Ikeda,A.,Luders,H.O.,Burgess,R.C.,Shibasaki,H. (1993) Electroenceph. clin. Neurophysiol. 89:269-277.
3. Scherg,M. (1990) In: Grandori,F.,Hoke,M.,Romani,G.L. (eds) Advances in Audiology vol. 6. Karger, Basel, 40-69.

AUTHOR'S ADDRESS

Dick F. Stegeman, Ph.D., University Hospital Nijmegen, Inst. of Neurology, Dept. of Clinical Neurophysiology, P.O. Box 9101, 6500 HB Nijmegen, The Netherlands
Phone: int.-3180615284, Fax: int.-3180541122, Email: NEURO_JM@AZNVX1.AZN.NL

THE EFFECTS OF SELECTED PARAMETERS OF MUSCLE ENERGY TECHNIQUE ON LUMBAR PARAVERTEBRAL MUSCLE ACTIVITY IN HEALTHY INDIVIDUALS

Lisa Ash, Roy Bechtel, Gene Ford, Margaret Kelly, Michael Marks,
Tami Radov, Steve Strunk
Physical Therapy Dep't. University of Maryland, Baltimore, MD, USA

INTRODUCTION

Muscle energy techniques (MET) are often used clinically to correct somatic dysfunctions of the lumbar spine (1). The purpose of this study was to observe the effect of certain parameters of MET on surface electromyographic (EMG) activity of lumbar paravertebral muscles. The goal of the study was to gain an understanding of the mechanism by which MET exerts its effects. Parameters of interest were trunk rotation, task (hip ab or adduction), and effort level (100% or 50% MVC).

METHODS

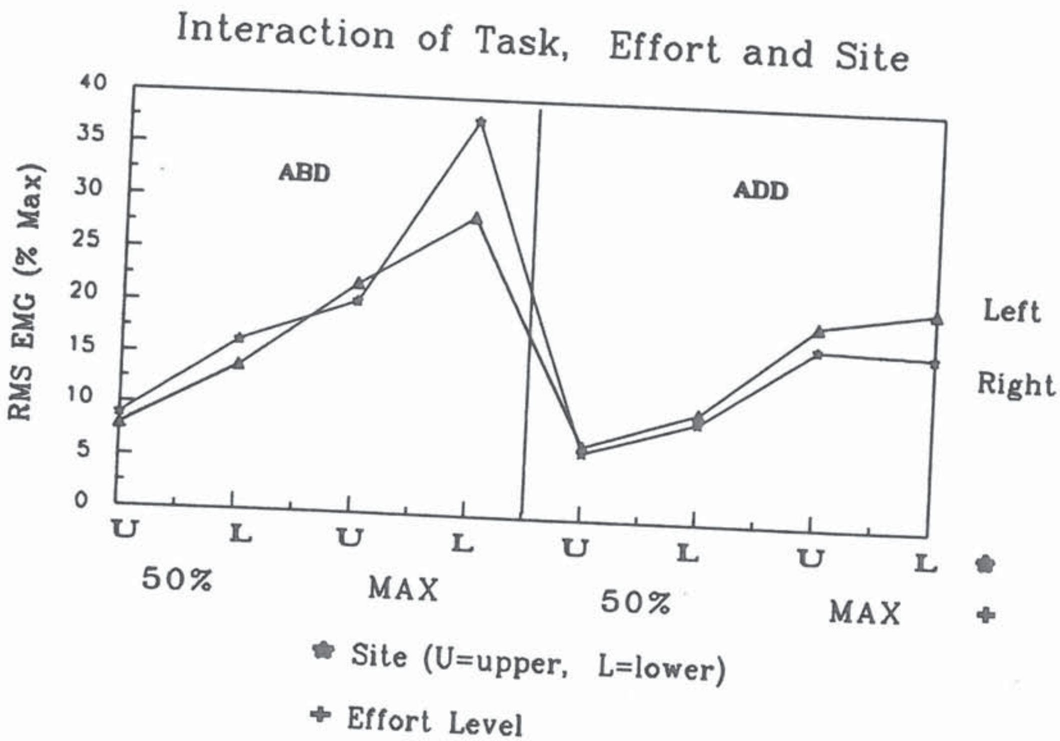
Healthy volunteer students (10 female, 7 male, age range 21 to 36, mean = 26, SD = 4.7) fitting the design criteria, were utilized for this study. Data were gathered with the subjects in sidelying, with hips flexed to produce lumbar flexion up to the L3-L4 level. One examiner positioned all subjects. Each subject performed hip abduction or adduction at maximum or 50% effort, with or without trunk rotation to the right and left. The order of trials was randomized. EMG recordings were taken from four active surface electrodes placed bilaterally on the erector spinae muscle mass lateral to L2-L3 and L3-L4 intervertebral joints. EMG data were sampled by computer at 1000 Hz per channel. RMS voltage levels were calculated and normalized using an MVC prone extension trial. The normalized RMS data were submitted to an ANOVA with an alpha level of .05.

RESULTS

Trunk rotation to the right or left as used in this study, had no significant effect on muscle activation. However, there were statistically significant effects of task and effort level. Hip abduction consistently elicited more paraspinal EMG activity than adduction especially at the L3-4 (Lower) level. Generally, trials at maximum effort (MAX) produced a greater difference in activity between the two sides, with the right side being more active during abduction (ABD), and the left side being more active during adduction (ADD).

CONCLUSION

The results of this study lend support to an indirect mechanism for MET. Since no difference in muscle activity between sides existed at clinical force levels, it is less likely that asymmetrical muscle force is responsible for changes brought about by MET. However, the possibility of asymmetrical force as a result of passive tension in ligaments and joint capsules cannot be ruled out. Additionally, the principle muscles studied in this protocol were multifidus and lumbar erector spinae (2,3,4). It is possible that asymmetrical muscle activity could occur in muscles not included in this study.



REFERENCES

1. Greenman PE: <u>Principles of Manual Medicine</u>. Williams and Wilkens, Baltimore, MD, 1980.

2. Bogduk, N., Macintosh, J., and Pearcy, M. "A universal model of the lumbar back muscles in the upright position". SPINE 17(8): 897-913, 1992.

3. Macintosh, J. and Bogduk, N. "The attachments of the lumbar erector spinae". SPINE 16(7): 783-92, 1991.

4. Valencia FP and Munro RR: An electromyographic study of the lumbar multifidus in man. Electromyography and Clinical Neurophysiology, 25: 205-221, 1985.

AUTHOR'S ADDRESS

Roy Bechtel, PT
University of Maryland
Department of Physical Therapy, Suite 115
100 Penn Street
Baltimore, MD 21201

FAR-FIELD POTENTIALS IN THE NEUROMUSCULAR SYSTEM

D.F. Stegeman, K. Roeleveld, S.L.H. Notermans
Inst. of Neurology, Dept. of Clinical Neurophysiology,
University of Nijmegen, The Netherlands

INTRODUCTION

Most potentials recorded in neuromuscular electrophysiology change in amplitude, wave shape and even polarity when the recording electrode is moved over a small distance. In the electroencephalogram, components are known which hardly change their characteristics with electrode position over the scalp. Jewett and Williston (1971) first used the term "far-field" in the description of auditory brain stem responses. Cracco and Cracco (1976) first described far-field components which were generated in the peripheral nervous system and could be observed all over the body.

A DIPOLE IN A FINITE VOLUME CONDUCTOR

In figure 1 two calculated potential distributions of a dipole source are shown. The source is equal in the two cases. However, in Fig. 1A the volume conductor around the source is infinite, as is often assumed in theoretical work, whereas in Fig. 1B the conducting extracellular medium has a finite extension.

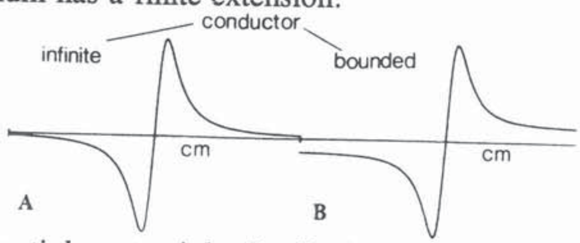


Figure 1: *Spatial potential distribution (A) in an infinite and (B) in a finite (cylindrical) volume conductor. The simulated observation is made at some radial distance from the source. Note the constant potential difference between left and right tails in part (B).*

From Fig. 1B it will be clear that we are dealing with an essential aspect: when the volume conductor is limited, the field of a dipole has a component that does not decrease with distance. Since an arm or a leg or the whole body are bounded volume conductors any dipolar source will generate a far-field potential distribution.

PROPAGATING ACTION POTENTIALS AND FAR-FIELDS

Propagating action potentials of nerve and muscle fibers are composed of two equal dipoles of opposite direction (often called tripoles, e.g. Poul Rosenfalck, 1969). For such a source the far-field contributions cancel. Therefore, nerve and muscle fibers are expected not to generate far-fields. However, when the constant propagation of an action potential is disturbed in some way, far-field potentials are generated (e.g. Stegeman et al., 1987; Dumitru, 1991; Deupree and Jewett, 1988). Such a disturbance can be caused by:
1) A change in the size of the extracellular medium, 2) a changing extracellular conductivity, 3) the generation or the blocking of an action potential and 4) a change in the direction of action potential propagation.
One can imagine that situations described in the above are numerous all over the

neuromuscular system. We will illustrate this by showing the large "far-field content" of surface EMG.

FAR-FIELDS AND EMG

Figure 2 shows twenty traces of surface EMG. The upper group of nine traces are recorded bipolarly with ten electrodes at six milimeter electrode interdistances over the muscle belly of the biceps brachii along fiber direction. The second group of ten traces applies to the same electrodes, but now with the elbow as a common reference. The lower trace is the "far-field" measurement of the muscle activity recorded between shoulder and elbow. Note the numerous components in the unipolar muscle records can also be recognized in the shoulder record. As can be expected, these far-field components are almost completely lost in the bipolar traces.

It is shown by Roeleveld et al. (1994) that the far-field components in these EMG records are almost exclusively caused by the blocking of action potentials at the fiber-tendon transition.

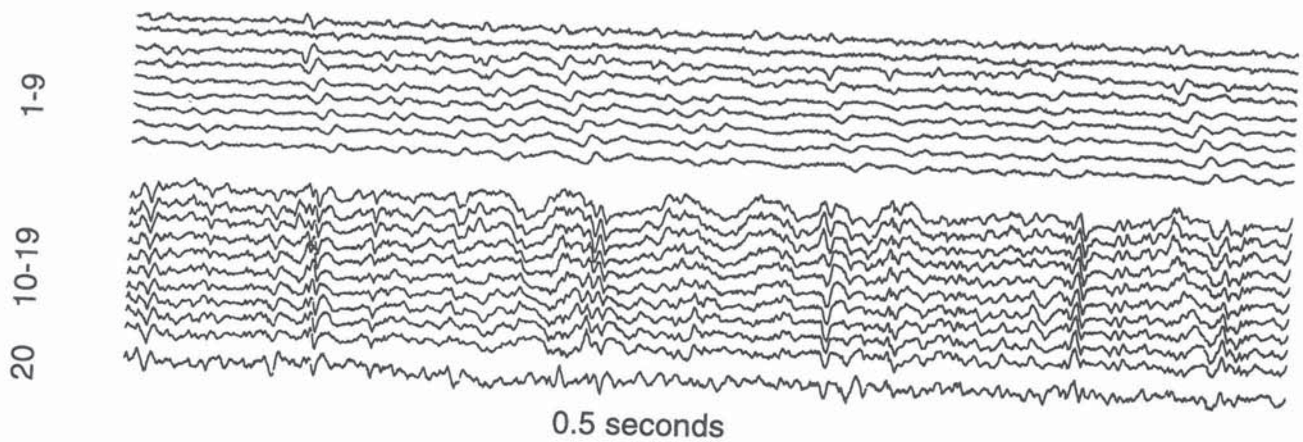


Figure 2: *Twenty traces of surface EMG from the biceps brachii. Traces 1-9 are recorded bipolarly in fiber direction over the muscle belly. Traces 10-19 are simultaneously recorded unipolarly with reference at the elbow. Trace 20 is recorded between an electrode on the shoulder and one on the elbow.*

REFERENCES

1. Jewett,D.L.,Williston,J.S. (1971) Brain 94:681-696.
2. Cracco,R.Q.,Cracco,J.B. (1976) Electroenceph. clin. Neurophysiol. 41:460-466.
3. Rosenfalck,P. (1969) Acta Physiol. Scand. (suppl.) 321:1-168.
4. Stegeman,D.F.,Van Oosterom,A.,Colon,E.J. (1987) Electroenceph. clin. Neurophysiol. 67:176-187.
5. Dumitru,D.,King,J.C. (1991) Muscle & Nerve 14:981-989.
6. Deupree,D.L,Jewett,D.L. (1988) Electroenceph. clin. Neurophysiol. 70:355-362.
7. Roeleveld,K.,Stegeman,D.F.,Vingerhoets,H.M.,Van Oosterom,A. (1994) this issue.

AUTHOR'S ADDRESS

Dick F. Stegeman, Ph.D., University Hospital Nijmegen, Inst. of Neurology, Dept. of Clinical Neurophysiology, P.O. Box 9101, 6500 HB Nijmegen, The Netherlands
Phone: int.-3180615284, Fax: int.-3180541122, Email: NEURO_JM@AZNVX1.AZN.NL

ELECTROMYOGRAPHICAL FEATURE OF HUMAN REACTION ON INFLUENCE OF NON-IONIZING AND IONIZING RADIATION.

E. A. ANDREEVA, O. E. KHUTORSKAYA
Institute of Control Sciences of the Russian Academy of Sciences. Moscow, Russia

SUMMARY

The authors developed a computer analysis method to investigate electrical activity of muscles (EMG) based on spectral-statistical processing of the envelope of the EMG (EEMG). The method made it possible to detect preclinical signs of human reaction under the influence of electromagnetic fields, ionizing radiation , reaction under displays radiation and ultrasonic fields as well.

INTRODUCTION

The method of spectral-statistical analysis of EEMG has been developed in the Institute of Control Sciences, the Russian Academy of Sciences. The method allowed to obtain new characteristic features of human reaction under influence of electromagnetic fields, ultrasonic fields and radiation.

The method was successfully used for examination of patients which underwent the Chernobyl's accident and participated to its liquidation, service staff of two nuclear power station (NPS) and of operators working for a long time with displays, ultrasonic devices and electromagnetic fields from mm. to dm. range (VHF).

METHOD

A method is based on spectral-statistical analysis of electrical activity of muscles (EMG). The main informational characteristics used are electrical muscular activity (EMG) and its envelope (EEMG); the principal mathematical apparatus of the method is spectral analysis and statistical processing of great arrays of spectral parameters of EMG and EEMG. The EMG are registered by standard electromyographic devices with aid of surface electrodes and transmitted to a computer by an analog-to-digital converter.By the spectral characteristics got on the computer a number of parameters are defined, stored and then the array is submitted to statistical processing. Results of statistical processing of spectral parameters allow to reveal objective statistically reliable criteria.

The main stages of the method: registration of EMG; transformation of EMG in EEMG; creation of arrays of EEMG spectral parameters and finding a characteristic parameter, whose value determines the presence of organism reaction to non-ionizing and ionizing radiation.

RESULTS

1. Non-ionizing irradiation of VHF-range

A group of physicians-volunteers (4 specialists) was subjected to special local irradiation of hands by an electromagnetical fields ($F= 2450$ MHz, $N=50\frac{Mw}{cm^2}$). Irradiation of each patient was carried out 4 times with an interval of several days. Irradiation duration was 20 min. The patients' examination was realized before the procedure series, one hour after the irradiation procedure end, a year and two years period after irradiation. After the first time some modifications of EEMG characteristics were observed. During the subsequent times the revealed deflections were not altered. The modified picture was conserved for 2 years.

All the examined were subjected to detailed clinical examinationat the Institute of Labor Hygiene and Professional Diseases. There were no clinical deviations observed. The examined patients did not express any subjective complaints.

2. Ionizing radiation.

In december 1986 a group of people (34) involved in the Chernobyl NPS accident and participating in elimination of its consequences were subjected to examining. In 1988-1990 a two groups of people working in NPS (85 and 47 persons) were examined. At the same time and within the same period the reference group of people (110) who were not professionally connected with the radiation factor were examined.

Table 1 contains values of the characteristic parameter, obtained for three groups. 1. The people involved in the accident or those participating in elimination Chernobyl NPS accident effects. The groups were twice examined: in 1986 and 1990; 2. The NPS personnel (N). (The joined groups NPS1 and NPS2) 3. The reference groups (R).

Table 1

Ch	N	R
-0.45	-0.36	0.0

Conclusion

1. The new objective sign of human neuromuscular system reaction to non-ionizing VHF radiation and ionizing radiation has been obtained, remaining distinct for not less than 2 years.
2. This sign allows us to determine a reaction to small and medium doses of ionizing radiation with no specific clinic deviations.
3. Individual estimation of a sign can be used for monitoring of human reaction to small radiation doses and detection of "risk" groups.
4. Average estimate of the sign with respect to a group can be used for estimating the environment state.
5. This sign may be used for biological dosimetry and development of objectively admissible doses.

INFLUENCE OF BODY POSITION ON MOTOR EVOKED POTENTIALS OF THE LOW BACK MUSCLES

M.A. LISSENS, A. DECLEIR, G.G. VANDERSTRAETEN
Dept. of Physical Medicine and Rehabilitation, University Hospital of Ghent, Ghent, Belgium.

SUMMARY

Motor evoked potentials (MEPs) of the low back muscles were studied through magnetic transcranial motor cortex stimulation in healthy humans and the influence of body position was evaluated during lying, sitting and standing. The MEP latency time was found to be shortened and the MEP amplitude to be increased significantly during sitting and standing as compared to the lying position.
These results reveal a direct projection from the motor cortex to the paravertebral low back muscles. Moreover, a different neurocontrol dependent from the body position was shown, the lumbar back muscles being significantly facilitated during sitting and standing compared to lying (inhibition). These findings might be important in the rehabilitation of several orthopedic and neurological disorders.

INTRODUCTION

Since 1985, it became possible to investigate the function and integrity of the corticospinal tracts through magnetic transcranial motor cortex stimulation to obtain motor evoked potentials (MEPs) of the recorded muscles (1). This technique has revealed already several clinical applications (2). So far, the low back muscles were not much studied with magnetic transcranial stimulation and the influence of body position on the motor evoked potentials was never evaluated. The aims of this study were to describe the MEP characteristics of the low back muscles in healthy subjects, and to determine if the body position influences neurocontrol, reflected by MEP latency times and amplitudes.

METHODS AND RESULTS

Ten healthy subjects (5 males and 5 females, age 20 to 40) were studied. Transcranial magnetic stimulation was performed with a magnetic stimulation coil of 90 mm diameter at the vertex (C_z according to the international 10-20 EEG system). Recording electrodes were placed on the lumbar paravertebral muscles at the L5 level. Supramaximal stimulation was applied during quiet lying backward on an examination bed, during unsupported, relaxed sitting and during unsupported quiet standing. Latency time and amplitudes of the recorded low back muscle MEPs were measured for the three evaluated body positions.
The MEP latency time significantly shortened and the MEP amplitude significantly increased (see figure 1) during sitting (14.5 msec, 450 µV) and standing (13.5 msec, 575 µV) as compared to the lying backward position (15.5 msec, 250 µV).

DISCUSSION

Firstly, these results have revealed a direct projection from the motor cortex to the human low back muscles (i.e. m. erector spinae). This projection shares many of the properties which have been demonstrated for the corticospinal projection to upper and lower limb muscles. Furthermore, the significantly increased MEP amplitude and

shortened MEP latency time of the paralumbar muscles during standing and sitting show a facilitation during these body positions and an inhibition during backward lying down, accounting for a change in neurocontrol in different body positions.
These findings might play an important role in the rehabilitation of several orthopedic and neurological disorders, in which body position is a crucial factor.

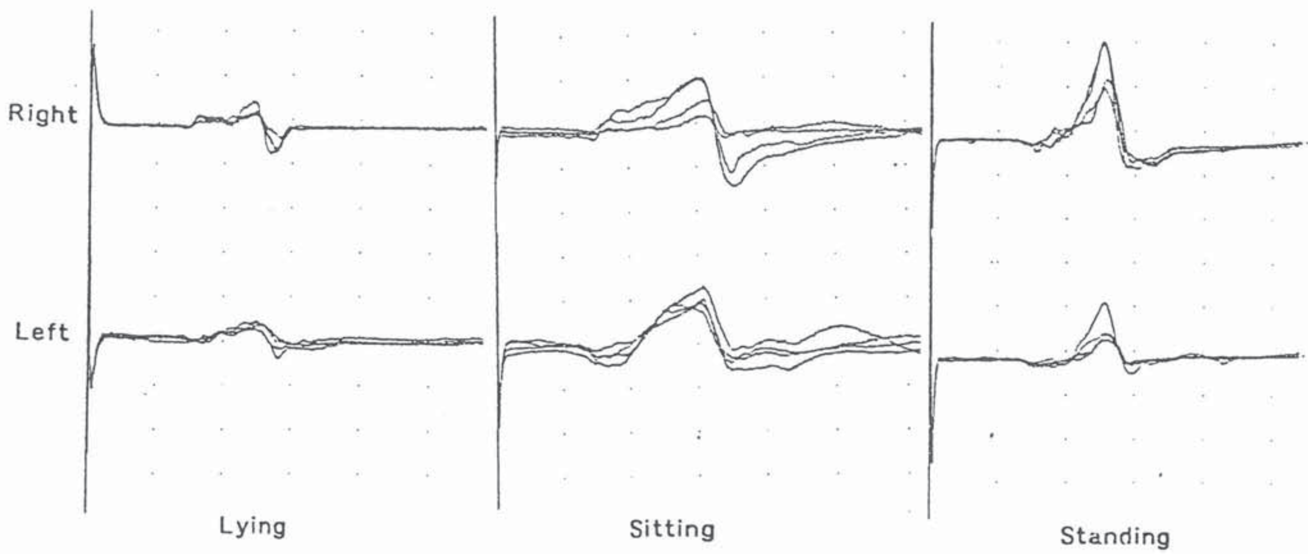


Figure 1 : Motor evoked potentials of both right and left paralumbar low back muscles at level L5 during lying, sitting and standing. Horizontal calibration is 10 msec/division; vertical calibration is 100 μV/division during lying and sitting, 200 μV/division during standing.

REFERENCES

1. Barker A.T., Jalinous R., Freeston I.C. (1985) Lancet, i : 1106-1107.

2. Lissens M.A. (ed.) Clinical applications of magnetic transcranial stimulation, Peeters Press, Leuven, Belgium, 1992.

3. Hummelsheim H., Hoppe S. (1994) Eur. J. Phys. Med. Rehabilitation (in press).

AUTHOR'S ADDRESS

Mark A. Lissens, M.D.
University Hospital of Ghent
Department of Physical Medicine and Rehabilitation
De Pintelaan 185
B-9000 GENT
BELGIUM

CERVICAL MUSCLE OVERUSE AS A FACTOR IN CHRONIC HEADACHE: A COMPARISON OF HEADACHE PATIENTS VS NORMAL SUBJECTS

S.J. Middaugh, J.J. Halford, W.G. Kee, J.A. Nicholson
Dept. of Physical Medicine & Rehabilitation, Medical University of South Carolina, Charleston, SC, U.S.A.

SUMMARY

Excessive use of muscles in the head and neck area may contribute to development or maintenance of chronic headache. Upper trapezius muscle activity during eight tasks was compared for 22 headache patients and 31 normal subjects. Statistical analyses of electromyographic (EMG) data (surface recording) found that headache patients had high resting baseline levels (sitting and standing), incomplete spontaneous recovery of baseline following contraction, and less ability to voluntarily relax with instructions compared to normal subjects ($p < .001$). This suggests that muscle factors contribute to chronic headache and treatment that includes muscle retraining, such as with EMG feedback, may be appropriate.

INTRODUCTION

There is considerable clinical and theoretical interest in the possible contribution of muscles of the head and neck to the development or maintenance of headache. While much of the interest has centered on tension type headache, muscle symptoms are often found in migraineurs as well. The upper trapezius ms. was selected for study since it is a common site of musculoskeletal pain in patients referred to a medical university headache treatment program.

METHODS AND RESULTS

Subjects were 22 consecutive headache patients (mean age = 43 yr, 6 male, 16 female) diagnosed by a neurologist as chronic tension type or mixed headache (n = 16, 72.6%), migraine (n = 3, 13.7%) or other headache diagnoses (n = 3, 13.7%) and 31 normal volunteers (mean age = 37 yr., 15 male, 16 female) screened to exclude those with head or neck pain. EMG was recorded over the surface of the upper trapezius muscle at the shoulder, tape recorded and quantified by analog to digital conversion for consecutive 1 second periods (RMS) for eight test activities: sitting and standing baseline, 3 repeated shoulder shrugs and 3 repeated shoulder abduction to horizontal (peak contraction, mean for a 3 second hold), spontaneous recovery of baseline after contraction (15 seconds) and ability to voluntarily relax the target muscle with instructions (15 seconds, sitting and standing).

Since maximum voluntary contractions could not be reliably obtained for the patient group due to pain, a dual strategy was adopted for statistical analysis of EMG data. (1) Group means for the absolute, non-normalized, EMG values were used for initial comparisons between groups using Analysis of Variance procedures and (2) the validity of these initial findings were further tested by selected within-groups analyses and selected comparisons using normalized EMG based on peak EMG during shoulder shrug and shoulder abduction.

EMG values did not differ by gender so data for males and females were combined. Right vs left side comparisons were included in all analyses, but were non-significant (data combined for figures).

Figure 1 presents data for sitting and standing baseline, recovery of baseline (spontaneous muscle relaxation) following test contractions (3 repetitions, shoulder shrug, shoulder abduction), and specific instructions to voluntarily relax the target muscle (instructed relaxation sitting, standing). For all 6 test phases in the figure, headache patients demonstrated elevated EMG compared to normals ($p<.001$). These between-group findings were supported by within-group tests; i.e., normals spontaneously recovered initial sitting baseline values and were able to decrease high standing baseline EMG with brief instructions to relax ($p<.001$) while headache patients did neither. The between-group findings were also supported by analysis of normalized EMG; i.e., headache patients still had inferior spontaneous muscle relaxation as a percentage of the mean peak EMG of the preceding contraction ($p<.001$).

CONCLUSION

The headache group showed elevated upper trapezius EMG compared with the normal group. This was most evident as slow, incomplete spontaneous muscle relaxation following moderate contractions in the headache group, and poor ability to voluntarily correct this problem with brief verbal instructions. These findings suggest that disruption of normal work/rest cycles may be a significant problem in headache patients and that specific muscle retraining procedures, such as EMG feedback procedures, may be an important component of headache treatment programs.

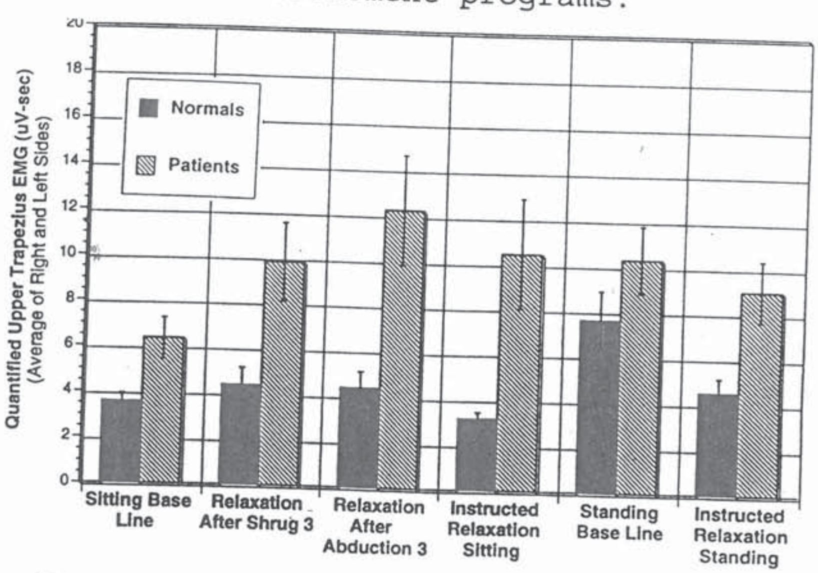


Figure 1. Upper Trapezius EMG in Headache and Normal Groups

Susan J. Middaugh, Ph.D.
Dept. of Physical Medicine & Rehab.
Medical University of South Carolina
171 Ashley Ave. Charleston, S.C., 29425
Phone (803) 792-2215 Fax (803) 792-5549

LONG TERM ASSESSMENT OF TRAPEZIUS MUSCLE LOAD USING AN AMBULATORY DEVICE

Sandsjö, L. (1,2), Öberg, T. (3), Kadefors, R. (1)
(1) Lindholmen Development, Göteborg, Sweden.
(2) Department of Applied Electronics, Chalmers University of Technology, Göteborg, Sweden.
(3) Department of Biomechanics and Orthopedic Technology, University College of Health Sciences, Jönköping, Sweden.

INTRODUCTION

In evaluation of manual work, the time-dependent loading on the trapezius muscle is often of high interest, since this muscle is commonly affected by pain syndromes. However, it is difficult to monitor trapezius muscle load over whole working days. In this report, we present a method for this type of assessment, using an ambulatory device (MyoGuard) for on-line monitoring and assessment of myoelectric signal parameters from the trapezius muscle (1,2).

METHODS AND RESULTS

Surface EMG from the trapezius muscle is recorded with a portable device. The device can easily be carried by the subject and computes the ARV (Averaged Rectified Value) and MPF (Mean Power Frequency) parameters from the myoelectric signal once every second and stores this information on a Memory Card. When the recorded session is ended the amplitude and frequency parameters are transferred to a PC for analysis.

A typical recording session starts with a calibration procedure followed by the desired number of trials (no more than 100). The calibration procedure consists of a number of repeated recordings, each 10 seconds long, from a standard position suitable for the work that is going to be monitored. In order to increase the comparability within subjects on different occasions, and between subjects, we normalize each recording session to the mean values of ARV and MPF obtained during the calibration procedure. The reliability of these initial ARV and MPF mean values is improved by the repeated recordings during the calibration procedure (3). The recording session can go on for an accumulated time of 16 hours.

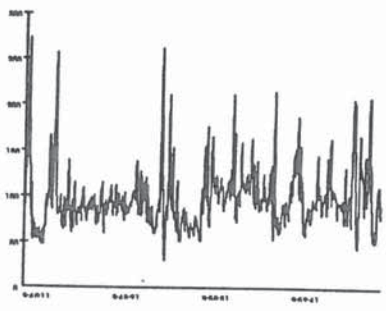
The most common use of the device is to show how the frequency and amplitude parameters develop with time when the subject performs his/her usual work. Furthermore, we can analyse amplitude and pause distribution. We have chosen to display the collected data in four different graphs. An example is given here showing the trapezius load pattern of a dental hygienist.

DISCUSSION

It is believed that the method described may prove useful in workplace evaluation and assessment of changes in the workplace to arrive at a reduced load on the trapezius muscle.

REFERENCES

(1) Sandsjö, L., Öberg, T., Kadefors, R. (1992), 9th International Congress of ISEK, 52.
(2) Kadefors, R., Sandsjö, L., Öberg, T. (1992), In: Mattila, M., Karwowski, W., eds, Computer Applications in Ergonomics, 433-437.
(3) Öberg, T., Sandsjö, L., Kadefors, R. (1994), Clinical Biomechanics, (in print).



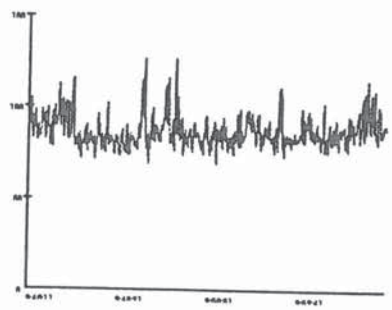
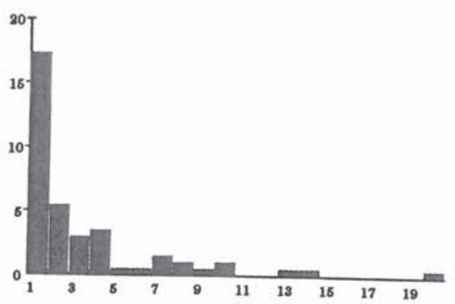


Figure 1: Shows the ARV parameter and its development with time during 5 minutes of a 2 hour recording. This sequence is selected from a display of the entire recording session by means of markers in the graph. 100% on the y-axis is equal to the mean ARV value obtained during the calibration procedure.

Figure 2: Shows the MPF parameter and its development with time during the same time interval and selected in the same way as the amplitude data in figure 1. 100% on the y-axis is equal to the mean MPF value obtained during the calibration procedure.



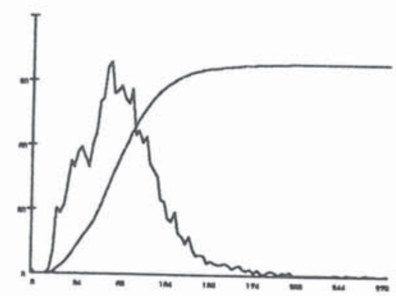


Figure 3: Shows the number of pauses (on the y-axis), and their length (along the x-axis), that are present in the 2 hour recording (ARV data). The pause treshold can be defined as a percentage of the mean ARV value obtained during the calibration procedure. In this figure the treshold is set to 25% of the mean ARV value.

Figure 4: Shows the amplitude distribution and the amplitude probability density function (APDF) of the 2 hour recording (ARV data). The y-axis shows percent levels regarding the APDF graph, the x-axis shows normalized ARV values where 100% denotes the mean ARV value from the calibration procedure.

AUTHOR'S ADDRESS

Leif Sandsjö, M.Sc.
Forskningsavdelningen
Lindholmen Utveckling
P.O. Box 8714
S-402 75 GÖTEBORG
SWEDEN
Tel: +46 31 50 70 39
Fax: +46 31 51 53 13

ACTION CURRENTS OF FAST AND SLOW MUSCLE FIBRES

Henk Wolters[1], Willemien Wallinga[1], Dirk L. Ypey[1,2] and Herman B.K. Boom[1]
[1] University of Twente, Institute for Biomedical Engineering, Enschede and
[2] University of Leiden, Department of Physiology, Leiden, The Netherlands

SUMMARY

Membrane currents of fast and slow skeletal muscle fibres of mice were measured during propagated action potentials. The membrane currents were always triphasic and depended strongly on the resting membrane potential. At their mean resting membrane potential (more negative for fast than for slow fibres) the action currents of fast and slow fibres showed different patterns. Action current and second derivative of the intracellular action potential differed significantly.

INTRODUCTION

Fast and slow muscle fibres show different intracellular action potentials (3). In an approximation, the membrane current can be represented by the second derivative of the intracellular action potential. Assumptions about membrane current strongly influence simulated single fibre action potentials and introduction of measured membrane current improved the simulation results (2). Consequently, it was of interest to determine the membrane currents during propagated action potentials for fast and slow fibres and to test the similarity between the membrane currents and the second derivatives of the intracellular action potentials.

METHODS

Mouse extensor digitorum longus (EDL) and soleus muscles were in mammalian Ringer solution, at 22-24 °C. Connective tissue was dissected. The number of EDL and soleus muscles was at least 4; the number of muscle fibres ≥ 13.

Muscle fibre activity was evoked with an intracellular pipette applying hyperpolarizing current pulses. The response occurred upon the termination of the stimulus.

For the recording of membrane current during propagating action potentials the loose-patch clamp technique was used (tip of the pipette clamped at 0 mV). From the triphasic membrane current two parameters were derived: the quotient of the amplitude of the first peak and the peak-to-peak amplitude (I_1/I_{pp}) and the interval between the first and second peak (δt).

In other experiments the propagated action potential was recorded with an intracellular pipette. The resting membrane potential (RMP) and the amplitude of the intracellular action potential (V_a) were determined. The second derivative of the intracellular action potential was compared with the recorded membrane current by means of I_1/I_{pp} and δt.

Statistical tests were the Student t-test (α=.01) and the analysis of covariance (F=.1).

RESULTS AND DISCUSSION

In second derivative representations of membrane current of an EDL and a soleus fibre at the mean resting potential of EDL and soleus muscle fibres the quotient I_1/I_{pp} differed, while the value of δt was comparable (fig. 1A). The mean value of the parameters of the intracellular action potential

(RMP: EDL -67.5 and soleus -60.5 mV, V_a: 98.5 and 86.0 mV resp.) and the I_1/I_{pp} parameter of the membrane current (.40 and .34) and of the second derivative of the intracellular action potential (.58 and .51) differed significantly between EDL and soleus fibres. Fig. 1B displays a recorded membrane current of an EDL fibre.

V_a, I_1/I_{pp} and δt depended on the resting membrane potential. If a linear relationship was assumed between the RMP and the parameters, there was no significant difference between the EDL and soleus fibres. Thus, the RMP appeared to be a dominant factor in the action current pattern (e.g. through inactivation).

I_1/I_{pp} and δt were significantly different for the transmembrane current and the second derivative transient.

In the near future we shall present action currents of human muscle fibres. This will contribute to optimization of simulations of human EMG patterns, because the approximation with the second derivative of the intracellular action potential can be avoided.

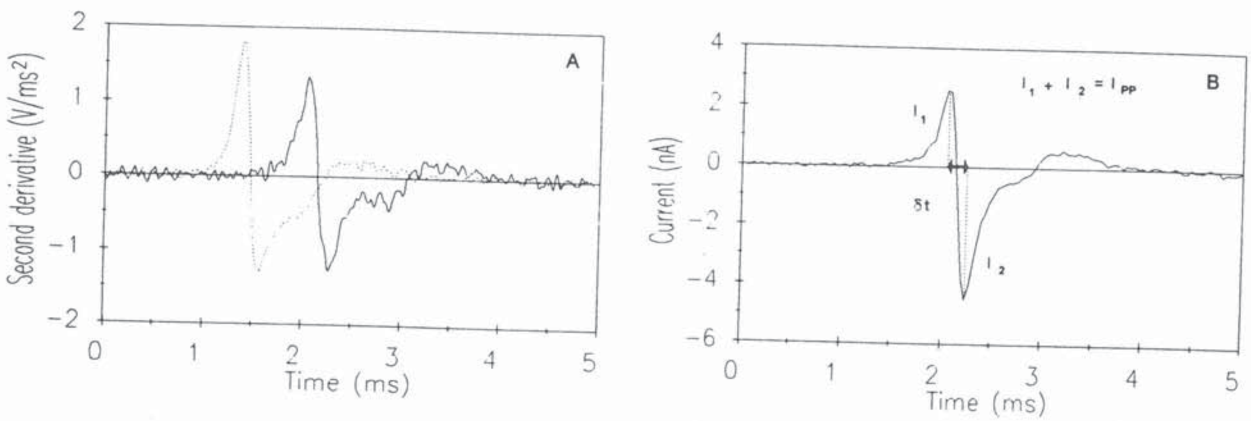


Figure 1. A. Typical calculated action currents of an EDL (dotted line) and a soleus fibre (full line) at the mean resting potential of EDL and soleus fibres. B. Recording of membrane current of an EDL fibre with the characteristics used.

REFERENCES

1. Ruff, R.L., Whittlesey, D. (1993) Muscle & Nerve 16:922-929.
2. van Veen,B.K.,Wolters, H., Wallinga, W., Rutten, W.L.C., Boom, H.B.K. (1993) Biophys. J. 64:1492-1498.
3. Wallinga-de Jonge, W.,Gielen, F.L.H., Wirtz, P., de Jong, P., Broenink, J. (1985) Electroencephalogr. Clin. Neurophysiol. 60:539-547.

Supported by "Prinses Beatrix Fonds".

AUTHOR'S ADDRESS

Willemien Wallinga, Ph.D.
University of Twente, Institute for Biomedical Technology
Department of Electrical Engineering
P.O.Box 217, 7500 AE Enschede, The Netherlands

LONG LATENCY EFFECTS OF TRANSCRANIAL MAGNETIC STIMULATION OF THE MOTOR CORTEX ON H-REFLEXES IN LATERAL GASTROCNEMIUS

D.L. Wolfe[1], K.C. Hayes[2], P.J. Potter[2],Neuroscience Program[1] and Dept. of Phys. Med. and Rehab.[2], University of Western Ontario, London, Ontario, Canada.

SUMMARY

The effect of transcranial magnetic stimulation (TMS) on the excitability of the motor neuron pool was examined by cortical conditioning of H-reflexes in the lateral gastrocnemius (LG) muscle. Of particular interest were the later effects following the initial period of facilitation mediated by the fast conducting corticospinal pathway. Most notable was a period of enhanced motor neuron excitability 70-120 ms following cortical stimulation. The appearance of late facilitation in the absence of short latency responses suggests that this increase in motor neuron excitability is mediated through descending tracts other than those subserving the early short latency facilitation.

INTRODUCTION

Stimulation of the motor cortex by a pulsed magnetic field (1) results in the appearance of a synchronous compound muscle action potential termed the motor evoked potential (MEP) which is mediated via fast conducting corticospinal pathways (2). Recently, evidence has been presented to suggest that there may also be longer latency excitatory inputs to the target motor neuron pool following TMS during specific time periods after the arrival of the early "fast" corticospinal input (3,4). In the lower limbs two distinct later periods of excitability have been identified although there appears to be some variability as to the precise timing of these responses. The pathways mediating these responses remain to be elucidated.

The present study was designed to examine these late effects of TMS on the excitability of the motor neuron pool as detected by the cortical conditioning of H-reflexes in the LG muscle.

METHODS

Test (T) H-reflexes elicited in the LG muscle were conditioned (C) by TMS at varying C-T intervals (0-300 ms) in healthy, adult subjects. Trials involving control H-reflexes were paired with trials at each C-T test interval with the order of these pairings being randomized. Two trials at each C-T interval were examined.

TMS was delivered by a Cadwell MES-10 stimulator through a focal point stimulating coil with an effective radius of 4.75 cm. Cortical stimulation intensity was set to the maximum comfortably tolerated by subjects. Test H-reflexes were elicited by electrical stimulation of the posterior tibial nerve at the popliteal fossa (0.5 ms rectangular pulse). Stimulation intensity was maintained at a level capable of eliciting a response of 25% of the maximum H-reflex amplitude.

The conditioned H-reflex peak to peak amplitude was normalized with respect to the adjacent paired control H-reflex amplitude and means were calculated to allow comparisons across subjects.

RESULTS

Normalized H-reflex amplitudes averaged across subjects are presented in Figure 1. Conditioned H-reflexes were facilitated at all C-T intervals tested by at least 2 standard deviations (SD) greater than the mean control H-reflex amplitude. Two profound periods of facilitation were evident at C-T: 0-40 ms and at C-T: 70-120 ms. Of particular interest in the present investigation was the later period of facilitation demonstrated in all subjects (n=5) and present even when the short latency facilitation was subliminal for evoking an MEP (n=2).

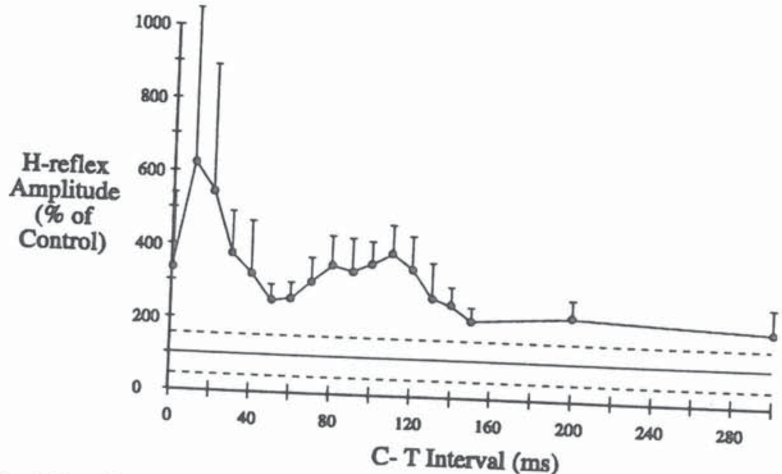


Figure 1: Normalized H-reflex Amplitudes (n=5). Dashed lines represent +/- 2 SD about the mean control H-reflex amplitude. Vertical lines reflect the mean + 1 SD of the conditioned H-reflex amplitude.

DISCUSSION

The present results demonstrate two distinct periods of facilitation of LG motor neuron excitability following TMS. The first of these is probably mediated by the fast conducting corticospinal pathway. The later period of excitability of the LG motor neuron pool is evident at C-T: 70-120 ms. This is consistent with the late responses to TMS noted by others which have been characterized with respect to their latencies by the designations "MEP_{70}" and "MEP_{120}" (4) or "S100" (3,5). The evidence accumulated to date suggests that the initial component of these late responses may be mediated through descending tracts such as indirect corticospinal and/or cortico-bulbospinal pathways via interneuronal circuits and not by activation of peripheral receptors responding to the initial muscle twitch associated with TMS (4). This view is substantiated by the present study in that the later period of facilitation was present even when short latency MEPs were not evident. Other investigators have suggested that this response may reflect a startle reaction (5). Further study elucidating the mechanisms generating these responses is necessary as these late responses may constitute an important new means for identifying preserved integrity of pathways other than the lateral or medial corticospinal tracts in cords compromised by trauma or disease.

REFERENCES

1. Barker, A.T., Jalinous, R. and Freeston, I.L. (1985) Lancet i:1106-1107.
2. Amassian, V.E., Quirk, G.J. and Stewart, M. (1990) Electroenceph. clin. Neurophysiol. 77:390-401.
3. Holmgren, H., Larsson, L.E. and Pederson, S. (1990) Electroenceph. clin. Neurophysiol. 75:161-172.
4. Dimitijevic, M.R., Kofler, M., McKay, W.B., Sherwood, A.M., Van der Linden, C. and Lissens M.A. (1992) Electroenceph. clin. Neurophysiol. 85:365-373.
5. Holmgren, H., Kadanka, Z. and Larsson, L.-E. (1992) Electroenceph. clin. Neurophysiol. 85:374-381.

AUTHOR'S ADDRESS

Dalton L. Wolfe
Research Department, Parkwood Hospital
801 Commissioner's Rd. E.
London, Ontario, Canada, N6C 5J1

Telephone: (519) 685-4000 (ext 2940)
Fax: (519) 685-4060

MOTOR EVOKED POTENTIALS IN THE LOWER LIMBS OF NORMAL AND SPINAL CORD INJURED SUBJECTS.

B.A. Conway, M. Delargy[1], F. Proudlock.
Bioengineering Unit, University of Strathclyde, Glasgow and [1]The Queen Elizabeth National Spinal Injuries Unit (Scotland), The Southern General Hospital, Glasgow, Scotland, U.K.

INTRODUCTION

It is now widely recognised that neurophysiological studies on patients with clinically complete lesions of the spinal cord can often reveal evidence of residual descending effects on motor behaviour caudal to a lesion (see ref. 2). Given the current interest in neural regeneration and transplantation in the spinal cord the above observations highlight the need for tests which can be used to assess the effectiveness of these therapies by detailing the viability of important descending spinal tracts to all level of the spinal cord. Since the first demonstration of magnetic stimulation of the motor cortex (1) there has been much interest in the use of magnetic stimulation as a non-invasive means of studying the corticospinal tract of normal subjects and of subjects with lesions of the central nervous system. However, due to the non-homogeneous nature of spinal cord injuries (SCI), studies on corticospinal innervation to the lower limbs in groups of SCI subjects have often provided conflicting views (e.g. see refs. 2,3). In studies on motor evoked potentials (MEP's) in such subjects it is essential that the stimulator and coil used can adequately excite the cortical areas innervating the lower limbs. We have therefore attempted to evaluate the effectiveness of magnetic stimulation directed to the leg area of the motor cortex in relaxed normal subjects by comparing MEP's recorded from the lower limbs with those obtained from arm muscles. This data was then used to provide controls for a pilot study on a group of SCI subjects.

METHODS

Experimental results were obtained from 20 normal and 8 chronic SCI subjects. Clinical testing revealed no observable voluntary movement in the lower limbs of the SCI patients. Muscles studied were (a) for the upper limbs: flexor carpi radialis (FCR), abductor pollicis brevis (APB) and abductor digiti minimi (ADMU), and (b) for the lower limbs; rectus femoris (RF), semimembranosus/semintendinosus (ST/SM), tibialis anterior (TA), lateral gastrocnemius (LG), extensor digitorum brevis (EDB) and abductor digiti minimi (ADML). To stimulate arm and hand areas of the cortex a 90mm circular coil held over the vertex was used while for the leg area a double cone coil (Magstim type-9902) was used. MEP's were recorded by use of surface EMG (Medelec Sapphire Premiere). Where appropriate, MEP's were characterised by measurements of MEP threshold, size (expressed as %Mmax), latency, central conduction time and morphology. All procedures were performed with Ethical Committee approval.

RESULTS

Normal subjects: Stimulation of the leg area of the cortex revealed that the threshold stimulus required to generate MEP's in leg muscles are only slightly higher than those observed to activate upper limb muscles when a standard 90mm coil is used. Of the leg muscles studied, threshold was lowest in ADML (42.9±7.8%) and highest in RF (52.7±5.8%). Stimulation at multiples of threshold (ie. from 1.1 to 1.5 times threshold) illustrate that when MEP's are expressed as a percentage of Mmax that the magnitude of responses in the arms and legs are comparable (Fig. 1).

Stimulation at 1.5 times threshold was close to the maximal stimulus strength deliverable by the stimulator. However, reference to Fig 1 would suggest that for both arm and leg muscles this stimulus remains submaximal. Nevertheless, the results illustrated in Fig. 1 show that in the relaxed subject large MEP's can be generated in the muscles of the leg.

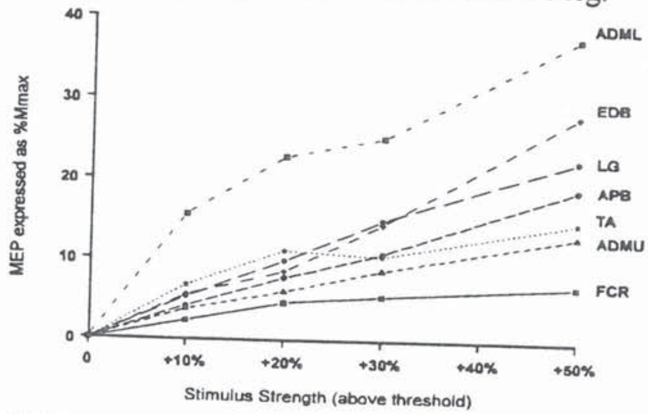


Figure 1. Variation in MEP size (expressed as % Mmax.) with stimulus strength (as multiple of threshold).

<u>Spinal Cord Injured Subjects:</u> Of the total number of leg muscles studied the majority showed no MEP. However, almost 25% did displayed a response to magnetic stimulation. These MEP's were highly variable and generally abnormal in at least one of the parameters used to characterise their behaviour. MEP's were generally small with prolonged latencies. In addition high thresholds and the appearance of a second long latency EMG burst were commonly seen.

CONCLUSIONS

The results from normal subjects illustrate that with the techniques employed that large MEP's can be evoked in relaxed muscles of the leg and foot. This highlights the adequacy of the stimulus in activating corticospinal neurones innervating lumbar and sacral segments of the cord. Consequently, the absence of MEP's in the leg muscles of SCI subjects is most likely to reflect disruption to descending pathways rather than to an inadequate stimulation of the motor cortex. The significance of the MEP's observed in the small group of SCI patients remains to be determined.

REFERENCES

1 Barker et al., Lancet, 1985,1106-1107

2 Lissens et al., In:Clinical Applications of Magnetic Transcranial Stimulation, chp4, pp42-56, Uitgeerij Peeters, Leuven,1992

3 Tegenthoff, In:Clinical Applications of Magnetic Transcranial Stimulation, chp3, pp33-41, Uitgeerij Peeters, Leuven,1992;

Supported by a grant from Action Research

AUTHOR'S ADDRESS

Dr. Bernard A. Conway
Bioengineering Unit
Strathclyde University
Glasgow, Scotland
Tel: 41 552 4400, Fax: 41 552 6098

CENTRAL MOTOR CONDUCTIVITY IN AGED PEOPLE

Yukio Mano, Takuya Nakamuro, Katsunori Ikoma,
Tadao Sugata, Shigeru Morimoto, Tetsuya Takayanagi and
Richard F Mayer [1).
Department of Neurology, Nara Medical University
Nara, Japan.
1)Department of Neurology, University of Maryland, School of Medicine. Baltimore, U.S.A.

SUMMARY

The conductivity of motor neurones in 26 aged females (mean age 79 years) was analysed by the pulsed magnetic stimulation and compared with 14 younger controls. In aged people slow motor conduction velocities were found in peripheral nerves. CMCT in relaxed muscle was shorter in the aged people, although CMCT was normal in mildly contracted muscle. These findings are coincidence with the results studied in Parkinson's disease. This may account for the present finding that aged people have neurophysiological abnormalities in CNS which are similar to those in Parkinson's disease.

INTRODUCTION

In most aged people motor functions decrease because of disturbances in motor programming, recognition, coordination, speed of movement, connective tissues, etc. In this study we analysed motor function in aged people from the perspective of the conductivity of motor neurones.

SUBJECTS

The 26 sound cases (all female, aged 72 to 90 years) were the subjects of this study. The mean± 1SD of age was 78.7± 4.8 years. Fourteen females, aged 31.2± 16.8 years, were studied as controls.

METHOD

The MEP from the abductor pollicis brevis (APB) and abductor digiti minimi muscle (ADM) were recorded by pulsed magnetic stimulation on the motor cortex and neck in relaxed and slightly contracted muscle in the sitting position. Muscle relaxation was instructed to the subjects and was defined as no muscle activity on the surface EMG recordings of the APB and ADM at least 3 minutes before stimulation. The central motor conduction times (CMCT) were calculated.

RESULT

In relaxed muscle, CMCT-f from the APB recording was 4.7± 1.5 msec in aged people and 6.9± 1.0 msec in younger people ($p<0.001$), CMCT-f from the ADM was 4.9± 1.4 msec in aged people and 7.2± 1.4 msec in younger people ($p<0.001$).

In mildly contracted muscle, CMCT-f from the APB was 3.9± 1.3 msec in aged people and 5.0± 1.5 msec in the younger group, CMCT-f from the ADM was 3.8± 1.5 msec in aged people and 4.7± 1.3 msec in the younger group. In mildly contracted muscle there were no statistically significant differnces between the two age groups.

DISCUSSION

This study confirms the previous report of slow conduction velocity in peripheral nerves of aged people. Although CMCT in mildly contracted muscle did not differ between aged and younger people, CMCT of relaxed muscle was shorter in the aged group.

The CMCT in non-contracted muscle in Parkinson's disease was found to be shorter than in controls. These studies indicate that central conduction time of motor neurones is affected by the extrapyramidal system. The excitatory inputs are alleged to affect the anterior motor neurones in Parkinson's disease. In this study, although subjects with definite tremor and rigidity were excluded, subclinical symptoms might have been present in the aged subjects and be related to the shortened CMCT in non-contracted muscle.

Auther's address : Yukio Mano, M. D., Ph. D.
Department of Neurology, Nara Medical University
840, Shijocho, Kashihara, Nara, 634, Japan.

H-REFLEX OF THE VASTUS MEDIALIS MUSCLE: A NEW DIAGNOSTIC TOOL IN THE TREATMENT OF RUPTURES OF THE ANTERIOR CRUCIATE LIGAMENT:

Wißmeyer, Th; Hülser, P.J.; Kutter, T.; Kinzl, L.
From the Departments of Orthopedic Surgery and Neurology, University Hospital, Ulm, FRG

INTRODUCTION

Up to now mechanical stability has been considered to be the most important factor in evaluating results after a tear of the anterior cruaciate ligament (ACL). Despite progress in surgical techniques of reconstruction and/or repair of the ruptured ACL clinical results have not always been satisfactory and many questions about this frequently injured ligament have yet to be answered [4].

In order to test not only the passive, but also the functional stability of the knee joint, it is necessary to obtain information about the control system of the muscles of the thigh. There has been increasing interest in the neurophysiological function of mechanoreceptors within the ACL and their role in motion control as well as their use in the diagnosis and treatment of the injured ACL. Since the early 1980's Johansson and co-workers have investigated the effects of mechanoreceptors within the ACL on the fusimotor-muscle-spindle system in animals (for review see [2]). Based on the integrative function of this system in motion-control and muscle-activity they postulated the theory of the "final common input" concerning the regulation effected by the Gamma-motoneurons.

To demonstrate function of the mechanoreceptors a new non-invasive method has been developed which uses changes in motor-neuron-excitability in the quadriceps muscle during external streching of the ACL. It is well known that the H-reflex represents an objective criterion to measure the excitability of motoneurons (3).

METHODS

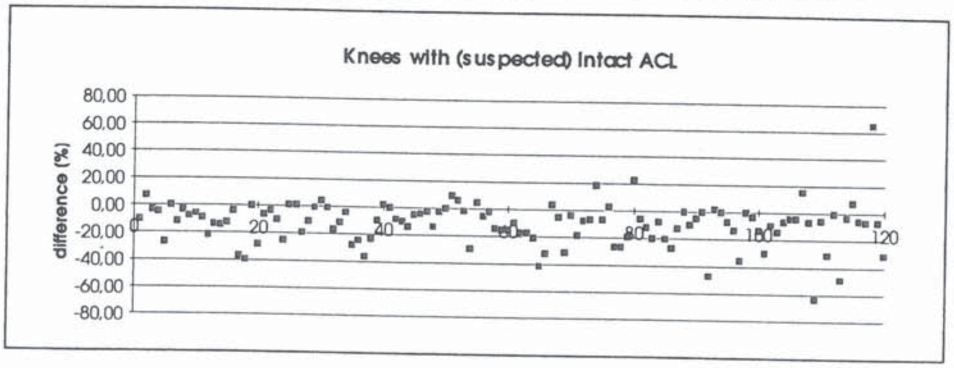
The human ACL is the primary restraint to anterior tibial displacement (ATD) at 30° of knee-flexion (1). In a position of 30° knee flexion and 135° hip-flexion the maximal amplitude of the H-reflex of the vastus medialis muscle is recorded by stimulating the femoral nerve in the groin using changing stimulus-intensity at a stimulation-frequency of 0.2 Hertz and using Ag/AgCl-surface-recording-electrodes and adhesive TENS-electrodes for stimulating. An optimum stimulus intensity is necessary to differenciate the M- from the H-response. Custom design instrumentation insures a relaxed and reproducible position of the lower extremity. Constant ACL tension is achieved by a 22 Kp anterior forward stress on the proximal tibia. The maximum H-response with and without forward stress on the tibia is registered and compared for each knee.

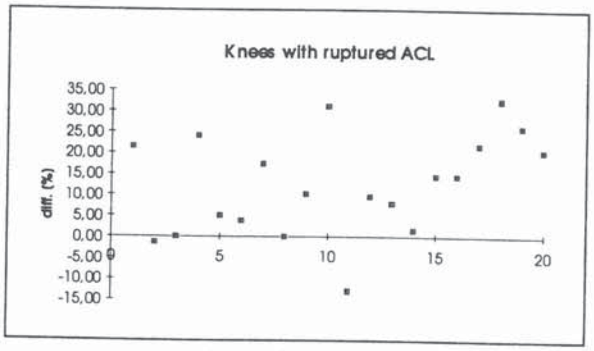
In 25 patients (mean age 29 ys; 17m 8f) with suspected unilateral rupture of the ACL the amplitude of the H-reflex was recorded with and without anterior displacement force 1 day before arthroscopic evaluation. Discomfort was tolerable and there was no evidence of muscle guarding. In 19 patients a torn ACL was found at the time of arthroscopy, six patients had a partial tear with parts of of the ACL still intact covered by a synovial membrane. 48 healthy volunteers (mean age 26; 29m 19f) without prior history of knee trauma or symptoms were used as a control group.

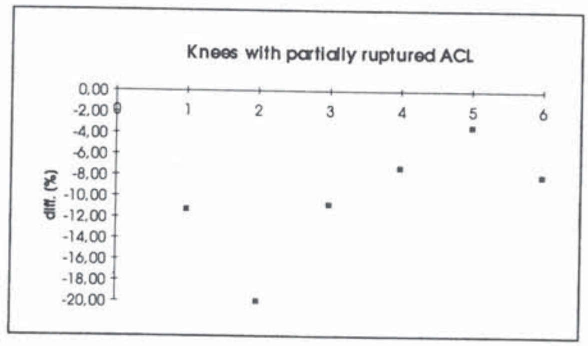
RESULTS

A reproducible H-response was recorded in 98% of cases. The maximum amplitude of the H-reflex decreased during passive stretching of the ACL in patients without injury to ACL and in the normal control group. In patients with a ruptured ACL the amplitude of the H-reflex tended to increase. A response similar to the control group was found in patints with a partially intact ACL.

Amplitude-changes of H-response during ACL-stress







	ruptured ACL (RA)	susp.intact ACL (IA)	pt.rupt.ACL (PRA)
average [%]	+11,9	-10,37	-10,15
Standard deviation	12,22	15,63	5,59
n	19	121	6

The proportional changes of the amplitudes between the groups are significantly different. RA / IA $p < 0,00001$ (T-test); PRA / IA $p < 0,0003$ (Wilcoxon-Rank-Sum-test)

CONCLUSION

1) Stimulation of neuroreceptors in the ACL leads to inhibition of motoneurons of the quadriceps muscle, which represents a major stressor of the ACL. According to the final-common-input-theory this states the effectiveness of the stress-controling system of the ACL.
2) The influence of ligamentous receptors on the motoneuronpool in humans is demonstrated for the first time by a noninvasive method.
3) This method allows the evaluation of the neurophysiological function of the intact and ruptured ACL.
4) The neuromuscular regulation by mechanoeceptors is preserved as long as parts of the ligament are intact and a covering synovial membrane is present.

REFERENCES

1. Butler DL (1989) Anterior cruciate ligament: its normal response and replacement. Orthop Res 7: 910-921.
2. Johansson H, Sjölander P, Soika P (1991) Receptors in the knee joint - ligaments and their role in the biomechanics of the joint. Biomed Engineering 18: 341-368
3. Paillard J (1955) Réflexes et régulations d'origine proprioceptive chez l'homme. Etude neurophysiologique et neuropsychologique. Arnette, Paris
4. Wroble RR, Brand RA (1990) Paradoxes in the history of the anterior cruciate ligament. Clin Orthop 259: 183-191

CORPORATE SPONSORS

As one of the premiere organizations devoted to the promotion and dissemination of knowledge in the field of electrophysiology and kinesiology, the 10th Congress of ISEK is delighted to welcome the support of the sponsors listed below. These corporations have a longstanding interest and commitment to this field of endeavor and we are happy that they have joined the Organizing Committee in supporting the 10th Congress. Individual corporations are listed in alphabetical order. The brief descriptions were provided by the companies as a service for our members and attendees. Please contact these corporations at the addresses indicated for further information about their products and services.

▼ ***Advanced Medical Technology, Inc.***
176 Waltham Street
Watertown MA 02172-4800
(617)926-6700
(617) 926-5045

AMTI will demonstrate the following three force platform systems:
1) BiovecTM - force vector analysis for gait, postural sway and lifting.
2) AccuSwayTM - postural sway, a new low cost balance measurement system.
3) PressMapTM - pressure distribution measurement while walking on a force platform

▼ ***ARIEL DYNAMICS, INC.***

6 Alicante
Trabuco Canyon,CA 92679
Phone: (714) 858-4216
Fax: (714) 858-5022

7920 Silverton, Suite E
San Diego, CA 92126
Phone: (619) 549-9717
Fax: (619) 549-9719

Ariel Dynamics, Inc. has pioneered research and product design in the fields of biomechanics and human diagnostic performance and training. This innovative company reflects the attitude of its president, Dr. Gideon Ariel, to develop the best, more technologically advanced, and objective tools for movement quantification. Dr. Ariel and his staff of scientists, engineers, and programmers have invented the most sophisticated computerized diagnostic and exercise equipment as well as developed the Ariel Performance Analysis System for the study of human motion and biomechanical quantification. Recent hardware additions have included a device allowing panning camera data to be included in the three-dimensional analysis and a 120 Hz camera. The biomechanical system was expanded to include kinetic capabilities from any force platform as well as extensive electromyography data collection and quantification. The motivating forces at Ariel Dynamics to develop these hardware and software additions are not the results of salesmen driven by marketing research but because more advanced technology became available, a customer needed or recommended an improvement, or a better method, algorithm, or technique was developed, discovered, or recommended. This company embodies the scientific ideals of the search for knowledge and always strives towards excellence.

▼ ***B & L ENGINEERING***
12309 E. Florence Avenue
Santa Fe Springs, CA 90670
(310) 903-1219
Fax: (310) 903-1221

B & L Engineering sells and develops products specifically for the Electrophysiology and Kinesiology markets. B & L's primary products are:

VICON - The premier 3D motion measurement and analysis system manufactured by Oxford Metrics Limited. It is fast, reliable, accurate and easy to use.

EMG Systems - Electromyographic Measurement Systems, Telemetry or Cable Transmission

QUANT - Comprehensive "EMG Analyzer" software program

Stride Analyzer - Basic entry level gait analysis system that measures velocity, cadence, stride length, gait cycle duration and single and double stance durations.

▼ *BTS - Bioengineering Technology & Systems*

Via Capecelatro, 66
20148 MILANO, ITALY
Tel: 39/2/26861048
Fax: 39/2/268861048

747 Althouse Street
Woodmere, NY 11598
(516) 295-2721
(516) 569-7586

BTS continues to be the preferred choice of the worlds' leading movement analysts. By responding to the need for an accurate, flexible, and time efficient movement analyzer, BTS has provided both the researcher and clinician a powerful tool for data collection and analysis. Minimal time is wasted on the complexities of data collection.

The foundation of the ELITE kinematic analysis is our tracking of passive reflective markers. Marker tracking is accomplished through shape recognition. This unique technology eliminates the problems associated with less sophisticated systems, namely environmental reflections and marker merging. Real time processing of the markers reduces time consuming and frustrating post data collection editing. The ELITE's ability to use markers as small as 1mm permits previously impossible applications. The Elite system is configured to integrate and synchronize a wide variety of peripheral analog technology, including our own TELEMG system and force platform analysis.

The TELEMG system is specifically designed for the electromyographic analysis of a moving subject. TELEMG incorporates telemetry and fiber optics allowing the subject complete freedom of movement. Data is transferred through highly reliable radio transmission. Supporting software PC-VECT provides the user dynamic software for user-defined data analysis.

In this time of rapidly developing technology, an investment in BTS will meet your changing needs of today and tomorrow. BTS is dedicated to providing cost effective upgrading and will easily accommodate the future challenges of your evolving profession.

BTS is supported in the United States by over twelve representatives and bi-coastal service locations and is the leading system for research. As a result of the ELITE's flexibility, BTS equipment has become popular for a wide variety of applications. The only limitation of the ELITE system is the users imaginations. Please accept our offer of a demonstration to fully explore this exciting technology.

▼ *BUTTERWORTH-HEINEMANN*

313 Washington Street
Newton, MA 02158
(617) 928-2500
Fax: (617) 928-2620

Linacre House
Jordan Hill, Oxford OX 2 8DP
44-865-310366
Fax: 44-865-310898

In addition to publishing the Society's journal, Journal of Electromyography and Kinesiology, Butterworth Heinemann publishes a number of journals of potential interest to ISEK membership, including British Journal of Sports Medicine, Gait and Posture (new in 1993), Clinical Biomechanics, Isokinetics and Exercise Science, Journal of Audiovisual Media in Medicine, and Neurorehabilitation. Free sample copies are available to qualified individuals upon request to one of the two addresses above.

▼ *DATAQ INSTRUMENTS, INC.*

150 Springside Drive, Suite B220
Akron, OH 44333
(800) 553-9006 Fax: (216) 666-5434
Contact: Roger W. Lockhart

Dataq Instruments has supplied the PC-based instrumentation needs of medical research and instruction for over 10 years. We offer a wide range of data acquisition hardware and software, as well as complementary products such as amplifiers for cardiovascular and electrophysiology applications. Our data acquisition products complement or replace traditional chart records, tape records, and oscilloscopes in applications that need computer-readable data. We offer the industry's...

...most dependable disk streaming software--digitize and record to hard disk as many as 240 channels of analog information for any length of time.

...most readable real time display--up to sixteen channels in a smooth scroll or triggered sweep format while you record to disk.

...most coveted graphical playback and analysis software--review any size data file at any degree of compression with statistical, frequency, and calculated waveform analysis.

...most unique multi-tasking environment for Window--record, playback, and analyze data (even from the same file) at the same time.

With thousands of systems placed world-wide, we've firmly established ourselves as the best alterative to traditional instrumentation. Contact Dataq Instruments and let us show you how to enhance the efficiency and flexibility of your laboratory or classroom using PC- based instrumentation.

▼ *KISTLER INSTRUMENT CORPORATION*

Biomechanics Group
75 John Glenn Drive
Amherst, NY 14228-2171
(716) 691-5100 Fax: (716) 691-5226

Kistler provides instrumentation to the Biomechanics industry. Our piezoelectric technology is used to measure force and acceleration in order to analyze, assess, rehabilitate and study humans in the area of gait and sport performance.

▼ *NORAXON USA, INC.*

13430 N. Scottsdale Road, Suite 104
Scottsdale, AZ 85254
(602) 443-3413 or (800) 364-8985
Fax: (602) 443-4327

Noraxon produces EMG instruments with proprietary signal processing that eliminates artifact. Signal processing is notchless without site preamplification, accommodating surface and invasive passive electrodes. Functions provided include multiple channel telemetry, fixed cable and hand held instrumentation. A full compliment of scientific data acquisition and analysis software is available. The Noraxon U.S.-manufactured telemetry will be introduced at this year's ISEK conference.

▼ *TEKSCAN, INC.*

Tekscan, Inc.
307 West First Street
South Boston, MA 02127
(617) 464-4500
(617) 464-4266

Tekscan manufactures the F-Scan Bipedal Dynamic In-Shoe Pressure Measurement System for diagnosis and evaluation of foot function during various physical activities. The F-Scan is a bipedal system incorporating a .007 inch thin film sensor which covers the entire plantar surface. It can be cut to the shape and size of the subjects foot and placed within their footwear. The F-Scan allows you to see foot function without interfering with your patient's normal gait pattern. As your patient walks or runs the F-Scan records actual real time forces which can then be assessed both qualitatively and quantitatively.

INDEX