Electrophysiological Kinesiology

Progress 4

Proceedings of the 11th congress
of the International Society of
Electrophysiology and Kinesiology
in Enschede, The Netherlands 1996

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RRD PROGRESS IN REHABILITATION SCIENCE: 4

ELECTROPHYSIOLOGICAL KINESIOLOGY

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PREFACE

It was an honour and a great pleasure to organise the Eleventh Scientific Congress of the International Society of Electrophysiology and Kinesiology (ISEK) in Enschede, the Netherlands.

The society aims to promote and teach the disciplines of electromyography and kinesiology in normal, pathological, and experimental conditions emphasizing interactive use of the two disciplines. In keeping with the tradition, the scientific program included many presentations as well as keynote speeches focused on areas like surface electromyography, movement analysis, motor control, posture control and rehabilitation. In addition, there were also sessions on low back and ergonomy. The following keynote presentations were made:

- Prof. Davis (United States) - Reflections on clinical gait analysis.
- Prof. Rau (Germany) - Principles of high-spatial-resolution surface EMG (HSR-EMG) single motor unit detection and the application in the diagnosis of neuromuscular disorders.
- Prof. Stegeman (The Netherlands) - Structural parameters of motor units detected from surface EMG topography: comparison with needle EMG techniques.
- Prof. Merletti (Italy) - Surface EMG signal processing.
- Prof. Marras (USA) - Spine loading during whole-body free dynamic lifting.
- Prof. Handa (Japan) - Current topics in clinical FES in Japan.

For the first time there was a very special lecture, the Basmajian lecture. It was given by Prof. Steven Wolf from the USA. He reflected upon the legacy of John Basmajian, one of the founders of the society.

The first chapter consists of all the keynote presentations. Following chapters are arranged according to themes of the scientific programme. The programme included various themes and the presentations made under the themes are compiled and presented in the chapters.

We hope that you enjoy reading the book as much as we have enjoyed compiling it together.

Hermie Hermens

Anand Nene

Enschede, october 1996

CONTENTS

CHAPTER 1: BASMAJIAN LECTURE AND KEYNOTE LECTURES

CHAPTER 2: MOTOR CONTROL

CHAPTER 3: REHABILITATION

CHAPTER 4: MOVEMENT ANALYSIS

CHAPTER 5: ELECTROMYOGRAPHY

CHAPTER 6: MUSCLE FATIGUE

CHAPTER 7: ELECTRICAL STIMULATION

CHAPTER 8: LOW BACK / ERGONOMY

CHAPTER 9: POSTURE

CHAPTER 1

BASMAJIAN LECTURE
AND
KEYNOTE LECTURES

THE FIRST BASMAJIAN LECTURE

REFLECTIONS ON THE BASMAJIAN LEGACY

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Introduction

During its tenth meeting in June, 1994 the International Society for Electrokinesiology announced the creation of the John V. Basmajian Lectureship, in honor of one of the Society's founders and president from 1968-72. It seemed only appropriate that the first lecture acknowledge the man who helped shape much of the basic and applied research efforts underlying contemporary use of electromyography. Indeed, few can question that John Basmajian's lists of contributions to anatomy, kinesiology, electromyography and rehabilitation medicine are nothing short of extraordinary. With over 370 publications, 64 books and editorship responsibilities for an additional 24 volume series, the Rehabilitation Medicine Library, this remarkably prolific writer and researcher has recently received his country's highest honor, Officer of the Order of Canada. In addition, Dr. Basmajian has received 17 international honors and is listed in more than 20 "Who's Who". As one of John's past graduate student and collaborators, I sometimes wonder what my life or that of my colleagues in ISEK would have been like had he not contracted pleurisy during his surgical residency training in Toronto, an event that dramatically redirected his professional orientation. I would like to believe that, even as a surgeon, his love of anatomy and the understanding of factors controlling movement would have steered him toward electromyography and kinesiology anyway.

John is viewed by his colleagues and friends in ISEK with tremendous respect. The many scientific contributions he has given to us are surpassed only by his sagacious and compassionate countenance. His leadership and counsel have left us the beneficiaries of his brilliance. It is only fitting that this first Basmajian lecture highlight John Basmajian as anatomist, electromyographer and clinical scientist.

The Anatomist

As an anatomist, John Basmajian represents a breed of lecturer slowly reaching extinction. Taught by two of the most distinguished anatomy and histology teachers of all time, John C.B. Grant and Arthur Ham, they imparted to Basmajian a fervor for teaching anatomy that grew to an inextinguishable passion.[1] In these days when molecular studies and genetic engineering dominate graduate and professional school science curricula, few faculty are trained to teach gross anatomy. Among those faculty who still perpetuate the art of teaching this subject, few can do so with the conviction and ingenuity that engrossed John when his presence invaded the lecture hall. His skill in building an entire lecture from one series of superimposed drawings left many medical and graduate students, including me, in absolute awe. Beginning with a faint outline of the relevant osteological structures, John would next superimpose appropriate vessels and nerves followed by muscles and lymphatics. Even more impressive were his dissection demonstrations. First appearing as a casual observer while students obsessively ventured into the dissection of a body segment, John could never resist the desire to approach the body and quickly usurp scalpel and hemostats while simultaneously providing a lecturer, not only of the material being studied but, at no additional prompting, a detailed history of the anatomists after whom relevant structures had been named. It was not surprising to see thirty to forty students soon flocked around the table anxiously hoping to gather a glimpse of this unique presentation.

Appropriately, John took over Primary Anatomy for its third edition in 1955 and co-authored it with H.A. Cates. Thereafter this text was nurtured by "Dr. B", as his graduate students liked to call him, through four more editions and ultimately translated into French, Spanish, Italian and

Chinese. The text was known for its simplicity and the ease in which anatomy was written for health related professional students. In 1965 Basmajian co-authored the seventh edition of Method of Anatomy with his mentor J.C.B. Grant, and honored the latter by changing the name of the text to Grant's Method of Anatomy. This text became quite popular among medical and dental students throughout the world and, in 1989, reached an eleventh edition in collaboration with C.S. Slonecker. What made this text so popular was predictably due to the Basmajian touch; for in the wake of traditional anatomy textbooks that were distinctive predominantly through reformatting of factual material, "Grant's Method" offered simplistic conceptualizations that reinforced ways to learn anatomy. For example, I still recall quite vividly, Basmajian's superimposition of a horse rider sitting atop the sella turcica, his legs straddling its sides and arms outstretched, the entire ensemble cleverly depicting the course of the internal carotid artery through the foramen lacerum. Or positions of the human wrist during hand grip to demonstrate the role of the forth and fifth metacarpophalangeal joints as powerful hinges.

For generations preceding the creativity afforded through learning anatomy from CD ROM or virtual reality computer simulations, the practical lessons and insights blending surface anatomy with astute observation readily extracted from Grant's Method served as a distinctive vehicle from which to learn human anatomy. Throughout the early 1960's, Basmajian also served as a stellar advocate for the importance of teaching anatomy with numerous publications on the subject,[2-4] culminating in a publication on the importance of quality teaching that appeared in no less a prestigious journal as Science.[5] Lest, one think that the Basmajian imprint was restricted exclusively to teaching the human anatomy, it should be recalled that this keen observer also contributed substantially to the literature underlying alterations in anatomical structure. Indeed, his publications included notations on variations in venous valve structures,[6] shape of the medial semilunar cartilage,[7] and even commentary on pharyngeal constrictor muscles of the rabbit.[8]

The Electromyographer

John Basmajian's fascination with muscles and their actions provided the necessary impetus to learn about how to record from them, first from the surface and then, more specifically from within circumscribed territories within muscle using fine-wire electrodes. Possessed with a voracious appetite for discovery, John used electromyography to help define the function of human: leg and foot[9], sphincter urethrae[10], elbow flexor[11], iliopsoas[12], back[13], forearm pronator[14] and supinator[15], pharyngeal constrictor and levator palati[16], iliacus[17], tongue[18], popliteus[19], teres minor[20], quadriceps[21], longus coli[22], and lip[23] muscles, all within the span of approximately 15 years.

Unbeknownst to most of his contemporaries or future colleagues, Basmajian's onslaught into the activity patterns of human muscle probably represented the most intense study of this subject by one individual within such a condensed time frame. In the midst of this excursion, John laboriously gathered his observations and those of other electromyographers into a systematically organized presentation first unveiled in 1962 as **Muscles Alive: Their Functions Revealed by Electromyography**.[24] Five editions later, and in collaboration with his esteemed past graduate student and colleague, Carlo DeLuca, this text stands as a crowning achievement and as a premier entry into the world of kinesiology and human dynamic motion for any aspiring clinician or applied scientist. During the creation of the third and fourth editions of this work, I had the pleasure of witnessing the Basmajian juggernaut at work. Driven by a desire to retrieve anything written on EMG, Basmajian would painstakingly place his notes into their appropriate locations within his most recent edition of **Muscles Alive** in preparation for the next edition. His compulsion to do so was surpassed only by that of his administrative assistant, Arlene Debevoise, who possessed an uncanny sixth sense to anticipate his every move and to have the next phase of this work ready almost immediately.

One of the great contributions to foster electromyographic study came from the Basmajian laboratory in Canada with the development of the fine wire electrode[25] and their connectors (source followers).[26] These 75 um wires with their 25 um uninsulated tips could be inserted into specific muscle locations using thin, hypodermic needles and were capable of recording from exceptionally confined areas of muscle. As a result, single motor units could be examined for their

wave forms and discharge characteristics. In the process, Basmajian and his coworkers were able to provide visualization of these potentials to subjects and audio cues from audio amplifiers. With progressively less effort, subjects learned to isolate and control motor units[27-30] under a variety of conditions. Soon individuals were identified as motor unit geniuses if they developed the ability to isolate several motor units and to recruit them upon command. The collective observations offered some initial insight into conscious control over single motoneurons, at least those within the recording circumference of the electrodes. By 1963, this startling revelation opened the doors to what eventually became known as EMG biofeedback; for if individuals could learn to isolate and control single motor units with appropriate audio and visual cuing, then certainly patients might possess the potential to control whole muscles with surface electrode recordings and appropriate shaping procedures derived from visual and audio representations of muscle signals.

The Clinical Scientist

The establishment of the Biofeedback Research Society in 1969 offered the interdisciplinary vehicle necessary to promote the notion that EMG signals representing whole muscles or muscle groups could be operantly conditioned. Basmajian's strong ties to rehabilitation and his move to Atlanta in the same year to direct the Emory University Regional Rehabilitation Research and Training Center made the application of muscle feedback to patients with neuromuscular or musculoskeletal disorders a natural evolutionary process. By 1974 Basmajian had published a paper[31] in collaboration with W.J. Newton on feedback training to buccinator muscles among clarinetists.

From that point forward, the shaping of whole muscle responses among neurological patients through on-line knowledge of results progressed rapidly. In one of the first clinical EMG controlled studies, Basmajian's group demonstrated that chronic stroke patients receiving muscle biofeedback could increase ankle dorsiflexion in their affected extremity more so than with conventional therapy.[32] Furthermore, this difference was also manifest in reduced measures of spasticity in targeted muscles[33] and in changes in gait speed.[34] From these studies emerged a full fledged dedication to establishing clinical guidelines for the application of feedback training paradigms for stroke patients.[35-39]

Prior to returning to Canada in 1977, Basmajian and Wolf identified differences in paraspinal muscle activity during dynamic sagittal movements in low back pain patients compared to normal subjects,[40] an observation that spawned two decades of research into biomechanical models for low back pain. Concurrently, Basmajian teamed up with two other Emory colleagues, Gary Debacher and Mike Brown to develop feedback goniometers for hinge joints.[41]

Upon assuming the directorship of the Rehabilitation Centre at Chedoke Hospitals in Hamilton, Ontario and continuing his research career as Professor of Medicine at McMaster University, John pursued his interests in muscle biofeedback with a new set of collaborators and, in 1982, they published the first in a series of papers that highlighted the importance of behavioral interventions in facilitating the role of muscle biofeedback for stroke patients.[42,43] As a result of these observations, clinicians began to realize that muscle feedback signals were not the sole domain of the patient; rather, by being aware of physiological responses on the part of patients, clinicians could also use that information to change treatment protocols on-line or to restructure verbal interactions with their patients. In his 1989 critical review of EMG biofeedback, Basmajian[44] notes with some concern that, while much had been accomplished in this field of surface EMG activity monitoring, the art underlying its usage had still not caught up with the promise of its theoretical constructs.

The Legacy

While Basmajian's review might connote a disappointing perspective, subsequent assessments were more benevolent. Very few rehabilitation interventions have been exposed to as many controlled clinical trials, nor subjected to as many meta-analyses ,[45-47] the results of which paint EMG biofeedback in a comparatively favorable light. Perhaps the greatest accolade that could be bestowed upon the father of this technology is embedded in the recent report[48] from the Agency

on Health Care Policy and Research of the U.S. Department of Health and Human Services on post-stroke rehabilitation. In this report examining thousands of research articles, the benefits of EMG biofeedback for motor improvement among stroke patients were equivocal. While this finding may seem pejorative, the reality is that evaluations of other interventions were virtually unequivocal. In fact little objective, quantitative evidence could be shown to suggest that any of these other neuromuscular reeducation approaches possessed any clear foundation for bestowing benefits to patients. In this regard Basmajian and his successors have left a calling card to gather more extensive data to demonstrate the value of this modality. This possibility is remarkable at a time when most health care providers seek to eradicate treatment options.

In the world of anatomy, the blackboard-laden drawings that link the mind of the artist with that of the student is today shrouded in a blanket of obscurity. The Basmajian touch is transmitted by the impact this anatomist had upon those of us challenged to project our creative teaching talents into computer simulations and virtual reality demonstrations. In this regard, the mastery in conveying knowledge about the human anatomy does not fade in the specter of the Basmajians, Grants, Cunninghams and Grays but becomes transformed in the minds of today's technocrats. Last, what is it that Basmajian leaves to us at ISEK and to the vast interdisciplinary audience who seek answers in the analysis of movement through the monitoring of muscles or the resolution of kinematic dynamics? Is it any one publication or any one speech? I think not. Is it the technology encased in wire electrodes or biofeedback machines? Perhaps not. The legacy, I believe, lies in the clear challenge reflected in the Basmajian dedication to the study of movement - to strive to discover better and more effective ways to characterize motion and to improve the lives of those in whom this motion is impeded. If we look introspectively and discover within each of us a spark of initiative that ignites a sense of inquiry remotely paralleling that of John Basmajian, then the strength deeply imbued within ISEK will expand and the legacy will live on.

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REFLECTIONS ON CLINICAL GAIT ANALYSIS

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INTRODUCTION

Clinical gait analysis involves the measurement of fundamental biodynamic parameters, the compilation of these basic data into an information set, and the systematic interpretation of the compiled information with respect to the identification of deviations from "normal" patterns or values and the understanding of the causation of these abnormalities that forms the basis for the recommendation of treatment alternatives for individual patients on a case-by-case basis. This approach has found good utilization in the assessment of pathological gait where motions are often complex, multi-planar, and distorted relative to a fixed observer. By far the most prevalent use of clinical gait analysis is in the assessment of children and adolescents with cerebral palsy (CP) in treatment (predominately orthopaedic surgery) planning.[1-4] Other examples of clinical pathologies currently served by gait analysis include amputation, degenerative joint disease, poliomyelitis, multiple sclerosis, muscular dystrophy, myelomeningocele, rheumatoid arthritis, stroke, and traumatic brain injury.

Clinical gait analysis is also useful in the documentation of gait-related changes that occur because of treatment. For example, surgical procedures that have been examined include rectus femoris transfer[5] and release,[6,7] hamstring lengthening,[8] tendoachilles lengthening,[9] gastrocnemius fascia lengthening,[10] osteotomy,[11] and selective dorsal rhizotomy.[12] This clinical research is vitally important in the ongoing development of the knowledge base associated with analysis of pathological gait. The evaluation of orthotic effectiveness is also an excellent utilization of gait analysis, both on a patient-by-patient basis and also in studies that examine the functionality a particular brace design.[13]

METHODS - DATA COLLECTION

The primary technology that underpins clinical gait analysis includes optica-electronic video camera-based systems that measure the displacement of markers (either retroreflective or light emitting) placed on the patient's skin and aligned in some fashion with bony landmarks and/or particular axes of joint rotation. Multiple cameras are configured around a calibrated measurement volume. Three-dimensional (3-D) marker trajectory paths then are stereometrically reconstructed from the two-dimensional camera image data. Examples of commercially-available motion measurement systems of this type include Vicon (Oxford Metrics Limited, Oxford, England), Orthotrak (Motion Analysis Corporation, Santa Rosa, California, USA), and Elite (Bioengineering Technology Systems, Milano, Italy). While this is the most prevalent method for gait measurement in the clinical setting other technologies that have been used include high speed cinefilm cameras to record marker trajectories and electrogoniometers, electromagnetic sensors and accelerometers that measure joint/segment spatial kinematics directly. Fundamental in this approach is the definition of the relationship between the markers (or transducers) placed on the skin's surface and the underlying bony geometry. Increasingly, clinical protocols have used the data capture systems to incorporate some type of subject calibration process whereby multiple bony landmarks are identified (through the use of additional markers or instrumented pointers) while the patient is standing still. In this way, the operator is able to establish a "technical coordinate system" associated with the externally placed markers and an "anatomical coordinate system" associated with the underlying bone(s) for each body segment under examination.[14] It is then assumed that these analytical relationships between these two segmentally-fixed coordinate axes remain constant during the movement activity, i.e., that the segment is rigid. The proper placement of markers on the patient in a repeatable, reliable and meaningful manner remains the principle challenge in routine clinical gait analysis today. Improvements in technology associated with

increased measurement accuracy and precision as well as data sampling rates and processing speed have not impacted this fundamental challenge.

Other technologies that are employed in clinical gait analysis data collection include force platforms for the measurement of the ground reaction loading to the patient's foot, dynamic electromyography (EMG) that allows an evaluation of muscle activity (either with surface or intramuscular electrodes), and foot switches to monitor foot/floor contact. Increasingly, arrays of miniature force sensors are finding application in the evaluation foot plantar pressure distributions during gait. Also, the assessment of metabolic energy during walking is being sought more often clinically to provide a measure of the overall ambulatory status of the patient. This technology allows a direct evaluation of one of the primary determinants of gait, i.e., energy conservation.

METHODS - DATA REDUCTION

The products of the gait data collection process include the 3-D trajectories of markers placed on the surface of the patient's skin, ground reaction values, and electromyographic signals as well as basic measures of when and where the patient's feet were in contact with the ground. Additional clinical parameters routinely collected at the time of the gait analysis include patient height and weight, the passive motion of the hips, knees and ankles, the strength and control of the lower extremity musculature, the presence of spasticity and muscular contracture, and static bony deformity.

The marker trajectories are typically used (in combination with the subject calibration data) to compute either the absolute rotational position of a segment (e.g., pelvis) in space or the relative rotational position of two juxtaposed segments (e.g., knee joint motion).[15] This segment/joint center position data can also be combined with the ground reaction values, estimates of segment mass and mass moments of inertia, and the segment/joint velocities and accelerations to compute the net joint moments about particular joint centers as well as the associated instantaneous joint rotational power.

Electromyographic signals can be provided for interpretation in a wide variety of formats, i.e., the signal might be presented unchanged (raw), filtered, smoothed, rectified, integrated, ensemble averaged and/or scaled to other references (maximum amplitude while walking, amplitude during maximum voluntary contraction or maximum voluntary exertion). Suffice it to say that there is no standard format by which EMG information (or gait data, in general) is presented for clinical gait analysis interpretation.

METHODS - GAIT INTERPRETATION

Although the specific approach varies between facilities, clinical gait analysis interpretation often involves not only the physical therapist or kinesiologist that has collected the patient data, but also a physician experienced in gait analysis and the management of the patient's medical condition. The information that is generally provided for pathological gait interpretation includes a bi-plane video recording of the patient walking, the results of the clinical/physical examination, stride and temporal parameters such as velocity, cadence, stride length, step length and percentage of stance/swing, plots of the rotational orientations of the lower extremity joints and selected segments (e.g., pelvis) as a function of the gait cycle, plots of the hip, knee and ankle net moments and powers as a function of the gait cycle, EMG plots of selected lower extremity muscles (e.g., rectus femoris, vastii, medial and lateral hamstrings, hip adductors, gastrocnemius, soleus, tibialis anterior, tibialis posterior, and peroneals) also as a function of the gait cycle.

The objectives of the interpretation process are to appreciate how the patient's gait is different from the normal ambulator and to understand the causes and significance of these differences, both of which may lead to the recommendation of a treatment plan. The first step in this process is to identify those characteristics of the patient's gait pattern that are deemed qualitatively and/or quantitatively different from normal. This involves a systemic evaluation of each of the parameters and plots described above, taking care to note corroborating and apparently conflicting information. An understanding of the gait model used in the computation of joint angles, moments and powers is essential to fully comprehend the significance of perceived gait anomalies. Then the team seeks to appreciate the impact that those perceived deviations might have on the patient's ability to walk efficiently and with adequate control, balance and stability.

Once the functionally significant abnormalities have been determined, then the goal of the interpretation process is to uncover the cause(s) of each gait deviation. Experience allows the interpretation team to recognize "secondary" gait deviations that are produced by "primary" gait deficiencies and to differentiate between primary gait problems (that perhaps need to be addressed) and compensatory gait mechanisms that the patients employ to walk. For example, in a patient with hemiplegia, one may see premature plantar flexion in mid stance (i.e., a vault) in order to compensate for reduced sagittal plane motion of the swing limb. This volitional compensation often appears similar biodynamically to a spastic stretch-reflex response of the ankle plantar flexors.

After the biomechanical etiology of the primary gait deviations has been established then the interpretation team may recommend treatment options that may include "do nothing, but continue to follow", physical therapy, spasmolytic medications (e.g., Baclofen), orthotic management, botulinum toxin, surgical (orthopaedic or neurosurgery) intervention. It is important to appreciate that a "gait analysis" does not specify a particular treatment direction, e.g., an involved surgical intervention. Treatment recommendations, while hopefully supported by the gait analysis findings, are produced in the context of treatment philosophy of either the interpreting physician, the patient's referring physician, or both.

This process is often challenging because of the complexity of the motion, the neuromuscular involvement of the patient, e.g., the possible presence of dynamic spasticity, the variability of treatment outcome, and on occasion, uncertainty about the quality of the gait data. Much of the success of this process is dependent upon the experience of the interpretation team with respect to gait biomechanics, a particular patient population, and the effects and effectiveness of different treatment modalities.

DISCUSSION

The use of gait analysis technology represents largely an outgrowth of the approaches employed in the clinic for observational gait assessments. The goal in the clinic is to observe and note gait abnormalities and where possible estimate the magnitude of the observed deviation. The clinician endeavors to both identify the cause(s) of the gait pathology and appreciate its consequences functionally. To explore this biomechanical etiology, the clinic observer typically attempts to correlate these observed gait characteristics with other clinical/physical examination measures such as passive joint range of motion and muscle strength. The initial thrust of technologically based gait analysis was to document parameters that could either be relatively easily obtained such as stride length and walking velocity or those that could also be "observed" such as segment displacements and joint rotations (particularly in the sagittal plane). In other words, the initial uses of this technology focussed on quantifying those variables that the clinical observer could only estimate previously.

The use of the quantitative gait information in the interpretation process is an evolving entity. The variables examined in the interpretation process are no longer limited to those which can "seen" as the patient walks or visualized easily. The use of joint kinetic (moments and powers) allows an appreciation of underlying biomechanical mechanisms. Ironically, this information takes more of a cognitive investment to be usefully integrated into the interpretation process as results are sometimes counterintuitive. In general, the analysis of

the gait data has become more comprehensive. For example, a decade ago, the interpretation team might have noted that a particular joint rotation magnitude (e.g., peak knee flexion in the swing phase of the gait cycle) was reduced or its maximum value reached relatively late in the gait cycle. The discussion at that point might then have focussed on the role of the muscles or tendons that cross that particular joint to ascertain the cause of the deviation. The clinicians now take advantage of an improved appreciation of the role of two joint muscles, the impact biodynamically of abnormal muscle tone and spasticity as well as the dynamic interaction between joints, between limbs, and between planes of motion. The cause of the reduced knee flexion in the swing phase might be sought by exploring the capabilities of the plantar flexors to accelerate the ankle (and thus the knee) upward and forward or the hip flexors to accelerate the thigh (and thus the knee) forward in late stance. This has come about because of an improved understanding of normal gait biomechanics, higher quality gait data, and years of experience associated with the distillation of the biomechanical details associated with countless clinical cases.

Of course, there is much more to learn about the gait biomechanics of both normal and pathological gait. Current interpretation practices involve the subjective correlation of various data types. For example, to evaluate the function of the gastrocnemius during gait, the ankle angular position, moment and power during early stance are examined along with the corresponding EMG data and appropriate clinical measures, e.g., passive range of motion of the ankle with the knee extended fully and flexed to ninety degrees. This approach is good, but could be strengthened with better quantitative tools that allowed for a more rigorous numerical determination of these correlations. Furthermore, an appreciation of the interrelationships between joints and the influence that a particular joint motion has upon segments away from that joint could be approached more systematically and quantitatively in the future. Clinical research that involves a thoughtful, systematic evaluation of pre- and post-treatment changes in pathological gait patterns continues to help broaden our understanding of the often complex interrelationships and synergies. A challenge for those involved in muscle mechanics research is to provide a better understanding of the relationship between electromyographic signal characteristics (e.g., amplitude, timing of onset, frequency content, phase shifts) and the generation of contractile force in muscles, particularly in the context of muscle "over activity" and spasticity. These types of contributions would be of immense value to those involved in the analysis of pathological gait patterns.

Although gait analysis approaches are well utilized at a number of clinical centers around the world, its incorporation as part of routine clinical decision making remains hindered by its association with a very aggressive treatment approach, i.e., complex surgical intervention. Unfortunately, too many potential users of clinical gait analysis fail to appreciate its role in other treatment approaches, e.g., bracing and less involved surgery. Furthermore the information supplied by gait analysis is not easily understood and requires a rather substantial commitment by the clinician in order to fully appreciate its value. Finally, the information supplied for interpretation is sometimes counterintuitive and may be discarded as inaccurate unless the interpretation team has sufficient confidence in the data collection and reduction processes.

Despite these ongoing challenges, gait analysis technology has continued to improve over the past two decades with faster systems that deliver more and higher quality data. Measurement protocols for pathological gait assessment have been significantly refined over the decade so that data can be collected reliably and practically. These improvements impact positively the interpretation process by allowing the clinical team to focus on better investigate the subtleties of each patient's gait pattern. The knowledge base associated with pathological gait has grown during this time as well. Much progress has been realized with respect to clinical gait analysis, providing a foundation upon which to explore its greater utilization in patient care.

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PRINCIPLES OH HIGH-SPATIAL-RESOLUTION SURFACE EMG (HSR-EMG) SINGLE MOTOR UNIT DETECTION AND THE APPLICATION IN THE DIAGNOSIS OF NEUROMUSCULAR DISORDERS.

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INTRODUCTION

Surface electromyography (EMG) is used extensively as a noninvasive procedure. It reflects the electrophysiological activity of a muscle while the population of the excited muscle fibres, as electrical sources, generate a potential field on the skin surface forming the boundary of the volume conductor of the human body. Since the motor unit (MU) is the smallest unit of excitation and contraction [2], the most detailed information about the structural and functional characteristics of the muscle, like the functional anatomy, the excitation spread or the innervation pattern, can be gained from the single MU action potential. Additionally, the information about the single MU activity is essential for the diagnosis of neuromuscular disorders, since they are often related to typical changes in the structural and functional properties of the single MUs.

The conventional surface EMG signal is a superposition of a large number of MUs located within the muscle in various geometrical arrangements and activated in a complex interference pattern. The difficulty herewith is to separate the activity of a single MU from simultaneous active adjacent ones. The conventional surface EMG regarding arrangement lead systems consists usually of large electrodes (Diameter 5 - 10 mm) in a bipolar arrangement (interelectrode distance at least 10 mm) [1]. Due to the low spatial resolution of the recording set-up, the standard surface EMG reflects only the compound activity of a high number of MUs. Therefore, the conventional surface EMG is used for obtaining information about the time and the intensity of superficial muscle activation with well justified application in biomechanics, biofeedback or rehabilitation. But the conventional surface EMG is not suitable for the detection of the single MU activity.

Invasive recording techniques, like needle- or wire electrodes, have a higher spatial resolution. In contrast to the conventional surface EMG they allow the detection of the single MU activity but the insertion of the needle or the wire causes discomfort, medical supervision and risks of infection. The repeatability of the measurements is worse, and long term monitoring is hardly possible [6]. Additionally, the conventional needle EMG provides usually no information about the excitation spread in single MUs. The wire EMG overcomes some problems of the needle EMG, e.g. reduces the painfulness and increases the information about the excitation spread. But this method is hard to use in practice because after insertion the wire electrodes tend to migrate within the muscle tissue during contraction. Therefore, the reproducibility of the EMG signal detected with wire electrodes applies more to parameters of the total EMG signal (such as amplitude) than to the more sensitive parameters of the individual MUs [1,6]. One major advantage of the wire electrodes is the possibility to record muscle activities fron deeper located muscles or muscle portions even during dynamic contraction.

The potential distribution generated on the skin surface by voluntary or electrically elicited muscle activity contains the information about the single MU activity. Consequently, the single

MU activity becomes noninvasively detectable when the spatial resolution of the noninvasive recording set-up is improved. A first step on this way was the development of active electrodes which allow the detection of very low signal amplitudes [1,11,14]. This technique makes the reduction of the interelectrode distance possible and improves in this way the spatial resolution of the recording set-up [8]. Another approach is based on concentric bipolar systems [21]. They consist of a central electrode surrounded by a concentric ring. The concentric systems as well as the bipolar lead with an interelectrode distance in the mm-range allow the noninvasive detection of the single MU activity, but the procedures are limited up to 20% maximal voluntary contraction of the muscle. Beyond this force level, too many MUs are activated forming the well known interference pattern, while decomposition of the time signal to isolate single MUAP is very difficult.

The excitation spread in single MUs becomes noninvasively detectable since electrode arrays, consisting of pin-electrodes arranged two dimensionally with an interelectrode distance in the mm-range are used. In electromyography such electrode arrays have been first used by Masuda et al. (1985) [9], De Luca et al. (1987) [3] and Reucher et al. (1987) [15] and become recently more and more important [13,19]. Some groups, as Stegeman et al. [17], have combined the already introduced electrode arrays with the invasive needle-EMG to get further information about the single MU activity. However, this procedure is invasive, because an inserted needle is necessary to pick up the activity of a single MU which is used as a trigger for the MU detection by the electrode array located on the skin surface. The averaging procedure used for signal processing is based on a considerably long observation duration which is a problem in voluntary contraction.

The noninvasive detection of the single MU activity, even during maximal voluntary contraction of the muscle, becomes for the first time possible when the electrode arrays have been used in combination with a two dimensional spatial filter processing [16]. This approach is, in fact, an extension of specialised bipolar [10] or tripolar [3] lead systems. The so called **High-Spatial-Resolution EMG** (HSR-EMG) opens access to the study and analyse of a variety of parameters of the single MU.

METHOD

A spatial potential distribution varying with time is generated on the skin surface by the excitation of the muscle fibres. When using a two-dimensional multi-electrode array, the potentials picked up instantaneous by the different electrodes can be regarded as spatial samples of the potential distribution. A theoretical model considering the muscle anatomy as well as the excitation spread is based on an electrical tripol [5,12,16] moving with conduction velocity along the longitudinal axis of the muscle fibres. The maximal interelectrode distance needed to get the complete information about the potential distribution can be calculated from this model [15]. It is shown, that for superficial muscles, like for example, the biceps brachii muscle or the abductor pollicis brevis muscle, this goal can only be achieved when interelectrode distances in the mm-range (e.g.: 5 mm for the biceps muscle; 2.5 mm for the m. abductor pollicis brevis) are used [15]. The interelectrode distance has to be adapted to the muscle and the volume conductor dimensions which influence the electrical field properties on the skin surface respectively.

The principle of spatial filtering is based on the fact that MUs located closely to the skin surface contribute to the potential distribution on the skin surface with a spatially steep potential gradient. In contrast to that, the activity of the more distant MUs as well as far field electrical disturbances generate a spatially flat potential course (Fig. 1). A bipolar lead with an interelec-

trode distance in the mm-range which forms in principle the lowest order of a spatial filter differentiates the potential distribution in the direction of the electrode configuration. In this way, the slope of the potential distribution generated by MUs located closely to the skin surface contributes more to the resulting difference signal than the low slope of the potential distribution of more distantly located MUs (Fig. 1). The bipolar lead with an interelectrode distance in the mm-range amplifies the activity of MUs located closely to the skin surface and reduces the activity of more distant sources.

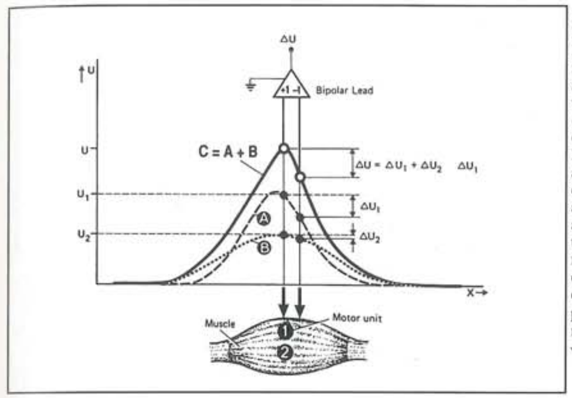

Fig. 1: MUs located closely to the skin surface (1) produce a spa-tially steep potential gradient (A) in the recording area; distantly located MUs (2) a flat potential course (B). Using a bipolar lead with a small interelectrode distance the flat potential course contributes only with a small part (ΔU_2) to the measured total value ΔU.

Due to the spatial filtering this bipolar lead allows the detection of the single MU activity only at very low voluntary contraction level of the muscle when very few MUs are active [11,16]. However, the spatial resolution of the bipolar lead is not sufficient to discriminate the single MU activity at higher contraction forces (Fig. 2). Therefore, this first approach of spatial filtering has been extended to three or more unipolar leads [3,16]. A weighted summation of three EMG-leads in a row performs a spatial double differentiation of the spatial potential distribution. Because of the double differentiation this filter separates more effectively the activity of single MUs located close to the surface from those MUs located more distantly from the skin (Fig. 2). The spatial selectivity of the double differentiating filter is high in line of the electrode direction and, therefore, in the orientation of the filter. Latteral distantly located sources may contribute in addition to the signal course (Fig. 2).

This disadvantage can be avoided when a spatial electrode arrangement with 5 electrode leads is used. Such a spatial filter is called 'Normal Double Differentiating Filter' (NDD-Filter) [16]. The five unipolar leads of the NDD-Filter are arranged crosswise (Fig. 2). To build a spatial filter each electrode of the filter must be weighted with a factor determined from the Laplace equation and approximated by the Taylor-series (the central electrode is weighted with a factor of -4 and the surrounding electrodes with a factor of +1). This NDD-Filter improves the spatial selectivity of the recording set-up in all directions and allows in this way the noninvasive separation of the activities of single MUs even at maximal voluntary contraction (Fig. 2) as has been shown previously [16].

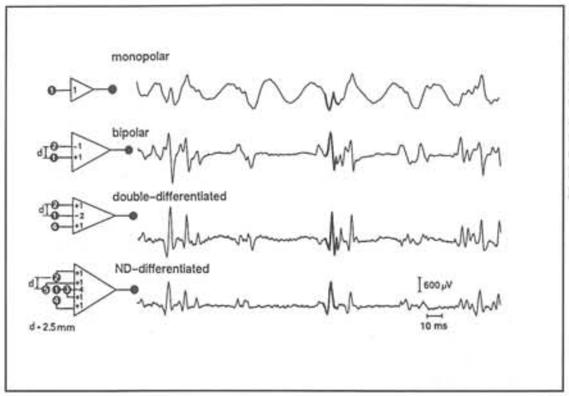

Fig. 2: Representation of 4 different EMG-leads recorded from the m. abductor pollicis brevis at maximal voluntary contraction. One excitation of a MU has been emphasised.

For the realisation of the spatial filters very small electrodes are required. The electrode array of the HSR-EMG consists of 32 - 64 special pin-electrodes with a very small surface (0.5 mm diameter, interelectrode distance 2.5 - 5 mm). This results in an extremely high impedance between skin surface and electrodes. To get still an acceptable signal to noise ratio, the electrode should be directly connected to a preamplifier with high integration and extremely low input capacitance. This technology has not been available and had to be specially developed for the HSR-EMG. The EMG-data of each channel of the array were recorded unipolarly at a sampling rate of 4000 Hz each. After a 10 Hz high-pass filtering of the unipolar data the NDD-Filter was applied using a software implementation [4].

BENEFIT PROVIDED BY THE HSR-EMG APPROACH

From one NDD-filtered channel the activities of superficial single MUs can be recorded as a function of time. To obtain additional information about the excitation spread along the muscle fibres of single MUs, electrode arrays are used which provide several NDD-filtered channels in a lined row [19] (Fig. 3). If this row of NDD-filtered channels is orientated in parallel to the muscle fibres the excitation spread in single MUs can be recorded noninvasively. Masuda et al (1986) using electrode arrays in combination with bipolar leads have shown the principle excitation spread detection before [10]. But due to the low spatial resolution of the bipolar recording set-up this approach is limited to low contraction levels of the muscle.

The parallel orientation of the electrode array in line to the longitudinal axis of the muscle fibres can be found and controlled easily: Because of the high spatial selectivity of the recording set-up the maximum amplitude of the peaks representing the activity of a single MU approximately has to be constant in all channels in line (Fig. 3). In this way, the detection of the fibre orientation can be performed in practical application by utilising this very sensitive criterion which can easily achieved by some experience with the set up. The accuracy and repeatability are very high.

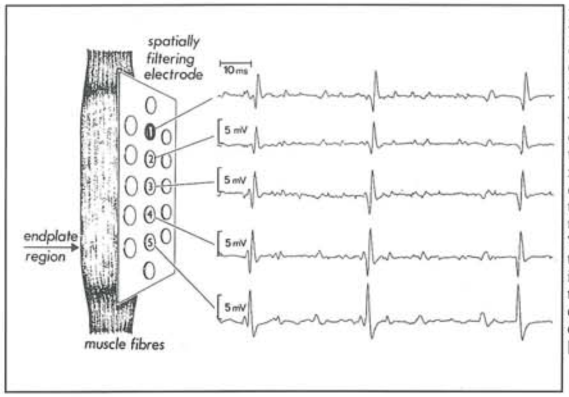

Fig. 3: Noninvasive recording of the excitation spread in single MUs. The conduction velocity can be calculated from the time delay of the response and the distance between two spatially filtered channels. The localisation of the endplate region is characterised by the spread of the excitation in two opposed directions. [Schneider 1989]

The knowledge about the excitation spread opens now access to some important information. From the time delay of the response of one MU in two different spatial filtered channels and their spatial distance, the conduction velocity in single MUs can be calculated (Fig. 3). By this, the HSR-EMG procedure is the only technology which provides the conduction velocity in single MUs in a noninvasive way even at maximal voluntary contraction of the muscle. Conduction velocity of a whole muscle is not defined since the individual MU's conduction velocity values differ considerably. Therefore, it is essential to separate the single MU activities. Conduction velocity measurements of individual MUs play an important diagnostic role [13]. For this reason temperature has to be controlled very carefully, since it has a marked influence [18,20].

The localisation of the endplate region is characterised by the spread of the excitation in two opposed directions. Based on this circumstance the endplate region can be localised with the HSR-EMG with an accuracy lower than the interelectrode distance of the array.

Since the HSR-EMG allows the noninvasive detection of the single MU activity even during maximal voluntary contraction of the muscle, the procedure is well suited for investigations of the innervation pattern of the muscle. The firing rate of single MUs as well as their recruitment can be determined by the HSR-EMG in a noninvasive way.

Neuromuscular disorders are often related to typical changes in the structure of single MUs. Therefore, the diagnosis of those disorders is essentially based on the information about the electrical activity of single MUs. The HSR-EMG provides this information and is, in consequence, suitable as a noninvasive tool for the diagnosis of neuromuscular disorders.

APPLICATION IN DIAGNOSIS OF NEUROMUSCULAR DISORDERS

The aim of needle-EMG in diagnosis of neuromuscular disorders is mainly to distinguish between patients without any neuromuscular disorder, patients with muscular disorders and patients with neuronal disorders. In first clinical studies it has been examined, if the HSR-EMG allows a distinction between the three patient groups in a noninvasive way.

The HSR-EMG has been recorded during maximal voluntary contraction of the m. abductor pollicis brevis in 117 volunteers (61 healthy children, 35 patients with Duchenne muscle dystrophy, 21 patients with spinal muscle atrophy) aged between the infancy and 25 years. The results show, that the HSR-EMG allows the detection of typical changes in the electrical activity of the MUs in neuronal and muscular disorders (Fig. 4) [13].

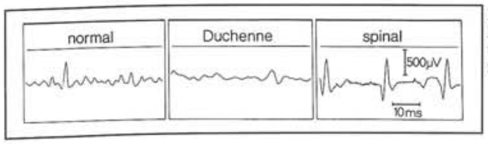

Fig. 4:
HSR-EMG Signal of children with different kinds of neuromuscular disorders.

The typical differences in the HSR-EMG pattern have been evaluated by 7 parameters regarding the excitation spread, the signal course in long signal segments, the shape of isolated peaks and the frequency distribution of all peaks within the signal (Tab. 1). A comparison between the parameter values of patients and a 'reference range' which contains at least 44 of the investigated 61 healthy volunteers show that there is a distinct difference in the parameter sets of the three groups (Tab. 1)

Parameter	Muscular Disorders		Neuronal Disorders	
	higher than reference	lower than reference	higher than reference	lower than reference
Conduction velocity	0	34	0	13
Signal Entropy	0	23	1	8
First zero crossing of ACF	35	0	4	3
Dwell time over RMS	22	1	0	16
CHI-Value of amplitude distribution	1	28	15	1
Gradient of peak edge	0	32	9	5
Peak-Frequency-Distribution	1	25	1	9

Tab. 1: Parameters for a quantitative evaluation of the typical HSR-EMG patterns and the number of patients belonging to an interval higher or lower than the 'reference range'.

These parameter sets can be used as a basis for the classification of the patients in healthy volunteers, patients with muscular disorders and patients with neuronal disorders. Here an especially developed classification procedure is used which is based on a Fuzzy-approach [7] and which has been adapted to this specific classification task. With this classification procedure 100% of all investigated healthy children, 100% of all investigated patients with muscular disorders and 87% of all investigated patients with neuronal disorders can be correctly identified. On the average the diagnosis by means of the noninvasive HSR-EMG was in 97% of all investigated children correct.

Conclusion

Using, electrode arrays in combination with the NDD-Filter the High-Spatial-Resolution EMG (HSR-EMG) provides information about the single motor unit activity in a noninvasive way even during maximal voluntary contraction of the muscle. With that, it distinguishes itself from all other noninvasive EMG-procedures. Additionally, it opens access to the excitation spread, which is usually not detectable even with needle-EMG procedures. Therefore it provides additionally the determination of the conduction velocity in single motor units and the localisation of the endplate region.

However, due to the spatial filtering the method is limited to superficial MU in superficial muscles. In spite of these limitations, the HSR-EMG detects changes in the electrical activities of the MUs which are typical for muscular and neuronal disorders. Based on a quantitative evaluation of these changes it was possible to classify correctly 97% of all investigated children. Therefore, the HSR-EMG promises to be suitable as a tool for the diagnosis of neuromuscular disorders in clinical application.

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STRUCTURAL PARAMETERS OF MOTOR UNITS DETECTED FROM SURFACE-EMG TOPOGRAPHY; COMPARISON WITH NEEDLE EMG TECHNIQUES

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INTRODUCTION

Motor units (MUs), the elementary functional units of muscle, can be characterised morphologically (cross-sectional area, fibre density, fibre typing, end-plate position, fibre length) and functionally (recruitment threshold, contraction force, firing frequency). A large number of known neuro-muscular diseases show substantial changes in MU structure and function. Since MUs are purely functional units of a muscle, both morphological and functional MU properties can hardly be estimated by other than electromyographical (EMG) techniques.

Different needle electrode techniques provide information about different structural aspects of the MU (Stålberg, 1986). Well-known are the "concentric needle EMG " and the "monopolar needle-EMG " with electrode areas of less than a square mm. With these electrodes MU activity can easily be discerned at low force levels. A general view of a MU can be obtained by "Macro-EMG". With its large recording area, it is expected to inform about the whole MU (Stålberg, 1980) and is often used to estimate the number of fibres in the MU, expressed as MU-size. A more elaborate technique to obtain structural MU properties is "Scanning-EMG" (Stålberg and Antoni, 1980; Gootzen et al., 1992).

Mainly due to the lack of easily available direct information on individual MUs, Surface-EMG techniques are hardly used to estimate MU characteristics in the EMG laboratory. The non-invasive character of Surface-EMG has advantages compared to needle-EMG: it is less stressful for patients giving the opportunity of studying more muscles, it is easily applicable in children, and there is no danger of infections. Therefore, we aim to explore where and how information at a motor units level can be obtained from Surface-EMG techniques and how possible results can be applied in clinical neurophysiological and also in routine kinesiological laboratories.

The problem in studying single MUs with Surface-EMG is twofold: First, the isolation of single MUs from the interference pattern, and second the dependency of the action potential's shape and amplitude on the MU position relative to the skin-surface. Averaging is necessary to study single MU activity in Macro- and Surface-EMG. Although, like in standard Macro-EMG, a Single Fibre electrode can be used for triggering purposes, the essence is, of course, to avoid needles in combination with Surface-EMG. A trigger signal from Surface-EMG can be obtained by applying a spatial filtering technique to a two dimensional array of Surface electrodes with small leading off surfaces (Reucher et al., 1987a, b). This search for a proper "needle-less" trigger signal needs ongoing attention. The second aspect, the basic knowledge of the representation of MU information in Surface-EMG, was a main topic in our recent research and is the subject of the present contribution. We studied motor unit potentials (MUPs) obtained with Scanning-EMG or Macro-EMG and compared them with MUPs from

Surface-EMG topograms, whereby a single fibre needle recording is used for triggering purposes.
The first series of experiments, using Surface-EMG topography in combination with Scanning-EMG, serves an understanding of the relation between a more detailed presentation of MU properties and Surface-EMG-MUPs. The Scanning-EMG results are then considered as a "golden standard" with respect to depth and size of the MU. Macro-EMG is generally used to give an estimate of what is called "MU-size" by calculating amplitude or area of the Macro-MUP. Surface-EMG can in principle serve that same goal. In the second experimental series, where Macro-EMG is combined, the results are used to compare the MUP area from Macro- and Surface-EMG of individual MUs.

METHODS

Subject selection

Healthy volunteers without signs of neuromuscular disorders were investigated. The m. biceps brachii was investigated, because of its well defined structure with fibres parallel to each other and to the skin surface. All subjects gave their informed consent. The Committee on Experiments in Humans of the Faculty of Medicine saw and approved the experimental protocol.

Electrodes and montage

Thirty gold coated screws, with a diameter of 1.2 mm, were mounted into an electrode holder and used as surface electrodes. The electrode holder was constructed from identical elements. Each element was made of a perspex body in which four recording electrodes could be placed with an inter-electrode distance of 6 mm. In the centre of each element a circular hole was made to provide simultaneous use of EMG needles.

Electrode placement was done without the use of electrode paste, because conventional electrode paste short-circuits the electrodes. The 30 surface electrodes were placed over the muscle, such that two electrode columns with 15 electrodes were placed in alignment with the muscle fibres. Double electrode rows were used to check signal changes for consistency (see Figure 1). Surface-EMG channels were amplified with a Neurotop-32® (Nihon Kohden) amplifier or an amplifier of Vision Research (Amsterdam).

There is a persistent habit in the world of electromyography to use bipolar recording montages. This is in contrast to most other electrophysiological procedures (EEG, ECG), for which a common "referential" recording is applied, allowing the selection of different montages during or after the measurements have been made. We will show that especially for topographical techniques, the information content in the signal is much better preserved by a referential recording with a far-away common reference electrode. So, all the presented results are primarily obtained in that way. Additionally, two conventional surface electrodes with a diameter of 1 cm were attached to the skin with electrode paste on the elbow to serve as common reference for the unipolar recording and to the hand to validate the elbow as a silent reference electrode.

Simultaneously, in the first experimental series a concentric needle was applied together with a single fibre needle to allow Scanning-EMG . In the second experimental series, a Macro-needle was applied together with the surface electrodes. The single fibre electrode or the small

surface in the Macro-EMG electrode were used to record action potentials from one muscle fibre to generate a trigger pulse on each MU firing. In the second experiment, the common reference electrode was also used to obtain unipolar Macro-EMG recordings. A conventional ground electrode was placed at the wrist.

Procedure

The experiments were performed on the voluntary, low force (up to 10%MVC), isometric contracting left biceps brachii. During the whole experiment, the subject lay on a bench with the upper arm 40 degrees abducted, the forearm supinated and the elbow angle at 100 degrees while the hand and lower arm were fixed. For displaying the needle electrode signals, and for obtaining a trigger pulse from the single fibre electrode, a Medelec MS-20 (Mystro®) was used. As soon as the single fibre electrode or the single fibre surface of the Macro electrode detected a single fibre action potential with stable firing characteristics, this motor unit was further studied.

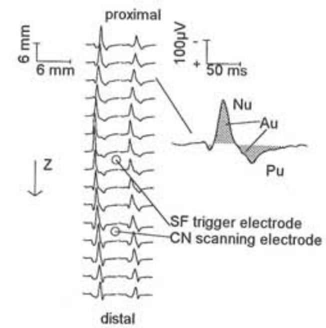

Figure 1: Spatio-temporal profile along the skin surface of a motor unit of the biceps brachii muscle (electrode spacing 6mm). Double electrode rows were used to check for consistency and to locate the MU more precisely. Negative potentials are plotted upwards.The parametrisation of the MUPs is indicated in the inset, as are the locations where the needle electrodes are inserted in the first experimental series.

Data Acquisition

The Scanning-EMG protocol was a slight modification of the procedure described elsewhere (Gootzen et al., 1992). Macro-EMG was also recorded in a conventional way (Stålberg, 1980). The surface-EMG signals and the Macro-EMG signal were obtained after amplification over a frequency range of 5-800 Hz and A/D converted with 2000 samples/s. The signals from the concentric needle and the single fibre needle were obtained in a larger (5 or 500 to 8000Hz) frequency band. The motor unit activity at the skin surface was elevated from the ongoing "background" EMG by using the single fibre signal as a trigger in an averaging procedure.

RESULTS

MUP magnitude and MU depth

Figures 1 and 2A show a typical example of the MUP distribution on the skin-surface generated by a rather superficial MU. For all the averaged surface MUPs, the following magnitude parameters were determined (inset to figure1): peak values of the (*u*nipolar) negative wave (*Nu*) and of the final positive wave (*Pu*) and total area (positive and negative) below the curve (*Au*). From the measurements, bipolar montages were constructed. An example of the result is presented in Figure 2B. From these *b*ipolar recordings the peak-to-peak amplitudes (*Ppb*) are measured.

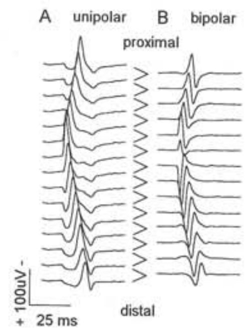

Figure 2: A: Spatio-temporal profile along the skin surface of a motor unit of the biceps brachii muscle (left row of Figure 1). B: bipolar recordings calculated from the same data. Negative potentials are plotted upwards.

In the first experiment, 55 MUs from nine healthy volunteers were investigated together with the Scanning-EMG protocol. The relation between the different magnitude parameters of the surface MUPs and depth (*D*) of the MUs is shown in Figure 3A, B, C for the unipolar surface MUPs recorded by the most proximal electrode in the electrode array (top Figure 2A) which had the largest average magnitude. The MU depth *D* was estimated with Scanning-EMG as an "electrical MU centre" (Roeleveld et al., submitted). A clear decrease in MUP magnitude with increasing depth can be observed. The solid line is the power fit performed in the (linear) log-log domain through the data resulting in the inserted power Q (for the positive peak amplitude: $Pu \approx D^{-Q}$). The standard error (se) of Q is also inserted and varied between 0.11 and 0.17. Although Q from the most proximal electrode was the highest, not a large variation could be observed along the muscle fibre direction (Roeleveld et al., submitted). Figure 3D shows that the bipolar MUPs recorded with a small inter-electrode distance (6 mm) decrease much faster with increasing MU depth than unipolarly recorded MUPs (Q = 2.4 for *PPb*). The *PPb* data were almost perfectly fitted with the log-log power model (linear correlation in the log-log domain >0.9 against 0.72 and 0.84 for the positive and negative peaks of the unipolar recordings

respectively). In Figure 3 E-H the presentation of parts A-D is repeated, but now with a normalisation of the individual data points for the size of the specific MUs. This MU-size (in mm) is also estimated from Scanning-EMG by defining a range in the scanning track where the MU has MUP amplitudes larger than a predefined level (Roeleveld et al., in preparation).

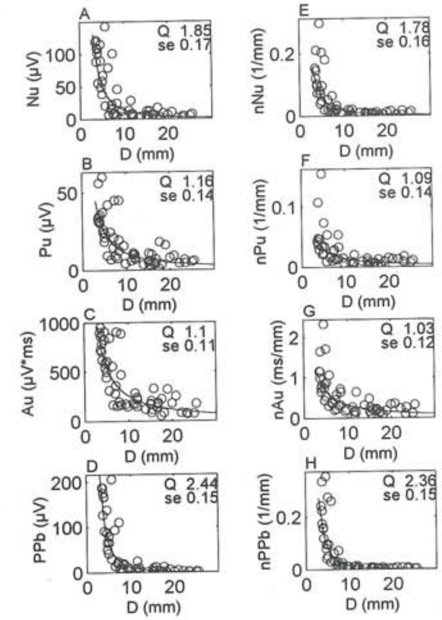

Figure 3. MU depth (*D*) - surface MUP magnitude relationship for three different unipolarly (*Nu, Pu, Au*) and the one bipolarly recorded surface MUP (*PPb*) parameters, absolute (A, B, C, D) and normalised to the size of the MU (E, F, G, H). Normalisation to the MU's size is performed by dividing the magnitude parameters by a motor unit size estimation obtained from Scanning-EMG. The "o" representing the data of individual MUs and the solid line representing a power function fitting the data. The power used (*Q*, see also text) and the standard error of this power (se) obtained from the log-log linear data are inserted in the plots. All parameters were obtained from MUPs recorded with the most proximal electrodes.

Surface-EMG versus Macro-EMG

In the second experimental series, 63 different MUs in six subjects were studied by means of simultaneous Macro- and Surface-EMG. MUPs were again averaged from the "background EMG". Generally, both the MUPs from the Surface-EMG and from the Macro-EMG needle-EMG technique is not as reliable as from the scanning-EMG experiments. These experiments were therefore extended with Surface-MUP recordings in the direction perpendicular to the muscle fibres in the m. biceps brachii. From the MUP decline in this direction an estimate from the MU depth was obtained in a way comparable to the technique described by Monster and Chan in 1980 (Roeleveld et al., in preparation). In Figure 5, the results of the MU-size expressed as MUP area is presented for the MUs investigated in the six individual healthy subjects in three ways: from the Macro-MUPs, from the unnormalized Surface-MUPS, and from the MU-depth-normalized Surface-MUPs.

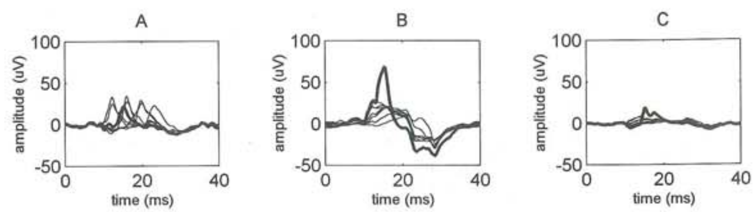

Figure 4. Examples of MUPs of different MUs. In each graph 7 MUPs recorded with surface electrodes (thin lines) placed parallel to the fibre direction are shown together with the Macro-MUP (thick line) of the same MU.

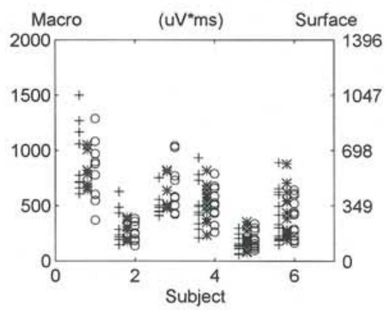

Figure 5. MU-size distributions for the 6 subjects expressed in the MUP area *Au*, obtained from Macro-EMG (o), Surface EMG (+) and the *Depth* normalised Surface EMG (*). The Surface-EMG parameters are scaled to the Macro-EMG ones, such that the average values over all (n=63) parameter values from Surface-EMG and Macro-EMG are equal.

DISCUSSION

The presented results show that it is possible to isolate motor unit activity at the skin surface and to characterise the resulting MUPs in terms of function and structure of the motor unit. The use of a unipolar montage has advantages. First, the uptake area of the electrical signal is larger, so that a deeper sight into the muscle is obtained. This can be concluded from Figure 3 and it is expressed in smaller values for the depth dependence Q for the parameters from the unipolar recordings.

Second, more details can be observed in the unipolar MUPs. For instance, the occurrence of a non-propagating final positive peak (see Figure 2A) is completely abolished in the bipolar recordings (Figure 2B). This peak is associated to the collision of the fibre action potentials at the muscle-tendon transition (Gootzen et al.,1991). Together with the information on the fibre propagation velocity, which can be calculated from the propagating peaks in Figure 2, this positive peak informs on the average length of the muscle fibres within this specific motor unit.

Third, as is illustrated with Figure 2B, unipolar recording allows the calculation of other montages. This would also apply to the high resolution spatial filtering techniques as presented by Reucher et al. (1987a,b), whereby the advantages of that technique in detecting motor unit activity from the surface-EMG itself can be combined with the advantages of the information from our unipolar MU distribution.

The observed dependency of the MUP magnitude on the depth of the MU quantifies how much deep MUs contribute to the total surface EMG than equally sized superficial ones. The different behaviour of the MUP parameters for the negative and positive peaks respectively ($Q(Nu)$=1.9; $Q(Pu)$=1.2), point to different source components for those components. For example, an increase in MU depth from 4 to 12 mm caused a decrease in *Nu* to 10%, while *Pu* and *Au* did not reach their 10% level within the measured range. The propagating negative peak is produced by a tripolar source, the positive end-potential is generated by a non-moving, but time varying dipole source at the fibre end during the collision of the action potential (Gootzen et al., 1991). From all relationships obtained, the bipolar recording decreased the fastest with distance (Q=2.4). In other words, when the objective is to study a small superficial area of the muscle, one should use the peak amplitudes from bipolar recordings with small inter-electrode distances. The signal area (*Au*) of the MUPs is least affected by the depth of the MU (Q=1.1), which can be understood from volume conduction theory as well.

These different Q-values should also be taken into account when the results together with the Macro-MUPs are considered. As can be observed in Figure 4, a Macro-MUP also consists of those two (positive and negative) peaks with different generation mechanisms. This can explain the largest difference between both techniques (Surface-EMG vs. Macro-EMG) for *Nu*. In Macro-EMG, the negative peak is generated very close to the needle surface, whereas the source for the positive peak is at the fibre end and thus far away from the recording surface in both techniques. With respect to the comparison with Macro-EMG, it can be concluded that, in principle, Surface EMG seems to be very useful. This conclusion is based on the facts that, first, the MUP area as MU-size parameter in Macro-EMG highly correlates with the parameter obtained from Surface MUPs and, second, that the variation in MU-size parameters between the subjects hardly showed any difference between the two measurement techniques (Figure 5). For some aspects, Surface-EMG might be even more useful than Macro EMG to gain information about MU characteristics. For instance, the speed of propagation along the sarcolemma (the negative peak) can easily be estimated from the electrode array in fibre direction (Figure 2, thin lines in Figure 4). Figure 3 shows that the Surface MUP is strongly dependent on the depth of the MU. Therefore correcting for the distance between the MU and the recording electrode improved the relationship between Macro and Surface parameters.

Finally, it can be concluded that Surface-EMG can be a powerful tool in clinical neurophysiology and kinesiology, also when the study of individual MUs is concerned. The presented results are obtained by combining two needle-EMG techniques with surface-EMG topographical methods. A next step is to combine our approach with methods of detecting motor unit firing characteristics from the Surface-EMG as already presented in literature (Reucher et al., 1987a,b). As a last point one has to realise in all signal interpretation that the understanding of the data is largely facilitated by the use of mathematical volume conduction models of the generated potentials as presented by Gootzen et al. (1991) and Blok et al. (1996).

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Surface EMG signal processing

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Abstract This paper provides an overview of techniques suitable for the estimation, interpretation and understanding of changes that affect the surface EMG signal during sustained voluntary or electrically elicited contractions. These changes concern amplitude variables, spectral variables and muscle fiber conduction velocity. The information obtainable by means of linear electrode arrays is discussed and applications of models for the interpretation of array signals are presented.

1. Introduction

In the last decade considerable progress has been made towards developing an objective non-invasive technique to evaluate the performance of a muscle with enphasis on the characterization of healthy and dysfunctional conditions [2, 9, 15, 27, 32, 33, 36]. This paper deals with the issue of surface EMG detection and processing for the purpose of extracting information concerning the histological, anatomical and physiological structure of the muscle under investigation. A considerable portion of this information is associated with the morphology of the EMG signals detected above the muscle and the time course of a number of signal variables monitored during sustained voluntary or electrically elicited contractions [25]. Recent reviews and tutorials describe the processing tools used to extract this information [6, 20, 26, 29]. The interesting possibility of non-invasive estimation of the fiber type distribution within a muscle has also been supported by experimental evidence [16, 36, 40].

2. Signal scaling

It is well known that during a sustained voluntary or electrically elicited contraction the surface myoelectric signal becomes progressively "slower". During electrically elicited contractions, this "slowing" is very evident and it appears to be a combination of scaling (stretching in time and in amplitude) and change of shape of the M-wave. During voluntary contractions, the "slowing" is also evident but more difficult to quantify since the signal is random. Fig. 1 shows an example of this phenomenon in both experimental situations.

A quantitative evaluation of this "slowing" during stimulation may be obtained by taking the initial narrow M-wave and "stretching" it until it matches (in the mean square sense) the subsequent and progressively wider ones. The scaling coefficient corresponding to each wave can then be plotted versus time [20, 29]. The resulting graph indicates how scaling evolves in time. This approach, however, cannot be applied to signals generated during a voluntary contraction since they never repeat and there is no reference wave to "stretch".

The same "stretching and matching" procedure can be applied in the frequency domain to the power spectrum of the signal, considering that dilation in time by a factor k implies compression in frequency, that is scaling by a factor 1/k. This approach is applicable to both voluntary and electrically signals and is widely used.

How can k be estimated ?

Various techniques for estimating the scaling factor k have been reported and discussed in the literature. Among these are the time domain matching mentioned above, the decrement of

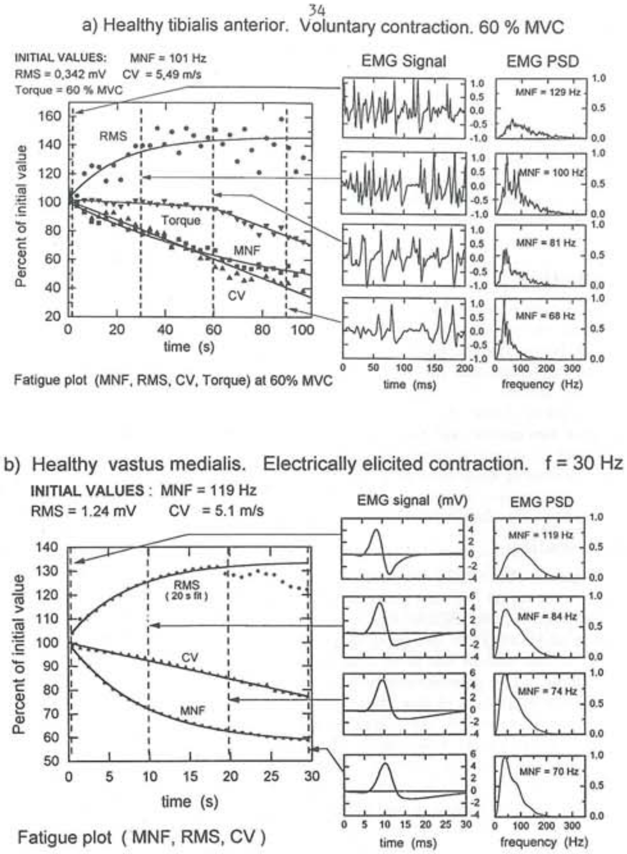

Fig. 1. Examples of fatigue plots showing the time course of the EMG signal variables during a sustained contraction, the EMG signal and its power spectral density (PSD) during specific time windows.

a. Voluntary contraction of a healthy tibialis anterior muscle sustained for 100 s with a target of 60% MVC. For the sake of clarity a three point moving average has been applied to the variables and one value every 3 s is displayed. Note the mechanical breakpoint at 60 s.

b. Electrically elicited contraction of a healthy vastus medialis stimulated for 30 s at 30 Hz. See [31] for a discussion of the MNF and CV behavior.

MNF=mean frequency of the PS, RMS=root mean square value, CV=conduction velocity.

specific spectral frequencies, such as the mean (MNF) or the median (MDF), scale coefficients of series expansions other than Fourier (such as the Hermite expansion, wavelet expansion, etc.), normalized integrals in either the time or frequency domain [6, 20, 26, 29].

In the case of electrically elicited contractions, in absence of noise and if scaling is indeed the only change, all these methods lead to equivalent results. However, these tools perform differently if the evoked signals are noisy or affected by either a change of shape or some tail truncation, as is often the case. Although partial reviews of these techniques have been published [6, 20, 26, 29], a comprehensive comparative analysis of their performance is not yet available.

In the case of voluntary contractions, the technique of scaling and matching of signals produced at different times during the same contraction cannot be applied since the signal never repeats. However, estimation of spectral compression is still meaningful and may be based on a single spectral frequency, such as MNF or MDF, few frequencies, such as the quartile or decile frequencies, or all frequencies of the discrete spectrum [21, 29], as shown in Fig. 2. The methods based on more than one frequency provide information not only on compression between two spectra but also on spectral change of shape. These frequency domain methods are of course applicable to electrically elicited signals as well. They allow the separation of effects due to scaling from effects due to other phenomena such as changes in the motor unit pool or muscle fiber conduction velocity (CV) distribution.

Why monitor spectral compression and spectral change of shape ?

As shown by Lindstrom [18], spectral compression (and therefore the decrement of MNF and MDF) reflects the decrement of muscle CV during a sustained contraction. Theoretically, this means that if CV decreases by 10%, MNF and MDF also decrease by 10%. However, in practice, this is rarely the case indicating that factors, other than CV, do indeed affect MNF and MDF. In most cases spectral variables decrease more than CV, in a few cases they decrease less (see Fig 1. and [31]). On the basis of this observation, it would seem useful to monitor both CV and spectral variables and see if, by this process, we could make interesting deductions about what these factors are and how we could quantify them. This brings out the need to estimate CV and spectral variables separately and then develop tools for the physiological interpretation of their different rates of change [31]. Further discussion of possible approaches will be presented in section 5.

What about dynamic contractions ?

During sustained isometric contractions the signals may be assumed to be *quasi-stationary*, that is stationary during short time intervals (0.5 s - 2 s). Under this assumption spectral analysis based on the Fourier Transform may be applied. However, in many cases the assumption does not hold and other methods must be used. This is always the case for burst activation or for dynamic contractions. In the latter case we are interested in the EMG of a muscle that is moving underneath the electrodes. This fact considerably complicates the signal, adding confounding factors [28]. Novel approaches have been described by Kaneko, Kiryu et al. who tried to account for the geometrical shifts of the innervation zone using electrode arrays [13]. These authors also defined a new fatigue index based on the instantaneous frequency of the myotatic reflex superimposed to background voluntary activity [14].

Quadratic time-frequency rapresentations do not require stationarity and are being investigated. Most of them introduce unacceptable artifacts due to crossterms [11, 12]. The Choi-Williams transform is promising because of the small amplitude of the crossterms. Good reviews of different approaches are provided by Lin and Chen and by Jones and Parks [17, 12].

3. The estimation of conduction velocity

Muscle fiber CV is an important physiological variable that is related to fiber diameter and membrane properties and reflects the size principle [1]. It may be estimated by placing two pairs of electrodes at some distance along the muscle fibers and dividing such distance by the delay between the two "single differential" signals. A better choice may be the use of two triplets of electrodes (double differential technique) that offer some advantages [3, 29]. If the two signals are identical, the definition and measurement of delay is trivial. Unfortunately, this is not usually the case. We are then faced with the rather subjective and fuzzy concept of delay between two signals that are similar but not identical. Common practice in neurophysiology is to measure the delay between two reference points on the two signals, such as peaks or zero crossings. This approach is somewhat arbitrary and the result depends on the particular reference point chosen. If the evaluation is done after A/D conversion, the estimate depends on the sampling rate and/or on the interpolation procedure adopted. Many other preferable approaches exist. Some are discussed and compared in a tutorial paper [20, 29] and one will be described shortly.

How can the delay between two similar signals be estimated ?

The estimate of delay between individual reference points on the two given waveforms uses only a fraction of the information contained in the two waveshapes. One may then consider using the information contained in the entire waveform. Methods that apply this approach are based on the waveform center of gravity, on the normalized integrals, on the cross-correlation, on the phase of the cross-spectrum [20]. The cross-correlation method consists in finding the amount of time shift that must be applied to one signal in order to minimize the mean square difference with the other [39]. This method can be applied in the frequency domain (that is to the Fourier transforms of the signals) where the delay is a continuous variable [23]. The quality of the estimate is then limited by the signal/noise ratio, not by the sampling rate (as long as the sampling frequency is above the Nyquist rate). In addition, the cross-correlation coefficient obtained with this algorithm provides information on the degree of similarity btweeen the signals being analyzed and the acceptability of the estimate. The technique seems simple enough, but the interpretation of the result is not trivial.

What are we really measuring ?

FIg. 3 shows a set of signals detected from a biceps brachii with 11 pairs of electrodes spaced 5 mm apart. Three motor units may be identified, two innervated between pair 8 and 9 and one innervated under pair 4. It is clear that if the detection electrodes are too close to an innervation zone, or between two innervation zones (i.e. between pair 4 and 9), they will sense the bidirectional propagation of the action potentials while if they are too close to the tendons they may sense the so-called non-travelling "far field potentials". In either case the delay estimate will be incorrect. It follows that the electrode location and the inter-electrode distance are critical [34]. In Fig. 3, only pairs 1, 2, 3 and 9, 10, 11 may be appropriate for a CV estimate. Smaller interelectrode distances may seem preferable at first sight, because of higher selectivity, however they yield smaller signals and lower repeatability, possibly due to greater sensitivity to local dishomogeneities [5, 38]. Regardless of these issues, the algorithms mentioned above yield a single number as the estimate of CV. But not all fibers have the same

CV and it is known that even within a single motor unit there is a spread of CV values. Then, how should the number provided by the algorithm be interpreted ?

The issue of CV distribution

The separation of travelling from non-travelling waves and the estimation of CV distribution in muscle fibers are extremely complex problems. Although some mathematical approaches have been developed [4, 35], their practical application has not been successful. The problem of how different velocities affect the global estimate does not yet have a clear answer, however it is intuitive (and can be shown) that the global estimate is a weighted average of the individual CV values with greater weight attributed to the motor units that are either larger or closer to the surface. The single CV value provided by the cross-correlation method may be quite unsatisfactory for the biomedical engineer as well as for the neurophysiologist. Two approaches may be considered to improve the quality of the information about CV:

1. we may try to separate and identify the contributions of individual motor units by using some selective device that would extract such contributions from the overall signal. This approach requires the use of electrode arrays, either linear or bidimensional, it is summarized in section 4 and described in detail in a companion paper.

2. we may reverse the approach to the problem and try to synthesize the signal at hand rather than analyzing it. To do this we must develop a mathematical model that simulates the generation of the individual motor unit action potentials that constitute the overall signal. We can then play with the parameters of the model and find out which combination of parameters allows an acceptable simulation of the experimental signal. The set of parameters providing a good matching between simulated and real signals is likely to be a good descripion of the real system. This approach is referred to as *model based signal understanding* and will be described in section 5.

4. Array detection of surface EMG signals

It is intuitive that the greater the number of signals that we detect from a muscle, the greater the amount of information that can be obtained. Instead of a single or dual pair of electrodes we could use many pairs and develop a multichannel approach as shown in Fig. 3. This approach can be split into two lines of investigation:

1a the use of bidimensional arrays as spatial filters that could "focus" on a particular region of space and detect the signal(s) generated by one or few motor units [37].

1b the use of linear arrays that show the generation of the motor unit action potential(s) at the innervation zone(s), their propagation along the fibers of the motor unit and their extinction at the fiber endings [22, 32].

Approach 1a is described in detail in another paper and is only summarized here. A bidimensional array is basically a bidimensional Finite Impulse Response filter that computes a weighted average of monopolar EMG signals detected from a set of points geometrically arranged on the skin surface. By proper selection of the location, number and weights of the signals it is possible to construct a "near-pass" filter and other types of filters that selectively sense signals generated in a specific region of space, thereby enhancing the contributions from generators contained in that space and attenuating those from generators located elsewhere.

Approach 1b is described in Fig. 3 as well as in previous works [22, 32]. It is clear that it provides a good description of the longitudinal events that evolve in space and time along the fibers of a few superficial motor units. This approach is particularly interesting in association with the modelling approach described in the following paragraph.

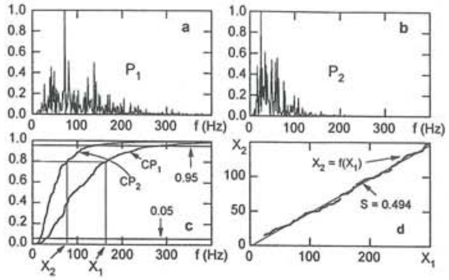

Fig. 2. A procedure for estimating scaling factors between two spectra (P_1 and P_2) based on the cumulative power functions CP_1 and CP_2 (a, b, c, and d). For each value of X_1 on the f axis the value of CP_1 is found. For the same ordinate, the value of X_2 is found from the CP_2 curve. The procedure is repeated for CP_1 and CP_2 between 0.05 and 0.95 and a curve $X_2 = f(X_1)$ is obtained [20]. If the curve resembles a straight line, as in (d), the slope s of the linear regression is taken as an estimate of the scaling coefficient. If the curve is not a straight line its features are descriptors of the changes between the two spectra.

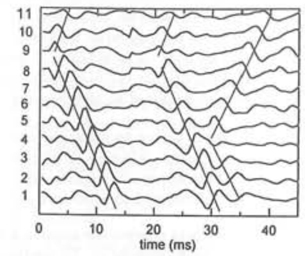

Fig. 3. Example of surface EMG signals detected with a linear array and showing individual motor units with different innervation zones. Note that under certain pairs of electrodes, (5 to 8) propagation in opposite directions is evident in different times. Interelectrode distance is 5 mm.

5. Model based signal understanding

A model is a mathematical rapresentation of a real system that allows simulation of its behavior or responses. Many physical parameters of a real biological system are not accessible and cannot be measured directly. However, they are accessible in the model, may be changed and the effect of the change on the observable variables may be studied. When the results of the simulation are very similar to those of the experiments it is likely that the parameters of the model have values similar to those of the real system. This conclusion must always be taken with caution since a) there may be more than one set of model parameters that provides a good fit of the experimental data, and b) the model always implies approximations and simplifications that may affect the results more than it may be expected. The model described in Fig. 4 is based on the work of Gootzen [7] and has been used to investigate and explain some experimental findings [8, 30]. Alternative approaches have been presented by other researchers [10].

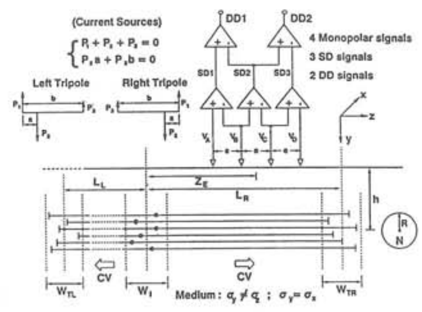

Fig. 4. Schematic structure of the model of a single motor unit. The motor unit has N fibers uniformly distributed in a cylinder of radius R at depth h. The axis of this cylinder may present an angle with respect to the skin plane and with respect to the z axis. The neuro-muscular junctions are uniformly distributed in a region W_I and the fiber-tendon terminations are uniformly distributed in two regions W_{TR} and W_{TL}. A right and a left current tripole originate from each neuromuscular junction and propagate to the fiber-tendon termination where they become extinguished. The conduction velocity is the same in both directions and for all fibers of a motor unit but may be different in different motor units. Each of the voltages V_A, V_B, V_C and V_D is the summation of the contributions of each tripole.

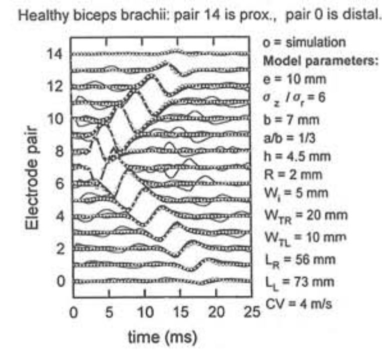

Fig. 5. Example of simulation of a single motor unit of a healthy biceps brachii. Ten firings of the same motor unit that show no superposition with firings of any other detected motor unit were identified and aligned. Note the excellent repeatability of the signals. Simulation was performed using the model described in Fig. 4.

Fig. 5 shows 10 firings of the same motor unit of a healthy biceps brachii during a low level voluntary contraction. The signals are detected in the single defferential mode from a 16 contacts linear array. The firings are those showing no superpositions with the firings of other motor units during a time interval of 1.5 s, are aligned and superimposed, are similar enough among themselves and different enough from the other firings to justify the assumption that they belong to the same motor unit. The results of the simulation (open circles) are superimposed. The parameters of the model used for the simulation are indicated in Fig. 5.

These results, as well as other results recently presented [30] suggest that a modelling approach may indeed be useful for the non-invasive characterization of superficial muscles and of their motor units. A computer assisted procedure that should simplify and accelerate the identification of the "optimal" set of parameters is being developed.

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SPINE LOADING DURING WHOLE-BODY FREE-DYNAMIC LIFTING

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INTRODUCTION

Occupationally-related low back disorders (LBDs) are currently and have been the leading cause of lost work days as well as the most costly occupational safety and health problem facing industry today. It is well known that most occupationally-related LBD risk is associated with manual materials handling (MMH) tasks. However, a major limitation in controlling the incidence of occupationally-related LBDs has been the inability to *accurately* assess the loading that occurs on the lumbar spine (the most common site of injury) during realistic, actual whole-body free-dynamic MMH conditions. An accurate evaluation of spine loading is necessary so that the loads imposed upon the lumbar spine can be compared with tolerance limits of the spine derived through cadaver studies as well as finite element models of the spine.

Historically, most biomechanical models used in ergonomics have made assumptions about the conditions under which work is performed which limit their applicability to realistic dynamic loading conditions. These models were originally static and did not consider the effects of motion. Latter models were dynamic but did not consider the effects of the collective influence of the trunk structures upon the loading of the spine. These models are deficient in that they were not able to explain how the trunk muscles work collectively (coactivity) to support the external load and simultaneously impose loadings upon the spine. Ignoring coactivity is undesirable in that the estimates of spine compression could be under predicted by 45% and estimates of spine shear loading could be underestimated by as much as 70% (4). In addition, such models are problematic in that they are not able to account for the variability in muscle recruitment and the subsequent variability in spine loading that occur from trial to trial during lifting bouts. Thus, neither of the previous classes of models were able to asses the risk of joint loading during dynamic workplace circumstances.

A key element of work, trunk motion, could be beneficial as well as detrimental. In some circumstances it is believed that ballistic motions could actually minimize antagonistic muscle activity thus reducing or at least changing the form of the joint loading. Therefore, it is imperative that one understands how the musculature behave under dynamic loading conditions. The significance of accounting for this trunk muscle coactivity has been recognized by many modelers. In order to include such activities some have used optimization techniques to estimate muscle activities. However, due to the indeterminacy of the muscle system these models are not able to account for the coactivation that occurs among the muscles surrounding a joint during dynamic motion.

In order to address these limitations of previous efforts, a free-dynamic three-dimensional biodynamic model of the spine has been under development in the Biodynamics Laboratory at the Ohio State University for the past 14 years. The model accounts for the collective coactive influence of 10 trunk muscles upon the three-dimensional loading of the spine. We have developed this model to the point where we can accurately predict three-dimensional loading of the spine under free-dynamic bending and twisting of the lumbar spine.

MODEL DEVELOPMENT

Our modeling efforts began with the development of descriptive network models of the time event changes that occur under various motion conditions (12). This research has investigated the time sequence activity of 10 trunk muscles and IAP as subjects produced trunk extension motions (lifting motions) in the sagittal plane. Based upon these observations, descriptive network models of dynamic exertions were developed (18). These networks described the sequence of trunk structure components that developed under various velocity conditions. This analysis indicated that there are indeed velocity dependent sequence changes that occur when the trunk moves at different rates of speed. This type of timing information becomes important in the development of dynamic models since dynamic models need to assess the impulse load imposed upon the spine during a work situation. Since impulse information is time dependent, the only way to relate that information is through time sequence information. It is believed that this impulse information represents a "worst case" assessment of spine loading which is imperative for cumulative trauma assessment. These authors have also used this information about muscle activities to create a crude EMG-driven model of the spine.

Since the EMG signals represent a continuous measure of muscle activity the EMG signal could be used as a basis for determining force generation history of each muscle surrounding a joint. This logic has become the basis for some of the new EMG-assisted models which have been introduced recently for the assessment of dynamic motion. One of the first models to use this information was developed by McGill and Norman (13,14). This model used the EMG signals from six muscles to estimate spine loading during sagittally symmetric dynamic lifting performed by three subjects. The model used theoretical relationships to adjust for muscle force based upon muscle length, cross-sectional area, and velocity. Marras and Sommerich (10,11) developed a 10-muscle EMG-assisted model which was capable of predicting spine forces under asymmetric dynamic bending conditions. This model used both theoretical relationships as well as imperical data from 100 subjects to adjust muscle force based upon EMG. EMG activity from 10 muscles was used to predict muscle forces based upon muscle length, EMG pick-up volume, velocity, gain, and cross-sectional area. For simplicity purposes, this model described EMG activity via geometric representations of the EMG time history signal. This signal was then adjusted for muscle length and velocity. This model predicted spine forces as well as trunk torque production. Measured trunk torque was compared to predicted trunk torque and used as a validation measure. Two validation experiments (3,4) involving over 30 subjects have shown that the model accounts for an average of over 85 percent of the torque variability. As asymmetry and velocity increased significant increases in spine compression were found. Similar trade-offs occurred with shear.

General Description Significant improvements have been incorporated into our EMG-assisted free-dynamic biomechanical model over the past several years. Compared to former EMG-assisted models this model is unique because multi-dimensional, trunk moments and spinal loads are determined from dynamic muscle force vectors and moment arms. Our current EMG-assisted, free-dynamic lifting model employs ten muscle equivalent vectors to approximate trunk anatomy and mechanics. The muscle equivalents are treated as simple cables tensioned between points of origin and insertion. Active muscle forces are assumed to be an adequate description of trunk mechanics without consideration of passive muscle, ligament, or disc forces. At trunk extreme flexion-extension angles passive forces may become significant, but within the design range of 45 degrees flexion to vertical, the trunk moments may by active muscle forces. Our field studies (9) have shown that this is a reasonable assumption.

Muscle fibers sampled by EMG surface electrodes are assumed to be representative of, and linearly related to the net muscle force. Lippold (8), Moritani and DeVries (16), and Yoo et. al. (20) demonstrated linear relationships between surface EMG activity and voluntary isometric joint torque. Conversely, Zuniga and Simons (21), Vredenbregt and Rau (19), and Komi and Viitasalo (7) measured EMG proportional to the square of the isometric joint torque. There is clearly, disagreement as to whether EMG is linearly or non linearly related to force. Hof and Van Den Berg (6) offered a resolution to this discrepancy by demonstrating EMG is linearly related to muscle force, but can be non linearly related to joint torque due to coactivity. Our model is valid only for those motions wherein the time delay between the onset of myoelectric activity and muscular contractile force is minimal, i.e. smooth lifting motions.

The model employs EMG and kinematic input to determine the dynamic, relative, muscle, force vectors of the ten, modeled, trunk muscles. Relative force vectors are scaled by a gain factor computed from input kinetic data. Predicted, multi-dimensional, dynamic, trunk moments and spinal loads are computed from muscle force vectors and muscle moment arms determined from subject anthropometry and kinematics.

Trunk Mechanics Whereas previous models assumed muscle vector directions were constant in space, the free-dynamic model allows each muscle orientation, length, and velocity to vary with the lifting motion and position of the trunk. Muscle origins are assigned a three-dimensional location relative to the spinal axis, coplanar with the iliac crest. Muscle insertions are located coplanar with the 12$^{\text{th}}$ rib. Muscle forces are represented as vector quantities between their two endpoints. In essence, the mechanics of this model can be visualized as two "plates" that can be allowed to move relative to one another. These plates represent the attachment point of the muscle in the pelvis and thorax. The muscles are represented by vectors between these plates that change their orientation as the trunk moves dynamically (Figure 1). Using this approach, muscle orientations and lengths change throughout a movement, thus, accounting for a muscle's changing mechanical advantage throughout the task. In addition, such a representation affords the opportunity to change the relative angles between the iliac and thoracic planes. Thus, pelvic tilt or asymmetric bends of the trunk could be simulated. Muscle origins and insertions are dynamically located via Euler rotation of the anatomically defined, three-dimensional coordinates relative to the measured trunk motion. Muscle vector directions, lengths and velocities are continuously determined from the instantaneous positions and motions of the muscle endpoints. Time and position dependent force vectors significantly affect the predicted trunk moments and forces generated by the musculature, by allowing the vector direction to move throughout an exertion.

Spinal loading (compression, right-lateral shear, and anterior shear forces) are calculated from the vector sum of validated muscle forces. Muscle generated moments about the spinal axis are predicted from the sum vector products combining dynamic tensile forces of each muscle, j, and respective moment arms.

$$\vec{M} = \sum_j \vec{r}_j \times \vec{F}_j \qquad \textbf{(1)}$$

Spinal compression, right-lateral shear force, and anterior shear force are displayed as a function of time. Measured and predicted values of the trunk moments, as well as predicted compression, anterior shear, and lateral shear forces are written to a file for post-modeling analysis. The task gain and correlation between measured and predicted moment profiles are recorded for model performance evaluation.

Figure 1. Schematic representation of trunk mechanics and muscle vectors used by model.

Measured and predicted trunk moments are compared, and must agree if the model is correctly simulating trunk mechanics. Statistical correlations between predicted and measured moment profiles (R^2) serves as the measure of model performance and indicates how well the model accounts for the variability in the motion moment. A high correlation implies the model generates an accurate simulation of dynamic spinal loading. Of course, there is no means by which one could actually measure spine loading in-vivo. However, we feel that if the model could accurately predict applied moments about the spine then the predictions of spine loadings should also be reasonable.

Muscle Force Estimates in the Model Relative, muscle, contractile, force magnitudes are computed from normalized EMG, modulated to account for muscle length and velocity, and scaled by muscle cross-sectional areas. Relative muscle activities are multiplied by an appropriate muscle force per unit cross-sectional area, i.e. gain, determined from the solution of dynamic equilibrium. Previous EMG-assisted models employed muscle areas representative of the cross-sections found at the lumbo-sacral transverse plane. The current model recognizes that the force generating capacity of the latissimus dorsi is poorly represented by the small slips of muscle that pass through the lower lumbar levels. Consequently, the maximum cross-sectional area of the latissimus dorsi, found near T5 (15), was employed in the model to scale the force generating capacity of that muscle.

The tensile force generated by each muscle, j, is described (eq 2) by the product of normalized EMG, muscle cross-sectional area, a gain factor representing muscle force per unit area, and modulation factors describing EMG and force behavior as a function of the length $f(Length_j)$, and velocity, $f(Vel_j)$ of muscle j.

$$Force_j = Gain \frac{EMG_j(t)}{EMG_{Max\,j}} Area_j\, f(Vel_j)\, f(Length_j) \quad \textbf{(2)}$$

EMG data are normalized relative to myoelectric maxima from each muscle. This was necessary to remove possible analytical errors related to electrode placement, skin abrasion, flesh resistance, muscle fiber density, and electronic channel differences.

Relative myoelectric activities are multiplied by a unitless function of length, f(Length), to account for the relation between tensile force and muscle length,. The modulation factor incorporates physiologic, length-strength relations and artifact due to variation in the myoelectric potential density picked up by the surface electrode. The functional coefficients were determined by minimizing the average variation in predicted gain as a function of length. The length modulation factor (eq 3) employs the instantaneous length of muscle, j, determined from the anthropometry coefficients and kinematic inputs.

$$f(Length_j) = -3.2 + 10.2Length_j - 10.4Length_j^2 + 4.6Length_j^3 \quad \textbf{(3)}$$

The empirically determined coefficients agree with the expanded form of the length-modulation factor proposed by McGill and Norman (14).

Relative muscle activity is also multiplied by a unitless function of contractile velocity, f(Vel), to account for the physiologic force-velocity relation and associated EMG artifact. Bigland and Lippold (1) demonstrated increased muscle contraction velocity yields increased myoelectric activity without a concomitant increase in force output. The modulation coefficients were computed by minimizing the average variation of gain predicted by the model as a function of velocity. The velocity modulation factor (Equation 4) includes the time-dependent, contraction velocity of each muscle, j, determined from anthropometry and kinematic data.

$$f(Vel_j) = 1.2 - 0.99Vel_j + 0.72Vel_j^2 \quad \textbf{(4)}$$

The coefficients for the modulation factor approximate the theoretical force-velocity relationship represented in the Hill (5) equation, and are appropriate for smooth, non-ballistic muscle contraction.

Normalized and modulated EMG data are multiplied by their respective muscle cross-sectional areas to account for the relative force generating capacity of each muscle. It has been demonstrated (2) that maximum muscle force is directly related to cross-sectional area. Therefore, scaling the EMG by muscle area provides larger muscles with greater, modeled, force generating capacity.

Gain, i.e. muscle force per unit area, is computed by comparing muscle-generated, trunk moments with measured, applied moments about the lumbo-sacral junction. To satisfy the equations of dynamic equilibrium, muscle generated extension moment must equal the measured moment. Gain is appropriately and automatically adjusted to satisfy this condition. To be physiologically valid, the predicted gain level must fall within the range of 30 to 100 N/cm^2 (17). Muscle force per unit area is highly variable between subjects, based on subject conditioning and natural ability. On the other hand, gain predicted for a given subject must be constant throughout each of the experimental trials. Examination of the gain value and its within subject variability provides a means for testing model validity.

Model Input Input data required by the model includes time-domain EMG, exertion kinetics and kinematics. Maximum exertion EMG levels and subject anthropometry are also employed to calibrate and format the dynamic data suitable for use in the model mechanics. The cross-sectional area of each muscle is computed from regression equations based on the subject's trunk depth and breadth.

Voluntarily applied external kinetics, including gravitational moments and acceleration effects on trunk mass are dynamically measured by a force plate and pelvic stabilization system. Translation of force plate mechanics is performed to compute three-dimensional force and moments about the lumbo-sacral spine. The pelvic stabilization system permits free-dynamic motion above the pelvis. More recently this system has been modified to permit free dynamic motion of the whole body. This is accomplished by mathematically correcting for the position of the pelvis relative to the forceplate.

Trunk velocity is computed from dynamic measures of trunk flexion, twist, and lateral angles collected from a lumbar motion goniometer (lumbar motion monitor or LMM) (9). The LMM offers no significant restriction to trunk mobility, while accurately recording trunk kinematics. Collected kinematic data are used in the lifting model to 1) describe the trunk motion as a function of time, 2) determine the muscle force and moment vector directions, and 3) modulate muscle EMG values to account for muscle length and velocity artifact.

Integrated, EMG data are collected from the right and left latissimus dorsi, erector spinae, rectus abdominus, internal abdominal obliques, and external abdominal obliques. The time-domain, myo-electric data represent muscle activity and are used to calculate relative muscle force. EMG signals are collected from surface electrodes, pre-amplified, high- and low-pass filtered, rectified, and integrated via a 20 ms sliding window hardware filter. Maximum and resting EMG values were collected from flexion and extension exertions to normalize the dynamic EMG signals.

All dynamic data, including kinetics, kinematics, and EMG, are smoothed via a Hanning weighted, time-domain filter within the model. Smoothing the data is necessary to remove digitizing noise and artifact from differentiation and calibration routines.

Model Performance and Validation The free-dynamic model has been tested in two separate experiments. The first experiment model was designed to test the ability of the model to assess symmetric and asymmetric lifting motions under isokinetic compared to free-dynamic conditions. The second experiment was intended to evaluate the ability of the model to assess spine loading during trunk twisting. .

In the first study, the EMG-assisted model was exercised and results generated from 703 separate lifting exertions designed to test its validity under free-dynamic conditions and to compare its performance with previous models of trunk mechanics. Ten subjects lifted loads of 0, 40, 80 lbs. at isokinetic trunk angular velocities (30, 60, and 90 deg/sec) as well as free dynamically (slow, medium, and fast) lift rates. Subject gain values averaged over all free-dynamic exertions was 47.4 ±12.1 N/ cm^2 and fell within the physiologically acceptable range. Because subject's muscle strength per unit area can not change from one exertion to the next, a subject's gain value must remain constant. Although gain changed significantly ($p<.01$) between subjects, the values did not vary significantly within subjects. Thus, model predicted a muscle force per unit cross-sectional area which was a physiologically-valid. Distributions of squared correlation coefficients indicating the association between measured and predicted external trunk moments were derived from the dynamic, lifting trials, and illustrate that over 88.6% of the trials performed with an R^2 greater than 0.80, and 68% greater than R^2=0.90. Statistical analyses demonstrated that the model performed well independent of the type of lifting exertion. There was no statistically significant difference between the isokinetic (R^2=0.87) and free-dynamic exertions (R^2=0.89). The model performed well at all dynamic velocities.

Model predictions of compressive, lateral shear, and anterior shear loading agree with trends cited in previous studies, however we believe our results more accurately represent loads that occur from primary and coactive muscle activity during realistic lifting tasks. Spinal loads generated throughout dynamic, lifting exertions increased as a function of trunk asymmetry and lifting velocity.

In the second study, twelve males 21 to 31 years of age participated in an experiment involving torsional moment, torsional direction, twisting position, and twisting velocity. Spine loading, as predicted by the EMG-assisted model, was significantly affected by many of the variables manipulated in this experiment. Statistical comparisons indicates that relative spinal loads (per unit of torsional

moment) changed as a function of exertion level, direction of the applied twisting torque, and twisting velocity. Twist velocity substantially increases spine forces in all three cardinal planes compared to isometric exertions. Predicted values of the calibrated subject muscle gains indicated that the model was also valid for twisting exertions. The model also performed well during both static and dynamic exertions. The squared correlation coefficients (R^2) between measured and predicted torsional moments from the 320 trials indicated that the average value was 0.80.

CONCLUSIONS

These efforts have demonstrated that we have been able to build a free-dynamic three-dimensional model of the spine that is capable of accurately assessing spine loadings during trunk motion during manual materials handling under laboratory conditions. Future embellishments of the model will endeavor to incorporate a better understanding of passive tissue mechanics on spine loading.

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CURRENT TOPICS OF CLINICAL FES IN JAPAN

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INTRODUCTION

Various kinds of FES has recently been applied to the paralyzed or paretic bodies of the patient. All the organs composed of skeletal muscles are the target of FES. Furthermore, FES can be also applicable to the disturbance of autonomic nervous system and smooth muscles such as the digestive or urogenital organs etc.

Most of FES studies, however, are still focused on the recovery or restoration of motor function of the impaired extremities. Demands for clinical usage of the extremity-FES are gradually increased. This implies that recent FES devices satisfies the needs of the patient, family and/or medical stuffs even in partial.

This paper describes recent topics of the extremity-FES especially in Japan.

CANDIDATES OF FES

Since muscle contraction is provided by electrical stimulation of motor branches of the peripheral nerve in Extremity-FES, the patients with upper motor neuron disorders where the lower motoneurons are not involved are the candidate of FES. Therefore, most of FES works are performed on stroke and SCI patients. Lower motor neuron disorders are rarely the object for TES but not for FES. However, we have recently applied FES to the patient with motor neuron diseases where both upper and lower motor neurons are involved.

FES SYSTEMS FOR CLINICAL USE

Transcutaneous and percutaneous FES systems are currently used in clinical fields in Japan.

Saito et al. [1] developed a hybrid FES system with surface electrode in combination with a new hip-knee-ankle-foot orthotic (HKAFO) system with a medial single hip joint (Walkabout unit, Polymedic, Australia) for gait control of the paraplegic patient.

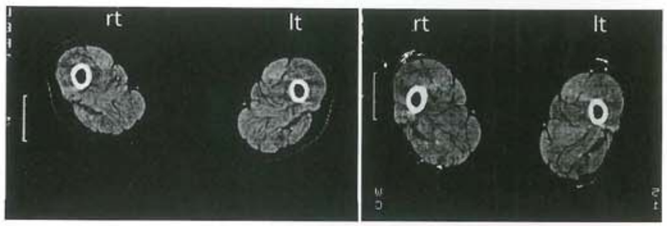
A commercially available percutaneous FES system with 30 output channels (FESMATE1000, NEC Medical Systems,Tokyo) and percutaneously indwelling electrodes (Nippon Seisen Co. Ltd, Osaka), which were developed by Handa et al. were used clinically for the FES control of the upper and lower extremities under the permission of Japan Ministry of Health and Welfare. Details of configuration of the system and electrode were described elsewhere[2),3)].

Kagaya et al. developed a hybrid FES system with the percutaneous electrode for the control of locomotion in paraplegics[4)]. The system consists of a stimulator with 18 output channels (a modified AKITA SYSTEM), two hand switches and a knee-ankle-foot orthosis with electrical knee unlocking mechanism achieved by solenoid.

TRAINING WITH TES BEFORE FES APPLICATION

Most of the patients with paralytic extremities show more or less disuse atrophy and hypertonus of the muscle. Hyperreflexia or hyporeflexia of the autonomic nervous system can also be seen especially in SCI patients. Improvement of such symptoms is required for accomplishment of FES control of the extremities for activities of daily living (ADL).

It has been reported that TES is effective for the reduction of spasticity and increase in muscle volume and force. Fig. 1 shows changes of CT image of before and after long-term TES therapy.



(a) (b)

Fig. 1 Effect of long-term TES on muscle volume of the thigh in a T7 quadriplegic
CT images of the thigh before (a) and after 6 months (b) TES therapy

As shown here, muscle volume were apparently increased through 6 months TES therapy. Simultaneously, 2.5 times increase in the torque of knee extension induced by FES was observed. Such an increase in FES-induced muscle force is essential in particular for the restoration of locomotive function in the disabled.

TES applied to a muscle branch of the nerve induces reduction of spasticity in the antagonists of the muscle stimu-.

lated. Since hyper-reflexia or spasticity often interrupt the movement restored by FES, enough TES therapy is also important for getting satisfactory results

FES FOR THE UPPER EXTREMITY

1. Control of the hemiplegics

As a general, we have applied FES to the hemiparetic patients whose Brunnstrom stages of the arm and hand are above Ⅳ and Ⅲ, respectively.

In the case of a hemiplegic whose Brunnstrom stages were Ⅴ and Ⅲ in his arm and hand, respectively, FES only for the hand was required because he could move his shoulder and elbow almost freely. He had a difficulty for hand opening while could grasp the object with mass flexion of the fingers. FES for hand opening, where the extensor digitorum and thumb extensors were activated, provided the patient hand tasks for ADL such as drinking a glass of water wipe his face with his both hands and so on.

We firstly succeeded to control all the joints of the left upper extremity in a hemiplegic who suffered from cerebral infarction four years before FES application[5)]. Since he was a

Fig. 2 Total control of the upper extremity in hemiplegia

painter, he wanted support his trunk by his left arm against the wall during painting. The Brunnstrom stage of the upper extremity including the hand was Ⅳ and he could not extend his left arm with enough power for supporting his trunk. Shoulder flexion, elbow extension and fist grasp with wrist extension were restored by FES with percutaneous electrodes. Thus the patient could used his left upper extremity for his job, the painting (Fig. 2).

2. Control of the quadriplegics

Functional stage of the upper extremity of the quadriplegia strongly depends on the level of injury. SCI more or less

Fig. 3 Control of the upper extremity
in C4 quadriplegia

Fig. 4 Control of the wrist and hand in C5 quadriplegia

a)

b)

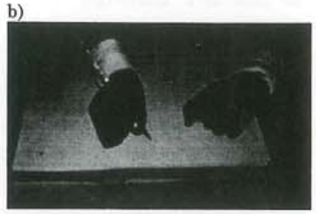


Fig. 5 Control of the bilateral hand in a C6 quadriplegic

accompanies peripheral motor neuron loss around the injury level (dead band). Control strategy of the upper extremity in quadriplegic patients, therefore, differs in the injury level.

Fig. 3 shows FES-control of a C4 quadriplegic. Under the usage of the orthosis "BFO", the patient could control his arm and hand voluntarily puffing and/or sucking a mouthpiece type controller.

Simultaneous control of the wrist and hand was required for restoring hand function of the C5 patient. Operation of the FES controller was achieved by contra-lateral arm of the FES-controlled arm. Thus ADL of the upper extremity was restored as shown in Fig. 4.

In C6 quadriplegia, prehension and release were restored only by control of the thumb and fingers. Wrist extension has been utilized for controlling ipsilateral hand movement[7]. Therefore, both hands were controlled separately but simultaneously by operating the wrist switch in each side as shown in Fig. 5.

FES FOR THE LOWER EXTREMITIES

1. Control of the hemiplegics

We have applied the percutaneous FES to the hemiplegic lower limb. For the stance phase, the hip and knee extensors and hip abductor were activated while the common peroneal nerve was stimulated in order to evoke the flexion reflex for the swing phase. And the quadratus lumborum was also coactivated during the stance phase for restoring the pelvic stability.

2. Control of the quadriplegics

The Hybrid FES system with either electrodes, surface and implantable, is much more clinically practical than the sole FES system for the control of locomotion in the paraplegics. Basically, the quadriplegic patients with the injury level below the middle thoracic cord can maintain standing-up position and walk under the usage of long leg braces. Therefore, assistive application of FES can boost their locomotion with orthosis.

A hybrid FES system consisting of Walkabout[R] and a 4-channel surface FES system developed by Saito et al. was applied this system to a T10 complete paraplegic[1] In addition, the Akita group also utilized recently the Walkabout[R] with percutaneous FES for restoring locomotion in a paraplegic patient (Fig. 6). It was found that walking parameters and energy cost better than those in the orthotic walking[1].

Another AKITA hybrid FES system with knee unlocking mechanism using solenoid was applied to a T8 paraplegic[4]. This system allowed knee flexion during the swing phase of gait by unlocking the knee of the orthosis electrically while the knee was locked in extension during standing and stance phase of gait. Therefore, stimulation of knee extensors was unnecessary and fatigue of the patient could lessen.

Fig. 6 A hybrid FES with Walkabout[R] and percutaneous FES

Fig. 7 Erect standing by percutaneous FES

We have applied the percutaneous FES system without using the orthosis. As for standing-up and erect standing, our FES was satisfactory when the patient received enough TES training.
The paraplegic patient in Fig. 7 had received TES training for two years and he could continue erect standing almost one and half hour without fatigue. The FES-controlled standing is completely clinically available and is utilized for ADL and activities parallel to daily living (APDL). It is also noteworthy that the percutaneous FES cosmetically acceptable for the patients and they can feel free through such implantable system.

Although the percutaneous FES still remains in feasible study as for walking in complete paraplegia, it is clinically applicable to gait disturbance in paraparetic patients. By using foot switches bilaterally, FES was applied to the bilateral common peroneal nerves to assist the volitional swing action and resulted in rapid flexion of the hip, knee and ankle joints with large amplitude. We could observe marked improvement of walking parameters accompanied by reduction of fatigue.

FES FOR AMYOTROPHIC LATERAL SCLEROSIS

We have reported TES effects on wasting muscles of amyotrophic lateral sclerosis[9),10)]. It was found through two years follow-up study that muscle deterioration was markedly slowed in some ALS patients. Recently FES was applied to the paretic lower limbs for the purpose of assisting standing up motion because the force of the hip and knee extension induced by electrical stimulation was much larger than the force by voluntary extension. His helper commented theat the force for assisting his standing up motion was markedly reduced, while the patient felt that standing up was much easier than that without FES.

NEXT STEP OF FES

FES is powerful means for restoring motor function of the stroke and SCI patients. However, the FES system currently used in the world is just like a newborn baby who can perform only minimum and primitive movement. However, It is commonly observed that the patient and his/her family often have excessive expectation for FES before application and are dis-appointed after application because of incompleteness of FES controlled movement.

It will be one of the most important problems for the next step how overcome the incompleteness of FES. Some breakthrough in control strategy of FES will be necessary. In addition, stimulating electrode is essential as an interface between the neuromuscular system of the patient and the FES system.

Although the percutaneous electrode realizes reliable control of the extremities, the patient often refuse it because of problems of its outlet, such as a possibility of infection and skin irritation and trouble of daily disinfection.

Development of total implant FES system will dissolve this problem. Groups in Cleveland[11)] and London[12)] etc. have already developed and clinically applied the total implant system. We are also developing a total implant system with 16 output channels[13)].

For development of the new born baby FES system, further and potent collaboration between researchers in engineering and medical fields will be required.

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CHAPTER 2

MOTOR CONTROL

Estimation of the position and size of active motor units by inverse analysis of surface EMG

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INTRODUCTION

In-depth investigations have recently been made on the cerebral nervous and cardiac muscular activities by using the inverse analysis of bioelectric signals like MEG, EEG and ECG. Application of the inverse analysis to surface electromyography may possibly provide information of active motor units which is equal to or more than that obtained by needle electrode, though such attempts seem to have been very rare. In this study, we have executed the inverse analysis by minimizing the difference between measured and calculated surface EMG, in an attempt to estimate the position and size of active motor units.

METHODS

Surface EMG measurement

Surface EMGs were recorded bipolarly from the short head of the biceps brachii muscle using a 16 channel array electrode with the array separated by 2.5 mm intervals. The electrode was placed on the muscle belly in the circumferential direction. The amplifiers had a gain of 60 dB over the frequency range of 53-1000Hz. We used an additional 50 Hz notch filter to eliminate the power line interference. Three male subjects carried out isometric elbow flexion for about 3s, with the elbow angle fixed at approximately 130° (complete extension of the joint being defined as 180°). The intensity of the contraction was adjusted to a level at which the motor unit action potential train could be discriminated.

Signal processing

The amplified EMG signals were A/D converted with 5kHz sampling frequnecy, 12 bit accuracy and 3s sample length. Averaging the sampled EMG, we obtained the wave form of the single motor unit action potential (MUAP).

Inverse analysis

We used the method of image for EMG simulation[1] and estimated for each parameter the value of parameters which minimized the difference between the averaged and simulated MUAPs. The parameters estimated were depth, position, dipole moment and conduction velocity of the current source and electric conductivity of the medium. Furthermore, error ratio, which was the ratio of the estimation error and averaged MUAP, was calculated as a measure of the degree of fitness.

RESULTS

Number of the action potentials averaged were 20-50 for 3s. The amplitude distribution of the averaged MUAP at peak time showed convexity like Gaussian distribution over the circumferential direction. On displaying the amplitude over the channel(circumferential position)-time plane, two peaks with lags in position and time were often found (Fig. 1a). Number of the current sources to be used in the inverse analysis was determined by inspecting the MUAP on the 3D display. Figure 1b shows MUAP distribution estimated by inverse analysis, using two current sources. In this case, error ratio was 36 % and estimated depths of the current sources were 2.1 mm and 2.3 mm. The radius of motor unit extent on muscle cross section was estimated to be 3-6 mm, because the distance between the current sources was about 6 mm. For the whole subjects and contractions, the depth of current source was estimated to be 1-7 mm and the radius of motor unit extent 0-7 mm.

DISCUSSION

The averaged MUAP often had two peaks with 1-5ms intervals. These peaks are most likely due to the divergent locations of end-plates belonging to single motor unit if the conduction velocity of muscle fibers within the motor unit are indentical. For example, when the conduction velocity of is 3m/s, end-plates are considered to be spreading 3-15mm along the muscle fibers.

The method of image we used in the inverse analysis assumes that muscle fibers are embedded in a homogeneous resistive medium of infinite extent and electrodes are placed on an infinite plane. In other words, this method does not consider the existence of boundary and subcutaneous tissues. Using the finite element method (FEM) that solves the above problems, we have found that existence of the boundary and subcutaneous tissues make surface myoelectric potentials higher than those calculated by the method of image [2]. The potential gradient calculated by FEM, however, is smaller and therefore the surface potential becomes wider in form. Consequently, the depth of the current source esimated by the method of image may be deeper than real. On the other hand, the depth of the current source estimated by using point source will be shallower than the depth of the center of motor unit extent, because of the stronger influence of muscle fibers nearer the surface on the surface potential. In order to verify the validity of position and size of the motor unit estimated by inverse analysis, we have to compare the estimation with the results of histochemical experiments using laboratory animals.

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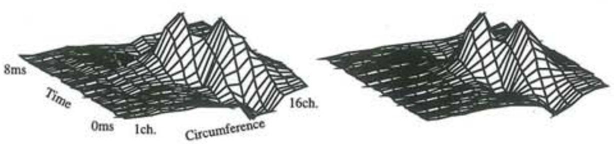

(a) (b)

Figure 1. Single motor unit action potentials displayed on circumferential position-time plane. (a) MUAP recorded by surface array electrode and averaged. Time lag between the two peaks is 1.4ms. (b) MUAP estimated by inverse analysis using two current sources which was determined by inspecting the actual MUAP shown in (a). The depths of the two currrent sources were estimated 2.1 mm and 2.3 mm, respectively.

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ACTIVITY PATTERNS OF LOW-THRESHOLD MOTOR UNITS IN THE HUMAN TRAPEZIUS MUSCLE

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INTRODUCTION

Recent studies indicate that some forms of occupational shoulder complaints, comprising the trapezius muscle, are associated with the activity pattern of low-threshold motor units, at least in work situations with sedentary work tasks: 1) Workers with short breaks in their trapezius activity pattern (EMG gaps) during work, as recorded by surface EMG electrodes, have less risk of developing shoulder complaints (Veiersted et al. 1993). 2) Workers who have or develop such complaints, show sustained, low-level EMG activity in situations that present an opportunity for rest (Vasseljen and Westgaard 1995, Veiersted 1994). 3) No association is found between the level of trapezius EMG activity in vocational recordings and the presence of shoulder complaints at activity levels below about 10% of the activity at maximal voluntary contraction (EMG_{max} ; Jensen et al. 1993).

This evidence is however indirect; there is no clear hypothesis as to pathophysiological mechanisms in the development of complaints at low activity levels, nor is much known about the activity pattern of low-threshold trapezius motor units during occupational work or in standardized muscle contractions. A study was therefore undertaken to gain insight in the firing behavior of such motor units.

METHODS

Seven healthy subjects, age 35-50 years, volunteered for the study and signed an informed consent form approved by the local Institutional Review Board. Four nylon coated Ni-Cr-Cu wires (diameter 50 μm), cut transversely to achieve a localized recording, were inserted into the right trapezius through a syringe needle. The insertion point was 10-15 mm medial to the midpoint between the C7 spineous process and the acromion. Two bipolar surface electrode assemblies (electrode 1 by 10 mm, 10 mm interelectrode distance) were placed 20 and 56 mm lateral to the midpoint of the C7 spineous process and the acromion.

Four fine-wire recordings were derived from the intramuscular recordings and stored on a FM tape recorder (TEAC XR-5000, bandwidth 1-10 kHz), together with the surface EMG recordings and a force recording, when appropriate. Three of the four intramuscular EMG signals were later analyzed by the *Precision Decomposition Technique* of De Luca (1993), to determine the activation patterns of individual motor units.

Each subject carried out a series of procedures: First, maximal voluntary force (MVC) and maximal, RMS-detected EMG activity in shoulder elevation and arm abduction were measured. Slow ramp contractions (3 min. duration, 10 to 40% MVC at end of contraction) and contractions with a trapezoid force profile (30 s total duration, highest contraction level 10 to 40% MVC) were carried out. Thereafter, 10 min. contractions were performed: 1) static contraction of about 5% EMG_{max}, controlled by visual feedback of the RMS-processed surface EMG signal, 2) 10 min. of typing using a word processor, 3) 10 min. of an attention-demanding task with minimal need of postural muscle activity (Wærsted et al. 1996), 4) the same task repeated, but with the added incentive of a money reward for good performance.

RESULTS

The following represents a summary of observations pertaining to the firing pattern of trapezius motor units, recruited at activity levels below 10% EMG_{max}:

1) Trapezius motor units show a stable firing rate of 10-20 pps (pulses per second) mean rate in prolonged contractions (1 minute or longer) with slow variations in force level, largely independent of force level within a range of 0-30% MVC.
2) Within the first 30 seconds after activation, the mean firing rate is typically higher, up to 25 pps.
3) Firing rates of 35-40 pps and modulation of the firing rate are observed in phasic contractions to 50% MVC or higher, similar to the activity pattern of extremity muscles such as the FDI. This is however not a typical usage of the trapezius muscle.
4) Motor units often fire doublets when first activated, with interpulse intervals of 20 seconds or less.
5) During static contractions, motor units may be active throughout the observed period (10 min.). A slight slowing of the muscle fiber action potential is sometimes observed.
6) If new motor units are recruited, the activity of already active motor units is suppressed. Units with long periods of activity may stop firing altogether, preferentially at times of short breaks in the surface EMG activity pattern down to no activity (EMG gaps). Motor units thus suppressed may become active at a later time.

DISCUSSION

The observed activity patterns of trapezius motor units are consistent with the hypothesis of some forms of occupational shoulder complaints being related to the activity pattern of low-threshold motor units, as inferred from the vocational studies cited in Introduction: 1) Since the activity of individual motor units is largely unrelated to overall activity level, no relationship between muscle pain development and the activity level observed by surface EMG in vocational recordings is expected. 2) EMG gaps provide an opportunity for long periods of rest/recovery of individual muscle fibers.

The mechanisms for achieving the observed suppression or displacement of motor units are unknown. Conceivably, a period of prolonged activity may increase the threshold for activation, and other motor units with a similar activation threshold are preferentially activated. Thus, the motor unit activity patterns in prolonged activation may be determined by a fatigue-like phenomenon at spinal level, independently of fatigue effects in the muscle.

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MOTOR UNIT CONTROL PROPERTIES IN CONSTANT-FORCE ISOMETRIC CONTRACTIONS

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INTRODUCTION

This study was carried out in order to 1) characterize the decrease observed in mean firing rates of motor units in the first 8 to 15 s of isometric constant-force contractions; and 2) investigate possible mechanisms which could account for the ability to maintain force output in the presence of decreasing motor unit firing rates.

METHODS

The decrease in mean firing rates was characterized by investigating myoelectric signals detected with a specialized quadrifilar needle electrode from the first dorsal interosseus (FDI) and the tibialis anterior (TA) muscles of 19 healthy subjects during a total of 85 constant-force isometric contractions at 30%, 50% or 80% of maximal effort. The firing times of motor units were obtained from the myoelectric signals using computer algorithms to decompose the signal into the constituent motor unit action potentials. Time-varying mean firing rates and recruitment thresholds were also calculated.

The role of agonist/antagonist muscle interaction in maintaining of the force output with decreasing firing rates was investigated using the muscles controlling the wrist joint. Myoelectric signals were recorded with quadrifilar needle electrodes from the wrist extensor muscles while concurrently monitoring myoelectric activity in the wrist flexor muscles with surface electrodes during constant-force isometric wrist extension at 50% of maximal effort.

RESULTS

Motor units detected from the TA muscle were found to have a continual decrease in their mean firing rates in 36 of 44 trials performed during isometric ankle dorsiflexion at force values ranging from 30% to 80%maximal effort and a duration of 8-15 s. Likewise, motor units detected in the FDI muscle displayed a decrease in firing rate in 32 of 41 trials performed during constant-force isometric index finger abduction for contractions ranging from 30% to 80% maximal effort. In 14 contractions(16% of total) firing rates were essentially constant, while in 3 contractions (4%) firing rates appeared to increase. Motor units with the higher recruitment thresholds and lower firing rates tended to display the greater decreases in firing rate over the constant-force interval, while motor units with lower recruitment thresholds and higher firing rates had lesser rates of decrease. Furthermore, increasing contraction levels tended to intensify the decrease in the motor unit firing rates.

Three possible mechanisms were considered as possible factors responsible for the maintaining of force output while motor units decreased their firing rates: motor unit recruitment, agonist/antagonist interaction, and twitch potentiation. Of these, motor unit recruitment was discarded first as none was observed during the 8 to 15 s duration of any of the 85 contractions. Furthermore, contractions outside the physiological range of motor unit recruitment (at 80% maximal effort) revealed the same decreasing trend in firing rates, ruling out recruitment as the means of sustaining force output. In the experiments performed to investigate the role of agonist/antagonist muscle interaction, firing rates of the motor units in the wrist extensor muscles simultaneously decreased while the flexor muscles were determined to be inactive.

CONCLUSIONS

All the findings of this study regarding the behavior of the firing rates could be well explained by the reported characteristics of twitch potentiation which have been previously documented in animals and man (Vandervoort, et al. 1983; MacIntosh, et al., 1994). The results of this study, combined with the results of other investigators, provide the following scenario to explain how a constant-force isometric contraction is sustained. As the contraction progresses, the twitch force of the muscle fibers undergoes a potentiation followed by a decrease. Simultaneously, the "late adaptation" property of the motoneuron (Kernell, 1965) decreases the firing rate of the motor unit. Findings of this study suggest that voluntary reduction in firing rates also can not be ruled out as a means to augment the adaptation effect in motoneurons. As the duration of the contraction increases and the twitch force decreases, the force output of the muscle would be reduced below the intended force level and a voluntary increase in the excitation to the motoneuron pool would be required to sustain the intended force. This excitation would manifest itself in an increase in the firing rates of the active motor units and a recruitment of new motor units.

ACKNOWLEDGEMENT

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MOTOR UNIT FIRING RATES IN ISOTONIC VOLUNTARY CONTRACTION

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INTRODUCTION: Motor unit firing rates were measured in isometric contraction obtained from various muscles. There were few reports in which isotonic voluntary contraction were investigated. The purpose of this study was to measure the motor unit firing rates in isotonic voluntary contraction from the human brachial biceps muscle using the surface electrodes developed by authors.

METHOD: We measured myoelectric signals from the human brachial biceps muscle in isotonic voluntary contraction. The subjects flexed their elbows in constant angler speed against a load. To measure myoelectric signals, we devised a surface electrode arranged as shown in Fig.1. Using six electrodes, we got five channel myoelectric signals. Channel 1, 3 and 5 are parallel to the muscle fibers while channel 2 and 4 are oblique.

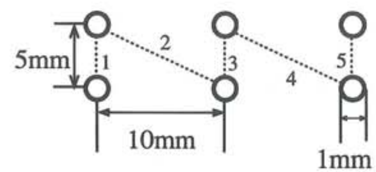

Fig.1 Configration of electrode

RESULTS: Figure 2-1 shows wave form examples. A, B and C indicate the time with upper arrows on the time axis in Fig.2-2. The action potential shown on channel 4 with closed circle indicates firing time of the same motor unit. Using identified wave forms, we calculated the firing rates by the inverse of inter-spike interval. These results are shown in Fig.2-2. The firing rates ranged between 12Hz and 16Hz. Figure 3 shows the relation between load torque and firing rates in various flection speeds. Each of them is the mean value of the firing rates which is obtained with isotonic contraction of one trial. As load torque increases, firing rate increases. Also as flection speed becomes faster, the motor unit firing rates increase.

CONCLUSION: We measured motor unit firing rates in isotonic contraction. The elbow angle flection speed was from 0 to 30deg/sec and the load torque was from 0.6 to 3.7Nm. Firing rate of a motor unit increased according to the increase of load torque and flection speed.

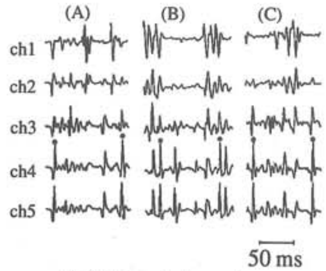

Fig.2-1 Example of wave forms

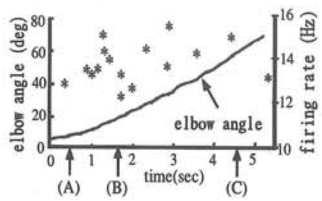

Fig.2-2 Example of firing rates

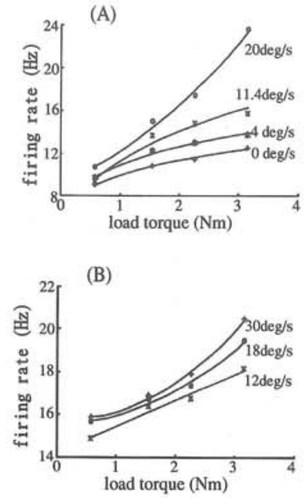

Fig.3 Relation between load torque and firing rates

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MANDIBULAR STRETCH REFLEX SENSITIVITY DURING CYCLIC JAW MOVEMENTS IN HUMANS

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INTRODUCTION — Reflex activity in the EMG of the jaw-closing muscles is produced when these muscles are stretched during isometric clenching or when a downward directed load is applied during voluntary closing of the jaw. This so-called jaw-jerk reflex is the result of stretching the muscle spindles. A linear relationship between jaw displacement and reflex amplitude has been observed for displacements up to 0.2 mm during isometric contraction of the jaw-closing muscles (Lobbezoo *et al.* 1993a). The reflex amplitude also depends on the excitability of the alpha motoneurons. A linear relationship between the reflex amplitude and the amount of EMG activity was observed for clenching levels up to 20 % of maximum voluntary clenching (Lobbezoo *et al.* 1993b). The reflex amplitude is further related to the amount of gamma drive of the muscle spindles, which adjusts the sensitivity of the spindles for different muscle lengths and during various motor tasks. Finally, the reflex amplitude will be affected by presynaptic inhibition of the input to alpha motoneurons from the muscle spindles. The muscle spindle feedback to the jaw-closing muscles is modulated during chewing. The jaw-jerk reflex is undesirable during the opening phase. Animal studies have shown that stretch of muscle spindles during jaw opening has little effect on the activity of the motoneurons of the elevator muscles (Chase and McGinty, 1970; Goldberg *et al.* 1982). Modulation of the jaw jerk reflex has never been studied under dynamic conditions in man, probably because of technical difficulties in applying reflex-evoking stimuli at different jaw positions in a controllable manner during chewing.

The aim of this study was to examine the modulation of the monosynaptic jaw-jerk reflex in man during various phases of rhythmic open-close movements. Furthermore, the relationships between the reflex amplitude and the displacement of the jaw and between the reflex amplitude and the amount of muscle activity were determined.

METHODS — Subjects made rhythmic mandibular open-close movements at their natural chewing rate controlled by a metronome. In some of the open-close movements stretch reflexes were elicited in the EMG of the jaw-closing muscles by force impulses exerted upon the dental arch of the lower jaw. The force impulse was supplied by a magnet-coil system (Ottenhoff *et al.* 1994). A coil could move freely up and down in a homogeneous magnetic field and was rigidly attached to the subject's mandible by means of a clutch cemented upon the teeth. A square shaped electric current impulse through the coil with a duration of 10 ms generated a force of 3, 6 or 9 N acting in a downward direction on the mandible. The onset of a reflex eliciting force impulse was dependent upon jaw gape. Light emitting diodes (LEDs) were attached to the upper and lower jaw and their position was recorded by means of an optical motion analysis system (Optotrak). The force impulse could be unexpectedly applied at one of the following 5 phases of the closing part of the cycle: at the beginning of jaw closing, early, halfway and late during closing and at occlusion. The amplitude of the reflex amplitude may depend on the amount of muscle activity needed to close the jaw. Therefore, the experiments were also performed with an external resisting force, which simulated food resistance. This force was present during the closing phase of each cycle. The force was computed on-line as a function of jaw gape. The amplitude of the force increased proportionally with the decrease in jaw gape up to a preset maximum. Experiments with force impulses of 3, 6 and 9 N were performed with food simulating resisting forces of 0, 10 and 20 N for each of the 5 phases. The experiments were repeated 30 times with and without a force impulse. The jaw movement and force as well as the activity of masseter and temporal muscles were recorded. The signals were averaged off-line with respect to the onset of the force impulse.

RESULTS — Fig. 1 shows an example of a recording of the jaw gape, the force and an EMG signal of an experiment in which a force impulse was given halfway through jaw closing. In both cycles jaw-closing was counteracted by a force of 20 N. In the second cycle a reflex was elicited. The force impulse disturbed the closing movement, which resulted in a small deflection in the jaw-gape signal of the cycles with an impulse (Fig. 1). The deflection depended linearly on the amplitude of the force impulse. The deflection was largest (0.5 mm) for the 9 N force impulse for the condition without a food simulating external force at the beginning of the closing phase. Clear bi-phasic jaw-jerk reflexes were elicited by the force impulses at all 5 phases. The first peak of the stretch reflex response occurred on average 11.7 ms after the onset of the force impulse. The reflex amplitude increased linearly with the amplitude of the force impulse. Furthermore, in each phase larger reflex amplitudes were observed for larger additional food simulating forces.

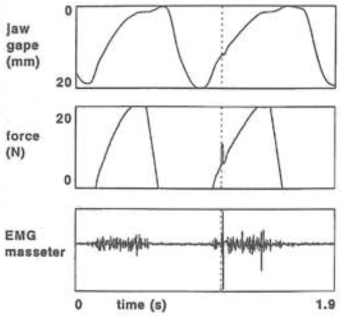

Figure 1 Jaw gape, applied force and EMG of masseter muscle.

DISCUSSION — The latency of the peak of the jaw-jerk reflex elicited during open-close movements is longer than the latencies reported for jaw-jerk reflexes elicited under static conditions (Lobbezoo *et al.* 1993a). The increase in the latency of the reflex of about 3 ms may be caused by the fact that the fusimotor drive to the shortening intra-fusal muscle is insufficient to keep the muscle spindles in their optimal range. The increase in reflex amplitude for larger force impulses is caused by an increase in muscle spindle stretch. The reflex amplitudes elicited during open-close movements linearly increase as a function of displacement just as was the case under static conditions. The background activity of the elevator muscles increased during jaw closing for the experiments without and with an external resisting force. This increase in muscle activity will lead to larger reflex amplitudes (Lobbezoo *et al.* 1993a). The reflex amplitudes were corrected for the background activity and for the mechanical effect of the force impulse in order to elucidate any dependency of the reflex amplitude upon the phase of closing. The corrected reflex amplitudes appeared to be largest at the very beginning of jaw closing. The larger reflex sensitivity at the beginning of the closing movement may be the result of an adjusted gamma motoneuron activity or of an adjusted presynaptic inhibition.

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A NEW EXPERIMENTAL SIMULATION TECHNIQUE TO ASSESS THE IMPACT OF MOTOR UNIT ACTIVATION STRATEGIES ON EMG

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INTRODUCTION

The electromyogram (EMG) is a complex interference signal resulting from the summation of trains of motor unit (MU) action potentials (MUAP). Its features are determined by peripheral (MUAP shape) and central (MU firing) parameters. However, the relative contributions of these parameters to EMG characteristics cannot be determined from EMG recordings alone. This question has therefore mainly been dealt with in theoretical and computer simulation studies (1,2). To complement and evaluate such approaches an experimental EMG simulation technique has been developed, which is based on multi-channel independent stimulation of single MUs in a cat nerve-muscle preparation. This technique allows the investigator to manipulate MU pool activation strategies in a controlled manner, and to design experiments to identify the contribution of firing statistics to EMG properties under conditions where motor drive is completely specified. In the case of EMG simulations, one of the crucial issues is the need of adequately representing the range of MUAP shapes which are encountered in a real muscle. The principal feature of the experimental simulation technique is that real EMG signals, which are generated by the activation of single MUs, are recorded experimentally under conditions where the investigator has full control of the strategy of activation of a substantial proportion of the MU pool.

METHODS

An acute muscle preparation was used in 10 cats maintained under general anaesthesia. With the exception of a single muscle, the left hindlimb was widely denervated, allowing for the functional isolation of single motor axons by a microdissection technique of the ventral roots (L7, S1). The experimental EMG simulation is based on independently controlled electrical stimulation of sets of functionally single and type-identified MUs (see below). Data acquisition consisted of high resolution force recordings, along with the recording of bipolar surface and indwelling EMG signals with a custom-designed patch electrode and insulated wire electrodes, respectively.

The essence of the method are the design features which enable the independent and repeatable stimulation of sets of single MUs. Individual stimulation profiles are generated with custom-designed computer software which defines lists of inter-stimulus-intervals (ISIs) and which in effect imitates a voltage-controlled oscillator (VCO) operating in real time. Once generated and optionally edited, these patterns are stored in mass-memory libraries and can be reproduced identically within and between experiments. Individual patterns have two main features, a background profile and a component of added Gaussian noise, to permit adequate imitation of physiological discharge rate variability. Specific algorithms have been developed to permit control of synchronization between parallel channels and/or to guarantee that strictly independent individual ISI patterns are generated even if background profiles are identical. With this general design the imitation of different recruitment and rate coding (activation) strategies is reduced to simple specification of appropriate families of background profiles. Finally, for experimental applications the use of high-speed microprocessors and interfaces permits strictly simultaneous generation of up to 16 ISI patterns, with a temporal resolution of 0.05 ms.

With the improved design of stimulation electrode arrays, simultaneous activation of sets of up to 10-12 MUs is now possible. A total of up to 48 independent stimulation patterns are used to activate four independent sets of single MUs. Normally, the MUs of each set are rank ordered according to the size (tetanic force), so that in typical simulations of activation strategies the smallest MUs are activated first (3). However, recruitment according to size is not a prerequisite. Estimates of EMG and force, as generated by activation of 40-48 MUs, are obtained by summing signals recorded successively during stimulation of four independent subsets (10-12 MUs).

40 Channel Pure Recruitment Stimulation Design

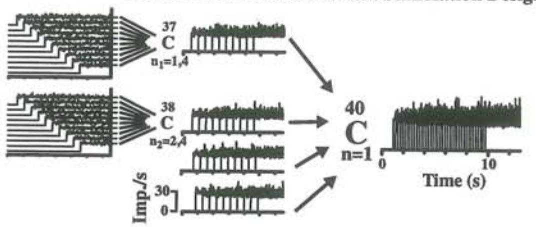

Figure 1: Combination of four 10-channel stimulation profile sets to permit simulations with activation of 40 independent channels, each with its individual activation pattern. Individual stimulation patterns for sets 3 and 4 not shown.

RESULTS

Qualitatively, typical time domain features of EMG are adequately simulated with combined activation of 10-12 separate single MUs. In addition, the frequency characteristics of voluntary EMG are readily reproduced, as power spectra calculated from simulated EMG are indistinguishable from those of physiological EMG. However, with such a small proportion (10-12) of the total MU pool (120-150 MU), the degree of interaction between the MUAPs does not accurately reflect certain temporal features of the EMG recorded during activations of larger numbers of MUs (4). For instance, the stimulation of a set of 10-12 MUs produces discontinuous gradations of both EMG and force signals due to the small number of recruitment steps, which is not observed during more physiological whole muscle activation.

The main issue that had to be addressed was whether linear summation of intermediate forces and EMG signals, as generated by small numbers of MUs, was permissible in order to simulate activation of larger sets of MUs. Experiments were carried out with multi-MU filaments which were micro-dissected and regrouped until - judging from the tetanic force generated - they contained on average 10 MUs, thus matching the forces and EMG signals produced by typical 10-MU sets. Linearity of summation was assessed by comparing the sum of the individual signals elicited by separate multi-MU filament stimulation, with the magnitude of the signals elicited by combined stimulation. The main finding was that under isometric conditions intermediate forces summed in an almost perfectly linear manner. Likewise, the EMG signals elicited by stimulation of these multi-MU filaments summed linearly. Furthermore, these summed signals no longer revealed the discontinuities attributable to recruitment of new MUs, and more importantly, the 40-channel simulated EMG could not be distinguished from voluntary EMG recordings.

APPLICATIONS

Several quite diverse applications of the EMG simulation method can be, and partly are, pursued: first, the assessment of theoretical models of EMG and of predictions on the effects of different central activation strategies on EMG characteristics; secondly, evaluation of the scope of inferences on underlying central MU pool activation strategies that can be made from quantitative (including spectral) EMG analysis; and thirdly, specific investigation of the impact on, and detectability from, EMG of various (putative) motor unit activation/adaptation strategies in muscular fatigue, such as the notion of recruitment of fresh motor units vs the concept of "muscular wisdom" of adapting activation rates.

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RESTORING GAIT BY FUNCTIONAL ELECTRICAL STIMULATION (F.E.S) FOR THE SPINAL CORD INJURED (S.C.I.) PATIENTS. A CLINICAL STUDY.

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INTRODUCTION

Various electrical stimulation technics have been developed, starting from hybrid orthosis , which associate F.E.S. to long leg braces (LLBs) , to implanted devices for the restauration of gait in spinal cord injury patient. We present here the results of a clinical study on 13 SCI patients conducted in the University Rehabilitation Departments of Rennes .

METHODS

13 paraplegic patients (9 males and 5 females), whose ages ranged from 17 to 42 years (mean : 29 years) , with a spinal cord lesion were studied. 11 patients were classified Frankel A,1 Frankel B and 1 Frankel C.The duration of the paraplegia ranged from 5 to 240 months with an average of 48 months. The etiology was a motor vehicle accident, in 11 cases, a fall in one case and infectious spondylitis in the last case. The level of the lesion was thoracic in all cases, between T4 and T12 .

The FES system consists in a microcomputer controlled neuromuscular stimulator . Six stimulation channels are avalaible : 2 for the quadriceps, 2 for the peroneal nerve stimulation sites, and 2 for the gluteus or lower spinal muscles. Surface skin electrodes are connected by lead cables to the unit.The device generates an alternating current monophasic symetrical wave form (with zero net charge), a pulse width of 300 μs, and a frequency of 24 Hz. The amplitude of stimulation can be adjusted from 0 to 300 mAmp.

A specially designed walker permits stability for ambulation . A propylene ankle-foot (short orthosis) was used by patients to provide ankle stability during ambulation.

RESULTS

The rate of the sessions was 3 to 5 per week. Each session lasted for about 2 hours. In most cases ,training was conducted on an outpatient basis; but for some patients living far from the center , it was necessary to admit them as inpatients during the course of their training.

The first standing with the Parastep was achieved quickly, most often within the two first sessions. The first steps with the walker were also quick, and took place around the tenth session, approximatively 2 or 3 weeks after the beginning of the programme.

One patient had not yet complete the training. The patient who were Frankel B , in spite of quick progress, gave up because of the pain induced by the electrical stimulation. The patient, who were Frankel C , has recovered a walk without orthosis, and also did not need to use parastep anymore . All the other patients have acquired independent ambulation with the Parastep,except one patient. She was unable to reach the stage of ambulating with the walker but she was able to stand up by herself in a safe way.

Walking distances showed large differences within the patients, with a maximum of 350 meters without rest. The mean walking speed was about O.2 m/s, with a maximum of 0.6m/s, which is slow. Incidental diseases, such as urinary tract infections, induce a decrease of the fatigue threshold, and, as a result , a decrease of independent walking distance.

A statistically significant increase of muscular strength was observed , showing noticeable variations amongst the patients. The cardiovascular tolerance was excellent.Complications affected mostly musculo skeletal system. 2 patients complained of back pain and 2 patients presented a benign ankle sprain . A fracture of the sacrum has been reported in another patient , resulting from a fall without any functional consequence . 2 patients out of the 10, who became independent have given up after a few months , althought they were independent for walking . For the seven patients who possessed their own device, we have now a mean follow up of 2 years. 5 are using their Parastep regurlaly . One patient has given up after a daily use of one year and another uses it for static standing.

In the cases where the Parastep is used regularly at home, we must notice that it is not used to increase ambulation autonomy in daily life , but as an active mean of exercise, in order to prevent complications of immobilization, and to answer the desire to stand and walk .

CHAPTER 3

REHABILITATION

This device appears as reliable. No technical failure was noticed throughout this study. The training was quick and the learning simple.

DISCUSSION

This panel of patients, although limited, provides interesting data . This device is reliable, easy to handle, and has a good cosmetic acceptance . Only one patient of our group has given up during the training program. Standing and gait are acquired in a few days . This observation confirms these of Graupe (1).

However, compared to normal gait, performance is modest. A great variability exists between patients : walking distance ranges from 4 to 350 meters . Marsolais (2) has reported a maximum distance of 330 meters, after a 3 years ' use of his implantable system . Krajl () and Graupe (1) mention performances close to ours , with, for one patient , a distance of 2 kilometers . The speed remain the main limiting factor , whatever the technic used, because it is far from normal speed(1,2,3,4) .

Therefore the goal of FES synthesized gait restoration should not be an attainment of normal ambulation, but rather the realization of benefits associated with biped locomotion as part of the management of spinal cord injury .We have noticed an increase of muscle efficiency during the training, with an improvement of the strengh of the quadriceps and of the motor performances.

These data are consistent with those reported in previous studies (1).

Various clinical parameters must be carefully reviewed before admission into the program : joint integrity, obesity, which can induce backache ,by overloading the posterior joint of the spine, and osteoporosis in old paraplegias.These parameters are not, for us, strict contraindications, but a rigorous supervision is needed.

None of the patients uses this device for daily life activities. This fact is observed , as a rule , for all ambulation technics for paraplegics (4).

CONCLUSION

The Parastep is a reliable FES technology , which patients can learn easily . The results of this study show a quick progression in the acquisition of gait performance, and noticeable psychological benefits. This method is not indicated for all paraplegic people, and requires the subjects to be highly motivated.

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RELATIONSHIPS BETWEEN LUMBAR DYNAMOMETRY PERFORMANCES AND PSYCHOLOGICAL FACTORS IN LOW BACK PAIN (LBP) PATIENTS.

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INTRODUCTION

Literature and clinical experience strongly indicate that many patients with chronic low back pain develop a "deconditioning syndrome". These patients develop pain and illness behavior which leads to a deconditioning of the back muscles. This deconditioning leads in its turn to pain and illness behavior. Patients land in a vicious circle with an ever decreasing physical condition of the back muscles (Mayer *et al*, 1985).

The objective of this study was to investigate possible relationships between some psychological aspects, pain experiences and low back muscle condition, as suggested by the "definition" of the deconditioning syndrome.

METHODS

Low back muscle condition was assessed by means of lumbar dynamometry measurements carried out with the Isostation B200 (Isotechnologies, USA) according to the OOC protocol. Range of motion, torque and velocity parameters were measured. The psychological complaints were assessed by means of the Symptom Checklist (SCL-90) (Arrindell & Ettema, 1986). Pain experience was assessed by the multidimensional pain questionnaire (MPI-DV) (Dingemans *et al*, 1993).

45 patients with chronic low back pain participated in this study.

Data were analyzed by means of correlation coefficients and analysis of variance. Investigated were the relationships between absolute scores on the questionnaires and different lumbar dynamometry parameters as well as relationships between absolute scores on the questionnaires and the number of lumbar dynamometry parameters below the fifth percentile with respect to a healthy population.

RESULTS

Results demonstrate significant relationships between lumbar dynamometry measurements and a number of psychological aspects of the SCL-90: depression, somatization, insufficiency and the psychoneuroticism score, which is a measure for psychic dysfunction. For the MPI-DV significant relationships are found for pain and interference of pain.

Higher scores on the SCL-90 and MPI-DV are correlated with more parameters below the fifth percentile and worse condition of the back muscles.

This means that, in general, more psychological complaints, more pain and more interference of pain correlate to a worse low back muscle condition.

DISCUSSION

From the results of this study can be concluded, that there is a relationship between lumbar dynamometry measurements and some psychological factors, pain intensity and interference of pain, as assessed with the SCL-90 and MPI-DV.

This suggest that the deconditioning syndrome can be described as two spirals which influence each other. One spiral is representing the decreasing physical condition, the other the increasing psychological complaints, increasing pain intensity and increasing interference of pain in everyday live.

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INFLUENCE OF PELVIC TILT ON BALANCE CONTROL IN SPINAL CORD INJURED PEOPLE

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INTRODUCTION

During rehabilitation spinal cord injured (SCI) people are trained to use non-postural muscles like the latissimus dorsi (LD) and the trapezius pars ascendens (TPA) to partially compensate the loss of postural muscle activity. These muscles have both a favourable biomechanical position relative to the trunk, spine and shoulder girdle and an innervation cranial to the lesion level. Closely related to the difference in postural motor control are the gross kinematic differences between SCI people and non-SCI people, especially concerning pelvic tilt and movement of the spine and trunk. This lead to the following research question: How does pelvic movement correlates with sitting balance control in SCI people?

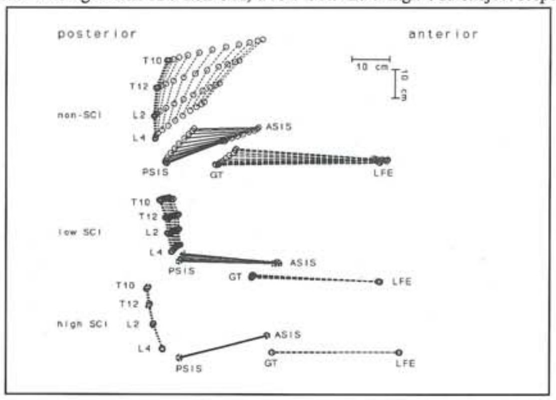
METHODS

The experiment included ten high thoracic (T2-T8) SCI people, ten low thoracic (T9-T12) SCI people and ten people without a spinal cord injury. The SCI people had complete lesions and had completed the active rehabilitation process at least six months. The three groups were matched for age, sex, height, weight and educational level. Subjects had to perform a bimanual submaximal reaching task in anterior direction towards button pairs placed on a table in front of them. Button pairs were placed at 15%, 30%, 75% and 90% of the individual maximal reaching distance, introducing systematically graded balance disturbance. All subjects sat in the same multi-adaptable chair. Three different chair configurations were tested: 1) a standard chair configuration (0° tilting and 10° reclining), 2) the same chair configuration fitted with a cushion that caused a 10° forward inclination of the seat and 3) a chair configuration of 7° tilting and 17° reclining fitted with the cushion that caused 10° forward inclination of the seat. Muscle activity of the bilateral erector spinae (ES) at level L3, T9 and T3, latissimus dorsi, trapezius pars ascendens, pectoralis major (PM), serratus anterior (SA) and the oblique abdominal (OA) muscles, were recorded by means of surface EMG. Sitting balance was monitored by measuring the change in location of the centre of pressure (COP) using an AMTI force platform mounted underneath the chair. Pelvic tilt and spinal movement were recorded using a 3D-movement analysis system (PRIMAS). Retroreflective markers were placed on the following anatomical landmarks: spinous processes at level L4, L2, T12 and T10, posterior superior iliac spines (PSIS), anterior superior iliac spines (ASIS), greater trochanters (GT) and the lateral epicondyles of the femur (LFE). Four infra red cameras stood in a semi circle behind the chair. To prevent the markers from being masked from the cameras the backrest of the chair was kept very small. The backrest gave trunk support cranial to the marker on T10. Spatial displacement of the markers was calculated. From the real world coordinates of the bilateral ASIS and PSIS markers a unit vector system, per image frame, was calculated for the pelvis. The first unit vector Z was parallel to the line through both ASIS. The second vector X was perpendicular to Z and its workline crossed the midpoint between both PSIS. The third vector Y was perpendicular to the first and the second one. The second and the third vector define the sagittal plane of the pelvis. Changes of the angle of the third unit vector in the sagittal plane correspond to the relative antero-posterior pelvic tilt during task performance. Preliminary kinematic and EMG results will be discussed.

RESULTS

Evaluation of marker displacement, an example of which is shown in Figure 1, revealed that non-SCI people use more pelvic tilt and trunk flexion than SCI people while making a reaching movement towards 90% of their individual maximal reaching distance. Data analysis of marker displacement of the low SCI subject showed an extension of the upper part of the spine and a backward movement of the head. This phenomena can be seen in all low SCI subjects. People with a high SCI keep their pelvis tilted posteriorly and use the backrest of the chair for support during task performance. These biomechanical findings are supported by EMG data of the ES, LD, TPA, PM, SA and OA muscles recorded simultaneously.

Figure 1. Side-view of marker displacement during forward reaching while sitting in the standard chair configuration of a non-SCI, a low SCI and a high SCI subject respectively.



DISCUSSION

Due to the growing lack of activity of the lumbar ES, at increasing levels of SCI, paraplegic people are not able to actively tilt their pelvis forward. To compensate for instability of the pelvis and the lower part of the spine, high SCI subjects tilt their pelvis backward and use the backrest of the chair for support, to increase the base of support. This position also causes the punctum fixum to rise relative to the base of support, thus creating more possibilities for more cranial muscles like the LD and TPA to become active in support of the erect posture of the trunk. People with a low SCI counteract the forward displacement of the upper limbs, which coincides with a disturbance of sitting balance, by extending the upper part of the spine and moving the head backwards. In this way the centre of gravity is kept within the base of support.

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POSTUROGRAPHIC IDENTIFICATION OF POSTURAL INSTABILITY AFTER TRAUMATIC BRAIN INJURY (TBI)

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INTRODUCTION

The aim of this study was to determine whether brain-injured subjects complaining of balance problems that could not readily be confirmed by standard clinical neurological examination could be discriminated from healthy subjects by using quantitative force-platform recordings. If so, the question was raised what conditions and parameters would be most sensitive to reveal these latent, but potentially disabling postural impairments. Indeed, reduced gross motor ability is a common complaint in TBI patients, even if they show no signs of paresis, ataxia or sensory loss.

METHODS

During one-and-a-half year, posturographic measurements were made in all consecutive subjects who visited a specialized out-patient rehabilitation department for advice regarding long-term TBI-related disabilities. Each subject had suffered a period of impaired consciousness and/or post-traumatic amnesia after the injury. All subjects had initially been hospitalized for at least 24 hours, the primary diagnosis being TBI. Only patients who (among other post-concussive symptoms) complained of postural instability since their head trauma were included. The following exclusion criteria were applied: (1) clinically abnormal muscle strength, muscle tone, sensation, coordination, balance, gait, as well as visual, vestibular or oculomotor impairments detectable by standard neurological examination, (2) vertigo, (3) pre-existent neurological or psychiatric disease, (4) a history of skull fractures, (5) established cervical injury, (6) musculo-skeletal abnormalities, and (7) medication affecting balance. Thus, 20 patients (11 men, 9 women; mean age 36.2±10.7 years; mean time since trauma 28.6±25.7 months) were included. Based on the lowest Glascow-coma-scale (GCS) scores *after* hospitalization, 13 patients had sustained mild (GCS 13-15), two moderate (GCS 9-13) and five severe (GCS 3-8) TBI. The control group consisted of 20 healthy volunteers who were matched for age and gender.

A dual-plate force platform recorded the center-of-pressure (CP) fluctuations in the antero-posterior (AP) and lateral (LAT) sway directions during quiet two-legged standing for 30 seconds (sampling rate 60 Hz; distance between heels 8.4 cm; toeing-out angle 9°). This balance task was combined with an arithmetic task to estimate the automaticity of the posture-control system, whereas visual deprivation was used to detect sensory integration deficits. Hence, 3 consecutive quiet-stance conditions were tested: (1) with eyes opened as a single task, (2) while simultaneously performing an arithmetic task ("dual task"), and (3) while wearing a pair of dark spectacles ("eyes closed"). The arithmetic task consisted of a verbal presentation of 8 single-digit additions equally timed over a 30-second period. The addition problems were selected at random and followed by a sum that could be either correct or incorrect. The subjects had to verbally indicate the correctness of each summation by "good" or "fault" responses. The number of arithmetic errors was noted.

Each balance assessment consisted of two identical test series, starting with the performance of the arithmetic task in a sitting position as a single task, followed by the 3 balance tests. Both the RMS amplitude (A_{cp}) and velocity (V_{cp}) of the CP fluctuations were calculated for the AP and LAT sway directions, separately. Ultimately, the comparable parameters derived from the two test series were averaged for statistical analysis. Each of the dependent variables was tested in a two-way MANOVA of Group by Condition with repeated measures on the last factor.

RESULTS

The TBI group generally showed 50% to 70% greater V_{cp} values than the control group for both the AP, $F(1,38)=14.05$, $p<.005$, and LAT sway, $F(1,38)=10.44$, $p<.005$. In addition, there was a main effect of condition on the CP velocity in both the AP, $F(2,37)=25.22$, $p<.001$, and LAT directions, $F(2,37)=8.80$, $p<.005$. Only in the LAT direction, a significant Group x Condition interaction was found, $F(2,37)=3.69$, $p<.05$, which was exclusively related to the eyes-closed condition. Hence, no indication of enhanced dual-task interference was found in the TBI group.

The A_{cp} values showed a similar pattern: main effects of group (AP: $F(1,38)= 8.90$, $p<.01$; LAT: $F(1,38)=8.61$, $p<.01$), condition (AP: $F(2,37)=9.61$, $p<.001$; LAT: $F(2,37)= 7.22$, $p<.005$) and the Group x Condition interaction only significant for the LAT sway, $F(2,37)=6.02$, $p<.01$. The arithmetic data revealed no significant effects.

DISCUSSION

This study was conducted to examine whether a (latent) reduction in postural control can be objectified as a long-term consequence of TBI. The results are part of a larger assessment of both static and dynamic aspects of postural dyscontrol after TBI.[1] Because postural instability has already been recognized as a consequence of severe closed head injury usually in subjects with clear sensorimotor deficits,[2] 20 TBI patients were included in whom no sensorimotor impairments could be demonstrated by standard clinical neurological examination. Nonetheless, all selected patients complained of postural instability. Thirteen patients were classified as having sustained mild TBI.

The results clearly indicate the possibility to identify latent balance problems in this subgroup in comparison to healthy controls. Amplitude and velocity parameters seem to be equally discriminative. The most sensitive condition appeared to be visual deprivation, particularly for LAT sway control, indicating that sensory integration deficits can be part of the aftermath of brain injury.[1,2] Loss of balance automaticity could not be demonstrated, although the basic level of postural control was clearly affected. It must be stressed that the finding that postural instability can be identified in this "self-selected" sample does not imply that all or even most persons will develop such problems following TBI. It merely indicates that balance problems *can* be present months or years (also) after mild TBI, even when clinical neurological signs have normalized.

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DEVELOPMENT OF NOVEL SITTING BALANCE STRATEGIES IN PARAPLEGIC SUBJECTS DURING CLINICAL REHABILITATION.

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INTRODUCTION

In non-sensorimotor impaired persons the erector spinae (ES) is used to stabilise the spine against e.g. gravitational forces during standing and sitting. In people with a thoracic spinal cord injury (SCI) this postural control function of the ES is impaired. It has been shown that in paraplegic subjects who had completed their clinical rehabilitation process this loss of stabilising effects of the ES is partially compensated by the use of the latissimus dorsi (LD and the trapezius pars ascendens (TPA), which are thought to be non-postural, phasic muscles [1, 2]. It was also shown that, in order for the LD and TPA to become stabilisers of the trunk in thoracic SCI subjects, the protractors of the scapulae (e.g. the pectoralis major (PM) and the serratus anterior (SA)) were activated to stabilise the shoulder girdle [2]. In this presentation the issue of the development of these new muscle coalitions for postural control during task execution in thoracic SCI patients during their active, clinical rehabilitation will be addressed.

METHODS

Twelve patients who were clinically diagnosed with a complete spinal cord lesion either at level T2-T8 (H-group, n=5) or at level T9-T12 (L-group, n=7) participated in this longitudinal study. Subjects were tested on their ability to control posture and restore sitting balance. During the initial 3 months of their stay in the rehabilitation clinic patients were tested every two weeks, starting approx. 1.5 weeks after admission. In the subsequent period experimental sessions took place every four weeks, until discharge from the clinic. Since not all subjects remained in the rehabilitation centre for the same length of time, data from 5 sessions which were identified as being equivalent for all subjects were used for further data analysis. These sessions coincided with: first participation after clinical admission (A), removal of spinal brace (B), the moment 4 weeks after B (C), clinical discharge (E) and the moment halfway between C and E (D). Postural changes during sitting were systematically elicited through bimanual reaching movements in antero-posterior direction over reaching distances of either 15, 30, 75 or 90% of each subjects individual maximal reaching distance. In order to remain or restore an upright sitting position subjects had to use muscles still controllable. Surface EMG activity of the bilateral ES (at level L3, T9 and T3), LD, TPA, PM, SA and oblique abdominal (OA) muscle was recorded, using a K_Lab MF-118 amplifier and SPA20/12 pre-amplifiers (K_Lab, Amsterdam, Neth.). Changes in the position of the centre of pressure (CP), induced by body movement during task execution, was monitored using an AMTI force platform (Biovec-1000, AMTI, Watertown, Mass.) underneath a chair in which subjects sat. Reaching distance covered by the subject was also recorded. Per session 2 x 48 trials were presented. Sample frequency for the EMG and CP recording was 500 Hz. Sample time per trial was 5 s. Data were initially stored on harddisc using MUCAPS software (IRV, Hoensbroek, Neth.). Mean EMG activity for all muscles during semistatic periods of maximal CP (CPmax) displacement per trial in all reaching conditions, during all sessions, was calculated. Statistical analysis for the within-group comparisons included Page tests for ordered alternatives, whereas Mann-Whitney U-tests were used for between-group analyses.

RESULTS

CPmax increased during rehabilitation only in the L-group (figure 1). At the end of the rehabilitation process the L-group showed higher CPmax displacements than the H-group. A reduction in CPmax displacement at completion of spinal bracing was observed in the H-group. In general EMG activity of the bilateral ES-T9, LD and TPA in the H-group increased significantly during clinical rehabilitation. In the L-group EMG activity increased in the bilaterai ES-L3 and ES-T9 across the rehabilitation period ($p<0.01$). No increase in SA activity over time was found in either of the groups. Between-group comparison of EMG activity showed higher muscle activity in the L-group in the bilateral ES-L3 during all sessions ($p<0.01$) except during the first session and in the ES-T9 at clinical discharge ($p<0.05$). The H-group showed more bilateral LD and TPA activity towards the end of the rehabilitation period ($p<0.01$). This was also the case for the left ES-T3 in the H-group ($p<0.05$).

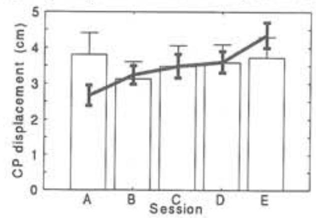

Figure 1.
Mean CPmax values and 99% confidence interval of the mean. Bar=H-group. Line=L-group. (For abbreviations see text.)

DISCUSSION

The aim of the study was to investigate the development of novel sitting balance strategies in SCI subjects during clinical rehabilitation. It seems that in both high and low thoracic SCI patients partial restoration of postural control functions of the ES at levels near to the SCI site occurs during rehabilitation. Also the use of the LD and TPA in balance restoration increased in patients with a high thoracic SCI, whereas in low thoracic SCI patients, whose ES was less impaired, compensatory postural muscle use was less. Present data are in agreement with the concept of alternative postural muscle use we proposed in earlier work [1, 2].

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THE EFFECT OF THE ROESSINGH BACK SCHOOL PROGRAM (RRP) ON LOW BACK MUSCLE CONDITION AND PSYCHOLOGICAL COMPLAINTS

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INTRODUCTION

Low-back disability is not only related to physical factors but also to psychological and social factors. A lot of patients become passive both in physical and social respect, with the consequence that their low back muscle condition decreases. Besides this, a lot of low back pain patients develop pain-and illness behavior (van Akkerveeken *et al*,1992).

Treatment of low back pain patients can only be effective if the intervention aims at both physical and psychological aspects. This means, it leads to an increase in low back muscle condition and to a reduction in psychological complaints. The Roessingh Back School Program (RRP) is a multidisciplinary treatment program for chronic low back pain patients with an integrated approach of the physical and social problems. The objective of this study was to investigate to what extend the RRP increases physical condition, decreases psychological complaints and changes pain experiences.

METHODS

Low back muscle condition was assessed by means of lumbar dynamometry measurements carried out with the Isostation B200 (Isotechnologies, USA) according to the OOC protocol. Maximum isometric torque, maximum and mean velocity parameters were measured. The amount of physical activity during work, sport and leisure time were assessed with the Physical Activity Questionnaire (FAV) (Baecke *et al*, 1982). Psychological complaints were assessed by means of the Symptom Checklist (SCL-90) (Arrindell & Ettema, 1986). Aspects of pain experience were assessed by the multidimensional pain questionnaire (MPI-DV) (Dingemans *et al*, 1993). 62 patients with chronic low back pain participated in this study. The lumbar dynamometry measurements and the questionnaires were taken both before and after the RRP. Wilcoxon tests and paired T-tests were used to investigate whether significant differences exist in mean parameter values between the measurements before and after the RRP.

The Reliable Change Index (RCs) was used (Jacobson and Truax, 1991) to investigate, for each subject individually, whether significant differences in parameter values exist between the measurement before and after the RRP.

RESULTS

Results of Wilcoxon tests and paired T-tests show significantly higher mean scores for the factor Sport of the FAV and for all lumbar dynamometry parameters after the RRP compared to before the RRP. Significantly lower mean scores after the RRP are found for the factors

Depression, Insufficiency, Somatisation and Total score of the SCL-90, Pain and Interference of pain of the MPI-DV and Work of the FAV.
The RCs of the lumbar dynamometry parameters show that 20 à 25% of the patients have significantly higher scores after the RRP compared to before the RRP. The RCs of the SCL-90 show that there are hardly patients with a significant improvement in the psychological complaints assessed with this questionnaire. For the MPI-DV, 12% of the patients show a decrease in pain intensity and 24% of the patients experience less interference of pain in everyday live. In contrast to this, there are hardly patients who show a significant improvement in control over pain and in general activities. For the FAV, the RCs show that 14% of the patients show significantly more physical activities during sport and that there are hardly patients who show significantly more physical activities during work and leisure time.

DISCUSSION

In conclusion, when mean parameter values are used, RRP leads to a significant increase in low back muscle condition, to a significant decrease in some psychological complaints and to a significant decrease in pain intensity and interference of pain. However, when individual scores are used, RRP leads to an increase in low back muscle condition in only 20 à 25% of the patients, to a decrease in pain intensity and interference of pain in 12% and 24% of the patients and to no differences in psychological complaints.

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CHAPTER 4

MOVEMENT ANALYSIS

MEASUREMENT OF UNCONSTRAINED WRIST AND ELBOW MOVEMENTS

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INTRODUCTION

Movements can be assessed by using commercially available motion analysis systems which provide objective and quantitative data of human movements. A detailed analysis of upper extremity movements was not possible until now, because the numerous degrees of freedom (D.O.F.) involved and the variability of the movements complicate the measurement and analysis.

A novel method has been introduced which will make it possible to measure the cardan angles of free, upper-extremity movements. In a first step, only wrist and elbow are considered.

METHODS

The VICON 370 motion analysis system was used for the measurement of free arm movements. This video-based system automatically tracks the 3-D trajectories of markers which are attached to the subject's skin.

For the calculations a simple model of the upper-extremities, consisting of rigid segments which are connected by ideal joints [1], [2] was established. Three markers per segment are sufficient to determine all 6 D.O.F. of each segment.

The markers were placed at the segments according to Fig. 1. In order to avoid errors due to skin movements no marker was placed directly at the joints. The skin movements were further reduced when the markers of one segment were connected by bandages of foam rubber.

From the measured 3-D trajectories, cardan angles were derived [3], [4]. For the calculation of the cardan angles a definition of joint co-ordinate systems is necessary. During an extra static trial additional markers were fixed lateral to the flexion / extension axes of elbow and wrist (Fig. 1).

Finally, for the calculation of cardan angles the order of the three rotations around the axes of the joint co-ordinate systems must be defined. In the case of the elbow, the chosen order is first flexion / extension, secondly a virtual 'ab- / adduction' and thirdly pronation / supination. In the case of the wrist, the order is first a virtual 'rotation', secondly ab- / adduction and thirdly flexion / extension.

In order to prove the reliability of the measurements the software was tested with simple, single joint, single axis movements. All movements were performed freely.

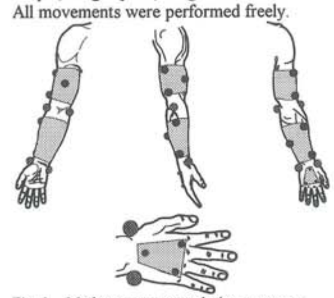

Fig. 1 Marker arrangement during movement (black markers) and additional markers (grey) for axes definition

RESULTS

Some examples of recorded movements are shown in Fig. 2-5. The measured elbow movements were flexion / extension with pronated forearm (Fig. 2) and pro- / supination with flexed elbow (Fig. 3).

The examples of wrist motions are the flexion / extension of the radial abducted hand (Fig. 4) and the abduction of the hand with flexed wrist (Fig. 5).

With the chosen order of rotations the elbow flexion / extension and pro- / supination can be measured separately. There is only little 'crosstalk' (Fig. 2, 3).

At the wrist a separated measurement of flexion / extension and abduction is not possible. The isolated flexion / extension of an abducted wrist can be represented with approximately one changing angle (Fig. 4). However, in general a motion of the wrist yields three changing angles (Fig. 5). Even with a changed order of rotations this can not be reduced to two angles.

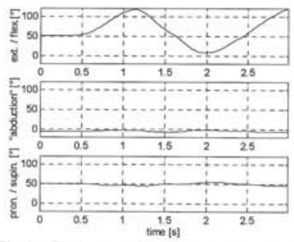

Fig. 2 Flexion and extension of a pronated elbow. Flexion and pronation are positive.

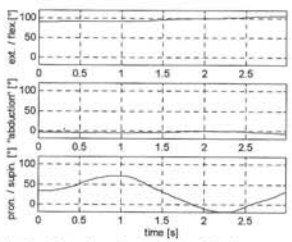

Fig. 3 Pronation / supination of the forearm with bent elbow.

DISCUSSION

With a simple model of the kinematics of the arm all D.O.F. of free wrist and elbow movements can be measured properly.

Deviations from the ideal results in the above angles do not necessarily indicate errors in the measurements or calculations. It is difficult to execute single-axes movements freely. Nevertheless, major sources of errors are skin movements and inaccurate placement of the markers which define the joint axes. The accuracy of the measured angles has to be investigated. Since the 'real' values are unknown the accuracy can only be estimated from simulations and measurements at mechanical joints.

The results show that the elbow can be described very well with two D.O.F.. This is not possible for the wrist joint. Three rotational D.O.F. are necessary to describe any joint orientation. This reflects the complex anatomy of the wrist.

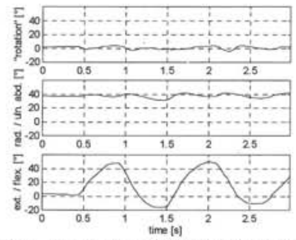

Fig. 4 Flexion / extension of the radial abducted wrist. Flexion and radial abduction are positive.

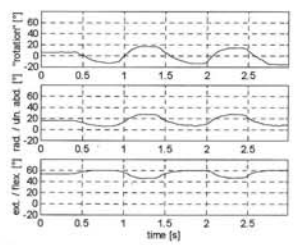

Fig. 5 Abduction of a flexed wrist

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MODEL OF THE MOVEMENT OF THE UPPER LIMB IN NORMAL SUBJECTS

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INTRODUCTION Aim of our study is to develop an adaptive model of selected mevements of the upper limb, in normal subjects. It could be the first step to verify the possibility to plan a functional electrical stimulation (FES) system, by means of a statistical-connessionist approach. In fact the idea is that the parameters of the our model could be used on an electrical stimulator, taking into account relationships between proximal and distal muscles, in paretic subjects selected on the basis of the motility of proximal muscles.

METHODS Ten normal subjects were invited to perform 3 movements of the upper limb: a medial, a frontal and a lateral grasp of an object positioned on a horizontal plane and return to rest position. Surface EMG signals have been recorded from the following proximal and distal muscles: pectoralis major, anterior, middle and posterior deltoid, upper, middle and lower trapezius, bicaeps and tricaeps, communis digitorum, ulnaris and radialis extensors and corresponding flexors. Furthermore kinematic trajectories have been recorded by means of the ELITE system. On these data, first, EMG signal parameters have been researched to discriminate among the selected movements and, second, the correlation between parameters have been studied. Furthermore the analysys of the relationships between proximal and distal muscles, for each movement, has been performed. The beginning and the end of the 2 phases of each movement have been individuate by comparing the EMG signals and the kinematic trajectories. The study of EMG signals variance has been done to define the beginning of the contraction for each muscle (Latwesen and Patterson, 1994). The EMG signals have been analysed for 130 msec (Zardoshti, 1993), and i) the integrals of rectified signal and ii) the coefficients of a 4th order autoregressive (AR) model (Knox et al., 1993) have been calculated. A Statistical analysis has been performed on these data for: i) to distinguish between the 2 phases of each grasp (Wilcoxon test); ii) to compare the 3 movements (Kruskal-Wallis test). The correlation between the values of integrals and the 4th order coefficients of the AR model have been studied by means of Spearman test.

RESULTS The coefficients of the 4th order AR model and the integrals of the rectified signal have been resulted the better parameters to discriminate among the movements. In fact, for these parameters, the Kruskal-Wallis test showed significant differences between: i) the medial and the frontal grasp; ii) the lateral and the medial grasp, and borderline between the frontal and the lateral grasp. The Wilcoxon test, on the other hand, has been showed significant differences between the two phases of each movement. Instead, the values of integrals and the 4th order coefficient of the AR model resulted not related for each grasp.

DISCUSSION Our results, completed with distal muscles study and with relatonships between proximal and distal muscles, will allow to ipotize a schema of grasp movements of the upper limb, in normal subjects. Next step will be the study of the same parameters of the EMG signals, in paretic subjects, to evaluate if these parameters will maintain their significance. On this basis, an electrical functional stimulator. driven by an especially developed neural network, will be designed.

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MANDIBULAR MOVEMENT RESPONSES TO ELECTRICAL STIMULATION OF VARIOUS PARTS OF THE HUMAN TEMPORALIS AND MASSETER MUSCLES

A.J. Zwijnenburg; G.W. Kroon; M. Naeije

INTRODUCTION

The human temporalis (TE) and masseter (M A) muscles are jaw closing muscles. Their complex architectural built allows for differential activation and variation in the line of action. Indications for this differential activation were obtained from experiments in which the EMG activity of different parts of the muscles was determined during various functional tasks. However, during voluntary activation not a single muscle part is solely responsible for the movement or bite force that is recorded, since other jaw muscles are active at the same time. Therefore, the aim of this study was to examine functional differences between various parts of TE and MA by means of recording the mandibular movement response to electrical stimulation of parts of these muscles.

METHODS

Subjects After giving written informed consent, five healthy male subjects participated in this study. The subjects ranged in age from 20 to 25 with a mean age of 22.

Recording technique The jaw movement responses were recorded with the OKAS-3D jaw rmovement analysis system (noise 0.08 mm, Naeije et al., 1995) at a sample frequency of 300 Hz, during an interval from 50 ms before until 250 ms after the stimulation pulse.

Stimulation technique Three stimulation locations in the right superficial masseter muscle were chosen, the anterior, middle and posterior section (MSA, MSM and MSP), one location in the deep masseter muscle (DM), and three locations in the right temporalis muscle, the anterior, middle and posterior part (TA, TM and TP). The required insertion depth of each electrode, in the center of the muscle part, was individually determined using MRI scans, in order to anticipate inter-individual differences in muscle and skin thickness. Stimulation was performed using 50 µm monopolar teflon coated hooked wire electrodes with a bare end of 2 mm and a current driven source (max. 10 mA).

Experimental procedure A stimulation electrode was inserted into MSA and the movement threshold level (minimal strength that caused a visible jaw movement) was determined. Series of 16 rectangular 0.5 ms pulses, with an inter-pulse interval of 1 s, were then ap-plied at this level. Stimulations were performed in four jaw positions (rest position, 1 cm to the left, 1 cm to the right and 50% maximum mouth opening). When this procedure was completed, the stimulation electrode was removed and the procedure was repeated for the other stimulation locations.

Data analysis The movement responses of the lower incisal point (IP) and both temporo-mandibular condyles were analyzed. The kinematic center was taken as condylar reference point (Yatabe et al., 1995). For each stimulation pulse, the jaw position determined during the 50 ms prior to the stimulus was subtracted from the post stimulus movement response. The responses of the 16 pulses of each series were then averaged. The IP movement response was expressed as a vector with an amplitude R_{max} and angles ϕ and θ (see fig 1). For the condy-lar response, only its excursion normalized with respect to R_{max} was analyzed. Statistical analysis was performed using analysis of variance (SPSS statistical software).

RESULTS

IP movement responses R_{max} ranged between 0.10 and 2.16 mm. The variation in R_{max} was partly explained by the factors 'stimulation location' ($p<0.001$) and 'jaw position' ($p=0.02$). Since experimental factors influencing R_{max} probably have a lesser influence on the direction of the response, the data analysis was further focused on this latter aspect. The variation in the direction was partly explained by the factors 'stimulation location' ($p<0.001$, illustrated by fig. 1) and 'jaw position' ($p \leq 0.006$). The directions of the response to stimulation of MSA, MSM and MSP were not different, but the direction significantly differed between the deep and superficial part ($p \leq 0.01$). When the stimulation location in TE shifted in an antero-posterior direction, the response changed from a vertical-lateral IP movement to a lateral-posterior movement with a smaller vertical component ($p<0.001$).

Condylar movement responses Due to TE stimulation, excursion of the right condyle was larger than excursion of the left condyle ($p<0.001$) and only the factor 'stimulation location' could partly explain the variation in excursion ($p<0.001$). In case of MA stimulation the factor 'stimulation location' could also partly explain the variation in condylar excursion ($p<0.001$), but there was no difference in excursion between both condyles.

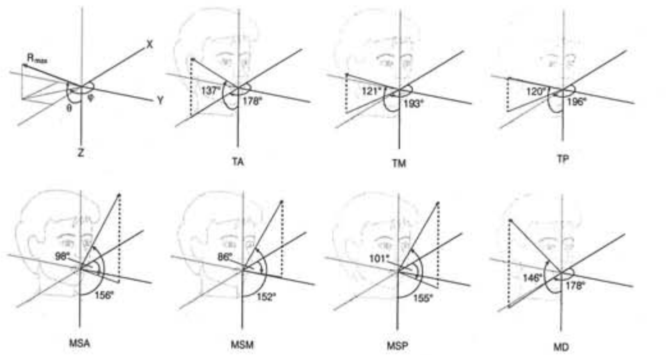

Fig. 1: Definition of polar coordinates (top left drawing) and the mean IP movement direction due to stimulation of the temporalis (TA, TM, TP) and masseter (MSA, MSM, MSP, MD) muscle, with the lower jaw in the rest position.

CONCLUSION

This study showed that the direction of IP movement due to electrical stimulation was different between TA, TM and TP and between the superficial and deep parts of MA. This is consistent with the findings of EMG studies. Condylar excursion differed as a result of stimulation of TE and MA. TE stimulation resulted in a larger excursion of the right condyle, whereas no side difference in condylar excursion due to MA stimulation was found. The variation in condylar excursion was partly explained by the factor 'stimulation location'. The results of this study provide an indication for the functional heterogeneity of TE and MA. They suggest a functional subdivision of the temporalis muscle in at least three parts and a subdivision of the masseter muscle in a superficial and a deep part.

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ERROR ANALYSIS OF INTERVERTEBRAL KINEMATICS PARAMETERS AUTOMATICALLY EXTRACTED FROM A VIDEOFLUOROSCOPIC SEQUENCE.

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INTRODUCTION

An "in vivo" analysis of intervertebral kinematics is attempted using digitized videofluoroscopic images of the spine [1]. This method can ensure useful diagnostic data, maintaining radiation exposure low enough to be acceptable for clinical application. Particularly, lumbar spine mechanics have been investigated. The work aims to provide quantification of spinal motion in vivo using an accurate and objective analytical system. This would be a very powerful tool for both the understanding of mechanical dysfunction of the spine and the assessment of its mechanical integrity. Previous work using manually identified vertebral landmarks were subject to small errors which could lead to large errors in the identification of the kinematic parameters [2]. Manual intervention, in particular, is regarded as the major contributor to the errors [3] and is very laborious. To overcome these limitation an automatic process of vertebrae position recognition, based upon cross-correlation, has been implemented. Intervertebral Angle (IVA) and Instantaneous Centre of Rotation (ICR) have been computed as kinematic indices. Statistical analysis have been performed to assess the repeatability of the algorithm and its sensitivity to the input data. The results show errors of order of 10^{-1} degrees for the IVA and 10^{-3} meters for the ICR location.

METHODS

Image sequences of the lumbar spine are grabbed from a videofluoroscopic device during patient's spontaneous bending. The operator have to select manually four landmarks per vertebra (corresponding to the four corner of the vertebral body) on the first digitised image of the sequence. The algorithm will find automatically the landmarks throughout the entire sequence. The algorithm is based upon cross-correlation which has proved to be effective and robust respect to the low signal to noise ratio of the images. Successive cross-correlations between templates and subsequent images estimate the displacement and the rotation of each vertebra. To further enhance the precision of the identified co-ordinates a correction is performed to ensure the assumption of vertebral rigidity. Finally, the intervertebral angles and the instantaneous centres of rotation are computed for each frame.

Fluoroscopic image sequences of the bending of a calibration model, constituting L3 and L4 lumbar vertebrae, has been used for validity assessment. The two vertebrae are linked at the disk level by an universal joint and their motion has been constrained to known values. The automatic detection procedure has been tested using these sequences. A repetitive computation of kinematic parameters was performed to assess the precision of the method, in terms of repeatability and sensitivity to the landmarks manually identified in the first frame. For each vertebra, not only the four landmark pixels selected by the operator, but also all the possible combination of the 8 neighbours pixels has been considered as inputs. This produced 9^4 (6561) input sets for analysis. The kinematic parameters automatically extracted from these inputs were analysed statistically.

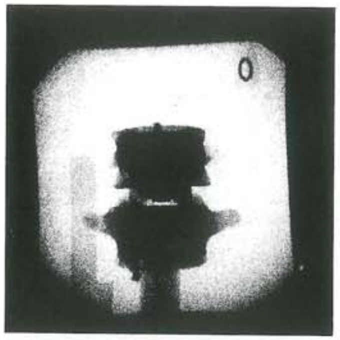
RESULTS

The kinematic parameters computed for a spinal segment are the Intervertebral Angle (IVA) and the Instantaneous Centre of Rotation (ICR). The means and the standard deviations of parameters computed using two subsequent frames of the calibration model coronal image sequence are presented. They refer to the 6 possible landmarks couple on which compute rigid kinematics.

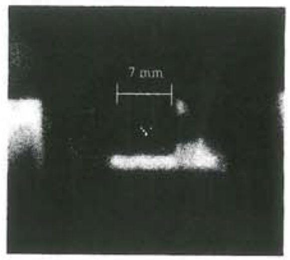
couple of landmarks	Intervertebral Angle [degree]		X coordinate of ICR [pixel]		Y coordinate of ICR [pixel]	
	Mean	*STD*	*Mean*	*STD*	*Mean*	*STD*
1	-5.6866	0.0532	242.3470	0.5894	302.6155	1.9914
2	-5.6867	0.0599	242.3549	0.9527	302.6065	2.3475
3	-5.6866	0.0573	242.4723	2.1968	302.6021	2.0214
4	-5.6865	0.0788	242.3266	1.4245	302.6247	1.8005
5	-5.6869	0.1045	242.3687	1.3542	302.6483	2.6981
6	-5.6866	0.0461	242.3507	0.8712	302.6117	1.7090

1 pixel corresponds to 0.25 mm

The parameters means correspond to the expected value on which the calibration model was constrained. Is due to remind that SNR of image sequences of the calibration model is higher then in patient images. Coronal and sagittal image sequences of the lumbar spine section of healthy and pathological subjects are now under study.



An image of the coronal videofluoroscopic sequence of the calibration model



Enlargement of the universal joint with the computed ICR throughout the sequence

DISCUSSION

The described method automates the tedious and imprecise manual landmark selection providing the computation and visualization of the kinematic parameters. Errors analysis suggest this method improves the accuracy of intervertebral kinematic calculation. In addition, if average values are used in place of landmark values the overall precision increase.

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MOVEMENT ANALYSIS USING ACCELEROMETERS AND GYROSCOPES

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INTRODUCTION

Human movement analysis has many applications in rehabilitation. It is especially useful for assesment of movement disorders. Body mounted inertial sensors (accelerometers and gyroscopes) have the potential to become an important alternative for laboratory bound optokinetic systems.

Micromachined rate gyroscopes [1] and accelerometers [2] have made it possible to make miniature inertial sensors suitable for applications in human movement analysis. It is our goal to develop a signal processing algorithm for a combined 3D inertial sensor in order to be able to track human movements accurately with a minimal number of sensors.

The signal obtained from accelerometers can be characterized by the following equation

$$\bar{s} = \bar{a} - \bar{g} \tag{1}$$

where $\bar{s}$ is the measured signal, $\bar{a}$ is the actual acceleration and $\bar{g}$ is the gravitational acceleration vector. Acceleration $\bar{a}$ can be estimated if the direction of $\bar{s}$ is known.

The direction of $\bar{s}$ relative to a reference frame (in which $\bar{g}$ is known) can be found from the gyroscopic signals. As a mathematical representation of attitude, the rotation vector can be used. This vector is defined in the reference frame and points in the direction of the axis around which the reference frame has to be rotated to obtain the sensor frame. Its magnitude and direction are related to the rotation angle by the right hand cork screw rule.

Bortz [3] derived the differential equation that expresses the rotation vector as a function of the measured angular velocity

$$\dot{\bar{\theta}} = \bar{\omega} + \frac{1}{2}\bar{\theta}\times\bar{\omega} + \frac{1}{\theta^2}\left(1 - \frac{\theta\sin\theta}{2(1-\cos\theta)}\right)\bar{\theta}\times\left(\bar{\theta}\times\bar{\omega}\right) \tag{2}$$

where $\bar{\theta}$ is the rotation vector and $\bar{\omega}$ is the vector composed of the measured angular velocities on the three orthogonal axes of the sensor frame. The second and third term together are called the non commutativity rate vector. They represent the effect of the non commutativity of rotations on reconstruction from gyroscopic signals.

The objective of this paper is to investigate the reconstruction of the sensor attitude from rate gyroscope signals on the basis of this theory.

METHODS

To obtain experimental data, three rate gyroscopes (Gyrostar, ENC-05S, Murata Manufacturing CO. LTD.) and three accelerometers (ICSensors, 3021-010-P) have been mounted orthogonally on a 4.5×4.5×4.5 cm cube. To account for construction errors and sensitivity axis misalignment, the calibration algorithm suggested by Ferraris [4] was used. A presampling filter of 30 Hz was used for all accelerometer and rate gyroscope signals.

Reconstruction of attitude from the rate gyroscope signals was done by integrating (2) at 100 Hz using the Runge Kutta integration methods. The algorithm was tested under coning motion. In this motion the path of the z axis of the sensor describes a cone, the sensor stays in place and does not rotate around its z axis. Coning motion is described as a heavy test for attitude reconstruction algorithms [5]. Simulation results showed good results for the attitude reconstruction algorithm.

As a preliminary test a hand imposed coning motion was chosen. As a reference, three markers for a Vicon™ system were attached to the sensor.

RESULTS

Figure 1 shows the position of the top marker as measured by the Vicon™ system, approximately 10 cm above the sensor. Figure 2 shows the position of this marker as calculated from the gyroscopic data. The data presented have been collectcd over 7.5 s.

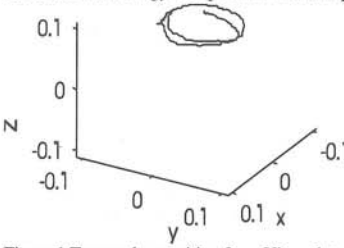

Figure 1 Top marker position from Vicon data.

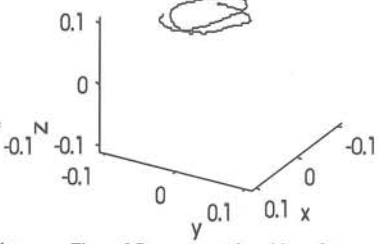

Figure 2 Reconstructed position of top marker from gyroscopic signals.

DISCUSSION

These preliminary results show some errors in tracking of attitude. Improvements are still to be made in calibration methods, reconstruction algorithm and signal processing (filtering).

Reconstruction of sensor (and therefore body) attitude from rate gyroscope signals requires a non straightforward approach as represented by equation 2. Depending on the integration algorithm the noncommutativity rate vector plays an important part in the reconstruction of attitude. Full precision was used in the current analysis, but for real time applications a faster, more efficient algorithm has to be derived. Especially then, the non commutativity rate vector plays an important role.

The results presented show the potential of 3D gyroscopic measurement in attitude measurement. Improved measurement procedures and the combined use of rate gyroscopes and accelerometers are expected to lead to a clinically relevant measurement method for human movement analysis.

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APPLICATIONS OF THE ANGULAR MOMENTUM METHOD TO THE STUDY OF HUMAN MOTION

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INTRDUCTION

A general algorithm for the analysis and simulation of a multi-segmental human body model has been developed (Shimba, 1995). The governing equation has been derived from the basic relations about the angular momentum of a rigid body instead of using the Lagrange's equation or Newton-Euler formulation. The general feature of the system of the equations allows applications to the direct, inverse and mixed dynamic problems of various aspects of motion with varying complexity of the structure of the model.

This paper presents some typical applications of this method to the motion study. The particular feature of each category of the motion or the problem will be also discussed.

METHOD

The basic starting equation of this method is represented as follows:

$$\frac{d}{dt} I_G\omega + r_G \times m\ddot{r}_G = N + \tau \,. \qquad (1)$$

Equation (1) is derived from the relation between the time rate of change of the angular momentum of a segment about the stationary origin in a reference frame and the sum of the total amount of the moments of the external forces about the origin and the external torque.

Equation (1) is applied to each subsystem of the model resulting in n equations for n-segment linkage system. A final form of the system will look like as follows:

$$\begin{bmatrix} A & \\ C & \\ E & 0 \\ R & 0 \end{bmatrix} \begin{bmatrix} \ddot{\theta} \\ \tau \\ \mathbf{F} \end{bmatrix} = \begin{bmatrix} B \\ D \\ G \\ S \end{bmatrix} \qquad (2)$$

where $\ddot{\theta}$ represents the angular acceleration of the segment, τ represents the joint torque and $\mathbf{F}$ represents the joint force, respectively. All other symbols in equation (2) represent the sub matrices or vectors. A particular feature of equation (2) is that all the variables in the vector $\left[\ddot{\theta}, \tau, \mathbf{F}\right]^T$ are treated in the same manner and the known variables are defined by the sub matrix [C] which is n x 4n matrix for n-segment model.

APPLICATIONS

Some particular features for various human motions are described below.

(1) Single Support Phase of Walking

In the foot flat case the angular acceleration of the foot is already known to be zero but the torque at the tip of the foot is not zero nevertheless there is no real joint. The torque can be interpreted as one that is required to maintain the segment in its position stationary.

In the case of the point contact of the foot the torque at the point is already known to be zero. In this case the motion of all the segments can not be defined arbitrary.

(2) Airborne Phase

The problem where no point in the system is in contact with the ground or its environment can be solved by introducing the time varying position vector of the tip of one foot.

(3) Double Support Phase

In the case where the system has a closed loop, equation (2) can be applied directly for the direct dynamic problem with the additional constraints relationships.

For the inverse dynamic simulation in the double support phase we need a relationship about the distribution of the terminal force components. There seem to be no definite rules for determining the relationship.

(4) The System with Branches

The multibody system with branches can be treated by dividing the system into main- and sub-trees. Usually a segment is represented by a link or a line segment, however, a massless link can be introduced to represent a massive segment such as the trunk where each limb diverges from different site of the segment.

RESULTS

Several computer programs have been developed on a personal computer (X68000, Sharp) using the C language. The main flow of the programs is as follows.

(1) Read a body parameter file or set the parameters manually. (2) Choose the mode of motion, and set the type of the problem by setting the sub matrix [C]. (3) Set the initial conditions and the kinematics of the segments as well as the time course of the joint torques. (4) Determine the elements of sub matrices [A], [B], [C], [D], [R], and [S]. (5) Solve the system of the equations for unknown variables using Gauss-Jordan elimination method. (6) Perform numerical integration for next time step using Euler's method or Runge-Kutta method. (7) Display and save the results.

DISCUSSION

Although the method has a general form, whether a specific problem can be solved or not depends upon the conditions such as the number of given values of the variables. In some cases, the value of an element of [A] in equation (2) may happen to become to be zero in a particular condition during motion. In such cases changing the type of the problem will be required. This is an example of the case where all the kinematics can not be defined arbitrary in a particular situation.

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EXTRACTION OF MOTOR COMMANDS BY MEANS OF PULSE DENSITY DEMODULATION PROCESSING OF SURFACE ELECTRODE ELECTROMYOGRAMS

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PULSE DENSITY DEMODULATION OF ELECTROMYOGRAMS

It is assumed that, in the process of controlling the contraction of each of the muscles by motor commands, an analog time pattern of the degree of contraction is generated in the motor area, and is transmitted through each motoneuron as a neural pulse train. The time change of the pulse density in the motoneuron superimposed over the whole muscle corresponds to the analog time pattern of the degree of contraction.

When muscle fibers connected to each motoneuron contract, the action potential propagates along the muscle fiber. The action potential is detected with surface electrodes, and the analog time pattern is reconstructed as follows: The number of pulses involved in the running time window are counted (pulse density demodulation) after correcting the difference in their amplitude caused by the passage from the motoneurons to the electrodes (pulse shaping). This principle of the pulse density demodulation of electromyograms (PDD-EMG) is schematized in Figure 1.

This study of PDD-EMG by computer programmed processing, with surface electrodes, is an extension of the previous study by one of the present authors and his colleagues, using electronic circuitry processing with hooked wire electrodes [Reference 1].

PROCESSING MULTI-CHANNEL SURFACE ELECTRODE ELECTROMYOGRAMS

As it is essential that the PDD-EMG collects action potentials from as many motor units as possible over a whole muscle, the electromyograms should be detected with multi-channel bipolar electrodes which are aligned sufficiently far enough from each other.

For PDD-EMG processing, it is also essential to extract as small an amplitude pulse train of action potential from diverse motor units as possible. The optimal threshold of pulse detection should be set, in order to separate small amplitude pulses from the noise component. Then, the pseudo pulses due to the bi-phasic and tri-phasic parts of the main pulse should be eliminated in order to minimize their effects on the analog time pattern of pulse density [Reference 2].

As the reconstructed analog time patterns of the motor commands, which were obtained in a preliminary experiment with leg muscles in running action, reflect a simple transmission characteristic in the neural passage, the on-off patterns in the higher neurological level motor commands can be estimated by compensating for the characteristic [Reference 3].

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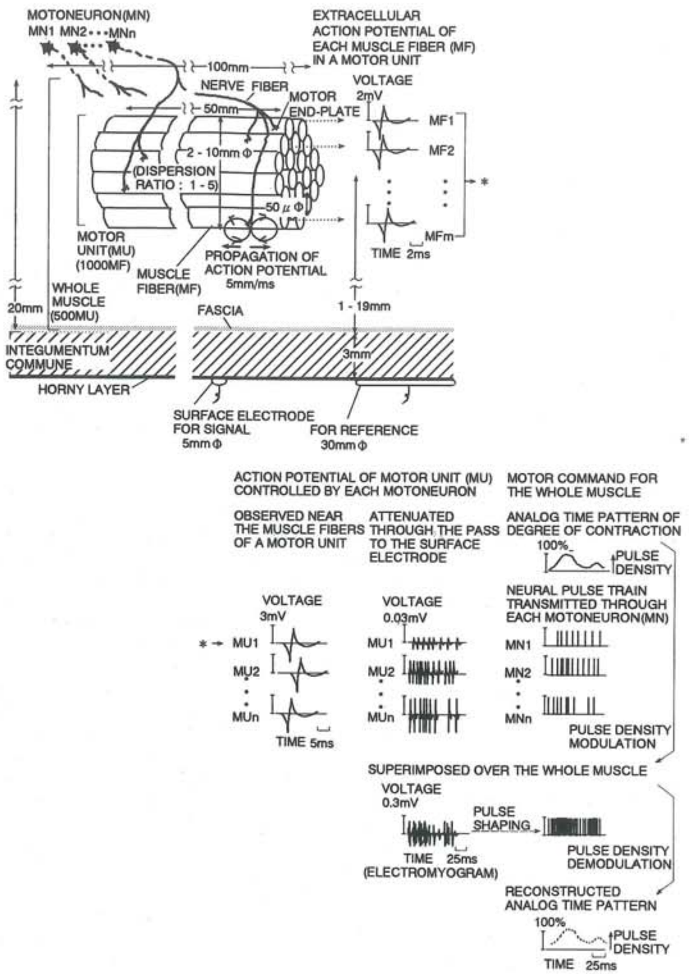

Figure 1. The principle underlying the processing of the pulse density demodulation of surface electrode electromyograms (PDD-EMG).

RECONSTRUCTION OF ANALOG PATTERNS AND ESTIMATION OF ON-OFF PATTERNS OF MOTOR COMMANDS FROM ELECTROMYOGRAMS

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PULSE DENSITY DEMODULATION PROCESSING FOR ELECTROMYOGRAMS

In order to reconstruct the analog pattern of the motor commands which control the degree of contraction of a muscle as a whole, computer programmed pulse density demodulation processing for surface electrode electromyograms has been developed.

The principle of the pulse density demodulation processing of electromyograms (PDD-EMG), the alignment of the multi-channel surface electrodes, the setting of the optimal threshold for pulse extraction, and the elimination of the pseudo pulses are discussed in References 1 and 2.

RECONSTRUCTION OF ANALOG PATTERNS OF THE MOTOR COMMANDS

Electromyograms used in this experiment were recorded from the soleus and tibialis anterior muscles during running action on a treadmill by a male subject. Four channels of bipolar electrodes, 5 mmØ in diameter, were aligned on the outer skin surface of each muscle in a direction at right angles to the muscle fibers. Electrodes in each pair and those in different channels were more than 2 cm apart from each other, so that they detected action potentials from different groups of motor units.

The sequence of the computer program for PDD-EMG is summarized in Table 1. The program was written for an Apple Macintosh Quadra 700 in Microsoft Quick Basic. From the electromyogram recorded by a pair of bipolar electrodes on the soleus muscle ((a) in Figure 1), the neural pulse train was reshaped (b) by extracting the pulses after eliminating the influence of pseudo pulses caused by secondary and tertiary phases of the large amplitude pulses. Then, the analog pattern of the motor command (c) was reconstructed by means of PDD, and the four channels of the patterns were summed (d). In this way, the identical analog patterns of the motor commands were reconstructed for the repetitions of the leg action (e).

ESTIMATION OF ON-OFF PATTERNS OF THE MOTOR COMMANDS

It was assumed that the analog pattern of the motor command was generated from the on-off pattern of the higher level motor command (f) through a time delay in the neural passage. The time delay was simulated by introducing an exponential function and a smoothing window.

From the electromyograms of the tibialis anterior muscle (g), the four channel summed analog pattern (h) was reconstructed, and, the on-off pattern of the higher level motor command was estimated (i). The differences in timing and amplitude of the motor commands between the agonist and antagonist muscles were clearly shown in the on-off patterns.

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1. Storage of data of the original waveform of the electromyograms.

2. Elimination of fluctuating component of the baseline of the original waveform.

3. Extraction of peaks of the primary pulses.

1) Set thresholds of the period and amplitude of peaks of the primary pulses.
2) When the increase changes to a decrease on the positive side of the waveform data, are the increasing or decreasing period and amplitude of the waveform data larger than the thresholds?
3) Register the amplitude and the sample number of the primary pulse.
4) The same processing for the negative side of the waveform data.

4. Extraction of peaks of the secondary pulses.

1) Derive distribution of the second derivative of the waveform data.
2) Set threshold of the change in slope of the waveform data from the distribution.
3) Is absolute change in the slope of the waveform data larger than the threshold?
4) Are the period and amplitude of the change in slope of the waveform data larger than the thresholds?
5) Register the amplitude and sample number of the secondary pulses.
6) Display the primary and secondary pulse train.

5. Derivation of analog patterns of the change in pulse density.

1) Set window width of pulse number counting.
2) Register sum of the pulses in the window width.
3) Display the analog patterns of the change in pulse density.

6. Estimation of the on-off patterns of the motor commands.

Table 1. Sequence of the computer program for the pulse density demodulation processing of electromyograms (PDD-EMG).

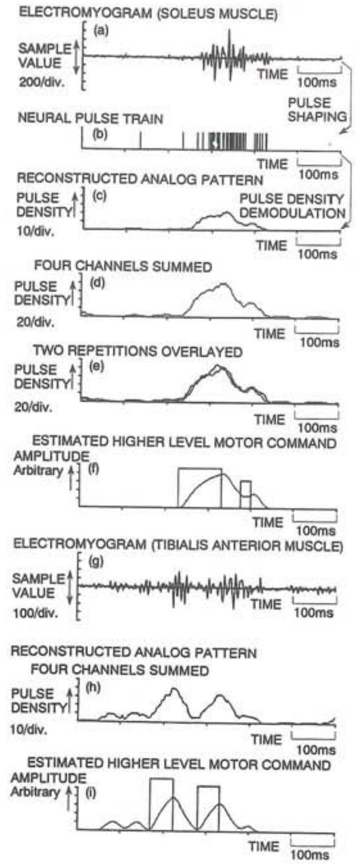

Figure 1. Examples of the reconstructed analog patterns and estimated on-off patterns of the motor commands for the soleus muscle (upper half) and tibialis anterior muscle (lower half) in running action. The motor commands also include the component for the reflex action.

DELAYED CONTRACTION OF TRANSVERSUS ABDOMINIS ASSOCIATED WITH LOWER LIMB MOVEMENTS

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INTRODUCTION: When a limb is moved reactive forces are imposed on the spine. These forces are equal and opposite to those producing the movement and are greater with movement of the leg compared to the arm. The central nervous system (CNS) deals with this challenge to stability of the spine by contraction of the abdominal muscles prior to movement of the limb [1,2]. Transversus abdominis (TrA) is invariably the first muscle active, preceding the onset of electromyographic activity (EMG) of the muscle responsible for movement of the arm by a average of 39 ms [2] and that of the leg by 109 ms [1]. Leg movement is associated with contraction of all muscles of the abdominal wall prior to the EMG onset of the prime mover. When people with chronic low back pain (LBP) perform movement of the arm the onset of TrA is delayed by between 50-450 ms, irrespective of the direction of movement of the limb [3]. This change in the way the CNS controls contraction of the muscles of the trunk may leave the spine unprotected from the reactive forces produced by the movement. The aim of this study was to evaluate if a delay existed in the onset of TrA and the other abdominal muscles in people with LBP with movement of the lower limb.

METHODS: The study involved 15 subjects with a history of chronic recurrent LBP and a group of age and sex matched control subjects. Fine-wire electrodes were inserted into TrA, internal oblique (OI) and external oblique (OE) under the guidance of ultrasound imaging. Surface electrodes were used for rectus abdominis (RA), erector spinae (ES) and the prime movers of hip flexion, abduction and extension; rectus femoris, tensor fascia latae and gluteus maximus, respectively. In standing the subjects shifted their weight onto the left leg to free the right leg to move. Subjects performed rapid right hip movement in response to a visual stimulus as fast as possible to approximately 20 degrees. Ten consecutive repetitions of movement in each direction were performed. The onset of EMG was evaluated using a computer algorithm. Onset of EMG of the trunk muscles was compared between groups using an analysis of variance (ANOVA).

RESULTS: *Non-low back pain response:* In the non-LBP subjects the onset of EMG of each of the abdominal muscles preceded that of the prime mover of the limb (Table I). TrA was the first trunk muscle active. The reaction time of RA was significantly longer with hip flexion than the other two movements and the reaction time of ES was shorter with flexion than the other two movements. There was no significant difference between the reaction time of TrA, IO and EO between movement directions. The latency between the onset of the TrA EMG and that of the prime mover was significantly shorter for hip abduction than the other two movement directions. *Low back pain response:* In the LBP subjects the onset of EMG of TrA followed that of the prime mover of the limb and each of the other trunk muscles with movement of the limb in each direction. In addition, the onset of EMG of RA occurred after the prime mover of the limb with hip flexion, all muscles were active after the prime mover with abduction and the onset of EMG of EO and ES followed the prime mover of the limb with extension of the hip. Similar to the non-LBP group the reaction time of RA was significantly shorter for hip extension than the other movement directions and the reaction time of ES was significantly shorter for hip flexion than extension. The reaction times of TrA, IO and EO were not significantly difference between movement directions. Unlike the non-LBP group the latency between the onset of EMG of TrA and the prime mover of the limb was not significantly different between movement directions. *Intergroup comparison:* When the EMG onsets were compared between the two groups the reaction time of the prime mover of the limb was not significantly different between the LBP and non-LBP groups. In contrast the EMG onset of TrA was delayed relative to the prime mover of the limb with movement in each direction. The EMG onsets of the other trunk muscles was delayed only with hip flexion, although ES was also significantly delayed with hip extension.

Table I Time between onset of EMG activity of the trunk muscles and the prime mover of the lower limb (mean $\pm$ SD) for the LBP and control subjects (negative value indicates contraction prior to the prime mover) (* $p < 0.01$).

	Flexion		Abduction		Extension	
	Non-LBP	LBP	Non-LBP	LBP	Non-LBP	LBP
TrA	-86 ± 40	36 ± 55*	-57 ± 38	80 ± 72*	-71 ± 35	48 ± 74*
IO	-48 ± 46	-10 ± 46*	-15 ± 43	21 ± 64	-22 ± 69	-14 ± 63
EO	-14 ± 44	-1 ± 46*	1 ± 70	52 ± 87	-20 ± 79	4 ± 62
RA	-27 ± 49	7 ± 41*	-12 ± 65	10 ± 55	-50 ± 32	-36 ± 51
ES	-69 ± 32	-23 ± 40*	13 ± 38	17 ± 31	11 ± 24	37 ± 31*

DISCUSSION: The results of the study confirm that EMG onsets of the abdominal muscles and ES precede the onset of EMG of the muscle responsible for movement of the leg. Furthermore, the onset of EMG of each of the muscles is delayed in people with low back pain, although this most consistently occurs in TrA. The failure of the direction of movement of the limb and the reactive forces acting on the spine, to influence the time of onset of EMG activity of TrA, IO and EO suggest that these muscles contribute to the control of a parameter consistent between movement directions. This may be the general control of spinal stiffness. When people have LBP this does not occur and the spine is exposed to unrestrained forces. It appears that the dysfunction is most consistently associated with TrA.

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EMG ENVELOPES DURING GAIT: FILTERING AND NUMBER OF STRIDES

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INTRODUCTION

The muscular synergy patterns generated in human locomotion are studied for a variety of purposes. The major problem for establishing representative patterns is that there are two general sources of pattern variability; one is inherent in the measurement process and the other is a reflection of the adaptability of the control process [1,3]. The use of linear envelopes (LE) for representing the electromyographic (EMG) measurements obtained during locomotion has become common practice [2,4]. The objectives of this work were to define the true waveform of a linear envelope and to determine the number of strides and envelope filter characteristics necessary to provide a meaningful estimate of the EMG profile.

METHODOLOGY

1. Measurement and Acquisition

Five normal healthy subjects whose ages ranged from 22 to 30 years participated in the study. They were asked to walk at their self-selected free, slow, and fast walking speeds over a metal covered walkway, 12 meters in length. A minimum of 30 strides at each walking speed for each subject was measured. Eight muscles were studied in the upper and lower leg. EMG signals were sensed with miniature Beckman surface electrodes, 2.5 mm diameter, in a bipolar arrangement with a two centimeter spacing. The signals were bandpass filtered from 10 to 200 Hz and amplified at a gain of 5000. The times of onset and offset of contact of the heel and forefoot were also measured. All signals were sampled at a rate of 500 Hz.

2. Analyses

A series of low pass Butterworth filters were designed as the envelope filters in order to create the linear envelopes of the EMG signals. They were implemented using an autoregressive-moving average filter with both parts being second order with cut-off frequencies of 2, 3, 4, 6, 8, 10, 20, and 30 Hz. Profiles of all of the EMG signals were calculated using each of these filters. The statistical characteristics of the resulting profiles were ascertained using the variance to signal ratio (V/S) and a bias measure (BIAS). The formulas are

$$V/S = \frac{\sum_{i=0}^{255} S(i)^2}{\sum_{i=0}^{255} M(i)^2}; \quad BIAS = \frac{\sum_{i=0}^{255} (M(i) - MREF(i))^2}{\sum_{i=0}^{255} MREF(i)^2}$$

where $S(i)^2$ is the variance of the ensemble average at stride point i, M(i) is the mean of the ensemble average, and MREF(i) is the ensemble average of the rectified EMG and is considered the "ideal" average. The "ideal" envelope was determined by performing an ensemble average on the rectified EMG signal.

RESULTS

1. Variability Characteristics

The magnitude of the variability of the EMG profiles is effected by all three of the factors considered; number of strides, walking speed, and cut-off frequency of the envelope filter. As the number of strides increases the variability increases up to 10 strides and then approaches the asymptotic value. In particular the variability of the lower leg muscles reaches from 90 to 100% of the final value at 5 strides and 100% at 10 strides. The other muscles only reach 50 to 90% of the final value at 5 strides but 90 to 100% at 10 strides. In general, as walking speed increases, the variability decreases. The cut-off frequency of the envelope filter has an appreciable effect on the variability. The variability increases linearly with cut-off frequency.

2. Bias Characteristic

The bias created in the EMG profiles is also effected by the same three factors. A 2 Hz filter creates a significant amount of bias. With respect to the major phases of activity there is an underestimation of maximum values, an overestimation of minimum values and an obliteration of undulations in activity. The 30 Hz filter causes none of these distortions. In general the bias in the EMG profiles is large when only a few strides are used and decreases dramatically within five strides. The bias decreases less rapidly with more strides and becomes approximately twice the asymptotic value at 10 strides. Walking speed seems to have little effect on bias when more than 6 to 8 strides are used in the average. The cut-off frequency of the envelope filter has a dramatic effect on bias. As the cut-off frequency increases, the asymptotic value of the bias decreases. At cut-off frequencies of 8 Hz the bias has decreased dramatically and is very close to its minimal value. Walking speed seems to have little effect on bias when a cut-off frequency of 6 Hz or greater is used.

CONCLUSIONS

The effect of signal processing on the ensemble average of linear envelopes has been studied in the normal population. The cut-off frequency of the envelope filter produces conflicting criteria that must be balanced, that is when bias is reduced, variability is increased and vice-versa. It is recommended for the best compromise and most versatility that one choose to maintain 95% of the signal power. Thus one can use an envelope filter with a cut-off frequency 8.9 Hz and thereby achieve a bias within 30% of the minimum value and a V/S ratio that is 12% of the maximum value. The number of strides is also important and in order to minimize the bias and stabilize the statistical stability, at least ten strides must be used in the ensemble average. If the less variable muscles are being studied, then possibly only five strides will be necessary.

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EVALUATION AND CLASSIFICATION OF SURFACE EMG SIGNALS IN GAIT ANALYSIS

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INTRODUCTION

A method to analyze the surface EMG signal recorded from the leg muscles of normal and cerebral palsied patients, during free level walking is developed.
Individuals with lesions of the central nervous system suffer from a lack of voluntary control and command of isolated movements. In fact, they may produce much more EMG activity during functional tasks, such as standing or walking, than when they are asked to, selectively, move a body segment in a specific direction. The lesions modify also the pattern of muscle activity and the interplay between muscles. These reasons have generated a growing interest in the analysis of the surface EMG signals during gait, to find out significant correlations with gait phases, i.e. the stance and swing phases of the gait cycle. Muscles' classifications as muscles of stance and/or swing phase is widely accepted. On the other hand the classification of the different EMG patterns, based on timing errors has been proposed for hemiplegic's gait by Waters et al. [1], considering footswitches as clock of the gait cycle. Usually the clustering of the EMG patterns by visual inspection and the criteria to decide where the limits of the bursts are, as well as to recognize filtered artifacts from true EMG, depend upon the clinician's experience. As a result, the obtained classification may be rejected because it can be considered too empirical. Shiavi et al. [2,3,4], Perry et al. [5,6], have proposed methods for computerized burst recognition. The main problem in these approaches is to set true criteria determining the beginning and the end of the bursts. For example, the EMG signal is considered relevant when its amplitude is greater or equal to X time the noise level or any pre-defined threshold but, according to the value of X, the number of bursts as well as their duration can be different. Moreover, to recognize the timing errors and possibly to classify them for each muscle, would allow the clinician to quickly assess the trends of muscle behavior during gait, to count the number of strides in each category and finally to compare the variation of trends before and after a treatment.
In this paper we propose a method to address the previous exposed difficulties, extracting EMG patterns recorded during gait, in normal and cerebral palsied patients, correlating them with the gait phases and the corresponding muscles, classifying automatically the EMG activity within nine pre-defined classes by a neural network.

MATERIALS AND METHODS

The analysis has been based on 125 EMG signals recorded on 4 females and 8 males (mean age: 16.8±11.3 years), supplied with information about stance/swing timing and relative to 13 different lower extremity muscles, for a total of 4330 valid gait cycles. Three of these subjects are normal, while the others are cerebral palsied. The data were recorded at the Kinesiology Laboratory of the University Cantonal Hospital of Geneva. All the EMG signal have been recorded with bipolar surface electrodes with a pickup diameter of 2.5 mm, placed 15 mm apart along the longitudinal axis of the muscle fibers over the area of minimal crosstalk between adjacent muscles. The recorded signal is transmitted to the fixed equipment by radio telemetry. The sampling frequency used to convert the analog signal into the correspondent digital signal is near 2000 Hz per channel.

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The extraction of the characteristics of the electromyographic bursts is based on a statistical temporal analysis of the raw EMG signals considering global and local signal characteristics. First, a moving average of the rectified raw EMG, and a moving average of the rectified derivative of the original EMG signal, are calculated. Then, the parameters computed by the algorithm are the following: 1) the mean value of the raw absolute EMG signal and its standard deviation 2) the mean value of the absolute derivative of the raw EMG signal and its standard deviation. The statistical analysis of the distribution of these parameters, based on a polynomial interpolation, has allowed to establish an approximation of the EMG bursts position. Afterwards, using a local analysis based on the evaluation of the distribution of the raw signal amplitude peaks, the algorithm refines the localization of onset and cessation of the activity patterns. The results of the feature extraction algorithm, after parametrization, become the input to a constructive neural network based on the Cascade Correlation algorithm [8], whose task is the classification of the activity patterns in respect of the gait phases.

RESULTS

The evaluation of our algorithm is made comparing the obtained results with the visual inspection on the same EMG signals peformed by a trained expert clinician, amd it reachs a 97.25% of success considering the whole experimental population. The main skills of the algorithm for feature extraction are: low time consuming in detecting the activity bursts (about 0.7 sec/stride on a 80486 DX4-100 MHz with 8 MB RAM), that can allow to perform on-line daily electromyographic analysis; low space consuming for EMG tracks and resulting storage; wide range typology of muscle activity recognition (from continuous to fragmented and short activities). The clinician has individuated nine classes depending on the onset and cessation time shifting and the number of the bursts, obtaining a slightly different version of Perry's definition of quantified electromyograms and EMG timing errors [7]. The network developed has reached a good discrimination levels between classes with fifteen hidden units.

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GFE: A TOOL FOR A COMPUTER-AIDED IDENTIFICATION OF GAIT DISORDERS

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INTRODUCTION: Gait functional evaluation is a demanding task, asking for the preliminary identification of gait deviations with respect to the "normal" behaviour and then for the identification of correlations among the variables showing the deviations from normality.
Aim of GFE (Gait Functional Evaluation) software tool is to engender a verbal description of gait analysis data and the comparison of available data with reference data. GFE provides both a visual-interactive screen presentation of the various gait variables and an output text, written in natural language, describing every signal and its deviations with respect to normality. The tool can be used: 1) by expert users, facilitating the analysis of the large amount of data coming from a gait analysis test; 2) by unexperienced users, as a didactic tool; 3) in a teleconsultation scenario, providing a written and standardized description of gait analysis tests performed in a remote laboratory. In this latter context, a pre-requisite is that the labs involved in teleconsultation share agreed clinical functional evaluation protocols and a common vocabulary for the description of the relevant variables.

METHODS: The design of GFE sw-tool is based on results of two CEC projects: CAMARC-II (AIM A2002) and BRITER (AIM A2050). In particular the Cerebral Palsy protocol developed within CAMARC-II has been taken as reference.
The program has been written in LabVIEW and runs on a PC-486 with 16 Mb RAM.
The gait variables taken into account are kinematic data and EMG time patterns. These latters have been determined according to "Rancho" algorithm described in [1]. As far as kinematics is concerned, reference has been made to [2]. Terminology used in the natural language description of gait deviations has been taken from [3].
GFE allows comparisons among a given trajectory and: 1) normal data, 2) data relative to the controlateral side, 3) same kind of trajectory recorded at a previous time (serial analysis).
GFE provides three levels of comparisons for kinematic data:
- I level: it evidences when, and how much, the given trajectory is out of its normal range;
- II level: it evidences differences in the shape of the trajectories being compared;
- III level: it evidences differences (in terms of mean value and peak-to-peak variation) in the shape of trajectories during each gait phase.

Second and third level comparisons ask for a conventional syntactic description of the waveforms. For EMG data, GFE provides two levels of comparison: the first one evidences, on the whole gait cycle, if the on-off patterns are longer or shorter than normal ones; the second level describes the differences existing within each gait phase.

RESULTS: The program allows to show in an interactive form, on the PC screen, the various gait variables or to save the description of the gait deviations as written text on a file. These descriptions are available for each variable and for each gait phase.
Figure 1. shows GFE's control panel where the text window describes the deviations among the trajectories visualized in window n.1.
Figure 2. shows an example of construction of a sentence written in natural language as appears in a text window.

DISCUSSION: GFE has been thought as a help for the analysis of gait data valuable both for experienced and unexperienced users. It can be further improved introducing a more detailed description of EMG data and extending the analysis to dynamics too.
The qualitative/quantitative description of gait deviations can be thought also as a primary

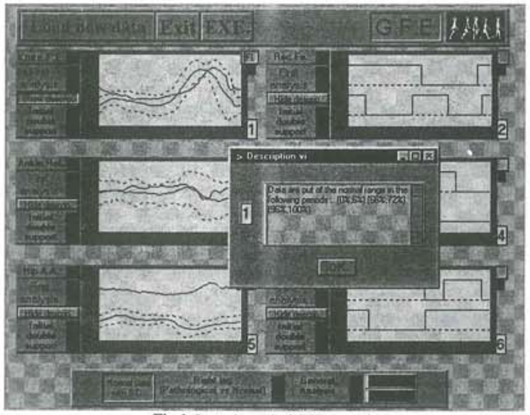

Fig.1 Control panel of GFE sw tool.

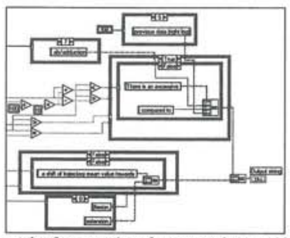

Fig.2 Example of construction of a sentence in natural language.

step towards the construction of clinical decision making systems once one is able to find in an automatic way correlations among the identified gait deviations. There is the need of an agreed thesaurus describing gait deviations; this is a part of the large and well known problem of homogeneity of medical languages.

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STIFFNESS CONTROL FOR LOWER LEG MUSCLES IN DIRECTING EXTERNAL FORCES

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INTRODUCTION

The question how to control the large number of degrees of freedom in the musculo-skeletal system has received considerable attention since Bernstein formulated this problem. In movements where external forces are applied on the environment, the so-called contact control tasks (CCTs) [2], the direction and magnitude of the force vector have to be controlled. These two dimensions of the external force alone do not directly determine the combination of net joint torques about hip, knee and ankle since the position of the point of force application can vary over the range of contact surface. The first question of this study therefore is how this degree of freedom is used to accomplish the task. A recent study of standardized CCTs [2] showed that mainly the biarticular muscles of the thigh are responsible for the control of the force direction. Considering all relevant joints and muscle groups of the complete lower extremity a second question arises: which stereotyped activation pattern with respect to the lower leg muscles is used to solve the redundancy of the musculo-skeletal system? Or, in other words, how does the CNS deal with this redundant system in order to accomplish the goal of the task? In order to answer this question, experiments were designed on a special dynamometer, which allowed the execution of standardized CCTs with strong emphasis on force control at similar angular displacements.

METHODS

For the experimental protocol, subjects (n=9) were instructed to exert a constant force of 300 N with their foot in a prescribed direction and subsequently change the force direction from 45 to 135° ***or*** from 135 to 45°. The force vector was shown as visual feedback for the subjects.All tasks were executed while the forceplate moved downwards with a velocity of 2 $cm.s^{-1}$. In the first 4 seconds of each trial, the force had to be directed in the initial direction and within 6 seconds, the force direction changed at about 15° per second while the forceplate moved downwards. During each trial, joint position and reaction force were recorded and muscle activation was registered by means of surface EMG. A linked segment model of the human body was applied to calculate the net torques about the hip, knee and ankle joints. Electromyographic activity was recorded from seven leg muscles. The data of the corresponding tasks for all subjects were averaged and used for further analysis.

RESULTS

The CCTs were performed very accurately for force direction as well as force magnitude. Figure 1 shows a typical example of the execution of the task, where the force had to be directed from 45 to 135°. In order to accomplish the change in direction of the external force vector, different combinations of net torques about the joints involved are required. The net torque changes were considerable for the knee and hip joint, whereas the ankle torque remained remarkably constant. The displacement of the point of application during the CCTs is illustrated in Figure 1. Figure 2 shows the mean EMG-activity of seven muscles normalized to the maximum value in each task, as a function of the exerted force direction during the tasks. The mean EMG-activity of the upper leg muscles (GU, HA, RF, VA in Figure 2), modulated with the changing force direction, while the EMG levels of the lower leg muscles SO and TA remain rather constant and are strongly co-contracting. In order to distinguish more quantitatively between the different types of activation, reciprocal activation or co-contraction, the EMG-data of two antagonistic pairs for each joint are compared with a measure for the type of activation(e.g.[1,3]); i.e. *A*ctivation *I*ndex:

$$AI = (A_{ag} - A_{ant}) / (A_{ag} + A_{ant}) \qquad (1)$$

with $(A_{ag} - A_{ant})$ and $(A_{ag} + A_{ant})$: the difference and the sum of the activity levels (EMG-data) of the involved antagonistic muscles. An *AI* of 1 indicates pure reciprocal activation, whereas an *AI* of 0 indicates pure co-

Figure 1. Stick diagram of a typical example performing the task 45->135. Reaction force vectors are shown with intervals of 30 ms.

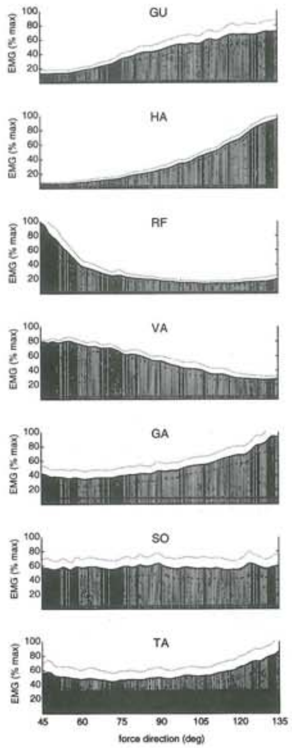

Figure 2. Mean EMG values (% of maximum value) of GlUteus Max., HAmstrings, Rectus Femoris, Vasti, GAstrocnemius, SOleus, Tibialis Anterior. Dotted lines indicate standard deviation; solid lines rest EMG levels.

contraction of the antagonists. For each subject, the mean value of *AI* over the whole range of force directions for each antagonistic muscle pair is calculated about each joint separately. The values of the antagonists working about the hip and knee are much closer to 1 (average: .69±.04 and .77±.05 respectively) when compared to the low values found for the ankle muscles (.19±11). For all subjects, the mean *AI* for the ankle muscles appeared to be significantly lower compared to the values of both hip and knee (Student's *t*-test, p<.01).

DISCUSSION

For the CCTs analysed in this study, the constraint on the third degree of freedom is accomplished by fixing the ankle torque, which might be qualified as a type of stiffness control. The required force direction is obviously regulated by changing the torques of the two proximal joints. The constant ankle torque during the CCTs is realized by a strong co-activation of the antagonists of the lower leg. Since other experiments of similar CCTs performed at higher velocities (40 cm.s^{-1}), also showed a strong co-activation of ankle muscles (mean *AI* .070± 12; n=9) and a high reciprocal activity for the upper leg muscles (mean *AI* .85 ± .08 and .71 ± .09 for knee and hip respectively), a difference in the organization of upper and lower leg muscles is present. By the mechanism of co-contracting antagonists, which act as two stiff springs, an equilibrium torque combination is maintained [4]. For the CCTs in this study, this might be necessary in order to facilitate a stable control of the imposed external force direction. In conclusion, the change in position of the point of application might be aimed at creating a virtual fixed point of force application at the height of the ankle joint, which allows the proximal muscles to control the direction of the force.

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INTRINSIC AND REFLEX CONTRIBUTIONS TO HUMAN ANKLE STIFFNESS: VARIATION WITH ANKLE POSITION

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ABSTRACT

A parallel-cascade system identification method was used to identify intrinsic and reflex contributions to dynamic ankle stiffness at different ankle positions. Intrinsic stiffness dynamics were well modeled by a linear second-order system between ankle position and torque. Reflex stiffness dynamics were accurately described by a linear, third-order system between half-rectified velocity and reflex torque. The gain of both the intrinsic and reflex stiffness increased as the ankle position was moved from plantarflexion to dorsiflexion. However, reflex gain increased more rapidly than intrinsic gain so that the relative contribution of reflex mechanics to ankle torque increased as the ankle was dorsiflexed. Nevertheless, the intrinsic mechanics contributed the majority of the torque at all positions.

INTRODUCTION

System identification methods have been successful in describing the overall dynamic stiffness of joints [1]. Recently, we presented a method for distinguishing the relative contributions of intrinsic and reflex mechanisms to overall mechanics. Intrinsic mechanics were estimated as a linear, position-dependent pathway while reflex mechanics were identified as a parallel, non-linear, velocity-dependent pathway [2]. The relative magnitudes of the intrinsic and reflex components varied with the amplitude and bandwidth of the perturbation; reflex stiffness decreased as the rms velocity of the perturbation increased. Intrinsic and reflex gain were also modulated by the level of voluntary contraction; intrinsic gain increased whereas reflex gain decreased with voluntary activation [3].

The objective of this study was to explore how the intrinsic and reflex components of ankle joint mechanics were modulated by ankle position.

EXPERIMENTAL PROTOCOL

Five normal (three male, two females) subjects were studied. Subjects lay supine with their left foot attached to the pedal of a stiff, position controlled, electro-hydraulic actuator by a custom fitted fiber glass boot. Joint position and torque were measured by transducers in the actuator. Electromyograms from the tibialis anterior and soleus muscles were recorded using bipolar surface electrodes.

At the start of each experiment, range of motion, dorsiflexing (DF) and plantarflexing (PF) maximum voluntary contractions (MVCs) were measured.

The perturbation amplitude which produced the maximum reflex torque was then determined. Pseudorandom binary sequences of ankle position were applied about different angles in the range of motion when subjects maintained 10% MVC in the ankle extensors.

ANALYSIS METHODS

Non-parametric Model.

Intrinsic and reflex contributions to the stiffness dynamics were separated using a new parallel-cascade identification method [2] as follows:

(1) Intrinsic stiffness dynamics were estimated in terms of linear, dynamic impulse response functions (IRF) relating position and torque. This IRF was convolved with position signal to predict the intrinsic torque, which was subtracted from the observed torque to leave the reflex torque.

(2) Reflex stiffness dynamics were estimated by determining the IRF between half-wave rectified velocity and the reflex-torque.

(3) These models were then used to predict the intrinsic and reflex contributions to ankle stiffness evaluated using the percentage variance of their relative torques accounted for by the total torque generated at the ankle (VAF_I and VAF_R respectively.)

Parametric Model.

Non-linear least squares methods were used to fit parametric models to the IRFs as follows:

(1) The intrinsic compliance dynamics were computed from the intrinsic stiffness IRF and well defined by a linear, second-order high pass system, having inertial (I), viscous (B) and elastic (K) parameters.

(2) The linear, dynamics of the reflex stiffness were well described by a third order system, describing by gain (G), damping (ξ), natural frequency (ω_n) and the other frequency parameter (p) [4].

RESULTS

Intrinsic Stiffness

Figure 1 demonstrates the modulation of intrinsic dynamic stiffness gain with ankle position in normal subjects. Intrinsic gain (K), represented by elastic parameter, increased with ankle dorsiflexion, whereas I remained constant and B did not change significantly.

The results were similar for all other subjects. These findings are consistent with [5].

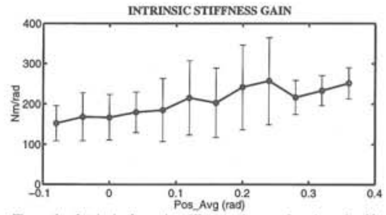

Figure 1: *Intrinsic dynamic stiffness gain as a function of ankle angle for normal subjects.*

Reflex Stiffness

Figure 2 shows the modulation of reflex dynamic stiffness gain with ankle position. Reflex gain increased with ankle dorsiflexion, while there was no consistent change in the other parameters.

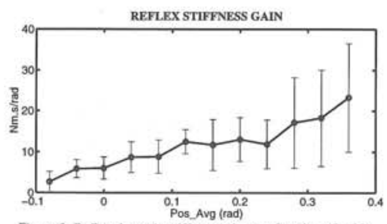

Figure 2: Reflex dynamic stiffness gain as a function of ankle angle for normal subjects.

Intrinsic and Reflex Contributions to Stiffness

Figure 3 shows the VAF_I and VAF_R as a function of ankle angle. VAF_I (solid-line) decreased with ankle dorsiflexion, whereas VAF_R (dash-line) increased. This indicated that reflex contributions to ankle stiffness became more significant as the ankle extensors were stretched. However, since VAF_I was significantly greater than VAF_R for all positions, the intrinsic relative contribution to ankle stiffness was always dominant in normal subjects. The results were consistent for all subjects.

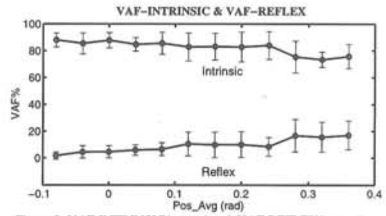

Figure 3: VAF-INTRINSIC(VAF_I) and VAF-REFLEX(VAF_R) as a function of ankle angle.

DISCUSSION AND CONCLUSION

The ankle became stiffer as it was dorsiflexed due to an increase in both intrinsic and reflex gain. Enhancement of intrinsic gain was probably due to the increase in passive torque caused by ankle dorsiflexion. Reflex gain increased probably because muscle spindles became more sensitive as the GS muscles were stretched. The findings also indicated that ankle stiffness was mostly due to intrinsic stiffness even in extreme of dorsiflexion where the reflex contribution was maximum.

ACKNOWLEDGMENT

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EVALUATION OF MOVEMENT PATTERN OF SPASTIC STROKE PATIENT USING EMG AND 3-D KINEMATICS MEASUREMENTS

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INTRODUCTION

About three-quarters of stroke patients could result in hemiparesis. The neurological deficits would affect the patient in many aspects, such as mobility, communication, activities of daily living, sensory function etc. An appropriate rehabilitation program may benefit the hemiplegic patients in functional improvement. However, scientific investigation supporting the effectiveness of rehabilitation programs and interventions is relatively absent. It is probably due to the lack of the convincing quantitative data and objective assessment method to show the functional improvement of stroke patients. In clinical tests, the status of motor recovery of the stroke patient is usually evaluated by the observation of experienced physicians or therapists. The aim of this study is to integrate the muscle activity patterns and the 3-D kinematics data of the lower limb for quantitative assessment of motor function of stroke patients.

METHOD

The hemiplegic patients were recruited from the Department of Rehabilitation and Department of Neurology of National Cheng-Kung University Hospital. The selection criteria include: (1) one side hemiparesis due to cerebrovascular attack, (2) consciousness clear and no aphasia, (3) can follow therapist command, and (4) affected lower limb has some voluntary movement. Normal subjects were also recruited as a norm data and for comparison. The motor function of patient was first evaluated by Brunnstrom assessment form [1]. For quantitative assessment, ten times of flexion and extension movements of the whole lower limb were conducted repeatedly in supine and side-lying position, respectively. These movements were frequently used in clinical assessment for the synergy pattern due to spasticity [1].

During the movement tests, the EMG and 3-D kinematic data were recorded simultaneously, synchronized by a trigger signal. The EMG signals were detected by the active surface electrodes (Motion Control, Utah, USA) from rectus femoris, biceps femoris, tibialis anterior, peroneus longus, and medial gastrocnemius. The raw signals were digitized through DaqBook (IOtech, Inc., USA), with sampling rate at 1 kHz, and then were processed into a linear envelop (LE) form which was further normalized as 100% by using the kinematics data as time reference. The ten normalized EMG LE were averaged to represent phasic activity patterns of movements. The 3-D kinematics data of lower limbs were obtained by using 3Space tracking system (Polhemus Inc., USA). The system provides a low-frequency electromagnetic field as source of reference to determine the position and orientation of the receivers. For the measurement of lower limb kinematics data, four receivers were placed on pelvis, thigh, shank, and foot. Before taking the movement data, the positions of ten anatomical landmarks of the lower limb were processed to establish a coordination system. During the movement, the signals were sent to a computer via RS-232 interface with a high-speed baud rate of 115.2 K, achieving a sampling rate of 30 Hz for each receiver. The kinematic data with a full six-degree-of-freedom can be calculated by reversing the transportation matrix between the sensor and digitizer [2].

Since the complex movements are rather encoded in "joint space" than in "task space," at least in the absence of external or cognitive constraints [3]. The calculated joint angle trajectories were then analyzed into joint space. We selected the typical movement components shown in flexion and extension synergy patterns to represent the characteristics of the movement pattern, which include hip flexion/extension, abduction/adduction, internal/external rotation, knee flexion/extension, ankle dorsiflexion/plantarflexion, and ankle inversion/eversion.

RESULTS AND DISCUSSIONS

Since the existence of consistent relationships between muscle activation patterns and kinematic or kinetic variables of movement may indicate which of those variables are important in planning the appropriate motor output [4]. Fig. 1 shows a representative EMG data of five muscles of a spastic patient performed flexion movement in supine lying. The Brunnstrom stage evaluation for the patient is stage IV. This indicates that the patient's limb was still affected by the abnormal program of the lesion brain. Fig. 2 shows the joint space of the same subject. We can observe the linear relationship binding all joint angles during the motion, as indicated by other research [3]. The ultimate goal is to develop a quantitative method for the assessment of motor function for stroke patient and to design therapeutic measure for each individual.

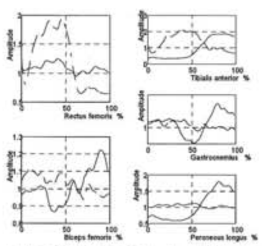

Fig.1 The EMG LE pattern of flexion of a spastic stroke patient. (—— sound side, ········ affected side)

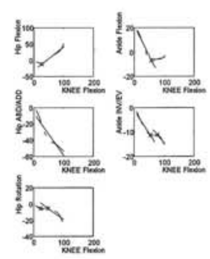

Fig.2 The joint space of flexion movement of a spastic patient.

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Moment-Angle loops at the ankle joint in hemiplegic walking

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INTRODUCTION

Preliminary investigations on normal and pathological subjects (Davis and DeLuca,1995; Frigo et al.,1996) have shown that moment-angle relationship can reveal interesting properties of the mechanical behaviour of a joint during walking. At the ankle joint, in particular, it was possible to identify different phases on the basis of different slopes, that correspond to absorption or production of energy, and represent different modalities of control of the state of the muscles. In normal subjects the moment-angle loop at the ankle joint follows a counterclockwise pathway.The first, rising phase, can be subdivided into two sub-phases: the first relatively constant at different walking velocities, and the second increasing in slope as far as velocity increases. In hemiplegic patients, our first observation was that a much more complex phenomena occurred, and the moment-angle loop was considerably modified.

The present work is a preliminar attempt to describe the gross morphology of the Moment-Angle loops at the ankle joint in these patients.

METHOD

Twelve hemiplegic patients (seven males and five females), all affected at the right side were analysed in our gait analysis laboratory at a post-stroke time ranging from 1 to 7 years. All of them walked without any assistive device or othosis and were analysed barefoot. Mean age was 51 years and the range was from 33 to 67 years.

Twelve normal subjects, without any history of musculo-skeletal disease, were analysed by the same experimental procedure and served as control group.

The experimental set-up included a particular arrangement of TV-cameras to detect retroreflective markers on the two sides and on the back of the subject (ELITE system, Ferrigno and Pedotti,1985, Pedotti and Frigo,1992), a force platform to detect the ground reaction forces (Kistler), and a system for telemetric recording of surface EMG signals from several muscles of the lower limb. A model of pelvis and lower limbs was implemented as to reconstruct from the external location of markers the internal points such as the centers of the hip, knee, and ankle joints (Romanò et al.,1996). Relative joint angles were obtained as the angles between adjacent segments projected on the sagittal plane. The inverse dynamics approach was used to compute dynamically the moments of the external forces at the different joints. The moment-angle loops were obtained by reporting in a graph the joint angles on the abscissa and the external joint moment on the ordinate. Quantification of the energetic aspects of the mechanical behaviour of the ankle joint was achieved by computing the work absorbed in the rising phase, the work produced in the descending phase, the net work produced (positive) or absorbed (negative) all along the stride cycle (area of the loop), the ratio between work produced and work absorbed.

The acquisition protocol consisted in walking at a comfortable speed on a 9 meters long level pathway where the force platform was embedded at approximately one half of the pathway length. A minimum of three trials was collected for each side. Anthropometric data were measured individually and served in the model for the estimation of inertia parameters.Left and right sides were both analysed, but data concerning the affected limb only will be presented here.

RESULTS

The first, extremely consistent phenomenon, that was observed in all the patients analysed, was

that the moment-angle loop was clock wise, at variance with our control population. This implies that the net work in the loop was always negative, and the ratio between work produced and absorbed was always lower than unit.
In absolute values, not considering the sign, the net work was lower in the hemiplegics than in the normal subjects, and the ratio between work produced and absorbed was closer to unit.
A second observation was that the rising phase was not monotonically ascendent, but was in most cases interrupted by two, three or even four steps. The electromyographic analysis revealed that people exhibiting the higher number of steps in the ascending phase frequently presented enhanced or clonic activity of Triceps Surae muscle.

DISCUSSION
The preliminar application presented above, shows that combining the joint moments and joint angles in a single graph makes very appearent a number of features connected to motor coordination that hardly could be evidenced by looking separately at the different biomechanical variables. The reverse of the course of the loop at the ankle joint seems particularly discriminant between normal and hemiplegic patients, and clearly represent the altered mechanism of muscle involvement, that shifts from being primarily energy supplier to primarily energy absorber. The reduction of the area within the loop, in addition, shows that hemiplegic subjects have a tendency to use plantarlexor muscles as passive springs, and this can be connected to modifications of the rheological properties of these muscles, or , in a more general sense, to a reduced capability of changing the state of contraction of the muscles. The presence of the steps in the rising phase might reflect the enhanced responsiveness of muscles to stretch and often correlated with the electromyographic data. All these observations require validation on larger populations and need to be interpreted at the light of the clinical and functional data. However it seems very likely that some of the synthetic parameters obtained from the moment-angle loops could become very useful to characterise the pathophysiologic profile of the patients, to improve the basic knowledge on the altered control mechanisms, and finally to contribute to clinical decision making.

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ELECTROMYOGRAPHIC ANALYSIS OF MUSCULAR ACTIVITY RELATED TO RIGID AND FLOATING PEDAL INTERFACES.

M. Dennis Bates and Lyndon J.K. Wood

INTRODUCTION:

Cycling is at first sight a simple circular and rhythmic leg movement, comprising a flexion extension motion in the saggital plane. However because of the simultaneous rotation of the hip, knee and ankle joints the general motion becomes quite complex (Hay 1985). The complexity of the motion can be compounded by the addition of differing pedal interfaces. During the power phase of cycling when the foot is in downward motion it may attempt to pronate. If the foot cannot produce this movement due to being locked in position, then compensatory motions are developed in other components. Pronation may be limited by a rigid shoe/pedal interface or stiff shoes, in which case the joints involved must absorb the motions and resulting loads (Wooten and Hull,1992; Francis, 1986).

Ruby and Hull, (1983) reported an increase in intersegmental loads at the knee by restricting the amount of movement available at the foot ankle complex. Wheeler *et al*. (1995) stated 'our general conclusion is that clipless pedal float designs quantifiably reduce applied moment at the shoe pedal interface without compromising power transmitted to the bike.' According to van Ingen Schenau (1982) and Gregor *et al.* (1985), the reaction forces of the pedal on the foot creates an extension moment at the knee joint which is countered by the knee flexor activity.

Many of the muscles involved in cycling are biarticular and therefore exhibit different flexor extensor activity relative to the joints involved. The work of Gregor *et al.*(1985) sheds light on which group is generating the majority of the torques involved at the joints, the hip extensors being active and responsible for the majority of the torque at the hip. When looking at the knee joint, the knee extensors are active from top dead center until about 135° of rotation. Therefore the hamstring group is active throughout the propulsive phase of the movement, with the vastus medialis and tibialis anterior bridging the gap between propulsive and recovery phases (Clarys and Cabri, 1992).

While this sheds light on the activity of the muscle groups during cycling it does not offer enlightenment on muscle activity that occurs when the pathomechanics of the cycling motion in the frontal and transverse planes are affected by the introduction of differing pedal interfaces of non-restrictive and restrictive design.

METHODS:

Nineteen competitive triathletes [17 male and 2 female (mean age 27)], participated in this investigation. The subjects cycled between 75 and 200 mile per week (mean 107 mpw). Subjects were tested via an airbraked cycle ergometer (King Cycle trainer tester) using the subject's personal cycle. Pedaling occurred at a constant rate of 110 rpm and at a constant load of 150 W. Preamplified bipolar silver/silver chloride surface electrodes were placed upon prepared skin in accordance with the proposal of Clarys and Cabri (1992). Where those placements were impractical (i.e. soleus and peronus longus) electrodes were placed in line with the recommendations of Basmajian (1983). The preamplified electrodes reported a CMRR of 106 dB at the sampling frequency of 500 Hz. The raw signal was then high pass filtered at 100 Hz and analysed. The linear envelope was normalised in time and amplitude based upon the 100% peak ensemble average (Yang and Winter, 1984) attained over each of the four revolutions for each muscle and an average integral produced for each muscle and each pedal group over the four revolutions.

RESULTS:

The results of the integral values are reported in table 1.

Table 1. Integral values of investigated muscles and subject groups (% of total area in mV • s)

Muscle	Floater group	Toe clip group	Rigid group
Vastus medialis	6.26	3.96	6.26
Vastus lateralis	6.94	2.98	4.85
Rectus Femoris	29.36	29.00	28.90
Semitendinosus	9.74	9.50	9.83
Biceps femoris L.H.	8.03	7.62	8.59
Tibilis anterior	12.66	10.53	10.32
Soleus	3.93	3.13	9.35
Peroneus longus	6.76	5.70	7.15

Results of one way ANOVA and Kruskall-Wallis tests (where data do not meet criteria for a parametric test) indicate that in all cases for all muscles there is no significant variation between the groups ($\alpha = 0.05$).

DISCUSSION:

The results of this investigation indicate that the type of pedal interface does not significantly affect the quantity of muscle activity in a cycling motion. Studies on EMG in cycling therefore should not need to take into account the pedal type or the changes in the pathomechanics induced by the differing cycle pedal interfaces. Pedal interface effects on the qualitative pattern remains an unanswered question.

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A COMPARISON OF JAW-CLOSING AND JAW-OPENING MUSCLE ACTIVITY TO OVERCOME AN EXTERNAL FORCE COUNTERACTING JAW MOVEMENT IN MAN

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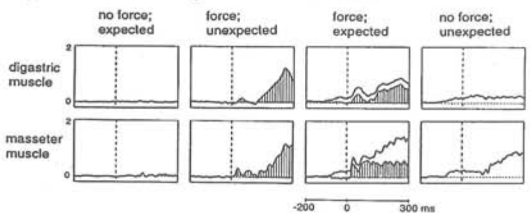
INTRODUCTION — Histological studies have revealed that the jaw-closing muscles contain large numbers of muscle spindles. Animal studies have showed that jaw-muscle spindles are very active during biting and jaw-movement (Appenteng et al., 1980). Muscle spindles sense changes in expected muscle length or shortening speed and probably play a role, among other sensors, in the generation of corrective action of the jaw muscles when jaw movement is perturbed. In contrast with other limbs where flexor and extensor muscles contain muscle spindles, the jaw-opening muscles, such as the digastric muscle, contain no or very little muscle spindles (Kubota and Masegi, 1977). Due to the lack of muscle spindles sensory feedback to this jaw-opening muscle and therefore its reaction to a perturbation may not be similar to that of jaw-closing muscles. This study compares the activity of the digastric muscle when jaw opening is counteracted by an external force during rhythmic open-close movements of the jaw, with the activity of the masseter muscle when jaw closing is counteracted by a similar external force.

METHODS — Subjects made rhythmic open-close movements with the jaw, controlled by a metronome (1 Hz). In the jaw-opening phase, or in the jaw-closing phase in the other experiments, an external force counteracting jaw movement could appear, supplied by a magnet-coil system. A coil, moving freely in a permanent magnetic field, was attached to the subject's mandible by means of a clutch cemented to the teeth. By varying an electric current through the coil the force on the mandible could be adjusted. The jaw gape was measured with an optical motion analysis system and used for on-line computation of the force counteracting jaw-closing. The force started at a preset jaw gape, then it increased linearly with the increase/decrease in jaw gape until it reached a maximum at a second preset jaw gape, just before jaw-opening/closing. The maximum force was 10 N when jaw opening was counteracted, and 25 N when jaw closing was counteracted. During the experiments sequences of 12-18 open-close movements with the force counteracting jaw movement were unexpectedly alternated with sequences of 12-18 unperturbed movements. The electrical activity of the jaw muscles was measured with bipolar surface electrodes and fed to an AD-converter together with the jaw gape and the force signal. With each appearance or disappearance of the force the signals were recorded in two cycles before and two cycles after the force transition. At least 30 appear and 30 disappear transitions were recorded for each subject. Off-line the recorded signals were averaged using in each cycle the point at which the force started or could have started as a reference. The muscle activity in the last unperturbed cycle, just before the force unexpectedly appeared, was regarded as the activity needed to only move the jaw as such. It was subtracted from the activity in the other cycles to obtain the extra muscle activity generated to overcome the force.

RESULTS — **Perturbation of the open movement** When no force was expected and no force appeared (Figure, upper row, column 1) there was, by definition, no extra muscle activity. When the force appeared unexpectedly (column 2), a small tempory increase of extra muscle activity was followed by a second, large increase in muscle activity with a latency varying between subjects from 90 to 200 ms. When the force was expected (column 3) with 6 out of 8 subjects extra muscle activity was present even before the onset the force. After the force had appeared a tempory increase in extra muscle activity was observed, followed by a second increase. When the force was expected, but did not appear (column 4) on average extra muscle activity was present throughout the open movement. This was pre-programmed extra muscle activity only, as there was no peripherally induced extra muscle activity due to the absence of the force. With two subjects, however, no pre-programmed extra muscle activity was observed at all. By subtracting the muscle activity in this cycle from that in the preceding cycle, with the force, the extra muscle activity with a sensory origin in that cycle is found (column 3, hatched area). It shows that the first, temporary increase in reflex muscle activity was larger then when the force appeared unexpectedly. It started approximately 25 ms after the onset of the force. The average latency of the local maximum was 53.8 ms ± 2.6 ms s.e.m. Then reflex activity decreased to zero with all subjects before the second increase started, which became significant after on average 150.8 ms ± 10.1 s.e.m.



Extra muscle activity to overcome a force during jaw opening (upper row) and during jaw closing (lower row). At the vertical dashed lines the force was or could be triggered. (Grand average of 8 subjects, 30 recordings per subject)

Perturbation of the closing movement When the force appeared unexpectedly (lower row, column 2) a small amount of reflex activity was observed, reaching a local maximum at approximately 35 ms after the onset of the force. Then extra muscle activity remained about constant before it increased at about 130 ms after force onset. When the force was expected (column 3) with 5 out of 8 subjects extra muscle activity was present even before the onset of the force. After the onset of the force extra muscle activity increased steeply reaching a local maximum at about 35 ms followed directly by a gradual increase in extra muscle activity. Again the hatched area in column 3 shows the peripherally induced part of the extra muscle activity. It shows that reflex activity of the masseter muscle increased much faster when the force was expected than when it was not expected. The average latency of the local maximum was 36.8 ms ± 2.2 s.e.m. The second increase became significant after on average 72.5 ms ± 4.4 s.e.m. When the force unexpectedly did not appear (column 4) with 5 out of 8 subjects pre-programmed extra muscle activity was observed throughout the closing movement. The other three subjects showed no pre-programmed extra muscle activity at all.

DISCUSSION — With respect to pre-programmed extra muscle activity to overcome the force, the digastric muscle and the masseter muscle did not show differences. With both muscles extra activity with a sensory origin consisted of two phases: first a temporary increase was observed, starting approximatly 25 ms after the onset of the force. The maximum of this first increase was reached signicantly faster by the masseter muscle (36.8 ms vs. 53.8 ms). Then a second, larger increase was observed. Particularly when the force was expected, this second increase occurred much faster with the masseter muscle than with the digastric muscle (72.5 ms vs. 150.8 ms). These differences may be the result of more feedback from muscle spindles to the masseter muscle, than to the digastric muscle. Extra muscle activity with a sensory origin generated at short latency, < 100 ms, was larger when the force was expected than when the force appeared unexpectedly. This was particularly notable with the masseter muscle and may be the result of increased gamma motoneuron activity.

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QUANTIFYING SYNERGISTIC ACTIVATION PATTERNS DURING PLANTARFLEXION: AN ORTHOGONAL EXPANSION APPROACH

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INTRODUCTION Effective quantification of synergistic muscle activation is important for understanding neuromuscular control of movement. Agonist and antagonist muscles are not all active in the same relative proportion when movement parameters are altered [1,2,3,4] thus there is a need to measure electromyographic (EMG) activity from more than one muscle and muscle site to accurately represent synergistic activation patterns. The purpose of this study was to evaluate the effectiveness of a pattern recognition approach based on the Karhunen Loeve expansion [5] to quantify synergistic activation patterns from agonist (4 sites on the Triceps Surae **TS**) and antagonist (2 sites on the Tibialis Anterior **TA**) muscles during concentric plantarflexion. Orthogonal expansion theory has been used to study phasic EMG patterns during gait [6], however, not to study patterns of EMG amplitudes.

METHODS Ten healthy subjects (mean age = 30 ±7.5 years) participated in the study. Meditrace pellet surface electrode (10mm) were placed in a bipolar configuration over the Lateral (LGA) and Medial Gastrocnemius (MGA) muscles, over the lateral (LSO) and medial (MSO) aspect of the Soleus muscle and over the upper (UTA) and lower (LTA) fibers of the TA muscle. Torque and angular displacement data from a Cybex II isokinetic dynamometer (Lumex Inc, NY) and the six EMG signals (bandpass 10 - 500 Hz, CMRR = 90dB, input impedance 100 Mohms) were digitized at 1000 samples per second using a Tecmar Labmaster (Scientific Solutions Inc., Ohio) interfaced with an IBM PC.

The subjects' right knee was flexed to 160° (180° being full extension) and their foot was secured in the Cybex dorsiflexor/plantar flexor footplate. Subjects performed two maximal voluntary isometric contractions (MVIC) of the plantarflexor and the dorsiflexor muscles for nomalization purposes. The test trials consisted of three plantarflexion movements at each of three angular velocity settings on the Cybex: 30, 90 and 150 °/s. The order of the test trials were randomly assigned and subjects were given a two minute rest between trials.

The root-mean-squared (RMS) EMG values were calculated for each muscle site; then the test trials were normalized to the RMS values from the MVIC trials for each muscle site: RMS_{Ni}. The RMS_{Ni} formed a matrix $[X]$ dimensioned $(m \times n)$; m was the number of electrode sites and n the number of trials. A cross product matrix, $[C]$ was calculated and an eigenvector analysis was performed such that $[C_c] = [T][\Lambda][T]^t$ where $[T]$ is a 6×6 matrix of orthonormal eigenvectors $\mathbf{EV_i}$ [5]. The transform of the column vectors $\mathbf{x_i}$ defined as, $\mathbf{y_i} = [T]^t\mathbf{x_i}$ produced a vector of coefficients, $\mathbf{y_i}$, for each contraction trial. The percent trace of $[C]$ provided an estimate of the error of reconstruction based on less than n $\mathbf{EV_i}$s. The measured patterns were reconstructed using linear combinations of $\mathbf{EV_i}$ and $\mathbf{y_i}$. The relative reconstruction errors were calculated.

RESULTS The mean (SD) RMS_{Ni} for each of the 6 muscle sites are in Figure 1a. The mean RMS_{Ni} from all 4 sites on the TS muscles were greater than 100% of the MVIC, and were not consistent among sites for the maximal plantarflexion contractions at the three angular velocity settings. The mean RMS_{Ni} for the upper and lower sites of TA were 30 and 20% of MVIC respectively. The SDs indicate a large between trial variability for all six muscle sites.

The percent trace of $[C]$ was 94, 97 and 99% for 1,2 and 3 eigenvectors respectively, with the corresponding measured relative reconstruction errors of 5.0 (±3.1), 3.1 (±2.3) and 1.6% (±1.1). The mean ys for the sample were used to reconstruct the EMG patterns for each $\mathbf{EV_i}$ from 1 to 3 (Figure 1a). Note that only one $\mathbf{EV_1}$ was needed to accurately represent the mean pattern for the sample, with subtle changes in the pattern when $\mathbf{EV_2}$ and $\mathbf{EV_3}$ were included. On the other hand, the measured EMG patterns for one trial and the reconstructed patterns based on $\mathbf{EV_1}$ $\mathbf{EV_2}$ and $\mathbf{EV_3}$ are illustrated in Figure 1b. $\mathbf{EV_1}$ did not accurately represent the measured activation pattern for this trial; three $\mathbf{EV_i}$s were needed to reflect the measured EMG pattern.

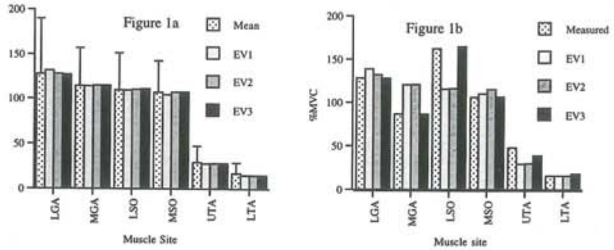

Figure 1a The mean EMG amplitudes and the reconstructed values for the sample and **1b** the measured EMG amplitudes and reconstructed values for a single trial.

DISCUSSION The large SDs illustrate that even for a controlled movement subjects use their synergists very differently. Since these were maximal efforts, a consistent percentage of MVIC was expected from all four agonist sites, but the results do not support sampling one TS muscle site, corroborating results for submaximal plantarflexion contractions [1,3]. The variability in the coactivation levels of TA during the plantarflexion task (Figure 1a) is also consistent with high variability reported for TA during stepping and walking [1]. The differences between sites on the same muscle may be explained by inhomogeneities in activation of motor units within the muscle [7]. Activity from subpopulations of motor units cannot be detected without sampling from multiple sites, and the measured EMG from this study supports the need to examine pattern recognition approaches to quantify activation patterns.

The percent trace estimated that 99% of the signal energy was contained in $\mathbf{EV_1}$ to $\mathbf{EV_3}$, from the eigenvector decomposition of $[C]$, thus reconstructions based on three $\mathbf{EV_i}$s were evaluated. Using the sample mean for $\mathbf{y_1}$, it would appear that $\mathbf{EV_1}$ would suffice in describing the synergistic activation pattern (Figure 1a). Figure 1b, however, illustrates that $\mathbf{EV_1}$ did not accurately represent the EMG pattern for this trial: three $\mathbf{EV_i}$s were needed. Although $\mathbf{y_1}$ was the largest coefficeint for the 89 trials, more than one $\mathbf{EV_i}$ was needed to effectively represent the combination of EMG patterns found among the 89 trials. In conclusion, healthy subjects used different relative proportions of their MVIC to perform a simple, controlled plantarflexion task at three angular velocities, and the orthogonal expansion analysis effectively quantified the principal EMG patterns with minimal error.

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EVALUATION OF EARLY SWING PHASE ACTIVITY OF THE RECTUS FEMORIS MUSCLE

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Introduction

Normal human gait cycle is divided into two phases, namely, stance and swing. The objective of stance is to provide support and stability and that of swing is to provide ground clearance and limb advancement. During swing knee flexion is essential to lift the foot off the ground for limb advancement. The complex mechanisms involved in producing limb advancement can produce excessive knee flexion at faster walking speeds. Under these circumstances the shank needs to be decelerated to reduce the amount of knee flexion. It is assumed that Rectus Femoris (RF) is active for a very short period at the beginning of the swing phase[1,2]. The amount of this activation is proportional to the walking speed and thus to the generated knee moment and the angular accelerations of the lower limb segments. However, there is very little evidence to support these assumptions. The objective of this study was to study this interrelationship. Human joint forces and moments can be estimated using the inverse dynamics approach, but it was thought necessary to direct measurements of linear and angular accelerations, angular velocity and displacements. Detailed description of the method of data collection and analysis is presented elswhere[3].

Method

Ten able bodied males, their ages ranging between 23 and 27 years, without any gait impairments took part in the study. Their height and weight were recorded to be able to calculate length and mass, position of the center of mass, radii of gyration for the thigh and shank segments. After several trials on a treadmill at various speeds, an average speed of 2.5 km/h was defined as the normal walking speed for the test group. Two slower (1.2 and 1.9 km/h) and two faster speeds (3.4 and 4.4 km./h) were defined for the group.

Four pairs of uni-axial accelerometers were mounted on the left upper and lower leg segments on two aluminum braces, fixed on the anterior aspect of the limb. The accelerometers measured the tangential and radial accelerations of each segment. Additionally, a gyroscope was attached in the middle of each aluminum strip to measure the angular velocity of the thigh and shank. Data obtained were recorded at a sampling frequency of 100 Hz, using analogue input channels provided by the VICON™ system. Calibration recordings were made before and after each subject to estimate gain and offset of the sensors.

Two pairs of adhesive silver-silver chloride surface electrodes were placed over the motor points of RF and VL to measure the EMG. The signals were high pass filtered with a cut-off frequency of 5 Hz, full wave rectified and the low pass filtered at 25 Hz to obtain a smoothed rectified EMG. These data were also recorded using analogue channels of the VICON™ system using a sampling frequency of 100 Hz. Foot switches under the heel and the head of first metatarsal were used to identify gait events.

The test subjects performed 2 trials of 12 seconds of treadmill walking at each speed making 10 trials for every subject.

All significant quantities of the gait cycle such as angular accelerations, EMG and knee moments were averaged for every subject for each walking speed. The mean values from these averaged data were calculated over the first 13% of the gait cycle (initial swing phase) starting at toe off. These data were checked to be normally distributed by using a graphical test from Matlab®.

Results

The results showed that RF and VL work independent of each other during initial swing phase. The amount of RF activity is clearly related to walking speed. The muscle activity increases with the speed of walking (Figure 1). The relationship between the angular acceleration of the shank and the amount of RF activity is linear (Figure 2). The active knee moment, as a function of the shank's angular acceleration, shows the same high correlation to the EMG signal of Rectus Femoris (Figure 3).

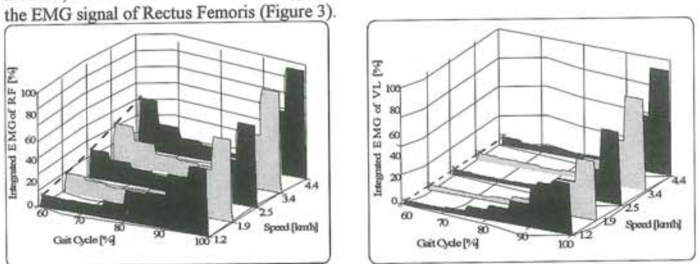
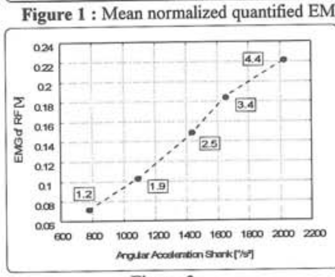


Figure 1 : Mean normalized quantified EMG of RF (left) and VL (right) during the swing.



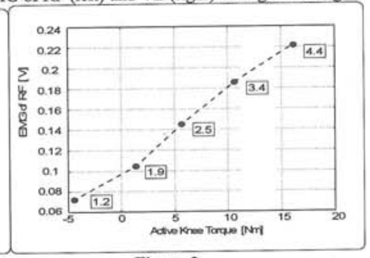


Figure 2. **Figure 3.**

Discussion

The results of this study confirm the few reports which were found in the literature about the activation of RF during the initial swing and provides the evidence needed to support this. However, the co-activation of RF and VL seen during the late swing needs to be further studied to differentiate more clearly between activities of these two muscles. A novel method for finding kinematics by direct measurement of angular velocities and accelerations is more accurate than the equivalent measures obtained from optical systems.

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Analysis of a model of limping: locomotor adjustments to walking on a split belt with unequal speeds.

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Adaptations of human gait are mostly performed through changes in the duration of the stance phase while the swing phase remains constant in duration. However, this constancy of the swing phase may be lost under asymmetrical gait conditions such as occur during limping. The latter can be studied using a split belt in which the speed of the right and left sides can be controlled independently (Dietz et al., 1994; Prokop et al., 1995).

Volunteers walked on a split belt treadmill with the speed on one side higher than on the 1other side with a factor of 2 (3 vs 6 km/h) or 4 (1.5 vs 6 km/h). Surface EMG electrodes were used to record the activity of various leg muscles (including TA) on both sides (8 channels simultaneously). Changes in the angles of hip, knee and ankle were measured with goniometers. Two force plates, one under each part of the belt, were used to detect foot contact.

All subjects adapted to the split belt situation primarily by reducing the ankle angle excursions on the slow side. For large speed differences the hip was slightly held more extended on the fast side. The range of knee excursions were relatively constant. On the fast side, the duration of the swing phase could be almost twice as long as on the fast side. In these prolonged swing phases the tibialis anterior (TA) showed an extra burst just prior to touchdown. During the stance phase there was an earlier onset of the activity in the medial gastrocnemius (MG) on the fast side.

The findings presented above illustrate that the human swing phase shows a high degree of plasticity. Most of the adaptations are made at the ankle and not at the knee, presumably to avoid imbalance because of differences in hip height on the two sides of the body. The changes in muscle activation reflect the increased need to generate larger braking moments at high speed.

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EMG OF THE PHASES DURING AN EXHAUSTIVE TEST IN FRONTCRAWL

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INTRODUCTION :

In 1971, Brown and Counsilman observed sweeping aquatic movements of the hand during the front crawl stroke. Schleihauf (1974) proposed a new decomposition of the stroke into phases in the three planes of the space.This stroke decomposition has recently been supported by an EMG approach (Clarys and Rouard 1995). In regard to these previous findings, the purpose of this study was to examine the evolution of muscular recruitments during the different phases of the stroke in a test up to exhaustion.

METHODS:

Nine french males swimmers participated in this study. The population was homogeneous in performance (mean time on 100m freestyle : 58.2s, s.d. 2.5), age (mean : 17.33years, s.d. 2.59) and height (mean : 180.66cm, s.d. 9).
Each subject was requested to swim 4 X 100m in front crawl. As mentionned by Edwards (1981), the fatigue could be defined by an "increased EMG activity for given performance". In this way, swimmers must sustain the same maximal velocity over the four trials. A 45 second rest period was allowed between each trial. At the end of the last trial, the swimmer was unable to realise another 100m at the same speed because of the " fatigue " condition. The total test corresponded to about 25O repetitions of a stroke cycle.
At the end of each 100m, the movement was filmed with two synchronised underwater camcorders (sony EVO 9100, 30Hz).Surface electromyography of the left M. biceps brachii, the M. triceps brachii, the M. flexor carpi ulnaris, the M. brachioradialis, the M. latissimus dorsi and the M. deltoidus anterior were recorded with a telemetric system (Rouard et al, 1988). Beeckman type electrodes (∅ 11mm) including a preamplifier (gain 1000) were fixed at the midpoint of the contracted muscle belly with an adhesive bandage. Each amplified signal was connected to a voltage-frequency converter and coded with a subcarrier frequency higher than the muscle frequency. The six coded signals were summed. The resultant modulated a radio frequency transmitter (frequency of 72MHz, power of 200MW). The receiving device was set at 72,455Mhz (maximal range of 800m, sensitivity of 2V). The received signal was stored on the HiFi soundtrack of the side underwater video camcorder to synchronise the EMG's with the kinematic data.

For the data treatment, the two views of the stroke cycle was digitized frame by frame using the software" Kinematic Analysis " (Schleihauf 1982). The stroke was divided into four phases, using the coordinates of the fingertip on the transversal axis relative to an absolute (pool-fixed) reference : (i) the downsweep : from the hand entry to its maximum external position, (ii) the insweep : from the maximum external to the maximum internal positions of the hand, (iii) the outsweep : from the maximum internal position of the hand to its exit of the water and (iiii) the recovery : from the hand exit to the hand input in the water.
According to the 1980's ISEK guidelines, the raw EMG signals of each muscle and each 100m swim were full wave rectified. After low pass filtered, the integral of the linear envelope was calculated for the cycle and each of the four phases. To normalise the results, the IEMG's were expressed in percentage of the highest IEMG value observed over the different phases in the four 100m trials for each muscle and each subject.
The Wilcoxon test ($p < 0,05$) was used to compare the different 100m trials and the different phases.

RESULTS :

The evolution of the muscular recruitments over the four 100m trials was different from one muscle to another. No increase in the IEMG of the whole stroke was observed during the four 100m repetitions.
The level of muscular recruitment from one phase to another remained similar over the four 100m trials. The fatigue condition did not affect the muscular patterns through the different phases. For each 100m trial, all the studied muscles presented the highest activity during the insweep phase and the lower during the recovery phase.

For each phase, no increase of IEMG during the four trials was observed excepted for the most activated muscle during the insweep phase (M. brachioradialis) and the outsweep phase (M. triceps brachii) (Fig.1). The antagonsit activities during the insweep and the outsweep phases did not increase over the four trials.

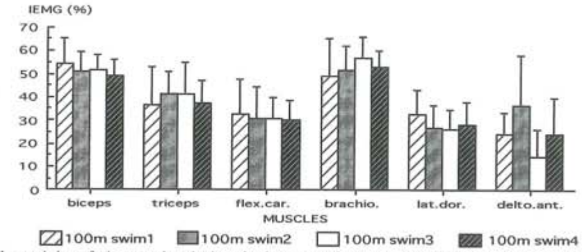

Fig. 1 The activity of six muscles during the insweep phase over the 4X100m front crawl test up to exhaustion.

DISCUSSION :

The repetition of the stroke up to exhaustion in a test of 4 x 100m front crawl was not associated with an IEMG increase during the phases excepted for the most activated muscles. To maintain a given swimming velocity, the swimmer would not recruit more motor units and/or would not improve the motor units synchronisation to produce the stroke cycle ? The exhaustion observed at the end of the 4 x 100m was not typical muscular " fatigue " as observed in previous studies (Bigland-Ritchie,1981, Nyland, 1993). In these previous studies changes in IEMG were associated with changes in force production. The stroking technique was a complex movement in which the length, the velocity and the load of the involved muscles changed continuously during the movement. We could suggest that the non-increase of the IEMG during the different phases over the 4 x 100m could be associated to a decrease of the forces produced by the upper limb.

CONCLUSION :

No typical muscular fatigue was associated with the exhaustion observed at the end of the 4X100m front crawl swimming test. Muscular patterns through the different phases were not influenced by the " fatigue " condition. New investigations will be necessary to control other parameters such as force production and energy supply.

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CLINICAL SIGNIFICANCE OF FUNCTIONAL EVALUATION FOR THE KNEE AFTER ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION

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INTRODUCTION

Anterior cruciate ligament(ACL) injury commonly leads to functional disability in athletic activities. However, it is believed that ACL reconstruction can improve total knee function including knee stability and functional performance. In order to monitor the functional performance of ACL reconstructed knee, several performance tests have been tried and reported. The International Knee Documentation Committee(IKDC)1) recommends the single hop test for the performance test. However, only this test may not be sufficient to accurately evaluate the patients' functional performance. In this study we analyzed four different hop tests in order to identify those with significance in a clinical setting.

MATERIAL AND METHOD

Sixty-five patients, with the mean age of 20.8 $\pm$ 4.6 years old, were studied one year or more after ACL reconstruction with the patella tendon autograft.

All the patients in this study were examined for their knee function by means of KT-1000 and Cybex II. Their performance was also evaluated by the functional ability test(FAT)2), which includes figure-of-8, up and down, side ways and single hop tests.

On the KT-1000 measurement, the IKDC evaluation form was used to classify the knee stability as normal, nearly normal or abnormal1). On the Cybex II evaluation, the method reported by Kannus3) was used to evaluate thigh muscular strength by means of a score. On the FAT, we utilized the criteria which was determined in a previous study of ours2). This criteria enabled us to discriminate between good and poor abilities of the ACL reconstructed patients.

RESULTS

Results of the KT-1000 measurement showed that 92% patients had regained normal or nearly normal knee stability. On the thigh muscular strength test, 54% patients scored more than 70 points, which was regarded as good or excellent. The results of the FAT indicated that the percentage of patients with good hopping ability was 54% for figure-of-8, 68% for up and down, 85% for side ways, and 71% for single hop tests. However, only 40% patients showed good

ability in all the FAT of tests. Moreover, the number of patients, who showed complete regain of their knee function in all the evaluations, was only 11%.

DISCUSSION

The goals of rehabilitation for the ACL reconstruction are to regain knee stability, to restore muscular function of lower extremities, and to improve functional performance in athletic activities. In consideration of these goals, we evaluated the patients with ACL reconstructed knee over one year after the surgery. The result of the KT-1000 measurement showed good in 92% of the patients, and that of the thigh muscular strength showed regaining in 54% of them, and that of the FAT showed improvement in 40% of them in this study. Moreover, only 11% of them totally regained in all the evaluations. These results indicate that knee stability, thigh muscular strength, and performance did not always show parallel relationships. This study suggests that performance test such as the FAT should be utilized for the evaluation of the knee after ACL reconstruction even if both the knee stability and thigh muscular strength were sufficiently improved. This study also suggests that the only single hop test is not sufficient to accurately evaluate functional performance. Therefore, it is recommended that at least three or four tests of the FAT will be needed for the evaluation after ACL reconstruction.

CONCLUSION

It was demonstrated that neither knee stability nor thigh muscular strength showed a parallel relationship with functional performance in patients after ACL reconstruction. Therefore, the conclusion was reached that in the clinical setting the knee function after ACL reconstruction should not only be evaluated by knee stability and muscular strength tests but also by performance tests such as the FAT.

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PHASE-DEPENDENT SWITCH IN STRATEGIES USED TO ANTICIPATE STEPPING OVER OBSTACLES ON A TREADMILL

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Recently, in our laboratory a method was developed to study obstructed gait on a treadmill (Schillings et al.,1996). The advantage of this technique is that the subjects remain in one position and that many trials can be made in a relatively short time. The goal of the present investigation is to measure the vertical ground reaction forces (GRF) for both legs, the movement of the center of pressure (CP) during obstructed gait on a treadmill.

A treadmill has been instrumented with force-transducers under each of the 4 supports to measure the local vertical force. The summation of all the forces delivered by the force-transducers gives the total vertical GRF. A 3D movement analysis system was used to collect the coordinates of the markers placed on the subject. The gait obstruction consisted in a board being dropped on the treadmill in front of the subjects at a predetermined time in the step cycle. The CP has been calculated by an algorithm which uses the signals from the force-transducers. A decomposition of the force signal was made into a left and right profile using the algorithm described by Davis and Cavanagh (1993).

Following an expected obstruction the subjects showed 2 types of strategies. In the first strategy the subjects showed active knee flexion to overcome the obstacle and a lengthening of the steplength (in confirmation of McFadyen et al., 1993 for walkway obstructions).

When faced with obstructing obstacles, subjects could also use a second strategy. This second strategy consisted of a rapid lowering of the ipsilateral limb and a shortening of the steplength. The ipsilateral foot was placed in front of the obstacle. To overcome the obstacle an active knee-flexion movement was made during the ensuing swing phase.

By manipulating the timing of the fall of the obstacle on the treadmill the time of switching between these two strategies was examined. It was found that the time of the switching between the two strategies corresponds to the moment when the subject would have stepped on the obstacle in case of unchanged steplength. The subjects were capable to estimate if the length of their corrective step was long enough to overcome the obstacle.

In conclusion, with the present method it is possible to measure the changes in vertical ground reaction forces resulting from reactions to obstructed gait on a treadmill. This allows the biomechanical study of highly reproducible anticipatory locomotor adjustments during obstructed human walking.

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CHAPTER 5

ELECTROMYOGRAPHY

SINGLE NERVE-FIBRE AND SINGLE-MOTOR UNIT ACTION POTENTIALS IN REHABILITATION

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INTRODUCTION: The outcome of central nervous system (CNS) lesions can be improved based on a better understanding of the human CNS function and with the therapy based on it. A new understanding of the functioning of the human CNS, i.e. self-organization of neuronal subnetworks to oscillators and the generation of macroscopic function by changing oscillator couplings, has been offered by a new method of simultaneous recording of natural impulse patterns of identified single afferent and efferent nerve fibres which enables CNS functions to be analysed under physiologic and pathophysiologic conditions [2,3]. The improved knowledge of the natural firing patterns of motoneurons can be expected to allow a better extraction of single motor unit action potential (MUAP) patterns from multi-unit electromyographic recordings.

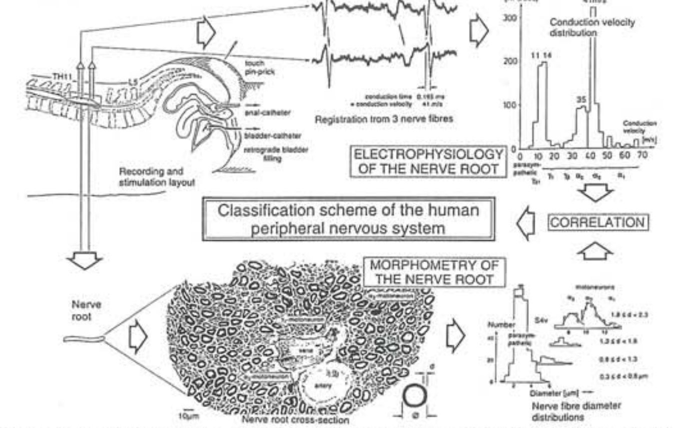

Fig.1 Schematic layout of the classification scheme for the human peripheral nervous system. By recording with two pairs of platinum wire electrodes from a nerve root containing approx. 500 myelinated nerve fibres, a recording is obtained in which 3 action potentials (APs) from 3 motoneurons (main AP phase downwards) can be seen. By measuring the conduction times and with the known electrode pair distance (10 mm) a conduction velocity distribution histogram is constructed in which the nerve fibre groups are characterized by ranges of conduction velocity values and peaks in asymmetrical distributions. After recording, the root was removed, fixated, embedded and stained, light microscope cross-sections were prepared and used to measure the mean diameter and the myelin sheath thickness (d). Distributions of nerve fibre diameters were constructed for four different ranges of myelin sheath thickness. Nerve fibre groups were characterized by the peak values of asymmetrical distributions. By correlating the peak values of the velocity distributions with those of the diameter distributions obtained from the same root, a classification scheme was constructed of the human peripheral nervous system. Brain-dead human HT6.

METHOD: Single nerve-fibre action potentials (APs) were recorded extracellularly with two pairs of platinum wire electrodes (electrode pair distance 10 mm; electrode distance in each pair 4 mm) from undissected nerve roots at two sites, preamplified (x1000), filtered (RC-filter, passing frequency 100 Hz - 10 kHz), and displayed on a digital storage oscilloscope (Vuko Vks 22-16), and also stored using a PCM-processor (Digital Audio Processor PCM-501ES) and a video recorder. Conduction velocity distributions of afferents and efferents were

constructed, calibrated, and group conduction velocities were identified and defined. Subsequently, simultaneous impulse patterns of several single afferent and efferent nerve fibres were extracted from multi-unit recordings. Recordings were performed in individuals with complete spinal cord lesions during surgery for the implantation of an electrical anterior root stimulator for urinary bladder control. Single MUAPs were recorded with the same equipment, with the wire electrodes replaced by standard surface elelctrodes or using EMG equipment.

RESULTS: The data derived using the single nerve-fibre AP recording method enabled to develop a classification scheme for the human peripheral nervous system (Fig.1) for myelinated fibres thicker than 3.5μm, with the nerve fibre classes being characterized by group conduction velocities and group nerve fibre diameters. The discovery of premotor spinal oscillators, the FF, FR and S-type motoneurons can further be identified by their repetitive firing (10, 6-9, 1 Hz, respectively) with impulse trains. Oscillation period and phase distribution changes provide information changes in the coupling of the spinal oscillators. Following spinal cord lesion, these spinal oscillators change their firing patterns [1]. Since there is indication that spinal oscillators build up, via the γ-loop, an external loop to the periphery, they can be entrained by a rhythm training for an improved self-organization and thus function. We could improve the rhythm of a patient with an incomplete spinal cord lesion sub C5 to run [4]. Since oscillatory firing patterns of FF and FR-type motor units have been found in EMG recordings, it should be possible to partly analyze CNS dysfunctions with the non-invasive method using EMG surface electrodes, to follow up network organization changes of oscillators.

DISCUSSION: The single nerve-fibre AP recording method provides a powerful tool to analyze CNS functions. The oscillatory firing of motor units and locomotor patterns can be used to non-invasively follow up therapeutic interventions. Further identification of the natural firing patterns of single motor units will require direct simultaneous recording of single nerve-fibre APs and MUAPs. Since rhythmic dynamic stereotyped movements are mainly located in spinal cord neuronal networks (newborn babys can step automatically), little supraspinal drive is necessary to initiate and maintain rhythmic movements, especially under weight reduction. In spinal cord lesion [5], stroke [6] and cerebral pulsy, it should be possible to essentially improve the outcome with rhythmic training methods included in the treatment; these methods fit the organization principle of the human CNS, i.e. the self-organization of neuronal subnetworks to oscillators and the generation of macroscopic functions by changing oscillator couplings.

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MOTOR UNIT ANALYSIS OF STROKE PATIENTS USING NONINVASIVE MULTIELECTRODE ELECTROMYOGRAPHY

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INTRODUCTION

Previous studies on the needle EMG indicated that the measurement of motor units (MU) discharge pattern might be useful in clinical for the electrophysiologic detection of upper motor neuron disorders, such as stroke [1]. However, the clinical EMG measurement using traditional needle electrode suffers various disadvantages including invasiveness, discomfort to the patient, and lower measurement repeatability. Instead of needle electrode, recent development in EMG measurement techniques has made it possible to utilize the surface electrode matrix for the recording of motor unit action potentials (MUAPs) in superficial muscles [2]. Due to its non-invasive property, the derivation of so-called surface MUAPs from the multielectrode EMG have been applied for clinical studies in the last few years. However, previous research was focused on the electrodiagnosis of patients with neuromuscular disorders [2]. The aim of this study is to extend the application of surface MUAPs to quantify the MU firing variation for paretic limb in stroke patients [4].

METHODS

In this research, a multielectrode with 4 contact probes in 2 * 2 matrix arrangement, laying in 2.5 * 2.5 mm interelectrode distance for differential input of EMG signal, and a ground pin electrode, is developed for the measurement of surface MUAPs. The surface multielectrode consists of stainless and sharp pin electrodes, adopted from electronic contact probe [4]. The pin electrodes are attached to the base with spring which can provide cushion for better contact with human skin.

A three-stage approach for the analysis of surface MUAPs is developed in this study. First, the surface MUAPs are passed through a digital spatial filter for providing higher spatial resolution. For the overlapping surface MUAPs, a modified EMG decomposition technique which considers the spatial information of multielectrode configuration is developed. After localization of dominate MU, features for quantifying firing rate and variation are represented in joint interval histogram (JIH) plot [3]. The JIH plot reflects the variability of subsequent firing intervals. The quantitative descriptions of the firing variation can be observed from the slope of long axis (JIH slope), the area of ellipse ($ab\pi$), and the ratio between long and short axes (a/b ratio). For each of the six stroke patients recruited, sessions of the EMG signal from both sides of abductor digiti minimi are recorded when the subjects performed a low level isometric contraction. With the limited clinical cases, the firing variation measurement are only appropriate for intra-subject comparison for affected and unaffected sides.

RESULTS

In this study, only the dominate MUAPs, are used for firing rate analysis. The JIH parameters of six stroke patients are listed in Table I. All the JIH slopes of the unaffected side are smaller than those of the affected side. In addition, most of the JIH slope for unaffected side (4/6) are negative; while most of that for affected side (5/6) have positive JIH slope.. These observations indicate that the firing patterns of most affected sides are either with trend or with higher firing variation. For a/b ratio, one-half of the unaffected sides (S1, S2, and S3), show smaller a/b ratio's which indicate more consistent firing intervals. Although the other half of subjects (S4, S5, and S6) have contrary results, their a/b ratio's usually are very close to 1.0. Furthermore, for these three subjects, S4, S5, S6, their ellipse area are very small.

Table 1: The JIH Parameters for Affected and Unaffected Sides of Stroke Patients

	JIH slope		ab ratio		area(*0.01)	
	Unaff.	Aff.	Unaff.	Aff.	Unaff.	Aff.
S1	-0.053	0.153	1.012	1.049	1.324	0.643
S2	0.227	0.327	1.104	1.238	0.367	0.342
S3	-0.015	0.639	1.005	1.916	0.354	1.182
S4	-0.107	0.022	1.024	1.023	0.256	0.749
S5	-0.161	-0.025	1.053	1.004	0.265	1.077
S6	0.035	0.075	1.039	1.012	0.211	0.972

DISCUSSIONS

Although promising results for quantifying the motor control of stroke patient has been shown in this study, many factors could affect the clinical application of surface MUAPs. First, the MU firing characteristics might depend on the force output; however, precise control of a stable isometric contraction force is relatively difficult for elderly stroke patients. Second, whether the MU activity recorded at different cites of same muscle would exhibit similar firing characteristics is still under investigation. Third, the quality of the measured surface MUAPs is essential for the success of this study. Our on-going research is to establish a real-time EMG analysis system which could provide a direct visualization of the signal quality and immediate quantitative analysis of the desired features.

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High-Spatial-Resolution EMG (HSR-EMG) for a Noninvasive Detection of Changes in the Motor Unit Structure: A Comparison of Model and Experimental Results

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Introduction

Neuromuscular disorders are often related to typical changes in the in the structure of single motor units (MUs). An approach for a noninvasive detection of these typical changes represents the **H**igh-**S**patial-**R**esolution-**EMG** (HSR-EMG), which is based on the use of a multi-electrode array in combination with a spatial filter processing [1]. The HSR-EMG allows in contrast to other surface-EMG procedures the recording of the single MU activity up to maximal voluntary contraction of the muscle [1]. First investigations in patients suffering from different neuromuscular disorders have shown, that there is a distinct difference between the HSR-EMG patterns of healthy volunteers, patients with muscular disorders and patients with neuronal disorders [2]. In this study the relationship between the typical HSR-EMG patterns and the characteristic pathological changes in the structure of the MUs is considered.

Method

In a first step a muscle model has been developed which is adapted to the physiological properties of the *m. abductor pollicis brevis* (Table 1). The muscle fibres have been assumed as infinite long cylinders arranged parallel. Each muscle fibre has been assigned to one MU regarding the physiological distribution of the MU in the muscle (Table 1).

Distance between skin surface and muscle tissue	3 mm
Conductivity of the volume conductor	0.08 1/Ωm
Density of the muscle fibres	256 Fibres/mm^2
Density of the motor units	16 MU/mm^2
Spatial extension of the motor units (circular)	50 mm^2
Muscle fibres belonging to one motor units	804 Fibres/MU
Firingrate of the motor units	8 - 20 Hz
Conduction velocity in single motor units	3.5 m/s
Orientation of the muscle fibres	parallel

Table 1: Physiological properties of the muscle model.

The potential distribution generated on the skin surface by each muscle fibre has been calculated from a tripole model, introduced by Griep et al. [3]. This tripole model represents the intramuscular single muscle fibre action potential. The volume conductor has been assumed as isotrop and infinite in two dimensions. The resulting potential distribution generated on the skin surface by a non synchronous excitation of all available MUs has been calculated by a superposition of the contributions of all active muscle fibres. Afterwards, the simulated potential distribution has been spatially filtered.

The effect of a loss of complete motor units (neuronal disorders) and of a loss of single muscle fibres (muscular disorders) on the HSR-EMG pattern have been considered by different computer simulations. The resulting HSR-EMG patterns have been compared with the typical HSR-EMG patterns of patients with muscular and neuronal disorders.

Results

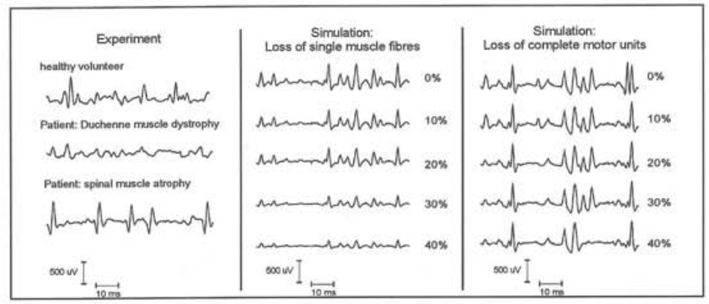

***Figure 1**: Comparison of experimental data and simulations of a progressive loss of single muscle fibres (Duchenne muscle dystrophy) or complete motor units (spinal muscle atrophy).*

The typical HSR-EMG pattern of patients with neuronal disorders shows isolated peaks with only a little activity between; the typical HSR-EMG pattern of patients with muscular disorders low amplitude peaks which are hard to recognise in the signal [2]. Simulating a progressive loss of single muscle fibres or complete MUs both patterns can be generated by the muscle model (Figure 1). An especially developed pattern recognition [4] detects the typical change in the simulated HSR-EMG patterns if more than 15% of all normally active muscle fibres or MUs became inactive.

Conclusion

The simulations of typical pathologic changes of the structure of the MUs show the same pattern as it has been seen in 97% of all investigated patient with muscular or neuronal disorders. From this correspondence it can be assumed, that the typical changes in the HSR-EMG pattern of patients with neuromuscular disorders are based on the pathological changes of the MU structure and as a consequence that the HSR-EMG allows the noninvasive detection of these changes.

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Distribution of innervation zones in the upper limb muscles identified by using multi-channel surface EMG

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INTRODUCTION

It has been recently documented that the surface EMG wave forms are influenced by the location of recording electrodes with respect to the neuromuscular junctions [1]. Therefore, identification of innervation zones prior to recording may be useful for surface electromyography. To date, however, attempts of mapping the innervation zones have been limited to those made by Masuda and colleagues [2, 3], in which the biceps brachii, trapezius, and latissimus dorsi muscles have been examined. In this study, we have attempted to identify and map the innervation zones in the upper limb muscles other than the above using multi-channel surface EMGs.

METHODS

Three male subjects participated in this study. Myoelectric potentials were recorded bipolarly from the deltoid, brachial biceps and triceps, forearm muscles, and intrinsic hand muscles. A 14-channel active electrode array, composed of 15 stainless steel bars of 5 mm long arranged at 2.5 mm intervals, was applied to the muscle in parallel with muscle fibers. The myoelectric potentials detected were digitized at a frequency of 2kHz, fed to a personal computer and displayed on a monitor screen. The propagation pattern of the potentials thus displayed enabled identification of the sites from where the potentials originated, which were taken as neuromuscular junctions. Mapping of the junctions to delineate the innervation zones was achieved by repeating these procedures while shifting the electrode locations over the whole muscle belly.

FINDINGS

While the propagating potentials could be obtained from the upper arm, intrinsic hand, and deltoideus muscles, identifications of the propagation on individual muscle basis were difficult in most muscles in the forearm. The following are brief descriptions of the results in each muscle group.

1) Deltoid: Despite that the potential propagations were often interrupted on account of the architectual complexity in this muscle including pinnations, a number of records were obtained in which the neuromuscular junctions could be identified with relative ease. Mapping of the junctions revealed that innervation zones as a whole distribute in a convex strip upward, although expanding rather widely in the longitudinal direction.

2) Upper arm muscles: The propagation patterns were fairly clear in the biceps brachii within the whole muscle belly. The innervation zones extend transversely in the middle portion of the muscle. Although the propagation was relatively clear in the triceps brachii,

perturvations were sometimes observed due to helicoid arrangements of muscle fibers. The innervation zones were mapped as a parabolic strip surrounding the aponeurosis.

3) Forearm muscles: Propagating potentials could be identified in the brachioradialis and flexor carpi ulnaris. Innervation zones were found to locate in the middle portion of these muscles, though scattered rather widely. Since it was difficult to demarcate individual muscles except the above ones, the innervation zones were identified collectively for the flexors and extensors, respectively. As a result, the innervation zones, while showing a tendency to scatter diffusely, were found to primarily distribute in the proximal half of the muscles in the flexors and in the middle half in the extensors.

4) Intrinsic hand muscles: The propagations were relatively clear in the abductor pollicis brevis, abductor digiti minimi, and interosseus dorsalis I muscles. Innervation zones in these muscles were identified to aggregate in the middle of the muscle belly.

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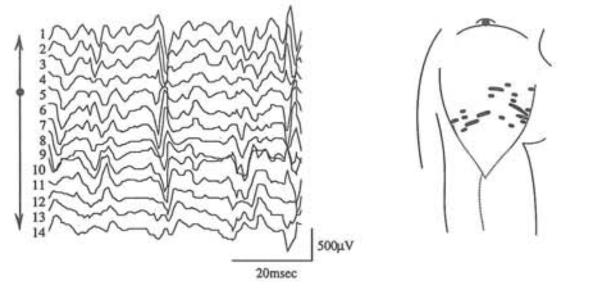

Fig. 1 *Left figure*: Propagating myoelectric potentials recorded from the clavicular portion of deltoid. Black circle indicates the channel including the site from which the potentials propagate to opposite directions as shown by arrows. *Right figure*: Innervation zones mapped in the deltoid. Black circle indicates the acromion.

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A NEW QUANTITATIVE NEURAL METHOD FOR THE INDIVIDUATION AND CLASSIFICATION OF MUAPS WITHIN A HYBRID SYSTEM FOR THE NEUROLOGICAL CLINICS

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INTRODUCTION

One of the hardest tasks formulating a neurological diagnosis is the interpretation of the EMG signal. The most accepted quantitative method for the evaluation of the EMG signal is the study of individual MUAPs and quantitation of the MUAPs parameters, such as amplitude, duration, and number of phases is used to discriminate between different neuromuscular disorders. Unfortunately, the analysis of the EMG signal in order to individuate and classify the MUAPs is a time consuming process and requires specific skill; consequently only a few number of clinical departments have used it for the daily routine EMG analysis. In the last generation of Electromyographs new specific modules for the automatic analysis and classification of the MUAPs have been presented [4]. This modules are based on pattern recognition, template matching, spike-triggered averaging, statistical and syntactical methods.

In our study a new quantitative method based on Artificial Neural Networks (ANN) for the analysis of the MUAPs has been developed to single out and classify the MUAPs recorded with an intramuscular needle electrode during a weak contraction. At each recording site 2 to 6 MUAPs may be recorded and almost no operator interaction is requested. The method has been implemented on a module which may be used as a stand alone system, called Neurap. On the other hand, Neurap has also been integrated with a tutorial Expert System (ES) called Neurex [2,3] for the diagnosis of the neurogenic diseases, realising the hybrid system [1] Hynex.

METHODS

Neurap is made of four modules: the first one visualises the signal with the recording information (date, sample frequency, sweep, sensibility, low filter, high filter,..) and the anamnestic data, the second module does the segmentation of the signal, while the third module, the neural one, decides weather the selected segments from the second module are true MUAPs and classifies them in clusters. The forth module extracts the relevant features of the MUAPs belonging to the individuated clusters and passes on these parameters to the inferential engine of Neurex, in order to formulate new diagnostic inferences.

The first module for the off-line visualisation of the signal, has been developed in order to allow the electromyographist to review-reprocess the signal providing the user with a friendly Window like interface.

The segmentation of the EMG signal and the individuation of the possible MUAPs (the second module) is based on an innovative algorithm that takes into account not only threshold criteria (amplitude and rising time), but also heuristics resulting from the clinical experience, acquired and assessed during years of study of the shape of the MUAPs.

The MUAPs classification is done with an unsupervised competitive ANN (Kohonen based) that creates clusters of identical MUAPs, with no *a priori* information about the number of the clusters. The segments singled out by the second module are presented to the ANN in an incremental way. The ANN matches the segment with the patterns memorised in its

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topological map and finds out the pattern which is most similar to the current sample. The two last modules work together, in cascade, during the problem solving task of the individuation and classification: the second module has been developed using the assessed experience and skill of the clinicians (deductive reasoning), while the ANN finds out new information in an inductive way.

The forth module visualises, for each selected cluster, the samples in it classified. A cluster has to have more than five identical samples in order to be considered a real MUAP class. The MUAPs features passed on to the ES are the amplitude, number of phases, duration, number of turns, area, rising time and thickness (area/amplitude ratio). The clinician may view for each cluster which MUAPs have been put together and if a MUAP does not appear to be exactly as the others, may discard it. The mean of the values of the meaningful features are then estimated and passed to Neurex.

RESULTS AND DISCUSSION

Hynex has the following functional flow: Neurex gets in input the signs of an Objective Neurological Examination, formulates the initial hypothesis of the site of lesion and guides the user in planning an optimal sequence of NG tests. After the interpretation of the results of these instrumental tests, the EMG session begins in order to confirm or reject the current hypothesis. Neurex suggests a muscle to be examined, the user records the EMG signal during a weak contraction with a deep electrode and runs Neurap to analyse the signal recorded. Neurap finds out the MUAPs (usually around 20 different MUAPs are analysed), extracts characteristic features of the retrieved MUAPs and passes them to the knowledge base of Neurex that interprets them on the basis of its knowledge and the current hypothesis. If the Neurex considers the information acquired to be sufficient, it stops the examination session and communicate its final diagnosis, otherwise suggests a new muscle to be examined.

Hynex meets the two main requirements of the tutorial/assistant hybrid systems: to tutor unskilled people to become experts and to increase the performance/cost ratio both for patients and health-care organisations being a time saving system. Neurap has been trained and tested for the Tibialis Anterior muscle and it seems to perform better than current traditional methods. Further work has to be done to make the whole system robust and validated. Since the system is used with conventional apparatus, displays data in a rapid and correct way we hope it could be accepted by the clinicians for daily routine EMG analysis and for tutoring less experienced doctors.

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A COMPUTER SIMULATION OF EMG USING REALISTIC NUMBERS OF EXPERIMENTALLY RECORDED MOTOR UNIT ACTION POTENTIAL WAVEFORMS

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INTRODUCTION

The electromyogram (EMG) is an aggregate representation of the individual action potential (AP) contributions from active motor units (MUs) detectable, within a critical distance, by the recording electrode. As a result, the relative magnitude of a MUAP contribution is highly dependent on its topographical location with respect to the recording electrode (1). In contrast, it is widely accepted that force output from a single MU is essentially independent of its location in the muscle. A number of studies, using different muscle preparations, have investigated the relation between the magnitude of the AP and the force produced by isolated single MUs (2-4). In each of the studies, the relations between the MUAP and force measures differed considerably; however, the amount of dispersion of the individual data points about the regression line describing the relation was markedly large. Similarly, there is large body of conflicting evidence regarding the linearity of the aggregate EMG -force relation for various preparations during isometric contraction (voluntary activation) (5). All these results taken together indicate that more sophisticated methods of signal analysis and/or novel methodological approaches are necessary, to understand and accurately quantify the EMG signal recorded during voluntary muscle contractions, as the differences in preparation and recording conditions may dramatically alter the composite EMG signal. One potentially beneficial approach is to theoretically fabricate MUAP shapes, or alternatively to record them experimentally, and use this data in a computer simulation of muscle contraction and EMG. Nevertheless, this method is dependent upon the assumption that the individual MUAP contributions interact in a consistent manner throughout the range of physiologic contraction intensities. In support of this assumption, recent evidence convincingly demonstrates that the individual MUAPs contributing to the aggregate EMG summate algebraically, i.e. in an essentially linear manner (6). The current study uses a computer simulation to assess the effects of different MUAP shapes on the characteristics of the EMG signal for various activation strategies of the MU pool during mimicked isometric muscle contraction.

METHODS

In acute cat experiments, a substantial proportion of the soleus MU pool was isolated and their MUAP shapes were recorded from the perspective of bipolar surface electrodes arranged in parallel and perpendicular to the longitudinal axis of the muscle (i.e. the predominant orientation of the muscle fibres, given the unipennate architecture of the muscle). Data from several experiments were pooled to generate a library of experimentally recorded, realistic, diverse and often complex MUAP waveforms. For some applications the complexity, duration and area of the recorded MUAPs were altered, in order to determine the effects of these parameters on the aggregate EMG signal generated by the computer simulation. Furthermore, a few single MU waveforms were fabricated based upon mathematical models (1), to compare the EMG produced with theoretical MUAP shapes, with the signal generated using the large numbers of recorded MUAPs. For each condition, a realistic number of MUs, mimicking the pool in the soleus muscle (~150), were used in a computer simulation which replicated various recruitment and rate modulation strategies. The activation pattern of each MU was defined as a unique sequence of intervals and an individual AP waveform. The resultant MUAP trains were summed producing the aggregate EMG signal. For each activation strategy, the order of MU recruitment followed the size principle (7), with the smallest MU recruited first and the largest last.

A pure recruitment design featured the sequential activation of each MU at a fixed rate of stimulation, while a combined recruitment and rate modulation strategy consisted of successive MU recruitment with subsequent increases in the stimulation rate. The pure recruitment design was used to assess the spectral properties of the EMG signal for the various conditions with differing MUAP characteristics. In contrast, the combined activation design was used to evaluate the properties of the raw and quantified aggregate EMG signal in the temporal domain.

RESULTS

Preliminary investigations with smaller pools of MUs (30 or 60) indicate that the characteristics of MUAPs can have considerable effects on the rectified, window averaged EMG signal (AEMG). Illustrated in Figure 1, and consistent for simulations with systematically varying MUAP characteristics, is the striking gain compression of the aggregate EMG (lower traces). At the highest ensemble activation rates there is a 52.6% and a 65.8% reduction in AEMG magnitude (for the 30 and 60 MU simulations, respectively), compared with the estimates of AEMG magnitude obtained in simulations with full wave rectified MUAPs (upper traces). The degree of AEMG reduction (shades areas in Fig. 1) is a measure of signal saturation attributable to cancellation of overlapping MUAP waveforms with opposite polarities. In the present simulations, the signal cancellation is strongly dependent on numbers of MUs recruited and their rates of activation.

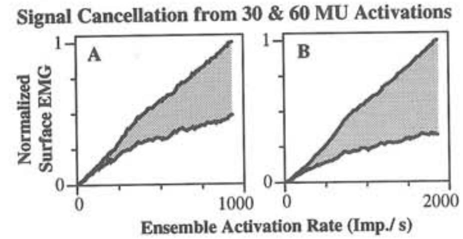

Figure 1: The rectified and averaged EMG from the simulated activation of A) 30 and B) 60 MUs with a physiologic recruitment and rate modulation design (lower trace). The MUAP shapes were input into the computer simulation after rectification (top trace) to illustrate the degree of signal cancellation (shaded area).

DISCUSSION

The results from the preliminary computer simulations indicate that it would be beneficial to use representative numbers of experimentally recorded MUAPs in further simulations of muscle activation, as it is apparent that the wide variety of MUAP shapes obtained for any electrode - muscle configuration, can not be represented by a small number of theoretically fabricated waveforms. Furthermore, a simulation using this approach may be even more advantageous for attempts of characterizing the aggregate EMG signal for certain neuromuscular disorders, where identifiable changes in the properties of the MUAPs are manifest.

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TIME-FREQUENCY ANALYSIS OF SURFACE EMG SIGNALS USING WAVELETS

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INTRODUCTION

Analysis of frequency spectra of surface electromyography (EMG) signals is a commonly used method to determine local muscle fatigue, force production, and myoelectric signal conduction velocity. The most common method for determining the frequency spectrum of the surface EMG signal is the Fast Fourier Transform (FFT). To produce a time-frequency representation of the signal, the short-time Fourier transform has recently been used. A major drawback of both these methods is that they assume a stationary signal. Even when there is no voluntary change of muscle state, myoelectric signals are non-stationary simply due to the inherent physiology of the organ. It is a common understanding that during a static contraction the surface myoelectric signal is locally stationary. During isokinetic contraction the frequency parameters can be questioned because the EMG signal is no longer stationary and therefore the Fourier transform is not applicable. A relatively new technique, the wavelet transform [1], is well suited to non-stationary signals, and could be an interesting method for isokinetic conditions.

METHODS AND RESULTS

Six healthy male volunteers (20-35 years) performed isometric knee extension at 70% of MVC with the knee flexed 90° until exhaustion. Surface EMG was recorded from vastus lateralis using two Ag/AgCl bipolar electrodes. The EMG signal was amplified (filtered between 15 and 700 Hz) and A/D-converted with 12-bit accuracy. The signal was sampled at 2kHz and stored digitally on hard disk for later analysis. All the analysis in this study was done using S+WAVELETS and Uvi_Wave Matlab toolbox (from Universidad de Vigo, Spain).

For the initial analysis, one simulated isometric surface EMG data set (from Dip. Elettronica del Politecnico di Torino, Italy) was used to test the applicability of wavelet transforms. Since myoelectric manifestations of muscle fatigue have been described by monitoring the mean frequency (F_{mean}) of the power spectral density (PSD) during voluntary or electrically elicited sustained contractions [2], several F_{mean} estimates based on different methods of PSD estimation are made for comparisons. The classical estimate of PSD based on FFT is shown in Fig. 1(a). For a stationary signal, it is asymptotically unbiased, but unfortunately not consistent. The wavelet shrinkage estimate [3] is shown in Fig. 1(b). This technique is based on wavelet decomposition of the FFT-periodogram and reconstruction of the spectrum, and it can estimate spectra which are non-smooth at a near-optimal rate. Their F_{mean} is shown in Fig. 1(e).

Concerning the further study on non-stationary signals, a new technique for analysing the spectral distribution, by using wavelet packet (WP), is introduced and tested for the simulated data. Fig. 1(c,d) shows this method and the wavelet shrinkage estimate based on it. The comparison of F_{mean} between FFT- and WP-periodogram is shown in Fig. 1(f).

In Table 1, correlations between different F_{mean} estimates for the volunteer surface EMG data are calculated.

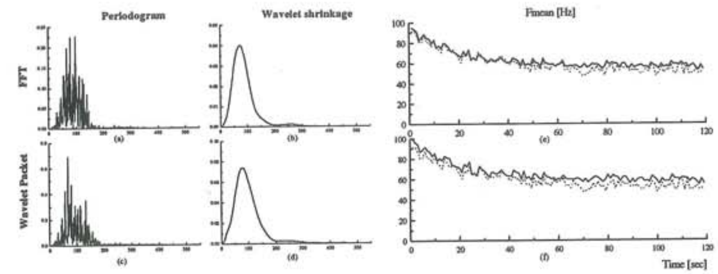

Figure 1: The four graphs to the left show estimators of PSD: (a) FFT-periodogram (b) wavelet shrinkage based on FFT (c) WP-periodogram (d) wavelet shrinkage based on WP. The graphs to the right show the F_{mean}: (e) FFT (dotted) vs smoothed FFT (solid); (f) FFT (dotted) vs WP (solid).

	#1	*#2*	*#3*	*#4*	*#5*	*#6*
FFT-WP(D6)	0.999	0.960	0.882	0.982	0.987	0.925
FFT-WP(D8)	0.999	0.967	0.921	0.989	0.991	0.941
WP(D6)-WP(D8)	1.000	0.994	0.989	0.996	0.998	0.995

Table 1: Correlation factors for the six subjects between F_{mean} using FFT and WP with Daubechies's wavelets D6 resp. D8.

CONCLUSIONS

This study presents an alternative time-frequency analysis method for locally stationary surface EMG signals. The results show that WP-periodogram performs estimates similar to the classical FFT-periodogram. Moreover, this method has a great potential to be used in analysing isokinetic (non-stationary) signals, since the wavelets allows us to analyse signals into both time and scale.

The table shows that the measured surface EMG signals can be analysed by WP-periodogram, which is highly correlated to the classical FFT-periodogram.

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A MOTOR UNIT SYNCHRONISATION INDEX BASED ON ELECTROMYOGRAPHIC SPECTRAL MOMENTS OF DIFFERENT ORDERS - A PILOT STUDY

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Introduction

In the discussion of muscular fatigue and its relation to surface EMG parameters, EMG spectral moments like median frequency (MF, moment of order 0) and mean power frequency (MPF, first order moment) play a central role. Also the zero crossing rate (ZC, expected number of ZC:s is a second order moment) has been used as a fatigue index. These indices mainly mirror alterations in action potential velocity (APV) decrease and synchronization of motor unit firings (SMUF) (Hägg 1992). Even if there are controversies regarding the existence and origin of SMUF (De Luca, et al. 1993), progressive peaks in the low frequency end of the EMG spectrum supports the idea of dominating firing frequencies at fatigue. It has been shown earlier that while there is a proportional relationship between a spectral moment of any order and APV alterations, the sensitivity of spectral moments to low frequency alterations related to SMUF depend on the moment order; the lower spectral order, the higher sensitivity to low frequency alterations (Hägg 1991).

In order to find a specific "synchronisation index" based on surface EMG, the followig index

$$I_S = \frac{E[ZC]/E[ZC]_0}{MF/MF_0}$$

has been suggested (Hägg 1993). E[ZC] is the expected number of zero crossings calculated as a second order moment from the EMG power spectrum (Rice 1945) and the 0 index referres to a well rested reference. This index will remain unity as long as no additional low frequency peaks occur independent of action potential velocity fluctuations. Specific increase in the low frequency band will imply an increasing index.

Material and methods

In a pilot experiment, eight male test subjects performed a static elbow flexion endurance test at 40 % MVC. The subjects were seated with the upper arm resting on a horizontal support, 90° elbow flexion and with a force transducer attached to the wrist with a horizontal strap. Surface EMG electrodes (20 mm interelectrode distance) were attached over the belly of m. biceps brachii and EMG was stored on tape. At the following analysis, EMG power spectra were calculated over two second periods each fourth second and from these spectra, an index I_S according to the definition above was calculated.

Results

The median endurance time was 148 s. Normalised MF and E[ZC] decreased to varying degrees, MF always a little more. Minimum readings at exhaustion were 0.41/0.60 for MF/E[ZC]. The resulting I_S curves plotted versus time (start value=1) show readings >1 with initial linear increasing trends (see figure 1). Six subjects had a linear trend during the whole experiment with final values in the range 1.03-1.19 while two subjects showed a marked progressive increase during the last third of the experiment with final readings 1.47 and 1.55.

Discussion

The results confirm earlier observations of a larger relative response in MF compared to E[ZC] (Hägg 1991). A comparison of average spectra at the start of the experiment and at exhaustion illustrates the two effects; a parallel spectrum shift (logarithmic scales) representing a APV decrease and a specific increase between 15 and 40 Hz representing firing statisitics (possibly synchronisation). It is suggested that this increase is due to progressive

recruitment of phasic type II motor units and thus not due to a synchronisation of type I motor units recruited from the begining of the static contraction.

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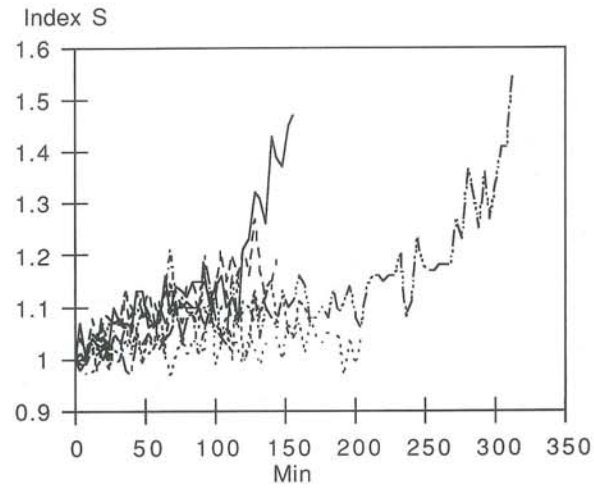

Figure 1. Development of Index S for each subject

NON-INVASIVE ESTIMATION OF MUSCLE FIBER CONDUCTION VELOCITY: COMPARISON OF TWO DETECTION SYSTEMS

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Abstract. Muscle fiber conduction velocity (CV) may be estimated either with needle or surface electrodes. The double differential technique was used in this work to estimate a) the repeatability of non-invasive CV estimates in the normal human biceps brachii and, b) the effect of interelectrode distance e (5 mm and 10 mm probes). Electrical stimulation of a muscle motor point was used to elicit M-waves of maximal amplitude in isometric conditions. CV was estimated as the ratio between e and the delay between the two double differential signals. Measurements were repeated 5 times in each of 4 days, for 5 normal subjects and for each probe. It is concluded that the estimates obtained with the e=10 mm are comparable with the needle estimates and are affected by a variance smaller than that in the e=5 mm case.

Introduction. Muscle fiber CV is affected by a number of pathophysiological conditions that are clinically relevant [1]. Needle estimates are specific but are time consuming, painful and cumbersome. Non-invasive estimates are more global and painless, may be repeated often and require little time. It was the objective of this work to estimate the effect of e and the repeatability of CV estimates in normal subjects.

Methods. Five normal subjects (three males and two females, age range 26 to 36 years) partecipated in the experiments after signing an approved informed consent form. The elbow was set in an isometric brace at about 120° and skin temperature was mantained at 32° ± 0.5°. Monopolar electrical stimulation of the lateral motor point of the right biceps was used to elicit a maximal amplitude M-wave. A round stimulation electrode (1 cm diameter) was placed on the motor point and a large rectangular electrode (8 cm x 12 cm) was placed on the triceps. Constant current stimulation pulses of 0.2 ms duration and 2 pps frequency were used. A previosly described stimulation/detection system with artifact suppression was used [3]. The first electrode of the 4-bar array was located about 2 cm distally with respect to the stimulation point. Two active probes were used for EMG detection: the first had four silver bars (Ø = 1 mm), 10 mm long and 10 mm apart, the second had Ø = 1 mm bars, 5 mm long and 5 mm apart. Stimulation was applied five times (trials) for 5 s each time at 4 minutes intervals. This protocol was repeated in four days. Each day the stimulation current was adjusted to elicit the M-wave of maximal amplitude. The low frequency (2 pps) and the short stimulation duration (5 s) were chosen a) to avoid any manifestation of muscle fatigue and b) to compare results with literature data obtained with needle detection [5]. The McGill-Dorfmann algorithm for delay estimation between the double differential (DD) signals, extensively used in previous research [2], was used in this work as well. Estimates of CV from the middle three seconds of each contraction were averaged and tabulated. Mean and standard deviations of the estimates were computed for each subject. Analysis of variance was applied to estimate the fractions of the total variance that could be attributed to trials, days and subjects.

Results. The probe with e = 5 mm consistently led to CV estimates higher than those obtained with the 10 mm probe. The difference between the means of the two sets of data was statistically significant for each subject ($p < 0.001$, t-test, N=20). The global (N=100) mean and standard deviation for the 5 mm probe were 5.52 ± 1.04 m/s (coeff. of var. = 18.8 %), while, for the 10 mm probe were 3.75 ± 0.34 m/s (coeff. of var. = 9.3%). The coefficient of variation within each subject was between 6.8 % and 17.2 % with the 5 mm probe, and between 3.1 % and 6.4 % with the 10 mm probe. Fig. 1 shows the results for the five subjects. The ANOVA led to the results shown in Fig. 2. The portion of variance attributed to the subjects is the ICC [4] which is considered "good" when it is above 80%. In our case the ICC is 41.6% (5 mm probe) and 51.7 % (10 mm probe). However, it should be underlined that in this work we are dealing with normal subjects of similar age and expected similar values of CV. An ICC value near 50% and a fraction of variance of 37 % attributed to days (see Fig. 2) suggest that the technique is indeed sensitive to small day-to-day and subject-to-subject variations.

Discussion and conclusions. We are aware of the fact that, even in isometric conditions, estimates of CV in response to individual twitches (rather than during tetanic contractions) may be affected by the initial shortening velocity of the muscle underneath the electrodes. Indeed, the time between the stimulus and the end of the DD M-waves was in the neighborhood of 25 ms which is a substantial portion of the twitch rise time [2]. However, this biasing factor was present for both probes. Despite this potential source of error, the estimates of CV obtained with this method appear to be more reliable than those obtained as the initial value of short tetanic contractions [3,4]. The 10 mm probe provides CV estimates that are closer to those obtained with the more complex and painful needle technique. In addition, the 10 mm probe allows the detection of individual differences undetectable with the 5 mm probe. For example it shows that subjects 2, 3 and 4 have three CV values significantly different ($p < 0.01$, t-test) and subject 2 has a CV significantly lower than that of any of the other four subjects ($p < 0.001$). The CV estimates obtained with the 10 mm probe are very similar to the values of 3.42 ± 0.33 m/s and 3.81 ± 0.34 m/s reported by Troni et al. respectively in two groups of 25 females and 25 males using the needle technique [5] and by other authors reviewed by Arendt-Nielsen and Zwarts [1].

It is concluded that, on the biceps brachii, the surface technique provides CV values that are comparable with those obtained by means of needles, and that a 10 mm probe is preferable to a 5mm probe.

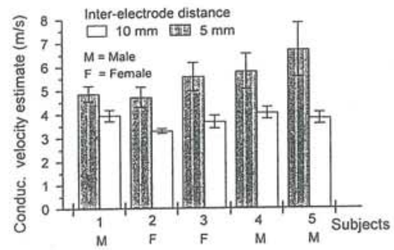

Fig. 1. Mean and standard deviation of the 20 CV estimates obtained for each subject and each probe.

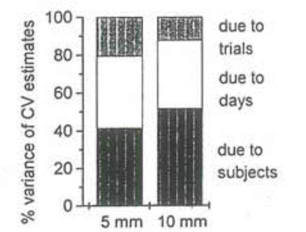

Fig. 2. Distribution of variance due to trials, days and subjects. Total variance is 1.1 $(m/s)^2$ for the 5mm probe and 0.11 $(m/s)^2$ for the 10 mm probe.

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SUPERNORMAL CONDUCTION VELOCITY OF MUSCLE AND THE WARM-UP EFFECT. GENDER DIFFERENCES.

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INTRODUCTION

We studied muscle fatigue and recovery in men after isometric maximal voluntary contraction (MVC) lasting one min. In a previous experiment we found during recovery, after a subnormal period, an overshoot of the MFCV and the median frequency of the power spectrum, reaching a steady state at supernormal values after 10-12 min (4).We hypothesized that these changes form part of the 'warm-up' effect. However, it is unclear if such an increase can be found in both sexes, and whether there is any functional significance. To answer these questions we performed additional experiments in biceps brachii muscle in men and women, in which we investigated variations in MFCV, force and EMG parameters during 1 min. MVC and subsequent recovery for the following 15 min.

METHODS

The experiments were performed on 21 healthy females (ages 29-40 years, mean 33.6) and 15 healthy males (ages 29-40 years, mean 33.6) who had given their informed consent. The experiments were performed on the left biceps brachii muscle as described previously (4). The MFCV, the median frequency (Fmed) and the mean integrated value of the EMG (IEMG) were calculated. The skin temperature and the circumference of the upper arm were simultaneously measured before the exercise and during recovery. All measurements were performed during short-lasting (2 seconds) contractions. The MVC was determined twice and the highest value was accepted as 100% MVC. The EMG was measured at the beginning of the experiment and after the recovery period in duplicate both at 20-30-50 75 and at 100% MVC. The exercise part of the experiment consisted of 1 minute MVC. Recovery was studied at 20% MVC after 1, 2, 4, 6, 8, 10, 12 and 15 minutes. Measurements were performed at every 10 s throughout the fatigue phase and at each contraction during recovery.

Table 1 Mean values (SEM) of force, MFCV and Fmed before fatigue and after 15 min. recovery at 20% and 100% MVC for men and women.

		20% MVC				100% MVC			
		before	after	change	P*	before	after	change	P*
Force (N)	M					205	194	-5%	0.15
	V					106	113	7%	0.24
MFCV (m/s)	M	4.10	4.44	8%	<0.01	4.19	4.66	11%	<0.01
	V	4.11	4.53	10%	<0.01	4.25	4.47	5%	0.02
Fmed (Hz)	M	103	114	11%	<0.01	90	99	10%	<0.01
	V	88	100	14%	<0.01	86	87	1%	0.44

* *Student's t-test (paired, 2-tailed)*

RESULTS

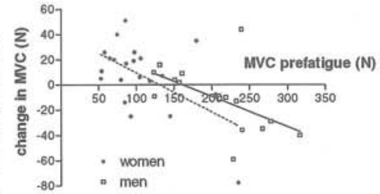

Fig. 1. Change in maximal force after 15 min. recovery (linear regression analysis).Solid line: men, dotted line: women.

The mean MFCV, Fmed and force decreased during 1 min. MVC most pronounced in men. During recovery the MFCV reached supernormal values, the relative increase for men and women was equal. The increased MFCV persisted during the remainder of the observation time in both sexes. After 15 min. recovery significantly higher values for MFCV and Fmed were found in men as well as in women. Force values in men were decreased. In contrast, a (non-significant) increase of force was found in women, see table 1. The rise in temperature and circumference was most pronounced in men. When the change in force was plotted against the initial MVC a significant inverse relation in both sexes was found; a low MVC corresponds with an increase in force and a high initial force with a decrease after 1 min. MVC and 15 min. recovery (fig. 1).

DISCUSSION

During 1 min. MVC both sexes showed a significant decrease of force, MFCV and Fmed, which was most pronounced in men. This could be an effect of the higher initial force of men, resulting in a higher metabolic demand. In combination with the higher intramuscular pressure, obstructing the blood supply, both would result in a higher lactate production. A major result of the study was the finding of a supernormal MFCV during recovery in women. In a previous study we found a supernormal MFCV during the recovery period after 1 min. MVC in men (4). The increase in MFCV postexercise had been related to a combination of muscle fiber swelling, temperature increase and altered membrane properties.The maximal force after 15 minutes recovery time changed inversely proportional to the initial maximal force, an effect present in both sexes. Previously we found an increase of force and MFCV after less strenuous muscle activity (3) possibly as a component of the warm-up effect. This increase in muscular performance is mainly attributed to an increase in body temperature. However, passive heating of the body is much less beneficial than exercise (1), which suggests that additional effects play a role. However, it is not likely that MFCV changes can account for an increase in force, since in hypokalemic periodic paralysis pathologically low MFCV values are combined with a normal force (2). An alternative explanation could be a conformational change due to the fiber swelling, resulting in beneficial mechanical properties. The decrease of force in case of high initial force could be due to opposing effects, such as membrane damage or metabolic depletion.

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MODEL BASED INTERPRETATION OF ELECTRICALLY ELICITED EMG SIGNALS

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Abstract. Computer simulations and experimental data from an electrode array placed on a human biceps brachii muscle demonstrate a non-uniform decrement of muscle fiber conduction velocity during sustained, electrically elicited contractions. The technique is promising for the non-invasive characterization of superficial muscles and for the estimation of muscle fiber conduction velocity (CV) distribution.

Introduction. It is well known that electrically elicited EMG signals (M-waves) undergo modifications during sustained tetanic contractions [1]. Such modifications consist mainly of a progressive widening of the M-wave and often a change in waveshape that is further affected by the electrode location along the muscle [2]. Spectral analysis has been the method most widely applied to describing such changes to the signal. The method, however, does not provide much insight into the underlying physiological mechanisms. We are proposing an alternative approach based on computer simulation of a set of signals detected with a linear array of surface electrodes.

Methods. A mathematical model has been developed that represents the surface EMG signal as a summation of contributions from the single fibers of individual motor units (MUs) [2]. Each MU has parallel fibers scattered within a cylindrical volume of specified radius in an anisotropic medium. Two depolarized zones move from the neuromuscular junction (NMJ) to the tendon endings. Each zone is modeled as a current tripole generated at the NMJ and extinguished at the tendon endings. NMJs and termination zones are uniformly scattered within regions of specified width. Average fiber length to the right and left of the NMJ and the CV are also specified. The signals produced by MUs with different geometries and CVs are superimposed. Monopolar signals are computed in equally spaced locations on the surface of the muscle. Bipolar (SD) and double differential (DD) signals are then computed from the monopolar signals [2].

A 16 channel EMG amplifier with artifact suppression circuitry was used together with a custom stimulator [3] to process the signals detected by a linear electrode array (12 contacts with interelectrode spacing = 5 mm). The array was placed distal to the motor point with the first electrode in close proximity to the motor point. Contractions were elicited by stimulation of the motor point of the biceps brachii muscle of a normal subject for 10 s. Stimulation amplitude was adjusted to elicit a maximal M-wave at a frequency of 28 pps and a pulse width of 0.2 ms. The stimulation frequency was selected to cancel 60 Hz interference by averaging the responses detected during each second.

Results. Simulated and real single and double differential signals, recorded during the 2nd and 10th second of stimulation, are shown in Fig 1. Signals were averaged and amplitude normalized.

Simulation of signals corresponding to the beginning of the contractions are shown in Fig. 1 a and b. The parameters used in the simulation are indicated in Table 1. Modification to the signal waveform following an 8 s contraction are shown in Fig. 1 c and d. At least three MUs (or groups of MUs) with two locations of innervation zones were necessary to simulate the signals. Notice the wider M-wave in the SD signals, the appearance of two peaks in the most distal DD signals, and the fact that some fibers *increased* their CV while others *decreased*. The simulations demonstrate that non-uniform changes of CV are sufficient to explain the observed changes of shape of the SD and DD M-waves. They also demonstrate that this technique may provide a means of estimating the distribution of muscle fiber CV values.

Conclusions. The proposed model-based approach may provide a useful and valid tool for characterizing and interpreting surface EMG signals detected from electrode arrays. It may also serve as a teaching tool for understanding the biophysical mechanisms underlying EMG generation and detection. The technique is currently being studied for the possibility of evaluating single motor unit action potentials during low level voluntary contractions.

Acknowlegement. The authors are grateful to P. Lovisolo, F. Figini, M. Knaflitz and A. Rodriguez for their contributions to the development of the hardware. The model was developed by E. Avignone, L. Lo Conte, P.

Guglielminotti. S. Roatta is now with the Department of Human Physiology of the University of Torino. This work was supported by NATO grant CRG 941101 and by grants from Camera di Commercio di Torino, Italy.

Table 1. Parameters of the simulation. N = number of fibers in the MU, X_i and Z_i = x and z coordinates of the center of the innervation zone, h and R = depth and radius of the MU, W_i = width of the innervation zone, Lr and L_l = length of the fibers to the right and left of the NMJ, W_{tr} and W_{tl} = width of the right and left terminal zones, CV_2 and CV_{10} = CV at the 2nd and 10th second of contraction. Fibers are parallel to the z axis. Balanced tripoles, with 12 mm length and symmetry = 0.33 [2].

	N	X_i mm	Z_i mm	h mm	R mm	W_i mm	L_r mm	W_{tr} mm	L_l mm	W_{tl} mm	CV_2 m/s	CV_{10} m/s
MU 1	30	0	14.5	12	10	3	100	40	40	30	4.5	5.4
MU 2	40	0	12.5	12	10	3	100	40	40	30	4.3	3.9
MU 3	55	0	-18.5	17	10	15	100	40	60	60	4.9	4.9

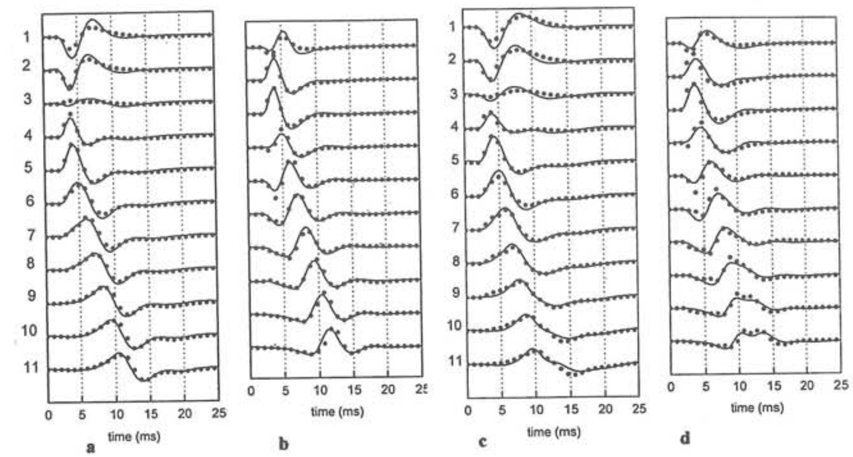

Fig. 1. Experimental (dots) and simulated (solid lines) electrically evoked EMG signals obtained during supramaximal stimulation (20 Hz, 0.2 ms) of the main motor point of the biceps brachii of a normal subject for 10 s. Detection is with a linear array of 12 contacts spaced by 5 mm and placed distally with respect to the motor point. Parameters of the motor units are defined in Table 1.
a and b: single and double differential signals averaged during the 2nd second of contraction.
c and d: single and double differential signals averaged during the 10th second of contraction.

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Computer method of EMG analysis.

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INTRODUCTION

A mechanogram is generally used to register the physiological tremor. A basic limitation of this method is that only the combined activity of different groups of muscles is registered but individual characteristics of a separate muscle are not taken into account. While the main frequency of human physiological tremor is from 0 to 15 Hz., the EMG signal frequency is located from 50 to 200 Hz. The envelope of EMG (EEMG) spectrum is used to get a signal checking with physiological tremor.

In Institute of Control Sciences the new computer method of EEMG analysis has been developed to study the physiological and pathological tremor. The computer analysis of signals allows to reveal objective statistically reliable quantitative characteristics of main EEMG parameters. The computer method of signal analysis permits to computerize science and clinic investigations.

METHODS

EMG signals are picked up by surface electrodes in standard method. It is possible to analyze up to 8 signals simultaneously. EMG signals are transmitted through analog- to-digital converter into a computer where they are detected and filtered with the pass bahd from 0 to 15-40Hz. Then the invelope EMG (EEMG) is formed. After that EEMG power spectra are constructed by the Fourier transform algorithm. The program allows to get 24 spectra of EEMG of one-minute EMG realizations. The power and frequency parameters which the researcher needs can be distinguished on the spectrum. For example: peak frequency, power, corresponding to this peak, square or density of selected spectrum rang. In addition the program permits to plot EEMG integral spectrum for each muscle under study. The whole matrix of parameters is stored and submitted to statistical processing. Accoding to the problem an envestigator may choose desired parameters and get their statistical characteristics. Statictics lets us get mean values and dispersions, plot probability distribution and obtain correlation coefficients which define the interrelation of parameters as inside EEMG spectrum so between EEMG spectre of varios muscles.

The analysis of statistical data permits to select the most important diagnostic features and get their quantitative values. This enables the diagnostics of various illness to have been computerized.

RESULTS

Computer method of analysis of EEMG is applied in neurological and neuro-surgical clinics. The diagnostic symptoms deduced from the method have been used in computerized system of diagnostics of Parkinson's disease clinical symptoms. The system allows to get symptom for each individual muscle under study. The complete symptoms received for each muscle let us determine the various forms of Parkinson's disease. This method can be used as for individualized selection of medicine so for evaluation the effectiveness of the treatment. In surgical treatment

on Parkinson's disease it allows is to estimate the results. Another application of the method is determination of influence of various physical factors on human beings. Objective characteristic features of human reaction on electromagnetic fields, on ultrasonic fields and display radiation are obtained. The investigation of the Chernobyl's accident consequences and service staff of a number of NPS allowed to get the sings of human reaction under small doses of radiation.

CONCLUSION

The computer method of EEMG analysis can be used for an experimental investigation of problems in statistics. In neurological clinic an application of the method makes it possible for a doctor to carry out diagnostics of disease and keep track of symtomatology dynamics more objectively using quantitative evaluation. Individual selection of medicine can be realized following the method. This method can be also used in orthopedics, in sports and cosmic medicine.

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MUSCLE STRENGTH AND EMG ACTIVITY OF THIGH MUSCLES IN STATIC LEG EXTENSION AND KNEE EXTENSION

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INTRODUCTION

The motor system in a bi-articular movement is not as simple as that for mono-articular activity. Torque production and muscle activity during knee extension, as an example of single joint movement, have been reported by many investigators. However, there are few studies that focus on those components of leg extension as a bi-articular movement.

This study investigated the torque-angle relationship and EMG-angle relationship in static leg extension and knee extension, and examined the correlation between these movements.

METHODS

Subjects

A total of 16 healthy volunteers, 6 males and 10 females, with a mean age of 24.6 years old, participated in this study. None of the subjects had a history of neurological or orthopedic disorders in their lower extremities or back.

Measurements of muscle strength

Isometric leg extension force was measured using Leg Power (TAKEI Co. Japan) at knee flexion angles of 15, 30, 45, 60, 75, 90 degrees (0° = fully extended leg) in the long sitting position. The Rehamate (Kawasaki Heavy Industries Ltd. Japan), which is a computerized strength testing apparatus, was used to obtain isometric knee extension and flexion torque at the same angles in the sitting position. At each angle and position, the subjects performed maximal leg extension, knee extension and knee flexion with one leg for 3 seconds. They were given a recovery period of 3 min between efforts.

Measurements of muscle activities

EMG activities in the vastus medialis (VM), rectus femoris (RF), semimenbranosus (SM) were recorded by a surface electromyograph. Miniature silver-silver chloride electrodes, measuring 8mm in diameter were attached to the subject's skin in line with the muscle fibers. A constant distance of 20mm was maintained between the centers of the electrodes. By bipolar recording using these electrodes, rectified filtered electromyography (RFEMG) was obtained. Skin resistance between electrodes was reduced to less than 10 kiloohms. From the RFEMG obtained, we calculated the mean RFEMG value of each muscle for 3 seconds during each movement. The mean RFEMG data of VM and RF were normalized to a percentage of the RFEMG voltage produced during maximal isometric knee extension at a knee flexion angle of 90 degrees (%RFEMG). The data from the semimenbranosus were normalized during maximal isometric knee flexion at a same angle.

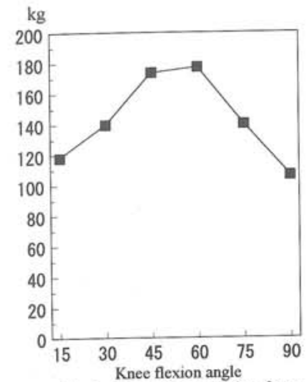

Fig.1. Mean leg extension force

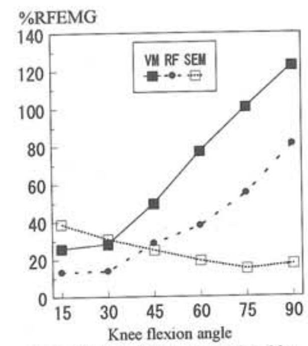

Fig.2. %RFEMG-angle relationship

Table.1. Correlations between leg extension force and knee extension torque

Angle	15	30	45	60	75	90
r	0.33	0.45	0.52	0.51	0.85	0.81
			*	*	**	**

*p<0.05 **<0.01

RESULTS

Maximal leg extension force at various knee angles was presented in Figure 1. The torque curves demonstrated a maximal peak value at knee flexion angles of 45° - 60° in static leg extension and of 60°-90° in knee extension. In static leg extension, maximal peak activity of the knee extensor was produced at a large knee angle, while that of the knee flexor was produced at a small knee angle (Fig.2). The %RFEMG of VM was higher at all knee flexion angles than that of RF. Table 1 shows correlations between leg extension force and knee extension torque at each knee flexion angle. There was a significant correlation at large knee angles, but not at small knee angles (15°, 30°).

DISCUSSION

The force-angle and EMG-angle relationship of leg extension were different from those of knee extension. In leg extension %RFEMG of RF decreased at a small knee angle while that of SM increased. Similar activity patterning has been observed by Eloranta [1].

At a small knee angle, there was no significant correlation between leg extension and knee extension. This result indicates that we should consider not only knee extension exercise but also leg extension exercise.

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Topography of Spectral Surface EMG Parameters in Patients with Centrally Caused Muscle Hypotonia

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Introduction

In patients with centrally caused muscular hypotonia (as a side-effect of stereotactic treatment of tremor syndromes) previous investigations demonstrated that during isometric muscle contraction the muscular resistance to stretching is initially decreased. It is supposed that the reason is a change of the innervation mode during tonic activity. It should be caused by a shift in the population of activated motor units from a more tonic to a rather phasic behaviour. In an EMG-mapping analysis these results could be confirmed [3]. - In the present study during a slight isometric muscle contraction (loading: 2.5%, 5%, and 10% of maximal voluntary contraction/MVC) surface electromyograms of right- and left-sided forearm flexors (especially the M. biceps brachii, monopolar recording) in eleven patients with centrally evoked muscular hypotonia (after stereotactic lesion in the area of Nucl. ventrolateralis thalami) and nineteen controls were analysed in relation to the topography of spectral surface EMG parameters above the biceps muscle and to their frequency properties. Eight of the eleven patients were reinvestigations of patients of the first study. Thus, one further main aim of this study was the confirmation of the topographic surface EMG results of the first investigations.

Methods

The myoelectrical activity of investigated forearm flexors was monopolarly recorded by a 16 channel surface EMG technique (sampling rate: 1000 Hz). On the skin above the muscles, the 16 electrodes were distributed adequate with respect to the morphological and functional conditions of analyzed area (Fig. 2). The reference electrode was localized on the forehead. Only EMG samples without artefacts were used to calculate the mean periodogram via Fast Fourier Transformation (FFT). On the basis of these parameters, the spectral EMG power of the frequency range 10-499 Hz and further bands (e.g. 10-47 Hz, 53-99 Hz etc.) were computed. Afterwards, the spectral EMG parameters between the 16 electrode positions were estimated by linear interpolation (4-nearest neighbours algorithm). The obtained matrix of spectral EMG parameters was fitted on a grey-tone scale with 10 intervals. In this way, myoelectrical activation processes can be demonstrated by a two-dimensional distribution of the spectral EMG power (EMG-map / details in [4,5]). Beside the monopolar EMG mapping investigation a bipolar surface EMG polygraphy of M. brachioradialis, M. extensor carpi radialis longus, as well as the medial and lateral part of M. triceps brachii was simultaneously performed.

Results

On the basis of the EMG polygraphy, it can be demonstrated that during the used load situation the activation of the biceps muscle distinctly dominates. During a load of 2.5% of MVC (activation of motor units with a more tonic behaviour) on the side influenced by the operation, the spectral EMG power (10-499 Hz) of biceps muscle is decreased, and during a load of 10% of MVC, the spectral EMG power is increased in comparison to the side without operation effect (valid for 85% of the patients, Fig. 1). On the side with the hypotonia, the higher frequency range of EMG signals between 53-299 Hz distinctly increases during all used loads. - In patients on the side without operation effect, the topography of biceps activation processes corresponds to the topography of EMG maps in controls (Fig. 2). On the side with operation effect in 63% of investigated patients, the maximum of activation is settled in the

area above the caput breve of biceps muscle. In comparison to controls this maximum is displaced in proximal direction. But in controls the same area is found, if the load is higher than 10% of MVC (mainly activation of phasic motor units / according to our earlier results).

FIG. 1: Spectral EMG power of several bands of biceps muscle in a patient with typical side-effects after left-sided stereotactic thalamus lesion (mean values of band power of the 16 electrode postions and 7 single trials; load: percent of MVC; left: absolute band power, right: band power in percent of total spectral power (right biceps muscle: side with operation effects)

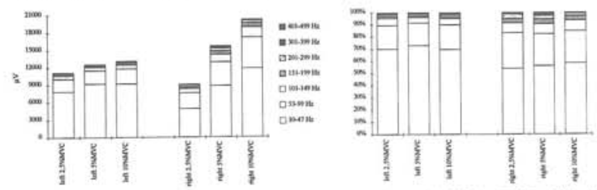
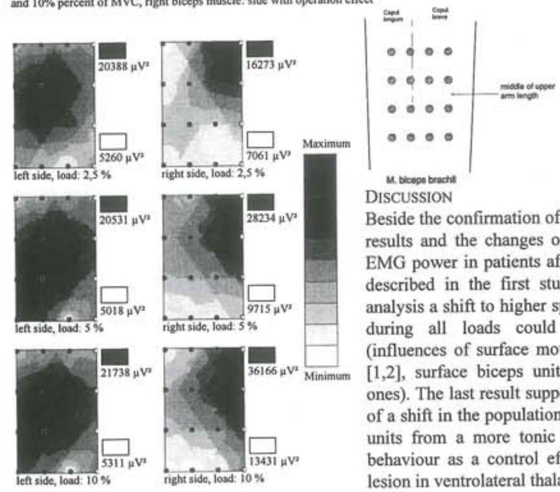


FIG. 2: Array of electrodes (right side of figure); spectral mean value EMG maps (10-499 Hz, n=7 single trials) of biceps muscle in a patient with typical side-effects after left-sided stereotactic thalamus lesion, isometric muscle contraction during different loads: 2.5%, 5% and 10% percent of MVC, right biceps muscle: side with operation effect



Discussion

Beside the confirmation of the EMG mapping results and the changes of the total spectral EMG power in patients after thalamus lesion described in the first study, in the present analysis a shift to higher spectral EMG bands during all loads could be demonstrated (influences of surface motor units / see also [1,2], surface biceps units are more phasic ones). The last result supports the assumption of a shift in the population of activated motor units from a more tonic to a rather phasic behaviour as a control effect caused by the lesion in ventrolateral thalamus of patients.

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MEDIAN FREQUENCY CHANGES OF TIBIALIS ANTERIOR DURING CONTINUOUS AND ARRESTED RAMP ISOMETRIC CONTRACTIONS

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INTRODUCTION

In the literature, both sudden increases in force up to a predetermined level (step contractions) and slow linear increases in force (ramp contractions) have been used to study electromyographic (EMG) power spectrum changes with respect to force. Using both types of contraction, Bilodeau et al. (1991) reported a significant interaction between the types of contraction and the force levels: the median frequency (MF) increased steadily throughout force levels for ramp contractions whereas it increased sharply at low force levels and levelled off or declined thereafter for step contractions. Two factors might influence these behaviors of the MF: the difference in the rate of force generation between ramp and step contractions or the possible difference in the motor command required for the two types of contraction. To test the latter factor, the present study assessed whether EMG signals obtained during arrested and continuous contractions present different MF behaviors throughout force levels.

METHODS

Isometric ramp contractions in dorsiflexion of the ankle performed with the same rate of force generation (20% maximal voluntary contraction (MVC)/s) were either allowed to progress throughout force levels or arrested voluntarily at predetermined force levels. The tests performed were the following: (1) Two initial continuous ramps from 0 to 100% MVC, (2) one arrested ramp at 40, 80, 20, 60, and 10% MVC, (3) two final continuous ramps from 0 to 100% MVC. MF spectrum (FFT, Hamming window, 256 ms windows, 2 kHz sampling rate) and root-mean-square (RMS) analyses were performed on EMG samples obtained from the tibialis anterior (TA). For arrested ramps, the windows were placed immediately after the force signal had stabilized. For continuous ramps, the windows were placed at the targeted force levels. Two-way ANOVAs for repeated measures were used (a) to analyze the differences in both MF and RMS values between force levels and the two types of contraction and (b) to contrast the possible differences in MF between mean initial and mean final continuous ramps across force levels. The latter analysis was performed in order to disclose the presence or absence of fatigue.

RESULTS

The results of the ANOVA revealed an absence of difference between the mean initial and mean final MF values across force levels. This suggests that fatigue did not occur during the experimental session. Figure 1A shows that RMS values increased with force levels in a similar way for both types of contraction. The ANOVA did not disclose any differences in RMS values across tasks. Figure 1B shows that the MF increased similarly across force levels for both types of contraction. There was no significant difference (ANOVA, $p>0.05$) in the MF values between continuous and arrested ramps for a given contraction level.

DISCUSSION

The results of this study suggest that changing the motor command from a continuous contraction to an arrested one while keeping the rate of force generation constant does not affect the behavior of the MF throughout the force levels. In contrast, Bilodeau et al. (1991) used different rates of force generation for their tasks (20% MVC/s for the ramp contractions and up to 100% MVC/s for the step contractions) and found a significant interaction between the types of contraction and the force levels. Other studies using either ramp or step contractions had previously reported different behaviors of the spectral statistics: the mean power frequency (MPF) was shown to increase progressively up to high force levels for ramp contractions (Moritani and Muro, 1987) whereas the MPF increased sharply at low force levels and then levelled off or decreased thereafter for step contractions (Hagberg and Ericson, 1982). Although it is not possible to rule out the influence of a change in the motor command, it seems more likely that the significant interaction observed by Bilodeau et al. (1991) is due to the change in the rate of force generation between the ramp and step contractions. Therefore, the behavior of the MF throughout force levels may possibly be more affected by the speed with which the force level is reached than by the arrest or continuation of the contraction once it is reached.

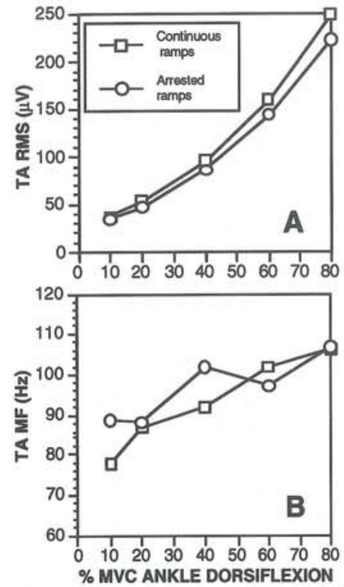

Figure 1. Averaged tibialis anterior (A) RMS (µV) and (B) MF (Hz) values for continuous and arrested isometric ramp contractions with increasing for level.

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EVALUATION OF FUNCTIONAL CHANGES IN BICEPS BRACHII MUSCLE OF HEMIPARETIC SUBJECTS USING EMG POWER SPECTRA

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INTRODUCTION

Weakness is a common clinical finding in hemiparetic subjects. Alterations in motor unit properties play an important role in the manifestation of this weakness. Evidences suggest that in association with general muscle atrophy, hemiparetic subjects present a selective lost of type II fibers [1] and a decrease in agonist motor unit firing rates [2].

Previous studies have shown that spectral parameters of EMG were correlated with conduction velocity [3, 4] and that increases in the median frequency (Md) of the EMG power spectrum during ramp contractions reflected recruitment of larger high conduction velocity type II muscle fibers [5]. With respect to these evidences, it is postulated that Md would be sensitive to muscle fiber atrophy and type II fiber loss in chronic stroke patients.

METHODS

In order to test this hypothesis, Md-Force and EMG-Force relationships in biceps brachii (BB) were studied in 10 hemiparetic subjects (HS) and 10 control subjects (CS) during ramp contractions ranging from 0 to 100% of the maximal voluntary contraction (MVC) in elbow flexion. Torques were recorded with a static dynamometer interfaced with a computer. Subjects were seated in a chair with their trunk secured and their arm flexed at a 90° position.

EMG activity was recorded from the middle portion of BB. Signals were pre-amplified and amplified (CMRR=120 dB, 10-500 Hz band) and digitized on-line with a sampling frequency of 1 kHz. Spectral analysis (Hamming window processing, 256 points, FFT) was performed on digitized BB over 256 ms windows extracted at 10% MVC intervals from 10% to 100% MVC. The EMG rms (μV) and MF (Hz) were calculated for each force level. The slopes of the EMG-Force and Md-Force relationships for each subject were computed from averaged (n=3) trials.

SUBJECTS

HS group consisted of 5 men an 5 women with an mean age and time post-stroke of 48.8 yrs (32-65 yrs) and 32 mths (5-80 mths) respectively. Their average score on the Fugl-Meyer upper limb scale was 42 out 66 and varied from 11 to 60. The CS group consisted of 7 men an 3 women with a mean age of 44.2 yrs (29-61 yrs).

RESULTS

Typical outputs of elbow flexion torque and corresponding raw BB EMG activity for one HS and one CS are shown in Figure1. For both subjects, gradual increases in BB EMG amplitude produced an augmentation in the force output recorded, with the HS producing less torque in elbow flexion than the CS subject.

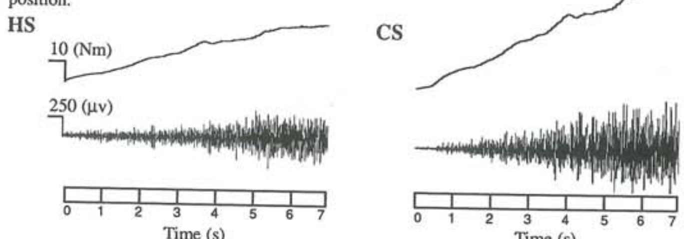

Figure 1. A) Force output (Nm) and B) raw BB EMG during a maximal ramp contraction in elbow flexion for one hemiparetic and one control subject.

Overall, mean voluntary elbow flexion torques were significantly lower (t, $p<0.05$) in HS (19.41 Nm) than in CS (47.4 Nm). While mean BB EMG rms generally increased with increasing elbow flexion torque for both groups (Figure 2A), BB EMG rms in CS was significantly higher than in HS and increased more rapidly across force levels (ANOVA, $p<0.05$). However, the mean slope for EMG-Force was not significantly different (t, $p>0.05$) between HS (0.221) and CS (0.186). The pattern of BB Md increases across force levels showed a similar trend between the 2 groups (Figure 2B) with CS presenting higher BB Md values than HS across force levels. However, this tendency did not reach statistical significance (ANOVA, $p>0.05$).

DISCUSSION

The purpose of this study was to evaluate the sensitivity of the Md of the EMG power spectrum in the assessment of physiological changes in muscle following a stroke. The present results partially support the use of Md in such an undertaking.

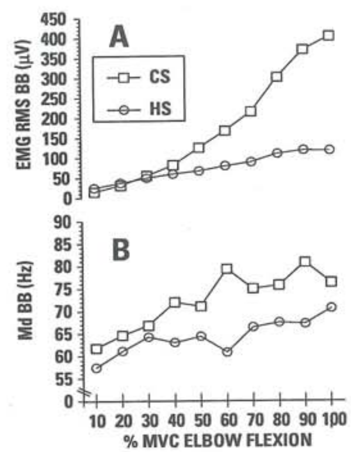

Figure 2. Mean biceps brachii A) EMG rms (μV) B) Md (Hz) during ramp contractions in elbow flexion.

The fact that lower Md values were observed in HS than in CS indicates that the average muscle fiber conduction velocity of the recruited fibers in HS was lower, and thus, the muscle fibers had smaller diameters. This could possibly reflect the selective atrophy of type II muscle fibers observed in HS following a stroke.

However, it appears that depressed descending drive in stroke patients (i.e. reflected in our results by the large differences in BB EMG rms across groups) is the predominant factor in explaining the differences in force output.

Even thought the EMG-force relationship for the BB in HS was similar to that of the CS, the interpretation of such results is limited by the fact that the contribution of other synergists to the external torque in elbow flexion was not assessed.

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INFLUENCE OF HIP ROTATION ON ACTIVATION OF INDIVIDUAL HUMAN THIGH MUSCLES DURING ISOMETRIC KNEE EXTENSOR EXERCISES

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INTRODUCTION

Because of the importance of knee extensor strength and the stabilising functions of muscles acting on the knee, conditioning of the knee musculature is an essential factor in effective sporting performance and rehabilitation. Although the influence of hip flexion on knee extensor strength has been studied before (see Kulig *et al.*, 1984), the effects of hip axial rotation, frequently used to produce specific effects in strength training, have been largely neglected. The purpose of this study was to examine the effects of hip rotation upon thigh muscle activation during isometric knee extensions in well-trained strength athletes.

METHODS

Five well-trained young male strength athletes performed isometric knee extensions seated in an adjustable chair that stabilises the position of the trunk. With a T-bar inserted between the ankles, knee extension torques were recorded. For standardisation purposes, the subjects performed maximal voluntary isometric extensor torques (MVIET) at knee flexion angles of 90, 120 and 160 degrees. Experiments were performed in random order, without hip rotation. The torque was displayed on-line for feedback. Subsequently, similar tasks were performed for three conditions of hip rotation: neutral, maximal medial rotation, and maximal lateral rotation. These exercises were performed at 50% MVIET for the respective knee angles. The level of 50% was chosen to minimise effects of fatigue and allow freedom in strategy to meet the task criteria. All tasks were performed twice in random order. Surface EMG was used to monitor muscle activity. EMG was rectified, integrated, and normalised to the maximal EMG level obtained from all muscles in standardised maximal voluntary contractions. Photographs of the lower limb were taken in the frontal and sagittal plane to determine the actual knee and hip angles under which the trials were performed.

RESULTS

Table 1. ANOVA results (significance levels) for normalised IEMG during the 50% submaximal efforts.

Muscle	Knee Angle	Hip Rotation	Interaction
biceps femoris (long head)	-	0.05	-
adductor magnus	0.05	-	-
tensor fascia latae	-	0.05	-
rectus femoris	-	0.01	0.01
vastus medialis lateralis	0.05	-	-
vastus medialis oblique	0.01	-	-
vastus lateralis	-	-	-
quadriceps	0.05	-	-

As a group, the *quadriceps* activity is only affected by knee angle. The IEMG activity of the *quadriceps* was around 50%, but differed widely from task to task (ranging from 56% to 32%). Considering the muscles of the *quadriceps* in isolation (Fig. 1), it appears that the medial *vasti* muscles are responsible for the EMG dependence on knee angle. The *vastus lateralis* activity does not show any particular relationship with joint angles. The *rectus femoris* activity depended on the hip rotation angle, being higher in external rotation than in the other positions.

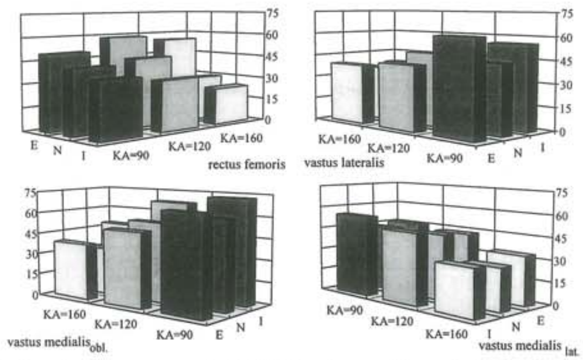

Figure 1. Average normalised EMG for the quadriceps (n=5) for all knee extension tasks (KA is knee angle; I, N, E indicate hip internal, neutral, and external rotation, respectively).

The *adductor magnus* activity depends on knee angle, the *tensor fascia latae* activity is enhanced at internal hip rotation up to 64%, and the *biceps femoris* (long head) showed activity levels ranging from 9% to 28%.

DISCUSSION

The obliquely oriented *quadriceps* muscle group is not affected by hip rotation. This result indicates that the *quadriceps* is active as a single, coordinated unit to exert the properly aligned force vector at the knee. However, the force alignment at the knee of the *rectus femoris*, the only *quadriceps* muscle crossing the hip joint, may be affected by hip rotation. Furthermore, the *rectus femoris* activity is influenced by hip rotation. Thus, apart from local effects at the hip, hip rotation may also affect *rectus femoris* activity because of implications at the knee. Changes in *rectus femoris* activity may have been counterbalanced by the *vasti* group (including the *vastus intermedius* which activity was not recorded).

External and muscle torques that are produced during a knee extension effort need to be in equilibrium at the hip. At all knee and hip positions, the external torque produced at the T-bar creates an extending moment at the hip. Thus, in net terms, hip flexor moments need to be produced to balance the required torque at the T-bar. Apart from the *rectus femoris*, the *adductor magnus* and the *tensor fascia latae* perform hip flexion (Dostal *et al.*, 1986) and thus may have helped in hip stabilisation. The actions of the long head of the *biceps femoris* are antagonistic at both the hip and the knee in the performed task. No explanation was found for its relatively high activity of up to 28%.

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AN IMPROVED STUDY FOR MULTIFUNCTION MYOELECTRIC CONTROL

Bekir Karlık* and Yüksel Özbay**

INTRODUCTION

The analysis of muscle dynamics has been employed for a variety of applications, including prosthesis control, discrimination of movements. Graupe has first shown that muscle contraction could be defined from myoelectric signals (1,2). He has also suggested and successfully implemented the idea of using a time series to identify a control strategy identical to the one created from EMG signals which are obtained from shoulder muscles (3,4). G.N. Saridis and T.P.Goote have done statistical analysis of the EMG signals occuring in the biceps and triceps of an amputee who has had his arm either cut off or parralized (5).

Kelly et. al have succeeded in producing a multifunctional control design based on myoelectric signal classification by using ANN and utilizing Graup's work (6). Later Huddings et. al implemented a multifunctional myoelectric control (7) based on Kelly et. al's work. In the last two works, classification of the four movements has been done after the time series parameters of the signals are produced by the biceps and triceps electrodes at a success rate of 91% recognition by using ANN. In our study classifications of six movements obtained with success rate of 97.6% for 5000 iterations using multi-layer perceptron network.

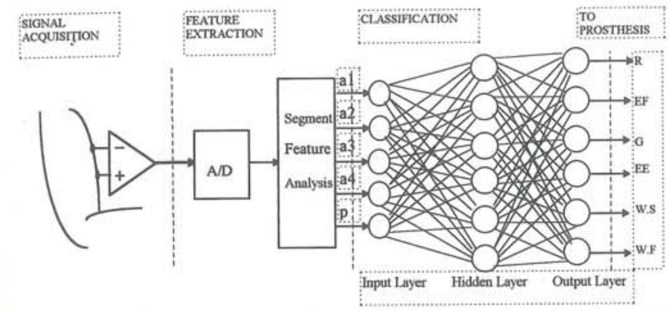
METHODS AND RESULTS

The aim of this work is to evaluate a logical strategy to drive the motors of the prosthesis. This strategy is obtained via EMG signals extracted from the biceps and triceps muscles of the front arm.Control design used to obtain the strategy is presented in Figure 1. As is seen from the figure, control design comprises the phases (steps) of initial signal processing, characterization, classification and decision making. In the first phase EMG signal optained from Ag-AgCl electrodes is amplified and passed through a 1kHz low-pass filter to be sent to the level determining circuit. In the second, A/D conversion (5 kHz), data recording and AR (Autoregressive) processes are carried out. At last stage, decisions are made by classifying the AR presented with ANN.

EMG signals are obtained from a 26 year old healthy man for elbow flexion and extension, wirst supination and flexion and grasp and resting. Each movement has been repeated 6 times and for each time 4800 samples have been taken within 1 second. Some data from the beginning and the end of the data of 4800 were extracted to have the rest linearized. During the experiment ultimate attention has been paid to use the same muscle force each time for the arm movement.

When the EMG signals are processed they are normalized and the dc levels are found. By subtracting the dc level from each sample the signals were made to have a mean value of zero. Samples of 4800 were spit into 12 segments of 400 samples of 80 ms. Each segment has been multiplied with a Blackman type window function to perform windowing. AR parameters of a1, a2, a3, a4 and their signal power were applied to the input of ANN which is of multi-layer perception and feed-forward. Training was done by back propagation algorithm. Recognition

percentage was found to be 96.1% for 3000 iteration assuming gain (learning rate) $\varepsilon = 4$ and momentum coefficient, $\alpha = 0.1$.



R:Resting, EF:Elbow Flexion,G:Grasp,EE:Elbow Extension,W.S:Wrist Supination,W.F:Wrist Flexion

Figure 1. Myoelectric Control of a Multifunction Prosthesis.

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SURFACE EMG SPECTRA OF VOLUNTARY, REFLEX AND SPINAL CORD STIMULATION-ELICITED ACTIVITY

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INTRODUCTION

Frequency domain characteristics of surface electromyographic (sEMG) signals have been well described in past reports. In particular, spectral parameters have been used to characterize muscle fatigue. LeFever and De Luca showed that the 0-40 Hz portion of the spectrum is primarily affected by the motor unit firing rate or inter-pulse interval (IPI), and the part above 40 Hz by the shape of the individual action potentials [LeFever and De Luca, 1976]. Although the overall parameters of a given sEMG signal are dependent upon a number of anatomic and recording factors (e.g., distance of the electrodes to the muscle and the inter-electrode spacing), we assume that the principles described for spectral parameters developed from needle EMG recordings [ibid.] may apply to sEMG data as well.

In voluntary muscle activity, the IPI is generated by activity of the central nervous system, combining brain-originated signals with peripheral feedback in the spinal interneuronal network to ultimately drive the motor neurons from the spinal premotor center. Muscles may also be driven reflexly via electrical stimulation of peripheral nerves and by vibration of muscle spindles. External stimulation of the central nervous system offers another way of activating motor units. The power spectrum may reveal differences not clearly seen in the interference pattern, which also reflects the degree of synchronization of the motor units.

METHODS:

Healthy subjects and subjects with spinal cord injury (SCI) above T10 neurological level were studied. EMG recordings from the triceps surae muscle were made with surface electrodes 3 cm apart along the long axis of and centered over the muscle belly after skin preparation with a gain of 2000 or 10000, and bandpass of 40 to 600 Hz. Subjects were requested to make specific voluntary movements and to execute reinforcement maneuvers, and were stimulated mechanically to elicit tonic and phasic stretch reflexes and plantar flexion reflexes. Several of the SCI subjects were stimulated at 50 pulses/s through electrodes placed in the epidural space at approximately L2 vertebral level in order to elicit activity of the putative central pattern generator (CPG) [Rosenfeld et al., 1995]. Power spectra of the sEMG were estimated in order to demonstrate differences among these sEMG signals.

RESULTS

Surface electromyograms recorded over leg muscles of spinal cord injured persons in voluntary activity showed interference patterns which appeared similar to those of healthy subjects when subjects were able to perform the requested maneuvers. Spectra calculated from such patterns likewise were unremarkably different from those of healthy subjects. sEMG activity could also be evoked in other ways in paralyzed subjects, including the elicitation of spasms, or involuntary activity which typically spread over multiple muscle groups with a repeatable pattern, and various reflexes. The application of strong vibration occasionally elicited a spasm or a sustained response called the tonic vibratory reflex (TVR) which appears superficially to be very similar to responses in voluntarily activated muscles. Power spectra from vibration occasionally show a peak at the vibration frequency, probably due to movement

artefact, and a suggestion of a shift of the spectrum to lower frequencies.

Finally, stimulation of the central nervous system through electrodes placed in the spinal canal just outside the spinal cord was used to excite activity directly or reflexly, which superficially appeared similar to reflex, voluntary or involuntary activity. However, close examination of the activity elicited by spinal cord stimulation (SCS) revealed underlying waveforms which were comprised of repeated compound muscle action potentials (CMAPs) [Lee et al., 1996].

By estimating the spectrum of the sEMG signal, differences in the frequency content became evident among those apparently similar signals. In the example shown in Figure 1, the same muscle had a very different power spectrum in the interval when stimulation was *on* vs. the period just after switching *off* the SCS. The spectrum of the sEMG recorded from the CPG study was dominated by peaks at the frequency of stimulation and harmonics. Presence of the spectral peaks could discriminate the synchronous, CMAPs arising from the SCS from central activation by other mechanisms. Spectra derived from voluntary activity or spasms showed no such peaks.

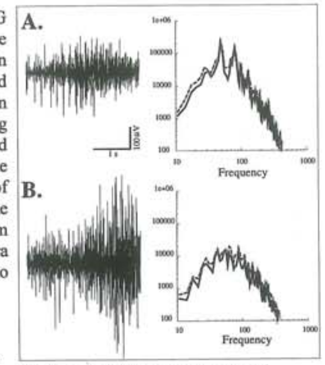

Figure 1. Interference patterns and power spectra from A) sEMG from triceps surae recorded during SCS at L2 and B) sEMG immediately after cessation of SCS. Dashed lines denote 95% confidence limits.

DISCUSSION

The intent of this preliminary study was to examine the characteristics of sEMG signals elicited by different mechanisms in order to examine the degree to which the origins of the signal contributed to differences in the details of the interference patterns so created. The CMAPs elicited by SCS created a sEMG signal which was markedly different from those arising from other mechanisms, with a correspondingly different spectrum. The degree of focusing of activity in those peaks at stimulation frequencies or harmonics offer a simple way of estimating the relative energy in the CMAP versus less organized activity, information which may provide clues as to the origins of the CMAPs themselves. Finally, comparison of spectral properties elicited by "natural" events such as voluntary activation or spasms with those elicited by stimulation may provide further insights into the meaning of the various portions of the spectrum of the full-bandwidth sEMG signal.

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COMPARISON OF METHODS FOR THE COMPUTER DETERMINATION OF EMG ONSET

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INTRODUCTION: The onset of electromyographic activity (EMG) is one of the most common temporal parameters evaluated in kinesiology research. Many methods have been described for the determination of this parameter. In the majority of studies this is done visually. In an attempt to increase the objectivity and repeatability of the evaluation of the EMG trace [1] and to exclude observer bias [4] an increasing number of studies rely on computer based analysis. Several different computer based methods have been developed using a variety of processing techniques and threshold criteria. No studies have evaluated the relative accuracy of the different computer algorithms or the influence of differences in trace characteristics on the accuracy of EMG onset determination. The aims of the study were to evaluate the accuracy of computer based EMG onset determination against visually determined EMG onsets and to evaluate the influence of differences in background EMG activity on the accuracy of the computer based methods.

METHODS: The study involved a random sample of 300 EMG recordings collected from a selection of trunk and limb muscles during a postural task for another study [2]. The EMG traces were separated into two categories on the basis of magnitude of background activity. An experienced examiner visually determined the onset of EMG on the basis of the earliest rise in EMG beyond the steady state [3] on two separate occasions. The EMG onsets were also evaluated by computer using an algorithm involving identification of the point where the mean of a specified number of samples exceeded the baseline activity by a specified number of standard deviations (SD). Twenty-seven different combinations of parameters were evaluated. The degree of smoothing of the EMG was varied using a sixth order elliptical low pass software filter at each of three different values (10, 50, 500 Hz). The other parameters evaluated were the number of samples for which the mean must exceed the threshold (10, 25, 50 ms) and the magnitude of the deviation from the baseline required to indicate the onset (1, 2, 3 SD).

RESULTS: The repeatability of the visually determined EMG onset times between days was found to be high with a coefficient of variation of 0.42 and a root mean square error of 24.79. The correlation between the computer derived EMG onsets and the visually determined EMG onsets was high for all methods ($r=0.999$, $p<0.0001$). Since the slope of the regression line approached 1.0 for all of the methods, the y-intercept value can be considered to accurately depict the degree to which the computer derived values vary from those determined visually. The y-intercepts calculated for the 27 parameter combinations is presented in Table I. The y-intercept for the majority of methods was significantly different from zero. Three combinations of parameters resulted in regression lines with a y-intercept that did not significantly deviate from zero, indicating a high degree of agreement between the computer and visually derived values. When the EMG onsets were evaluated for the low and high background activity groups the y-intercepts were not significantly different from zero for the 25ms/3SD/50Hz and 50ms/1SD/50Hz parameters for both groups. For the low background activity group the 50ms/2SD/50Hz and 10ms/1SD/500Hz parameters also satisfied this criteria as did 25ms/2SD/50Hz for the high background activity group.

DISCUSSION: EMG onsets determined by the computer using the majority of the parameters assessed varied significantly from those identified visually. However, several combinations of parameters did accurately identify the onset of EMG when compared to visually derived values.

Two methods accurately identified the onset of EMG irrespective of the magnitude of background EMG. These parameter combinations are recommended for future use. The 25ms/3SD/50Hz method is similar to that described by DiFabio [1]. Although the methods used in the current study accurately identified the EMG onset, computer based methods must be used with caution as the onset may be influenced by the rate of rise of EMG and interference by movement artefact or an electrocardiogram. For this reason any automated method for identification of EMG onset needs to recognise and reject traces in which the EMG onset is confused by interference and modify the criteria on the basis of the rate of rise of EMG. Currently no method exists with the complex pattern recognition skill required to perform this task. Until this is achieved it is necessary to visually check traces against the computer derived value to ensure that the onset is meaningful.

Table I. Y-intercepts of the linear regression lines for the computer values compared to the visually derived values (SD = standard deviation, LPF = low pass filter cut-off frequency)

10 Hz LPF		50 Hz LPF		500 Hz LPF	
10 ms, 1 SD	-144.87**	10 ms, 1 SD	-127.18**	10 ms, 1 SD	7.81
10 ms, 2 SD	-126.36**	10 ms, 2 SD	-89.01**	10 ms, 2 SD	60.64**
10 ms, 3 SD	-117.41**	10 ms, 3 SD	-61.80**	10 ms, 3 SD	81.39**
25 ms, 1 SD	-131.89**	25 ms, 1 SD	-50.28**	25 ms, 1 SD	119.15**
25 ms, 2 SD	-113.31**	25 ms, 2 SD	-20.6*	25 ms, 2 SD	120.54**
25 ms, 3 SD	-97.38**	25 ms, 3 SD	5.51	25 ms, 3 SD	146.73**
50 ms, 1 SD	-87.21**	50 ms, 1 SD	1.39	50 ms, 1 SD	142.88**
50 ms, 2 SD	-70.32**	50 ms, 2 SD	26.43*	50 ms, 2 SD	149.17**
50 ms, 3 SD	-56.44**	50 ms, 3 SD	42.81**	50 ms, 3 SD	153.49*

** = $p<0.001$ * = $p<0.05$

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IS SPASTICITY OF SCI PATIENTS ALWAYS SPASTICITY? TRANSCRANIAL MAGNETIC STIMULATION AND PERIPHERAL ELECTRIC STIMULATION IN SCI PATIENTS

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INTRODUCTION

With this short paper we only wish to introduce the protocol of a research we have in progress in our departments and to present some anecdotal cases just only to illustrate the aims of this work. *A set of cases and the detailed results will be presented from the platform of the ISEK Congress.*

The same Authors recently presented and published a paper (1, 2), in which they communicated their observations on what is commonly called spasticity in SCI patients, that manifests in different forms; these forms respond to electric stimulation with variant effects, likely having different pathophysiological origins; in addition, it results difficult to consider some clonic contractions, occurring in hypotonic muscles, as spasticity.

The aims of the present research are to investigate some electrophysiological aspects of the paralyzed limbs in SCI patients, particularly in those patients affected with clonic contractions, that seem to respond, in the majority of cases, to the electric stimulation of their so called spastic muscles with a transient reduction of the number, duration and force of the contractions. This could appear difficult to be accepted, in as much as theory and experience suggest that stimulating a spastic muscle results in an increase of spasticity, unless we count upon the local muscle fatigue or the Vedenski effect (neuronal depletion). In addition our study is directed to try to clarify if the transcranial magnetic stimulation (TMS) could be a predictable test of possible improvements and if it is possible to operate a sort of synergy between TMS and therapeutic electric stimulation (TES).

METHODS

SCI patients of A, B, C Frankel grade, without limits of age and period of time from the onset of the impairment, suffering for clonic contractions of low and moderate intensity at the lower limbs enter the protocol. Stimulating electrodes are positioned on the quadriceps muscles and on the peroneal nerves bilaterally; EMG electrodes are placed on both quadriceps muscles and on the soleus muscle (So) of the side most affected with spasms.The stimulating electrodes are connected with a 4 channel electronic stimulator FASE STM 100, supplying 5-s-trains of rectangular impulses of 600 μs at 30 Hz, with a 5-s-pause. TMS is performed with an electromagnetic stimulator ESAOTE BF 00116, delivering a 100 μs pulse with an intensity up to 5500 A and a magnetic field on the coil of 2.5 Tesla by a circular 90 mm coil . The motor potentials are collected and stored by a Nicolette Viking 4, setting 20 ms and 50 μV, high filter 100 KHz and low filter 2 Hz.

The procedure prescribes the following steps: 1) the coil is previously placed on the vertix and moved, while delivering pulses, in order to find the best stimulating point, then a sequence of 15 TMS pulses at 90% of the maximal magnetic power is performed while recording from the quadriceps muscles the eventual evoked Compound Muscle Action Potentials (CMAPs); even if no peripheral response is recorded, the stimulation is supplied in any case with the coil put on the vertix, where 7-8 pulses from each side of the coil are delivered; 2) the H and M responses to stimulation of tibialis nerve at the popliteal fossa with repetitive electrical

impulses of 1 ms at 0.5 Hz from 0 to 199 mV are recorded, then considering the Hmax/Mmax ratio, the H latency and the H threshold; 3) the points 1 and 2 have been previously applied 3-4 times to 4 healthy young adults, 3 males and 1 female, in order to can have control tests; 4) a 15 min session of electrical stimulation follows on the above mentioned muscles and nerves; 5) the TMS is repeated as well as the electric stimulation of the tibial nerve for H and M responses of the SCI patients; 6) a set of 30 sessions of TMS and TES of the quadriceps muscles and the peroneal nerves, twice a day, is performed; 7) examinations are repeated.

RESULTS

The results are here referred to the points 1, 2, 3, 4, 5 of the protocol and to only 3 of the assessed patients by way of example.

Normal subjects. The average amplitude of CMAPs recorded from quadriceps muscles was 1345 μV (SD 1.057) with a latency of 25.4 ms (SD 4.97); Hmax/Mmax = 0.23 (± 0.1); H latency = 28.45 (± 3.96); H threshold = 55.25 V (± 5.58).

SCI patients. Patient No. 1, T12 C. Before TES: no response to TMS from the quadriceps muscles (peripheral denervation); from the tibialis anterior muscle, CMAPs of 117 μV and latency 124 ms on an average; Hmax/Mmax = R.So 0.43, L.So 0.64; H latency = R.So 32.3 ms, L.So 21 ms; H threshold 140 V. After TES of peroneal nerves: Hmax/Mmax = R.So 0.67, L.So 0.75; H latency R.So 31.3 ms, L.So 31.5 ms; H threshold = 92 V.

Patient No. 2, C4 B. Before TES: no response to TMS; Hmax/Mmax = L.So 0.41; H latency = L.So 20.8 ms; H threshold = 83 V. After TES: no response to TMS; Hmax/Mmax = L.So 0.32; H latency = L.So 28.5 ms; H threshold = 94 V.

Patient No. 3, T4 A. Before TES: no response to TMS; Hmax/Mmax = R.So 0.22; H latency = R.So 27.35 ms; H threshold = 48 V. After TES: no response to TMS; Hmax/Mmax = R.So 0.16; H latency = R.So 32.7 ms; H threshold = 45 V.

CONCLUSIONS

Even if no conclusion can be drawn from the above presented examples, by the examination of these and some other patients, we can observe, as a first feeling, that some patients, affected with clonic contractions of moderate intensity (frequency of the spasms, between 7 and 12; duration, from 11 to 30 s; observation from 09.00 a.m. to 03.00 p.m.), <u>do not show electrophysiological parameters peculiar to spasticity</u> and that the same patients are sensitive to the positive influence of TES: see patients No. 2 and 3, that have normal values of H/M ratio and H latency, the first-one further lowering and the second increasing after TES; they have also an increased or unchanged H threshold. Some other patients (see pat. No. 1) seems worsening after one session of TES. As for the completeness of paralysis, one session of TMS combined with TES has not been able to show latent pathways; we have to wait for the final results and for more patients to be tested.

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A MICROCOMPUTER-CONTROLLED ABOVE-KNEE PROSTHESIS, USING SURFACE EMG-SIGNALS OF THE STUMP MUSCLES

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INTRODUCTION

Various above-knee prosthesis (AKP) designs exist nowadays. The most critical component is the knee joint. A natural knee joint is powered by muscle action in two ways. Muscles provide active force by contraction and they apply variable stiffness. Only the latter muscle activity is simulated in an artificial knee joint. It is usually implemented as switching between high resistance to flexion in the stance phase and free rotation in the swing phase of the step.
Recent attempts have been made to more closely duplicate the biological knee joint mechanism, to enhance the appearance, safety and comfort of the AKP user's gait. Since the natural knee joint is acting as a brake for most of the time, a continuously varying braking moment is applied on the artificial knee joint.

DESCRIPTION

Our prosthesis [1-4] is a modular state-of-the-art assembly consisting of a custom-made suction socket, a knee mechanism, a lower leg pylon and a SACH-foot (see Fig. 1). The knee mechanism consists of a magnetic particle brake. Sensors measure the contact pressure under the prosthetic foot, the knee angle and the EMG-signals from the stump muscles. These transducers provide selected information about prosthesis behaviour, user intention and environment interaction to the control unit.

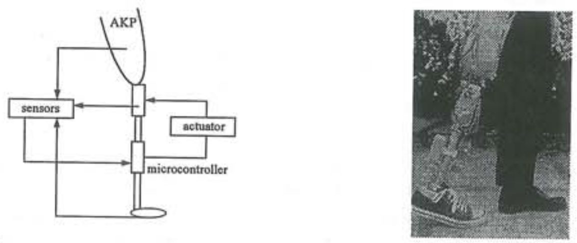

Figure 1 Block scheme and amputee walking with microcomputer controlled AKP

The control algorithm calculates continuously the new, appropriate knee braking moment and applies it to the brake. A finite state approach on two levels has been chosen to tackle the control problem:

- On the highest level, states represent the different *modes* of prosthetic gait. Level walking, ramp ascent or descent, stairs descent, stance, sitting and stumbling are the modes that can currently be distinguished. Also the recognition of transitions between modes belongs to

this mode identification step. Basic pattern recognition techniques [1,3] are applied to derive the indicators for the gait mode from the surface EMG signals at the remaining hip flexors and extensors of the stump.
On the lowest level, each specific mode is subdivided in a finite number of distinct *phases*, based on the detection of *gait anchor points* (like e.g. heel contact, maximum heel rise,...). For each phase, a specific control action is implemented to simulate the normal muscle activity for this phase in this particular gait mode.

RESULTS

Extensive performance and acceptance tests were carried out with one normal (by means of an orthosis system) and three prosthesis users. A training protocol has been elaborated in order to teach the amputees how to use this new prosthetic system in different walking environments on a specially designed gait platform with stairs and a small ramp. Purpose was to get them a feeling for the new feature of a brake controlled flexion of the knee joint and to tune the user dependent control parameters.
Once the training period was completed, AKP users were encouraged to walk freely around. With great success and satisfaction, they explored the new functionalities offered by this prosthesis.

DISCUSSION AND CONCLUSIONS

The new prosthetic system is generally well-accepted by the users, due to the user specific fine-tuning (of control parameters) and to the great amount of new walking possibilities.
Extensive training is mandatory, since all amputees are biased by their everyday prosthesis: they are not readily accustomed to knee stability in flexed position during stance phase.
The technical feasibility of the microcomputer-controlled above-knee prosthesis has been shown clearly. However, user feedback on the control algorithm becomes now very important. More field tests with prosthetic users in daily life are planned in the near future to further optimise the control algorithm.
The social and economic benefits are obvious. This microcomputer-AKP is optimally adapted to the user's needs and is flexible to cope with all circumstances the user encounters. It surely enhances the amputee's quality of life and his/her integration in a normal life.

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PHASE-DEPENDENT MODULATION OF REFLEX RESPONSES DURING REPETITIVE MANIPULATIVE TASKS RELATED TO HANDWRITING

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INTRODUCTION

Interaction between central control and peripheral regulation of rhythmic motor behaviour has been studied in humans and animals by analysing phase-dependent modulation of reflexes (Lennard & Hermanson, 1985; Dietz, 1992). Thus, for example, stretch reflexes or H-reflexes vary in amplitude or gain during the step cycle in walking and running (e.g. Capaday & Stein, 1987). Handwriting and other rhythmic manipulations of the fingers must similarly involve dynamic integration of sensory feedback with central motor programmes. However, despite many studies of reflexes evoked in human hand muscles by either electrical or mechanical stimulation under steady state conditions (e.g. Doemges & Rack, 1992), reflexes elicited during rhythmic finger movements have not hitherto been investigated. The aim of this study, therefore, is to examine the phase dependency of reflex responses of four hand muscles to mechanical perturbation during three repetitive tasks for the fingers related to writing.

METHODS

Twelve healthy right-handed volunteers (7 female and 5 male) participated in this study, which was approved by the local medical ethics committee. Subjects performed a standard protocol incorporating three rhythmic tasks, following a 'metronome' pulse at 0.5, 1 or 2 Hz for 3 minutes in each task whilst monitoring on an oscilloscope the force exerted by the right index finger. The 3 tasks were: (1) abducting/adducting the index finger, almost isometrically between two force limits, with the lateral side of the PIP joint pressing against a 12 mm disc mounted on a force transducer in series with a stiff electro-mechanical prodder (vibrator); (2) pressing up and down with the tip of the index finger on the prodder, keeping within the same force range; (3) squeezing with a dynamic tripod grip a specially constructed pen with strain gauges on three sides, so as to vary the force exerted by the index finger cyclically between 10% and 20% of MVC, whilst holding the pen-tip on the prodder with a nearly constant force of about 10% of maximum. Mechanical stimulus pulses 100 ms in duration and ca. 1 mm displacement were delivered via the prodder at pseudorandom intervals between 0.4 and 1.2 s. The changes in resultant force produced by the index finger (or pen-tip) and prodder in each task, and also by the index finger on the pen in task 3, were digitised along with surface EMGs recorded by paired Ag-AgCl electrodes from four hand muscles: flexor pollicis brevis (FPB), extensor pollicis brevis (EPB), flexor digitorum superficialis (FDS) and first dorsal interosseous (1DI). Rectified and filtered EMGs for each muscle and task were averaged over successive movement cycles (Xia & Bush, 1996), taking the beginning of each cycle as the trough of the rhythmic force signal. 'Mean amplitudes' of clear EMG peaks identified in the rectified records as reflex reponses to the mechanical stimuli were computed for selected time windows. To examine whether these responses showed any phase-dependency, each movement cycle was divided into 8 or 16 equal parts, and all responses within the same segment were averaged together.

RESULTS

In each of the three tasks, the mechanical 'prod' stimuli elicited distinct peaks in the EMG recordings from some or all of the four muscles. These reflex EMG responses varied in form and amplitude (absolute or relative to the background EMG level) with the task and muscle, and in most cases also with the phase of the movement cycle. Two examples are illustrated in Fig.1A and B, which show reflex responses of 1DI (A) and FDS (B) during the pen-tip task (task 3), averaged over all 180 movement cycles irrespective of stimulus phase. Clear sharply rising peaks are seen in both muscles, with a consistent difference in latency of response onset (ca. 35 ms for 1DI and 25 ms for FDS). In both muscles the mean amplitudes of these reflex responses varied markedly with the phase of the stimulus within the rhythmic contraction task ('movement') cycle, as shown for FDS in Fig.1C. This can be compared in Fig.1D with the background (control) EMG activity, averaged (± SEM) over 30 undisturbed cycles (divided

into 16 equal phases to facilitate comparison). The reflex amplitude in this example is clearly modulated in a phase-dependent manner, and is not just a reflection of the background EMG activity. Phase-dependent modulation of the responses to prod stimuli was also observed in the other tasks and muscles studied. The extent of this modulation and its relationship to the background EMG level, however, differed between muscles, tasks and subjects. In some cases the correlation between reflex size and EMG level was greater, in others less.

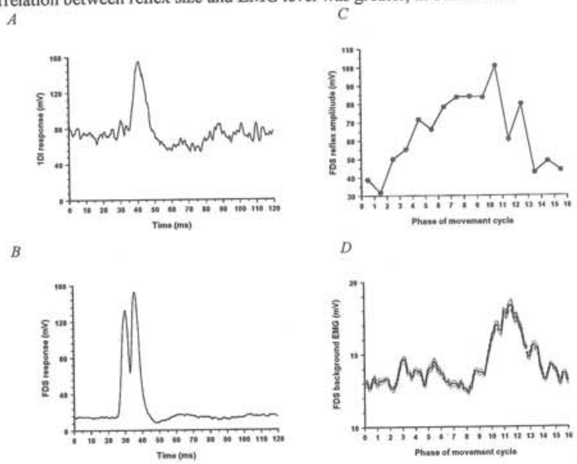

Figure 1. A, B: Average EMG responses of 1DI and FDS to 257 prod stimuli applied via the pen-tip. C: Mean amplitudes of FDS responses for 16 equal phases of the pen grip cycle. D: Average EMG activity of FDS for 30 unstimulated pen grip cycles (task 3).

DISCUSSION

Reflex responses to mechanical stimulation of the index finger during the three rhythmic manipulations studied here have been found to vary in a distinctly phase-dependent manner. As implied also by comparable studies on locomotory rhythms, this might be interpreted as indicating a dynamically fluctuating role for proprioceptive feedback, from muscle and/or joint receptors, in the control of the muscles performing the tasks. This view is reinforced by the observation that the variation in amplitude of the reflexes with stimulus phase is not a simple reflection of the background EMG level. It is also noteworthy that the pattern of this phase-dependent modulation differs between tasks for any one muscle, and between muscles in a given task, consistent with an interplay of sensory feedback and central programming, presumably adapted in characteristic ways to the particular task in hand.

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CHAPTER 6

MUSCLE FATIGUE

MYOELECTRIC MANIFESTATIONS OF MUSCLE FATIGUE: A SIMULATION STUDY

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Abstract. Computer simulations in which muscle fiber conduction velocity (CV) is reduced at different rates in different motor units (MUs) may partly explain why electromyographic (EMG) spectral variables decrease more rapidly than CV during sustained contractions.

Introduction. EMG spectral variables and muscle fiber CV decrease during isometric sustained voluntary or electrically evoked muscle contractions. Theory predicts that the decrement of these variables, normalized with respect to the initial value of each, should be the same [1]. Empirically, however, the normalized decrement of mean or median frequency (MNF or MDF) is often greater than the decrement in CV [1]. EMG signals undergo a spectral compression that results from the progressive widening of the M-wave in the time domain. Such a widening may be caused by a) a decrease in CV, b) an increase in the length of the depolarized zone, or c) an increase in CV dispersion among the active fibers. A decrease in CV implies a relatively equal percentage decrease in MNF and MDF. Increased length of the depolarized zone should imply an increase in signal amplitude, however this is not typically observed. An increase of CV dispersion leads to time dependant changes in MNF and MDF that are not easily predicted. Computer simulations were performed in this study to investigate the effects of CV dispersion on evoked and voluntary EMG signals.

Methods. A model [2] described in Fig. 1 was used to simulate monopolar surface potentials from the superposition of single fiber action potentials propagating in opposite directions from the neuromuscular junction. Each action potential was modelled as a current tripole [2]. For simulating evoked (i.e. stimulated) potentials, three identical MUs with the same geometrical parameters were assumed to fire synchronously. Each of the three MUs contained 100 fibers with half-lengths of 70 mm. Fibers occupied a cylindrical territory 10 mm in diameter oriented parallel to the z axis at a depth h = 10 mm. Innervation and termination zones were each specified as being 10 mm wide. The model included a 4-contact electrode system (interelectrode spacing = 10 mm) positioned on the muscle surface with the center of the array (Z_E) specified for locations at 10, 20, 30, 40, 50 and 60 mm along the z-axis. Simulations were calculated for MUs having CVs that decreased by different rates during the contraction. The first MU had a fixed CV_1 = 5 m/s, the second decreased its CV_2 from 5 m/s to 4 m/s, and the third decreased its CV_3 from 5 m/s to 3 m/s in 10 equal steps. A different combination of CV values, corresponding to the three MUs, was identified for each of the steps resulting in 11 CV triplets .

MDF and MNF were computed from the differential signal obtained from the central pair of contacts on the array (SD2) and CV was estimated from the two double differential signals (DD1 and DD2) [1,2]. The simulation was repeated for the 11 CV triplets and for the 6 electrode locations indicated above. For each location, the estimates of MDF, MNF and CV were normalized with respect to their initial values which corresponds to the initial CV triplet. Plots of the normalized variables are shown in Fig. 2 a, b, and c for the 11 CV triplets. Each curve corresponds to one of the six electrode locations.

The same model parameters were utilized for simulating EMG signals from voluntary contractions except that the three MUs were specified as firing randomly at an average rate of 20 pps with a uniform distribution between 17 pps and 23 pps. Signals of 0.5 s duration were generated from each simulation and used to estimate MDF, MNF and CV. Plots of the normalized variables are shown in Fig. 2 d, e and f for the 11 CV triplets. Each curve corresponds to one of the six electrode locations

Results. For evoked signals, the action potentials are assumed to start at the same time at each neuromuscular junction and propagate in the direction of the tendons. Because the CVs for each MU were specified to change at different rates during the contraction, the sources progressively separate in space and generate a wider potential distribution on the surface of the muscle. This, in turn, leads to decreasing estimates of MDF, MNF and CV. Near the tendon (position 6), and for very different CV values within a triplet, the surface wave becomes polyphasic, the estimated MNF and MDF increase, and the estimate of CV becomes erratic because of the fiber end effect. It is clear from Fig. 2 a, b, c that the percent decrement of MDF and MNF is greater than that of CV. For locations away from the tendon, the estimate of CV calculated from DD1 and DD2, is the average of the three CV values specified in the triplet. Also, the normalized decrement of the calculated CV is the average of the three decrements since the three MUs have the same weight on the surface signal. For voluntary contractions the results are different. The action potentials may occasionally overlap, but, in general, are separated in space

and time. MDF and MNF show a smaller decrement and reflect the fluctuations of signal shape due to random superpositions. MDF is more sensitive than MNF to such fluctuations (Fig. 2 d,e,f).

Conclusions. We have shown, by means of computer simulation, that a non uniform decrease of CV values among different MUs may explain the different rate of decrease often observed between spectral variables (MDF and MNF) and CV. This phenomenon is particularly evident in electrically elicited contractions. These findings should be considered when interpreting myoelectric manifestations of muscle fatigue or comparing EMG results for evoked and voluntary contractions.

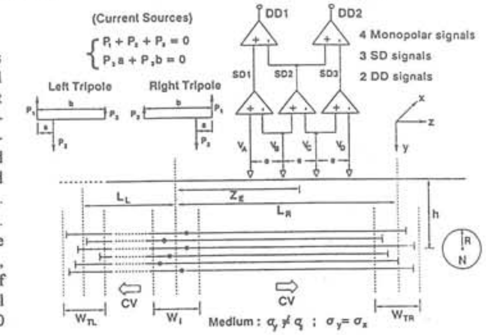

Fig. 1. Model diagram. Each MU has N fibers distributed in a cylindrical territory of radius R. A right and a left tripole originate from each neuromuscular junction (NMJ) and propagate to the fiber termination. NMJs and terminations are uniformely distributed over widths W_I , W_{TR} and W_{TL}. Medium conductivity ratio σ_x/σ_z = 6. Tripole width b = 7 mm, tripole symmetry a/b = 1/3. Each voltage V_A , V_B , V_C and V_D is the superposition of the contributions from individual tripoles. Z_E = 10, 20, 30, 40, 50 and 60 mm (locations 1 to 6 in Fig. 2).

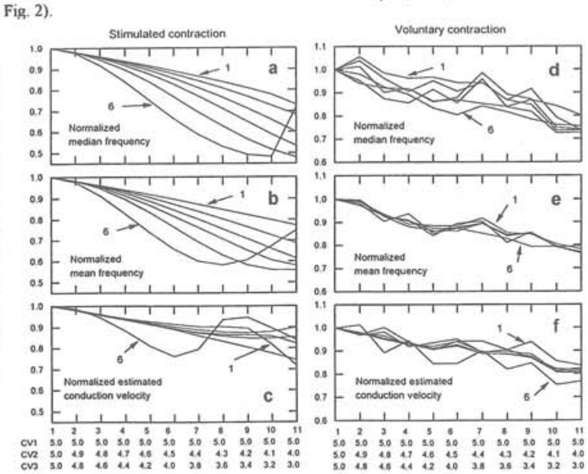

Fig. 2. Simulations of electrically evoked and voluntary EMG signals generated by three identical MUs. The plots show the values of MDF, MNF (computed from SD2) and CV for six electrode positions (Z_E = 10, 20, 30, 40, 50, 60mm) and 11 combinations of the three CV values. In Fig 2d,e,f the three MUs fire at 20 pps ± 15 %.

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EFFECTS OF MUSCLE FATIGUE ON SERIES ELASTIC STIFFNESS AND MECHANICAL EFFICIENCY IN SKELETAL MUSCLE

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INTRODUCTION

In muscle function, the contractile machinery and series elastic structures continuously interact. Cross-bridges generate force, extending the intra- and extracellular series elastic structures, which in their turn allow length change of sarcomeres. These length changes affect force generation. In this process, some of the mechanical energy that is generated is stored in the series elastic structures. This energy may be re-utilised or is lost by internal waste (i.e. stretching sarcomeres). Recently, Ettema (1996*a*) studied the process of storage and release of series elastic energy and mechanical efficiency of skeletal muscle during stretch-shorten cycles. In this study the effects of fatigue on series elastic properties and mechanical efficiency were investigated in stretch-shorten cycles that resemble *in vivo* locomotion contractions. Fatigue effects on stiffness and efficiency may explain changes in kinematics and muscle activity patterns in locomotion that occur as a result of fatigue.

METHODS

Five rat gastrocnemius muscles were used under *in situ* conditions with the animals under general anaesthetics. The muscle-tendon complexes were placed in a muscle-puller. Series elastic stiffness was measured by 210 Hz small amplitude vibrations, superimposed onto a stretch-shortening contraction during which muscle force decreased from force levels above F_o (maximal isometric force) to approximately zero (stimulation 100 Hz, 3 mA, 0.1 ms pulse train on the severed nerve). This information allows separation of contractile and series elastic behaviour (Ettema, 1996*a,b*). Subsequently, 250 sinusoidal stretch-shorten cycles were performed at a rate of 1 Hz. During each cycle, the muscle was stimulated for 300 ms by a 60 Hz pulse train to mimic high-intensity but submaximal stimulation conditions. The amplitude of the stretch-shortening cycle was 2 mm with a small active stretch period. This cycle resulted in high work production and other aspects that resemble *in vivo* locomotion conditions (Ettema, 1996*a*). To determine stiffness in fatigued muscle, the stiffness measurement was repeated within one minute after cessation of the fatiguing cycles. The stiffness measurement and one series of 15 stretch-shortening cycles were repeated after 30 minutes rest to investigate if the muscle condition was returning to the fresh state.
The third to fifth cycle (fresh muscle) and last three cycles (fatigued muscle) of the 250 fatiguing contractions were analysed to determine mechanical efficiency (Ettema, 1996*a*):

$$Eff_{mech} = E^{+}_{mt} / (E^{-}_{mt} + E^{+}_{ce}),$$

mt and *ce* indicating muscle-tendon complex and contractile element, respectively. The '+' sign indicates work release and the '-' sign indicates external work done upon the structure.

RESULTS

Figure 1 shows the force-stiffness relationships for fresh and fatigued muscle. At intermediate to high force levels, series elastic stiffness is significantly increased (Student t-test, $p < 0.01$). Figure 2 shows a typical example of the length-force tracings of the imposed stretch-shorten cycles. The net power, calculated for the duration of activation, amounted 26.5 ± 3.4 mW in fresh muscle and 5.0 ± 0.5 mW in fatigued muscle. Mechanical efficiency in fatigued muscle was significantly higher than in fresh muscle (mean ± s.e.m.: fresh 0.71 ± 0.06, fatigued 0.81 ± 0.05, Student t-test, $p < 0.01$). This was caused by changes in contractile properties (reduced force, power, and relaxation rate), changes in series elastic stiffness, and an altered interaction between the two components. The combination of changes resulted in less elongation of the

contractile machinery during contraction and relaxation, processes that waste energy (Ettema, 1996*a*). The contractions after 30 minutes of recovery showed a partial return towards fresh series elastic and contractile properties.

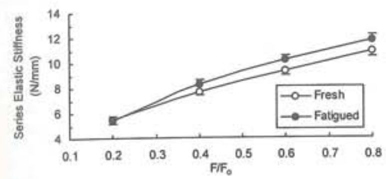

Fig. 1. Series elastic stiffness as a function of relative muscle force in fresh and fatigued muscle. Note that the curve of the fresh muscle extends beyond the graph borders (forces well above F_o).

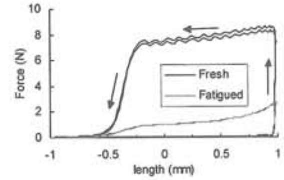

Fig. 2. Length-force curves of fresh and fatigued stretch-shorten cycles. The area enclosed by the loops indicate the net work production.

DISCUSSION

The most likely location of the fatigue induced changes in force-stiffness properties are the elastic links of the active cross-bridges as opposed to changes of tendon and myofilaments. Increased stiffness in fatigued muscle indicates that more cross-bridges are required to generate a particular amount of force compared to fresh muscle (Edman and Lou, 1990). It is probably not caused by the shift in the stretch-shorten cycle phase at which the same force levels were obtained for fresh and fatigued muscle (Ettema and Huijing, 1994). Haan et al. (1996) found no change in overall efficiency (metabolic • mechanical efficiency) as a result of fatigue. The current findings indicate that a reduction of metabolic efficiency may still occur, but are balanced by opposite changes in mechanical efficiency. It is suggested that, in locomotion, the increase in mechanical efficiency during fatigue may work as a compensatory mechanism for the reduced muscle power. In a similar way, kinematics and muscle activity patterns may change during fatigue (e.g. Nummela *et al.*, 1994) and improve mechanical efficiency. Thus, some compensation for the reduced muscle power may be gained. Furthermore, the increased stiffness will reduce the electromechanical delay, which is important for interpretations of EMG data in kinesiology.

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THE MECHANISM BEHIND NORMAL MUSCLE CRAMPS: INSIGHTS FROM SURFACE EMG

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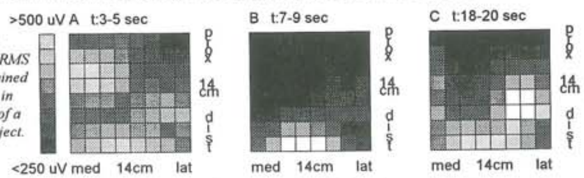
INTRODUCTION

Muscle cramps are sudden non-voluntary painful muscle contractions. They may occur after a trivial movement or after forceful contraction. Muscle cramps have a neurogenic origin, but there are still different views about their precise origin. Two main hypotheses emerge from the results of preliminary research. The first theory proposes cramp to be the result of abnormal excitation of the terminal branches of motor axons (e.g. Layzer, 1994), while the second theory states that cramps result from some form of hyperexcitability or bistability of motor neurones (e.g. Ross & Thomas, 1995). Each theory fails to explain all observations concerning cramp or can provide conclusive evidence. To gain more information about the underlying mechanism of muscle cramps, muscle cramps and maximal voluntary contractions (MVC) are studied electrophysiologically.

METHODS

Eight subjects who were able to voluntarily induce cramps in their calf muscles and considered themselves as healthy participated in this experiment. All subjects tried to exert cramp and performed MVC tests in the calf muscles while EMG recordings were made continuously. For the EMG measurements, 64 electrodes with a diameter of 7 mm were fixed on the skin surface over the calf muscles in a square 8 by 8 matrix with inter-electrode-distances of 2 cm (row and column wise). The recordings were made unipolarly. To study the spatial-temporal development of the different contractions (cramps and MVC), the root mean square (RMS) was obtained for all 64 channels in blocks of 2 seconds. The electrode with the highest average RMS value was considered to be the centre of the contraction. Additionally, the median frequency (MF) of the EMG recorded by this electrode and the mean fibre conduction velocity (CV) around this electrode (in proximal-distal direction) were calculated in blocks of two seconds.

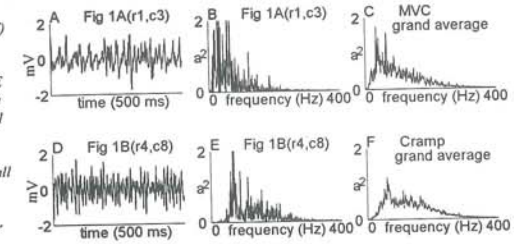
Figure 1. *Typical example of three RMS distributions obtained at different times in the development of a cramp in one subject.*



RESULTS

From all subjects at least 2 cramps could be measured, while only 4 subjects were able to produce MVCs without getting muscle cramp. Figure 1 shows an example of the development of a cramp in terms of a change in RMS distribution over time. First, the subject contracts the muscle voluntarily (1a), then she tries to relax the muscle, but a small area develops a cramp (1b). Thereafter, the cramp spreads over a larger region and its centre changes position (1c). Like in this example, cramp involves often a smaller part of the muscles than MVC and it can change its position rather slowly over the muscle (in cms per second). Compared to MVC, the MF was 10 to 50 Hz higher during cramp (71±20 Hz in MVC vs. 105±19 in cramp), while the CV in both states of contraction was similar.

Figure 2. *EMG (A,D) and EMG power spectra (B,C,E,F) of a muscle in voluntary contraction (A,B,C) and in cramp (D,E,F). 1A,B,D and E are obtained with parts of the same data as used in Figure 1 (row,column). C and F represent the grand average spectrum over all MVC and all cramp recordings, respectively. All spectra are normalised to the total power content.*



DISCUSSION

The power spectrum of an EMG signal and therefore its median frequency is determined both by the shape of the individual action potentials and by the firing process. Ross and Thomas (1995) tried to study these two aspects before and during muscle cramp using a needle electrode with a small recording surface. They observed an increase in firing rate from 12±7 pulses per second (pps) during MVC to 16±14 pps with occasionally firing rates up to normally not occuring values of 80 pps during cramp. Because the shape of the action potentials recorded with the needle electrode during MVC and cramp did not show differences, they concluded that in cramp like in MVC whole motor units are active. An average difference in firing rate of 4 pps cannot explain the shift in MF of 30 Hz in the present study. Looking in more detail at the EMG and its power spectrum during voluntary contraction (Fig 2A, B), MVC (Fig 2C) and cramp (Fig 2D, E, F), it strikes that the signal during cramp lacks low frequencies. Such a low frequency "gap" can hardly be caused by other causes than a firing process with firing rates between 30 and 80 pps. Additionally, the spectrum reveals that the shapes of the individual action potentials have probably higher frequencies in cramp than in MVC. Because the CV did not differ between the two states of contraction, this can only be the case when smaller units than whole motor units are active at individual firings. Therefore, the results of the present study support the theory concerning abnormal excitation of the terminal branches of motor axons in explaining the origin of benign muscle cramps.

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NMR-SPECTROSCOPY IN FATIGUE: TWO DISTINCT pH-LEVELS IN THE TIBIALIS ANTERIOR MUSCLE

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INTRODUCTION

The goal of our research is to measure simultaneously central and peripheral electrophysiological changes and the energy metabolism of a fatiguing muscle[(1)]. In this context, phosphor nuclear magnetic resonance (^{31}P-NMR) spectroscopy was used to measure phosphor metabolites and pH-levels of the tibialis anterior muscle (TA). In this paper we adress pH changes during isometric exercise and recovery.

METHODS

Subjects laid back down with the left foot fixed in an ergometer. This ergometer was specially designed for use in the NMR spectrometer. It consists of non-ferro materials only (PVC, brass and aluminum) and contains a strain gauge bridge. The ankle angle was 90 degrees, the knee angle about 110 degrees. The left leg was supported by "vacuum pillows".

The exercise protocol consisted of two short maximum voluntary ankle dorsiflexions (MVC) of the left feet, followed by at least 5 minutes rest, followed by a sustained ankle dorsiflexion at a fixed level (20, 40 or 60 % MVC) till exhaustion. The subject was provided with visual feedback and was encouraged verbally through an intercom.

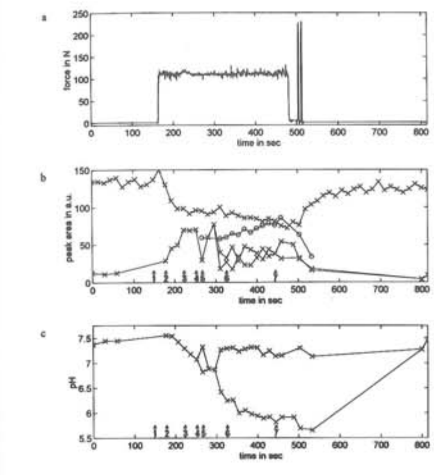
Spectra were acquired using a whole body NMR system (1.5 T, Siemens) and a U-shaped coil, specially designed for the TA. It consists of two turns, one (the smallest) is used for ^{31}P-NMR spectroscopy, the other for making images and proton decoupling. The centre of the coil was placed on the belly of the TA.

The sequence for ^{31}P-NMR contains an acquisition time of 256 ms and an adiabatic pulse (sincos, pulse length 2560 μs, amplitude for spin excitation 90 V). During acquisition a WALTZ-sequence was used for proton decoupling. The repetition time is 7 s. Two scans were averaged which resulted in a time resolution of 15 s. During the whole exercise protocol and during recovery spectra were measured continously.

For the post-processing of the data, VARPRO is used. This time domain fitting procedure calculates the peak areas, line widths and frequencies of selected peaks. Phosphorcreatine (PCr) and inorganic phoshate (Pi) were selected. Intracellulair pH was derived from the frequency shift of the Pi peak to PCr as an internal reference.

RESULTS

Figure 1 shows the results of an isometric ankle dorsiflexion at 40 % MVC until exhaustion, performed by a healthy male volunteer. Figure 2 shows examples of ^{31}P-NMR spectra (corresponding to arrows 1-7 in fig 1b and 1c). During exercise, the peak area of PCr decreases (first rapidly then slowly), the peak area of Pi shows the opposite behaviour, until the Pi peak broadens (4) and transforms in two distinct peaks (5-7). The increase of the sum of the two Pi peaks is still equal to the deficit in the PCr peak. After exhaustion, PCr recovers rapidly. Pi disappeared to an almost undetectable level (fig 1b). The intracellular pH decreases to 5.7 for one Pi peak and to 7.2 for the other. After relaxation both pH's recover to their resting values (fig 1c).



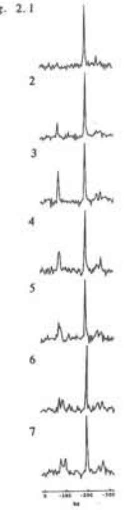


Fig. 1. Fatigue and recovery of an isometric ankle dorsiflexion at 40% MVC:
a. force, after relaxation the subject performs two MVCs
b. peak area of PCr (top line) and Pi (bottom line) and the sum of the two Pi-peak areas (o--o--o line), when the Pi-splitting occurs
c. intracellular pH's

Fig. 2. Raw phosphor spectra (the left peak is Pi, the right peak is PCr). The arrows 1-7 in figure 1b and 1c correspond to spectra 1-7. Note the undetectably small concentration of Pi in case 1, the broadening of the Pi-peak in case 4 (with an equal peakarea as in case 3), followed by the Pi-splitting in cases 5, 6 and 7. Each spectrum is amplitude normalised on the PCr-peak.

DISCUSSION

The results show the development of two distinct intracellular pH levels in the TA during isometric exercise at 40 % MVC. Both "pH-compartments" contribute to the generation of force, because they are derived from the Pi resonances. The low pH-compartment could be the contribution of the white fibres, the high pH-compartment that of the red fibres. At higher contraction levels ($\geq$ 60 % MVC) we did not observe the Pi-splitting. Presumably, the contribution of the red fibres will be undetectably small at such levels.

Ankle dorsiflexion is hardly possible without extension of the toes. The extensor digitorum longus muscle (EDL) is situated dorsolateral of the TA. Therefore, the possibility that the high pH-compartment belongs to the EDL, and the low pH-compartment belongs to the TA may not be discarded yet. Further ^{31}P-NMR spectroscopy measurements may clear up this possible complication.

Our results confirm earlier findings of Park et al.[(2)], obtained by a different exercise protocol, a different NMR set-up and a different muscle. They only showed averaged results of the last 3 minutes of exercises at different loads. The high time resolution enabled us to show the development of the Pi-splitting. This may be relevant in a correlation with EMG measurements.

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EVALUATION OF MUSCULAR FATIGUE USING SUPERIMPOSED M WAVE AND PRECEDING BACKGROUND ACTIVITY

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INTRODUCTION

In rehabilitation medicine, especially in the restoration of muscle activity by means of functional electrical stimulation, there is an acute demand for a method that can be used to predict the patient's functional endurance. This endurance is defined by the fatigability of the muscle for a given task. The fatigability of the muscle is also important in sport science in determining appropriate training tasks and in avoiding overtraining.

We have proposed a muscular fatigue evaluation method based on the relationship between a superimposed M (SM) wave and the preceding background activity [1]. The SM wave is the electrical elicited M wave superimposed on a voluntary contraction. If the preceding background activity and the SM wave reflect localized muscular fatigue in different manners, the relationship would be effective in developing a muscle fatigue index at an arbitrary time. This paper describes the experimental results of the relationship in the frequency domain as a function of the progression of muscular fatigue.

METHODS

Nine male subjects (20 - 23 years old) volunteered for this study and were informed of the experimental procedures and risks associated with the muscle fatiguing efforts. Working with the tibialis anterior muscle, we asked subjects to maintain a 70 % maximal voluntary contraction. We fixed a pair of stimulation pads on the motor point area and obtained the highest SM wave within pain tolerance. The active 4-bar electrode for measuring two-channel myoelectric (ME) signals was pasted on the skin parallel to the muscle fibers. One experimental set for each subject consisted of the first 40 s, the middle 100 s, and the last 40 s trials, separated by resting intervals of 1 min. Total exercise includes three experimental sets with each 5 min resting interval between consecutive sets.

Each epoch including the preceding background activity and the SM wave. The instantaneous frequencies (IFs) were estimated from each averaged SM wave by using the FFT algorithm. The mean power frequency (MPF) of each preceding background activity was estimated before the external trigger signal. Then the averaged MPF was obtained in each frame consisting of five epochs. After obtaining the time-series of MPF and IFs, the MPF-IF distribution was classified by the discriminant analysis using the Mahalanobis distance into two different classes. We studied the correlation coefficients of two different classes and the time-varying behavior of them in the MPF-IF distribution.

RESULTS

Feature vectors of the MPF-IF distribution in each experimental set were divided into the highly correlated class Ω_1 and the almost uncorrelated class Ω_2. Normalized MPF-IF distributions during each middle trial proved that separation of two classes was still meaningful through three experimental sets (Fig. 1). The results mean that the features of the MPF-IF distribution were similar to each other within experimental sets, even if the absolute values were different. The class Ω_1 usually appeared during the first part of a fatiguing contraction, then Ω_2 arose as muscle fatigue progressed (Fig. 2). That is, correlation coefficient $c_{MPF\text{-}IF}$ decreased as muscle fatigue progressed. This feature was meaningfully demonstrated through first to middle trials for IFs at the first peak or the zero crossing. The difference between Ω_1 and Ω_2 became apparent, when we selected the MPF-IF distribution at the specific time that showed the highest correlation coefficient for Ω_1 in each experimental set ($p < 0.05$).

DISCUSSION

The features of fast twitch motor units (FT-MUs) probably produced the highly correlated class Ω_1, because they appeared to contribute steep reductions in the MPF and IFs during the beginning phase of a fatiguing contraction. Both the degenerated muscle activity and the peripheral effect by the supramaximal stimulation on the motor point may have produced the uncorrelated class Ω_2. That is, the supramaximal stimulation seemed to sustain the original shape of the SM wave, even if MUAPs were continuously degenerated in the preceding background activity during the latter phase of muscular fatigue.

In our results the specific time was generally the first peak of the SM wave in the first experimental set. The change in the SM wave at the specific time can be interpreted as reflecting the MUs that can be fatigued and/or easily reactivated near the surface electrodes. Solomonow *et al.* [2] described the change at the first part of the M wave in relation to the recruitment of larger MUs with higher CVs, that is, the recruitment of FT-MUs. Conventional studies have also showed that the conduction velocity (CV) reduction influenced the latter part of the M wave in particular, assuming all the MUAPs had the same CVs. Recent studies, however, have showed that factors other than the CV reduction could be considered to explain the variation in the SM wave. By applying the scaling model to the successive change of the M wave, Merletti *et al.* [3] found that the variation around the latter part of the M wave was not completely attributable to the CV reduction. Because of the above uncertainties and the inadequate separation between Ω_1 and Ω_2, we consider that the second peak of the SM wave is not suitable for presenting the profile of the MPF-IF distribution. Consequently, selecting the specific time of the IF that showed the highest $c_{MPF\text{-}IF}$ among three Ω_1 classes in each experimental set and using $c_{MPF\text{-}IF}$ will be useful to evaluate muscular fatigue at a required time.

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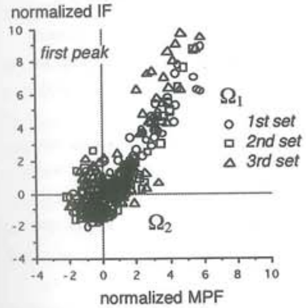

Fig. 1. Normalized MPF-IF distributions.

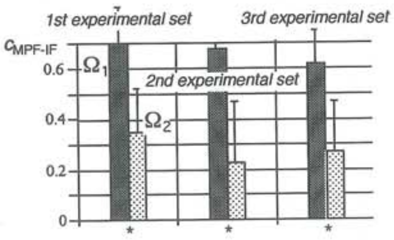

Fig. 2. The occupation ratios of (*mpf, if*) samples for Ω_1 and Ω_2

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PROGRESS OF MUSCLE FATIGUE ASSESSED BY USING EVOKED AND VOLITIONAL MYOELECTRIC POTENTIALS

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INTRODUCTION

Surface electromyography has been used extensively as a measure to evaluate muscle fatigue noninvasively. It has been well known that the power spectrum shifts towards lower frequencies and IEMG increases with progression of muscle fatigue[1]. Although the reduction in the muscle fiber conduction velocity have been reported to explain the frequency shift as peripheral factors, involvement of central mechanism including motor unit recruitment cannot be excluded.

Recently, evaluation of muscle fatigue using evoked EMG has been proposed[2),3)], which may provide possibilities of discrimination between central and peripheral factors in muscle fatigue. The purpose of this study is to examine the mechanism of muscle fatigue progression using evoked and volitional myoelectric potentials, and to identify the pattern of muscle fatigue progression as depending on the contraction task concerned.

METHODS

Seven healthy males aged 20-27 participated in this study. Subject seated on a chair exerted isometric voluntary contractions of the first dorsal interosseous muscle (FDI) at 70% MVC for 1 min, and at 30% MVC for 3 min. The contraction force was measured at the interphalangeal joints of index finger with strain gauge. Simultaneously with the voluntary contractions, the ulnar nerve was stimulated at the wrist with 1Hz square-wave pulses of 0.1msec duration at the supramaximal intensity.

Using a four-bar active electrode, bipolar surface EMGs and supramaximal M-waves were obtained from FDI during the isometric voluntary contractions. The myoelectric signals were fed to a personal computer, and mean power frequency (MPF) and instantaneous frequency (IF) were calculated for the volitional EMG and superimposed M-wave, respectively. The IF was determined at first peak , zero cross, and second peak of the superimposed M-wave.

RESULTS AND DISCUSSION

MPF of the volitional EMG decreased consistently during the isometric contraction in all subjects, indicating development of fatigue in the muscle. With progression of fatigue, waveform of the superimposed M-waves showed a tendency to decrease in amplitude and

increase in duration, while IF of the superimposed M-waves showed a tendency to decrease at each of their selected points (fig.1).

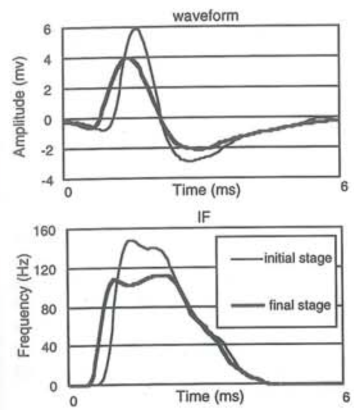

fig. 1 Waveform (top) and IF (bottom) of superimposed M-wave compared between initial and final stage of 70% MVC contraction.

MPF and IF declined greater in high intensity contractions than in low intensity contractions. This difference is considered to be caused by the fact that high intensity contractions are dependent upon activities of FT fibers of which FDI is primarily composed. MPF was highly correlated with IF at the first peak of superimposed M-waves. This high correlation in the early stage of contraction gradually declined with progress of muscle fatigue. Besides, the coefficient of the MPF-IF regression was greater in high intensity contractions than in low intensity contractions. These observations suggest that muscle fatigue progression in the final stage of contraction depends on central as well as peripheral factors.

CONCLUSION

The results obtained suggest the task dependency in the course of muscle fatigue progression, and the possibility of applying the evoked myoelectric potentials to the evaluation of muscle fatigue.

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MUSCLE UNLOADING AND THE EMG SIGNAL

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INTRODUCTION: The assessment of muscle adaptation by surface EMG signal analysis may provide a non-invasive method of monitoring physiological changes occurring in muscle. In this study, an in vitro technique was applied to compare muscle fiber morphology and histochemistry with spectral parameters of the EMG signal in muscles exposed to short- and long-term muscle unloading.

METHODS: A well documented hindlimb unweighting model was used to induce atrophic changes in hindlimb muscles of the rat [1]. Animals in the experimental group (n=12) were hindlimb unweighted for either seven or twenty-one days. Equal numbers of control animals were included. Neuromuscular preparations of the soleus muscle were extracted and placed in an oxygenated isothermal Krebs bath. Preparations were supramaximally stimulated via the nerve at 40 Hz to induce tetanic contractions. Contractions of 2 s and 30 s duration were studied. EMG signals were recorded using an electrode with three detection pins (O.D. 0.5 mm, interelectrode spacing 2.3 mm) located against the muscle belly. Signals were sampled at 2048 Hz and then averaged over 0.25 s epochs before analysis. After the experimental protocol, muscles were frozen for later histochemical analysis. Muscle fiber cross-sectional area (CSA) was measured and fiber type content was determined by myosin ATPase staining.

RESULTS: As shown in the graph in Figure 1, average muscle fiber CSA was significantly decreased when compared to controls at 7 days and 21 days of unloading. Figure 1 also shows a significant increase in the percentage by area of fast fibers after 21 days of unloading when compared to controls. The results of the EMG signal analysis are summarized in Figures 2 and 3. Average initial median frequency (IMF) significantly decreased after 7 days of unloading and was further decreased by 21 days of unloading whereas no significant changes in IMF were observed in the control group (Figure 2). The change in median frequency (ΔMF) for contractions sustained 30 s were not significantly affected by the unloading protocol (Figure 3).

DISCUSSION: Muscle unloading resulted in the atrophy of soleus muscle fibers as indicated by the significant decrease in average fiber CSA for the experimental but not control animals. Additionally, there was an increase in the percentage of fast fibers, due to an increase in the production of fast myosin [1, 2]. The degree of atrophy and change in fiber type percentage was similar to changes reported by others [1, 2]. The decrease in the mean IMF observed with 7 days of unloading most likely resulted from the decrease in muscle fiber conduction velocity in fibers with a smaller mean CSA. At 21 days of unloading the results are not as clearly interpreted. Mean IMF decreased further following continued unloading, as would be expected since there was a continued decrease in the average muscle fiber CSA. However, there was also a significant increase in the % fast fiber area, which according to previous work should have lead to an increased IMF [3]. These conflicting results suggest that the change in myosin type following adaptation may not have included a corresponding change in the type or densities of the Na-K membrane pumps. These pumps are likely to be a more important determinant of EMG signal shape than myosin ATPase which is more closely related to contractile mechanisms [3]. It may be that changes in Na-K pump densities lagged behind myosin-ATPase changes during the time course of the adaptation to loading.

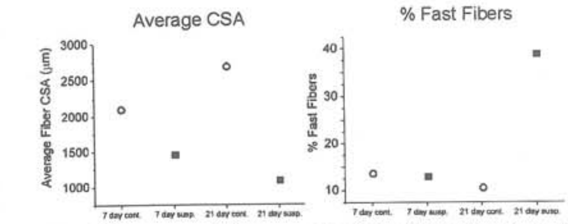

Figure 1: Average muscle fiber cross sectional area (CSA) and percentage of fast fibers by area are shown for control (open circles) and experimental (filled squares) groups at 7 and 21 days.

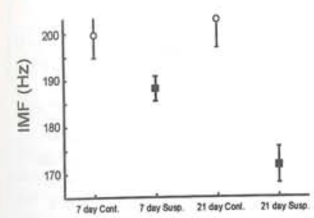

Figure 2: Mean initial median frequency (IMF) for M-waves elicited from control muscles (open circles) and unloaded muscles (filled squares). Standard deviation bars are indicated.

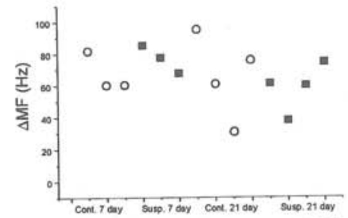

Figure 3: Change in median frequency (ΔMF) following the 30 s contraction is presented for several sample control (open circles) and unloaded (filled squares) muscles. No significant differences were noted between groups.

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ELECTROMYOGRAPHIC SPECTRAL CHANGES CORRELATE WITH SUBJECTIVE ASSESSMENT OF LUMBAR MUSCLE FATIGUE

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INTRODUCTION

Fatigue could be defined as "a failure to maintain a required or expected force" (1). Muscle fatigue could also be present as soon as a muscle contraction starts (2) and measured by mean power (MPF) - or median frequency shift of the EMG power spectrum to lower frequencies (3, 4). As a subjective assessment of fatigue, Borg CR-10 scale (5) has been used (6). The aim of the present study was to correlate objective measurements (EMG) of fatigue to subject's own assessment (Borg's scale) of fatigue and make a comparison between men and women.

METHODS

Forty-three healthy subjects (23 men and 20 women) with informed consent participated and performed a modified Sörensen's test to exhaustion. When the test was concluded, 5-second contractions were performed at 1 min., 2 min., 3 min. and 5 min. rest to test recovery from fatigue. Subjects' self rated fatigue was assessed every 15 s during the modified Sörensen's test and the tests of recovery, using the Borg CR-10 scale (ranging 0-10).

Surface EMG was recorded from the Erector Spinae muscle at L1-L2 and L5-S1 levels bilaterally. A telemetric system was used and the EMG MPF signal was calculated every second and was analysed initially, when the subjects rated Borg as 3, 5 and 7 and at end (Telemyo 16 and MyoSoft 2000 software, Noraxon, USA) and further correlated to subjects' own estimate of fatigue.

Students *t* test and one factor repeated ANOVA were used for comparison between the EMG spectral parameters. Pearson's correlation coefficient were used to measure strength of correlation between the subjective and objective measurements of fatigue. $p < 0.05$ was considered significant.

RESULTS

Mean endurance time of the modified Sörensen's test for men were 378 s (SD 128) and for women 378 s (SD 152). Women showed significantly higher mean power frequencies at end of the contraction than men at all electrode sites. Women had also less decline in MPF slopes but only significant at L5-S1 right side.

Table 1. Decrease in mean power frequency (MPF in Hz) at Borg ratings 3 (moderate fatigue), 5 (strong fatigue) and 7 (very strong fatigue), from initial MPF (n=42).

Electrode sites	MPF at BORG 3, mean (SD)	MPF at BORG 5, mean (SD)	MPF at BORG 7, mean (SD)
L1 right	9.5 (8.2)	18.2 (12.2)	21.8 (13.1)
L1 left	11.5 (9.3)	20.3 (11.3)	25.4 (12.2)
L5 right	11.1 (8.6)	20.3 (11.0)	26.8 (14.9)
L5 left	11.6 (9.3)	19.9 (9.4)	25.7 (10.4)

Significant correlation was present between Borg ratings at 195 s, endurance time and MPF slopes at left and right L5-S1 and left L1-L2. Significant differences were present between the mean power frequencies at Borg ratings of perceived fatigue 3, 5 and 7 for the

whole group (Fig. 1) but no significant difference was found between men and women. Table 1 shows mean power frequencies in Hz at different Borg rating levels.

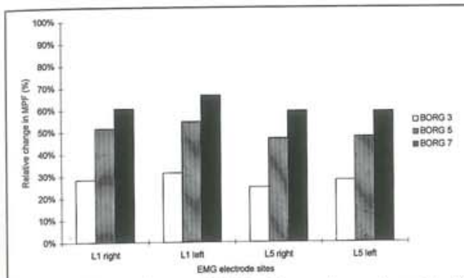

Fig. 1. Relative change in MPF at Borg ratings of perceived fatigue 3, 5 and 7 compared to the total change of MPF.

DISCUSSION

A Borg rating level of strong fatigue (5) must be considered as a biologically significant fatigue since it is experienced by the subject. Thus a 20 Hz decrease in MPF should be considered not only as statistically significant but also as biologically significant. One can discuss the interpretation of MPF and Borg rating level 3. This close relationship could be useful in clinical testing, i.e. to reduce the factor of motivation a person could perform sub-maximal time to Borg rating 5 and a 50% reduction of MPF could be expected.

MPF slopes between men and women differed significant at L5-S1 right and these results shows the same direction as Oddsson et al (7) where significant difference between men and women in median frequency slope as well as holding time to exhaustion have been reported.

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EMG POWER SPECTRAL CHANGES DURING FATIGUE OF THE BACK EXTENSORS: THE INFLUENCE OF MUSCLE FIBRE SIZE AND TYPE DISTRIBUTION

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Introduction

Determination of the rate of decline in median frequency of the EMG power spectrum is a method which is frequently used to objectively monitor and quantify the development of skeletal muscle fatigue. Employing such techniques, a pronounced fatigability of the back extensors of low back pain patients has repeatedly been observed[1,3,4]. This finding has proven somewhat difficult to explain, partly because, whilst the precise myoelectric changes which occur during fatigue are well established, the cause of such manifestations is still unknown. There is evidence that the quality of a muscle — in terms of its fibre size and type distribution — determines its resistance to fatigue and, in this respect, an association between the myoelectric changes observed during fatigue and the muscle's fibre type characteristics might be expected. We sought to investigate this hypothesis, using the erector spinae muscle group.

Methods

Thirty-one healthy volunteers (17 men, 14 women), 18 to 49 years old and with no history of serious low back pain, took part in the study which was approved by the local Medical Ethics Committee. Each subject performed two tests of back extensor fatigability (on separate days), each to the limit of endurance: (i) maintenance of 60% maximum voluntary contraction (MVC) (where MVC was given by the maximum extensor moment that subjects could generate, pulling up on a floor-mounted load cell positioned between their feet), (ii) the Biering-Sørensen test (BS test). Pairs of surface electrodes were attached to the skin overlying the belly of the erector spinae, bilaterally, at the levels of T10 and L3. The raw EMG signal was band-pass filtered between 5 and 300 Hz, amplified, and A to D converted at a sampling rate of 880 Hz. The EMG power spectral density was computed for 1.2 s sampling periods using a Fast Fourier Transform algorithm, from which the median frequency (MF) was calculated. The slope of the linear regression of MF on time (MFgrad, normalised by the intercept; $\%.s^{-1}$) gave the objective measure of fatigability. One week after the exercise tests, two percutaneous erector spinae muscle biopsy samples were obtained under local anaesthesia, from the same sites as described for EMG (left side only), using 6.5 mm Tilley-Henckel punch forceps. Samples were snap-frozen in isopentane then serial sections, 14 µm thick, were cut at -20°C and reacted for myofibrillar adenosine triphosphatase (ATPase) following acid (pH 4.3 and 4.6) and alkali (pH 10.5) preincubations. Each fibre in the section (average 1480) was identified as either type I, IIa, IIb or IIc, based on its staining intensity. Muscle fibre cross-sectional areas (CSA) were quantified on an average 280 fibres per biopsy using a computerised image analysis system. Linear regression analysis was used to examine relationships between EMG and muscle biopsy data (all type II fibres were considered together for these purposes).

Results

The mean fibre CSA, at each erector spinae region, showed a significant correlation with maximum back extensor strength and with body weight ($p<0.05$).

There was no significant relationship between the fibre type distribution (i.e. the percent of a given fibre type, by number) and any of the EMG parameters, in either of the erector spinae regions.

In the thoracic region, the relative area of the muscle occupied by type I fibres (% type I area, which accounts both for the relative size *and* relative distribution of the fibre types), showed a significant relationship with the rate of decline in median frequency (MFgrad) recorded during each of the two fatigue tests ($p<0.05$). The ratio describing the size of the type I fibre relative to that of the type II (type I:II CSA) also correlated with MFgrad, but for performance in the BS test only. In the lumbar region, MFgrad showed a significant dependence on the % type I area, but for the BS test only ($p<0.05$). In each case, a high % type I area and a high type I:II CSA, were associated with a less rapid rate of decline in MF, i.e. a better fatigue resistance of the muscle.

A significant positive relationship was observed between 60% MVC task endurance time and the % type I fibre area in the thoracic, but not lumbar, region ($p<0.05$). For the Biering-Sørensen test, the trend was reversed: % type I fibre area in the lumbar, but not thoracic, region was a significant predictor of task endurance time ($p<0.01$). For each fatigue test, combining the muscle biopsy data from both regions did not improve the ability to predict endurance time.

Discussion

The electromyographic changes recorded in the back muscles during fatigue appear to be related to the underlying muscle fibre type characteristics. The relationship was sometimes limited to a given level of the erector spinae (thoracic or lumbar), depending on the task under consideration. This may indicate that one region took greater responsibility for force generation in that particular posture or during that specific activity. The greater the relative area of the muscle occupied by type I (slow twitch) fibres, the less rapid was the decline in median frequency and (generally) the longer was the contraction sustained. The fibre type distributional area within a muscle is known to influence its ATP turnover rate during isometric contraction: a muscle with a type I fibre predominance displays a greater economy of force maintenance (i.e. use less ATP to maintain the same relative force for the same time), and accumulates inhibitory metabolites at a slower rate, than one in which the type II fibre prevails[2]. If the EMG power spectral shift results from changing metabolic conditions within the muscle, then this may explain the observed association betwen the rate of decline in MF and the muscle fibre type area distribution. Interestingly, the correlation appears to be crucially dependent on the relative *size* of the muscle fibre types, in addition to (and rather than solely) their distribution by number.

Conclusion

The results of the present study confirm the usefulness of electromyography, in reflecting the fibre type characteristics of the muscle in a non-invasive manner, when monitoring changes in function consequent to the development of, or rehabilitation from, skeletal muscle dysfunction.

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RELIABILITY OF SPECTRAL PARAMETERS FOR REPEATED MEASUREMENTS OF BACK MUSCLE FATIGUE

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INTRODUCTION

The reliability of the frequency spectrum parameters during muscle fatigue has been tested by authors using different recording sites, body positions and test protocols. Test-retest within one hour (2,4,5) and test-retest with five days to three months between the tests (1,2,3,6) has been reported for healthy subjects.
The aim of this study was to determine the variability for mean frequency parameters and torque over six repeated measurements in order to get a more extensive knowledge of within-day and day-to-day variations.

METHODS

Eleven backhealthy persons, three men and eight women, performed a 45-second contraction, twice a day, on three different days with 2-13 days in between.
The subjects were tested in un upright sitting position (0°) in a David Back Extension Device. In this device the pelvis is fixed by a rounded pad behind the lordosis and a hip belt. In addition adjustable footplates make the hip flexion more than 90°. The knees were fixed by a special knee fixation pad. The resistance roll was at scapula level and the subject sat in an upright position (0°). A measurement cassette (David Fitness & Medical Ltd) connected to the device displayed the torque in Newtonmeter (Nm). Maximal Volontary Contraction (MVC) was the mean of the two highest values out of three, and the 80% of MVC force was kept by visual feed-back from the display of the measurement cassette. Recovery contractions were also measured and are presently being analysed.
Surface EMG was recorded from the erector spinae at L1 and L5 spinal level bilaterally with a telemetric system (Telemyo 16; Noraxon, USA) using a 1000 Hz sampling frequency and 1000 Hz bandwidth. Mean frequencies (MF) were calculated with fast-Fourier transformation (MyoSoft 2000, Noraxon, USA) and the window was 500 ms. At the beginning (IMFm) and end (FMF) of the 45-second contraction, as well as for the recovery contractions, mean frequency was calculated as the average over four seconds.
Alternatively, for the 45-second contraction, the intercept of the regression line was used as initial frequency (IMFi).
Analysis of variance (ANOVA) with repeated measures were done to reveal within-day and day-to-day variations. Within subject calculations were done comparing within-day and day-to-day tests to define the degree of variations.

RESULTS

For all spectral parameters ANOVA showed no significant within-subject differences on repeated measurements within-day or day-to-day and no systematic variations were found. Total variation (coefficient of variation) for all recording sites over all six measurements ranged for IMFi 4.3-7.1%, IMFm 3.8-6.8% and FMF 5.7-9.0%. Torque measurements varied 8%. For individual variations in spectral parameters see table 1.

DISCUSSION

The results that within-day and day-to-day variation showed no difference in spectral parameters, indicate that the individual variability is more due to biological variation than to the location of the electrodes. Repeated measurements of the slope were variable which has been reported previously (2). The variation in torque might also affect the spectral parameters (3,6).

Table 1. Individual variations of EMG mean frequency parameters (n=11) over all six repeated measurements, within-day and day-to-day. Erector spinae muscle at spinal levels L1 and L5 right and left side. (IMFi=initial mean frequency intercept, IMFm= initial mean frequency average over first four seconds, FMF=final mean frequency average over last four seconds, Slope=regression line over 40 seconds)

		L1 right			L1 left		
		Variation over all six tests sd within•±sd	Within-day variation sd within•±sd	Day-to-day variation sd within•±sd	Variation over all six tests sd within•±sd	Within-day variation sd within•±sd	Day-to-day variation sd within•±sd
IMFi	Hz	4.9±1.4	5.7±2.4	5.0±1.5	6.4±3.2	7.6±5.0	6.0±3.0
IMFm	Hz	4.3±1.6	5.2±1.8	4.0±1.8	4.3±1.6	6.9±6.1	5.8±3.5
FMF	Hz	5.5±3.5	5.6±3.6	5.4±3.5	7.6±4.7	8.5±7.5	7.5±3.8
Slope 40s	Hz/s	0.14±0.08	0.16±0.16	0.14±0.08	0.14±0.04	0.14±0.08	0.12±0.06
		L5 right			L5 left		
IMFi	Hz	7.7±3.2	7.3±4.1	7.6±3.5	8.4±4.2	8.2±7.8	7.8±4.1
IMFm	Hz	6.6±2.6	6.6±3.6	6.3±2.9	8.1±3.5	7.9±7.1	7.5±3.7
FMF	Hz	9.0±2.3	10.2±6.9	8.7±2.5	8.2±3.0	9.7±4.5	7.6±2.7
Slope 40s	Hz/s	0.18±0.12	0.22±0.24	0.18±0.12	0.20±0.10	0.20±0.14	0.20±0.10

• Within sd for each individual: sd within = $\sqrt{(\sum (x_i - \bar{x})^2) / (n-1)}$ $i = 1,\ldots\ldots 6$ (6 tests)
Mean within sd over all individuals: $\overline{sd}$ within = $(\sum sd_i$ within$) / n$ $i = 1,\ldots\ldots 11$ $(n = 11)$

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TIME-FREQUENCY ANALYSIS OF EMG SIGNALS FROM REPETITIVE LIFTING

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Introduction. Estimates of EMG median frequency have provided researchers and clinicians with useful tools for monitoring the electrical manifestations of muscle fatigue [1]. The technique has traditionally been limited to isometric and constant-force contractions because of the need for wide-sense stationary signals when classical spectral estimation procedures are utilized. In this study, we overcome this limitation by adopting a time-frequency transform to analyze EMG signals recorded during dynamic contractions.

Methods. Surface EMG signals were acquired from six paraspinal muscle sites of a healthy 22 y.o. male subject while performing a repetitive lifting task. The subject lifted and lowered a 10.2 kg. box (10% of his maximum) a distance of 0.9 meters, at a duty cycle of 12 lifts per minute, for a total of 5 minutes. A Lido-Lift® dynamometer was used to implement the task and monitor the position of the box. EMG signals and box height were sampled at 2048 Hz using a PC workstation with a 12-bit A/D card. Three segments containing 3000 points of the EMG signal were isolated from the beginning, middle and end of each 1 minute interval, respectively. EMG segments were limited only to the lifting portion of the exercise, based upon the box height. Each EMG segment was analyzed using a discrete-time implementation of a Choi-Williams transform [2]. The Choi-Williams distribution is a modification of the Wigner-Ville distribution and both are subsets of the larger Cohen Class. These distributions are quadratic and time-frequency shift invariant and can reliably measure the changes in time of the frequency content of nonstationary, monocomponent signals. The Choi-Williams distribution uses an exponential kernel as a 2-D low-pass filter to attenuate interference terms in the ambiguity function domain that can prevent the accurate identification of the true frequency components in the signal. Median frequency values were calculated from the Choi-Williams distribution by applying a moving average window with respect to the time axis and treating each time slice as an instantaneous power spectrum. The frequency resolution was equal to 4 Hz.

Results. Changes in the EMG power density spectrum during the lifting exercise can be represented as 2-D contour plots of frequency components above a relative threshold (Figures 1A,1B). The figures are the result of analyzing EMG signal segments from the beginning and end of the exercise using the Choi-Williams transform. The compression of the EMG spectrum to lower frequencies are represented in these figures by the different distributions of the frequency contours. Another representation of the change in the EMG signal spectrum for this same muscle site is represented in Figure 2A as a linear regression plot and 95% confidence interval of median frequency estimates. The percentage decrease in the median frequency estimates during the lifting protocol are summarized for all 6 muscle sites in Figure 2B.

Discussion. A measurable decrease in the spectral content of the EMG signal was observed for superficial paraspinal muscles during the dynamic lifting exercise. Although we cannot deduce from this experiment that the changes in the spectral content of the signal are a direct result of fatigue processes, the results are encouraging in that the technique describes spectral changes of a magnitude similar to reports of paraspinal muscle fatigue induced by static isometric contractions [3]. The impact of concomitant factors present during the lifting exercise will need to be analyzed in more controlled experiments. These experiments will need to consider such factors as muscle length, muscle force, and location of the electrode with respect to the underlying muscle fibers which are known to alter the EMG spectrum [1].

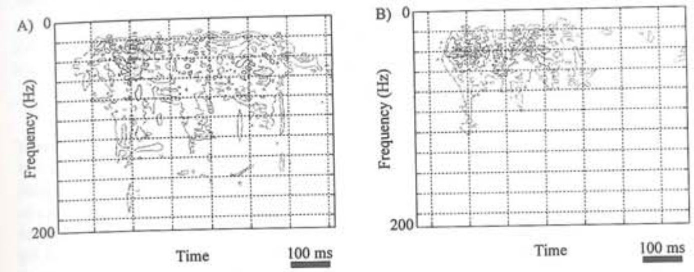

Figure 1. Time-frequency representation of a lifting burst (Ch. 4) occurring during the first minute of the lifting exercise A), and at the end of the fifth minute of the lifting exercise B).

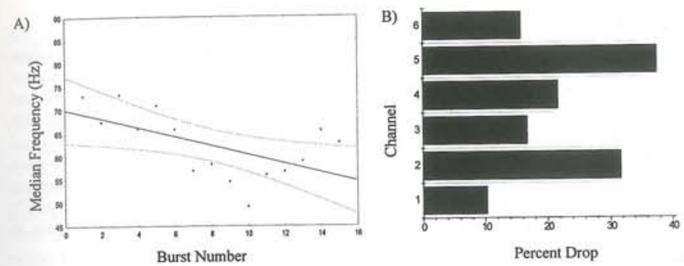

Figure 2. A linear regression plot and 95% confidence interval of the average median frequency obtained for each of 15 lifts obtained over a 5 min. period A). The total percentage drop in the median frequency estimate for each of the 6 EMG channels during the 5 min. lifting protocol B).

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DETERMINATION OF MUSCULAR EFFORT USING EMG AND ENDURANCE TIME

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INTRODUCTION

Rehabilitation professionals often use a measure of muscular strength as an indicator of restoration of muscular function and termination of treatment. Most measures of muscular strength are easily influenced by the motivation of the patient and can yield deceptive information to the clinician. The length of time that a muscular contraction can be maintained increases as the level of effort decreases. In addition to this relationship between effort and endurance time (ET) are established relationships between amplitude and frequency components of EMG and the fatigue state of the muscle. It was hypothesized that if a patient were to misrepresent the level of effort during a sustained contraction, it could be detected by an unacceptable ET and/or changes in the EMG parameters. The pupose of the present investigation was to examine the variabilty in ET and median power frequency changes in the EMG of sustained elbow contractions of highly motivated subjects.

METHODS

Seven healthy subjects maintained isometric elbow flexions of 40%, 50% and 60% of their maximum voluntary contraction (MVC) for as long as possible. The interpolated twitch technique was used to ensure maximum effort during the MVC and fatigue trials. Surface EMG was collected via computer and one second intervals were analysed every 15 seconds during the 40% and 50% efforts and every ten seconds during the 60% efforts. The average EMG (AEMG) and the median power frequency (MPF) were recorded for each interval along with the ET for each trial. A least squares method was used to fit an equation that would predict effort from endurance time. Minimum criteria for MPF and ET were set for the 60% efforts that would fail to identify any of these efforts as being less than a 60% effort. The effectiveness of these criteria at successfully detecting lesser efforts as being such were then evaluated.

RESULTS AND DISCUSSION

In spite of using well motivated subjects, the interpolated twitch revealed that some subjects were less capable than others at producing a maximum effort. It was noted that if a subject increased the ET by using different muscles, the AEMG of the target muscle would decrease yet the force would be maintained and the MPF would continue to decline. Whenever this happened, the test was terminated. When the actual level of effort was plotted as a function of ET, the data were predicted well with equation [1]. The standard error of the prediction was about 20 seconds (see Figure 1). The variability was likely due to differences in fiber type. The major weakness of implementing such a scheme in malingering detection is that a subject could cease the effort prematurely and avoid detection.

The EMG results revealed the common finding that amplitude increases and MPF decreases during a sustained submaxmimal effort. Figure 2 shows the coupled effect of ET and percent increase in AEMG. Once again, a malingerer could rather easily avoid detection by ceasing the effort prematurely. The MPF results showed a similar decrease (from 60 $\pm$ 8 Hz to

39 $\pm$ 7 Hz) regardless of the level of effort. Since the decline in MPF was similar but the ET was not, it was hoped that the rate of decline of MPF could be used to predict effort. Even though there was a positive relationship (r=0.53), large variability prevented its confident implementation.

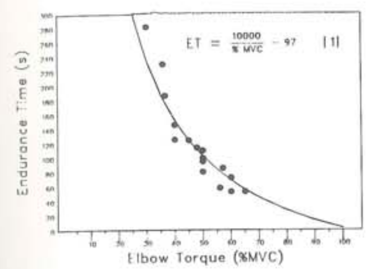

Figure 1. ET vs. %MVC for all subjects with the predictive function [1].

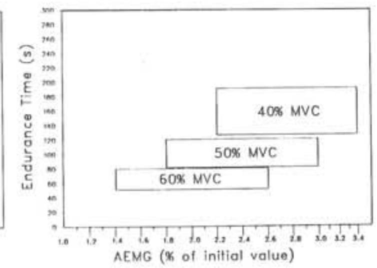

Figure 2. Range of ET vs. range of %AEMG increases for 40%, 50% and 60% efforts.

It was possible, however, to set an ET range of 50 to 80 seconds and the criterion that MPF must be below 50 Hz at termination and have decreased by at least 10 Hz from the beginning of a 60% sustained effort. These criteria were set as a malingering detection that would produce no false positives when applied to the 60% efforts. While these criteria were successful in detecting many true positives for efforts less than 60%, the false negatives included one 56% effort and one 50% effort. One subject could avoid detection with a 36% effort if the effort was terminated between 75 and 80 seconds. As unlikely as this may be, a 40% effort by another subject and a 50% effort by a third subject could avoid detection with a cessation after 60 seconds and another subject could also avoid detection by stopping after 60 seconds.

CONCLUSIONS

The possibility of this test failing to detect the 50% efforts of four of the seven subjects regardless of motivation indicates that the test requires further refinement. The unlikeliness of a 40% effort avoiding detection as being less than a 60% effort indicates that the test may apply to individuals who are grossly misrepresenting their muscular strength (less than two thirds capacity). Obviously, irrespective of motivation, subjects with a lower conduction velocity will have a lower MPF at the beginning of the contraction. In such cases, the MPF would drop below 50 Hz rather quickly and yield a poor detection of malingering and required the addition requirement of a change in MPF of at least 10 Hz.

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CHAPTER 7

ELECTRICAL STIMULATION

ELECTRICAL MUSCLE STIMULATION: STUDY OF THE STIMULATION PARAMETERS

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Introduction

In order to fully understand the optimal possibilities of functional electrical stimulation we have to study the response of different puls- parameters on the generated muscle strength.
In a first part we studied first the It curve before and after a maximal excercice session.
In the second part of this paper we studied also the effect of pulsduration and frequency on the induced force by using "Low frequency" and "Medium frequency" currents.

Methods

The first part of this study is done on humans.
On 10 normal - but not sportive - persons an It curve was made without any previous muscle activity.
This It curve was repeated several times on different days and showed constant values of both Rheobase and Chronaxy.

A second session of measurement was preceded by 10 minutes of intense physical activity of the muscles examined (M. Biceps Brachi and M. Tibialis Anterior).

In the second part of this study a mouse lower leg was connected to a force traducer while the upper leg was fixated.
The mice (N...30) 4ceps muscle was stimulated by means of different currents:

separate pulses with different pulsdurations and with constant intensity

monophasic low frequency pulses with different frequencies but with constant pulsduration and constant intensity

biphasic low frequency pulses with different frequencies but constant pulsduration and constant intensity

medium frequency currents were used with different AM modulation frequencies but with constant pulsduration and constant intensity

Results

All results were statistically analyzed.

In part one we found that an intensive physical excercice does not influence the Rheobase values but does change the Chronaxy value.

In the second part we could conclude that

Separate pulses of different pulsduration do have an optimal induced muscle activity with a pulse of 10 msec.

For low frequency currents (monophasic and biphasic) as well as medium frequency currents in an AM mode we found that the frequency is an important factors in relation with the obtained muscle force.

Frequencies of 50 to 100 Hz give the best result.
Frequencies below or higher do not generate the same muscle strength.

TRANSCUTANEOUS FES of the Upper Limb

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Abstract

The HANDMASTER NMS 1 is a non-invasive personal device providing therapeutic and functional microprocessor-controlled phased, patterned electrical stimulation to the upper limb. It is intended for sufferers from injury or disease to the CNS which has resulted in hemiplegia or quadriplegia. The device has two programmed exercise modes and three functional modes for hand postures and prehension-release patterns. The device is intended for unsupervised use after the initial clinical set up and a few training sessions. This has enabled intensive FES treatment to be self-administered for several hours daily. Results from ongoing clinical assessment have shown the device to be effective therapeutically in the tendency to correct certain abnormal neurological patterns in CVA and head injuries, leading to increased active range of motion, and often unmasking voluntary motion. It also has been used in CVA, head injury, and cervical (C5 and C6) spinal cord injury and shown to reinstate prehension-release function to the paralyzed hand, leading to the restoration of a variety of hand-related activities.

Keywords: Neuroprosthesis, upper limb, FES, non-invasive, CVA, therapy, spasticity

ADVANCES IN ARTIFICIAL STIMULATION OF MUSCLE USING A DIRECT CURRENT BLOCK

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INTRODUCTION

In muscle research and artificial muscle stimulation in clinical situations, electrical stimulation of the peripheral nerve is frequently used activate muscle. A supramaximal pulse train will excite all motor units synchronously. Submaximal electrical currents will initially excite the large motor units rather than the small ones. This is opposite to the normal order of recruitment according to the size principle (e.g. Milner-Brown *et al.*, 1974). Different blocking stimulus strategies, applied concurrently with the normal excitation pulses, have been used to reverse the order of recruitment (Petrofsky, 1978; Solomonov, 1984, Huijing & Baan, 1992). In the present study, we have investigated aspects of the blocking technique that determine the completeness and rate of development of the block. Thus, improvements of the technique may be achieved.

METHODS

Five rat gastrocnemius muscles were used for this study. Under general anaesthesia the muscle and nerve were exposed. The nerve was severed and the distal muscle tendon was cut at the calcaneus and attached to a force transducer. A tripolar needle electrode was inserted in the muscle to obtain (local) EMG activity. A bipolar excitation electrode was positioned most proximally on the nerve. A bipolar block electrode, used to enforce the "size principle" recruitment order, was positioned in between the excitation electrode and the muscle. A direct current (DC) rather than a high frequency pulse train (e.g. Solomonov, 1984; Huijing and Baan, 1992) was used to block the electrical conduction potential of any axon hit by the DC above threshold. Petrofsky (1978) used a constant-volt block and constant-volt excitation pulses. The voltage across the axons depends on the pulse voltage as well as on the variable resistance of other tissue in and around the nerve. The undesired variable effect is eliminated by using a constant-current block and excitation stimulus. We compared two blocking electrodes, one with the poles oriented in parallel with the nerve and one with the poles placed perpendicularly along the nerve. The level of the current of the blocking electrode was computer controlled and could be adjusted to 8 different levels during a single contraction. The electrodes were electrically isolated from the equipment and each other. The actual currents from the electrodes were recorded in all measurements.

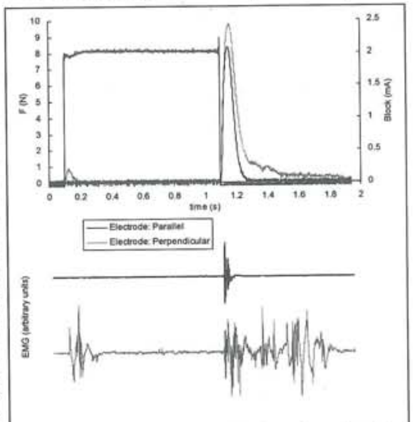

Fig. 1. Constant current DC block performed with a parallel and perpendicularly oriented electrode. Upper diagram shows force response, lower diagram shows the EMG pattern.

RESULTS

The parallel electrode performed better than the perpendicular one which is indicated by the EMG bursts and accompanying force during the ramps of the block (Fig. 1). The perpendicular electrode showed a higher level and longer lasting burst of EMG and force generation. Also, some EMG activity is still apparent during the DC. Reversing the polarity of the electrodes had no significant effect. The EMG pattern resembles surface EMG, unlike the effects of the excitation pulse train, which shows pulses that resemble distorted action potentials (Fig. 2). Figure 2 shows force and (integrated) EMG during a 100 Hz excitation pulse train in combination with a diminishing DC block. Force and integrated EMG increase step-wise with the reducing DC. We reversed the block pattern, changed the maximal block level, and performed different block amplitudes in separate contractions. All these treatments showed similar sigmoid relationship between the level of the block (mA) and force.

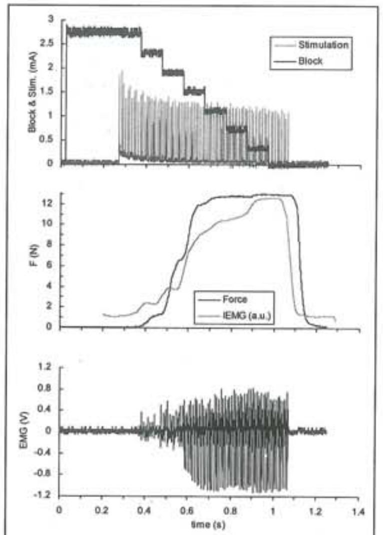

Fig. 2. Force and EMG as a result of a 100 Hz excitation pulse train in combination with a reducing DC block.

DISCUSSION

The most likely reason why the parallel blocking electrode performed better than the perpendicular one is because of altered direction of the current, which was perpendicular on the travel of the action potential. Thus, it may have less power in triggering action potentials during the DC ramps, that may act as an AC pulse. In terms of the development rate and completeness of block, the DC block performs better than the high-frequency-pulse-train block used in the literature (Solomonov, 1984, Huijing and Baan, 1992). Force production was sigmoidally related to block amplitude, indicating that initially small units are excited when reducing the block. However, the last force increments during the staircase block pattern were small as well, which is in contradiction with the size principle.

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MUSCLE ACTIVATION STRATEGIES IN SPINAL CORD INJURED PATIENTS. PRELIMINARY STUDIES.

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INTRODUCTION

The study of the modifications of muscle activity in postural control can shed a light on the comprehension of the recovery of the functionality in complete spinal cord injured patients [SCI]. The SCI patients, due to the impossibility of using the muscles controlled by the neurones below the lesion, try to increase stability by reducing upper body movement. This task is achieved by increasing the activity of the postural muscles and using other muscular groups not normally involved. As shown in previous work [1] this behaviour of SCI patients clearly results in a significant reduced Centre of Pressure variation. It is of interest to study in depth these phenomena. By monitoring the superficial EMG during the subject movements, not only the muscle strength, but also the muscular activation sequences can be investigated. These sequence are expected to be different in SCI patients. To extract EMG temporal features, each signal linear envelope is decomposed in terms of gaussian pulses related to different phases of muscle activity [2].

METHODS

The patient starting from a sitting position performed several movements of trunk, i.e. flexion and extension towards a desk, simulating an usual work situation. The patient sat in a multi-adaptable chair. One rectangular, central start button was placed on the desk about 25 centimetre in front of the seated subject. Also two round peripheral target buttons were placed in parasagittal direction. This target button pair was placed at 90% of the individual unsupported maximal reaching distance. The subject was asked to perform bimanual reaching movements in the sagittal plane towards the target button pair. During task execution force platform signals (Biovec-1000, AMTI, Watertown, Mass.) were recorded. Simultaneously, bilateral superficial EMG activity from latissimus dorsi (LD), trapezius pars ascendens (TPA), serratus anterior (SA), pectoralis major pars sternocostalis (PM), the oblique abdominal muscles (OA), and the erector spinae (ES) at level L3, T9 and T3 was recorded. Were used Ag/AgCl disposable electrodes with an 1 cm^2 contact surface, inter-electrode distance of 2.3 cm, arranged in the direction of the muscle fibres.

The electromyogram envelope was obtained by software processing. The EMG signal was filtered by a band-pass filter to remove both motion artefacts and high frequency noise. Then the signal was rectified and filtered by a low-pass filter with 10 Hz frequency. Centre of pressure (CP) during task execution was calculated from the force platform signals. The envelope was obtained in the balance perturbation phase of each cycle and interpolated to obtain a record of 256 points. To minimize inter-subject variability the average of the interpolated linear envelopes for 15 repetitions was calculated. According to the decomposition technique described by shavi et al., the envelope of the EMG signal is modelled as a summation of unnormalized gaussian pulses of various lengths called phases.

RESULTS

Figure 1 shows the averaged envelope of the EMG activity of the erector spinae at level L3 superimposed on the centre of pressure signal. We have three different phases: the reaching phase in which the subject move his trunk forward to reach the button; the semi-static phase in which the subject presses the peripheral button 5 times; and the balance restoring phase in which the subject moves his trunk backwards. The first phase is from 0 to 30% of the movement, the second phase from 30% to 60%, and the third phase from 60% to 100%. The EMG envelope shows two main peaks, one in the first phase and the second in the third phase; these two peaks are correlated with the peaks of the CP signal.

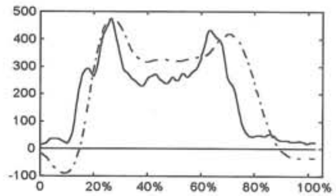

Figure 1. The averaged EMG envelope of the erector spinae at the level L3 in solid line and the centre of pressure (CP) in dash-dotted line. (X-axis percentage of the reaching movement, Y-axis arbitrary unit)

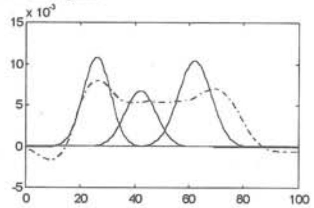

Figure 2. The Dominant gaussian component of the EMG envelope and the CP signal (MDM subject). (X-axis : percentage of the reaching movement, Y-axis: arbitrary unit)

Figure 2 shows the result of the decomposition of the EMG envelope of the erector spinae at level L3. The gaussian curves which are above the threshold of significance are shown. In this example only three components are shown but in all subjects not more than five or six components above the threshold have been found.

DISCUSSION

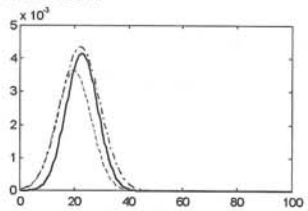

Figure 4: the dominant components of the EMG in the first phase(reaching movement). The solid line is the erector spinae level L3, dashed line is the erector spinae level T9 and dot-dashed line is the erector spinae level T3, (X-axis: percentage of the reaching movement, Y-axis: arbitrary unit).

It is interesting to notice that there are only three dominant phases and they are placed in the three main phases of the movement. This result is always true for all subjects. Only in the middle phase (semi-static phase) we can sometimes see more than one dominant component. The mean of the dominant gaussian represents the position of the electromyographic component. If we consider the first and the last dominant as the main activity efforts in the reaching phase and in the restoring balance phase respectively, we can get a quantitative temporal relationship between the EMG activity and centre of pressure displacement. In figure 3 the dominant phases in the reaching movement of the left erector spinae at level L3, T6 and T9 for a normal subject are shown. Figure 4 is a graphic example of the possibility to use the decomposition in order to get information on the temporal patterns of postural and non-postural muscles in patients and non-patienets.

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INFLUENCE OF STIMULATION PARAMETERS IN ELECTRICAL NERVE STIMULATION ON THE FORCE RESPONSE OF SLOW AND FAST TWITCH MUSCLE

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INTRODUCTION

Depending on the goal of functional electrical stimulation (FES) either slow twitch muscle fibers(STF) (for standing in spinal cord injured patients, tonification in postural disorders) or fast twitch muscles (FTF) (e.g. gait reciprocator in SCI patients, N. peroneus stimulator in CVA patients) should be preferentielly stimulated. The aim of this in vitro study was to observe the influence of different stimulation parameters (pulse rate, duration,configuration and current type) used in FES on the isometric contraction force (ICF) in FES.

METHODS

A 100% STF muscle (m.soleus) and a 95% FTF muscle (m. extensor digitorum longus) of Dunkin Hartley albino guinea pigs were dissected in toto and mounted between two platinum electrodes in organ baths. One set of experimemts evaluated the influence of different frequencies (25,33,50,66 and 100Hz) and 2 current types (low frequency current type and a midfrequency current type, with a carrier frequency of 2500 Hz in burst)(n=10). A second set evaluated the influence of 2 the pulse configurations (monophasic versus biphasic rectangular current) and different pulse durations (50µs to 600µs in sterps of 50µs)(n=10).(train duration: 10s, train interval:

5 m, intensity : 4 times the rheobase of the tissue). Electrical stimulation was done with a constant current stimulator(Gymna 400®) and registration was done with an isometric force transducer. Isometric contraction force was expressed in mN.s/mm^2. ANOVA followed by the Student t-test were used for statistical analysis.

RESULTS AND DISCUSSION

Our results shows a different frequency- force relationship for STF and FTF, the maximal responce being obtained at 33 and 100Hz. respectively, which correlates with firing pattern of the slow twitch and fast twitch motoneurons [1]. Althrough the energy delivered to the tissue is 6 to 25 times higher with the midfrequency current type in comparison to the low frequent current, the latter induces a significantly higher ICF, illustrating that there is no corelation between the energy delivered to the ICF obtained in the tissue.For both muscle fiber types, independent of the pulse configuration, a supramaximal pulse duration is obtained at 400 μs. The 5 to 8 times greater ICF recorded in the FTF muscle for biphasic current in comparison to the monophasic current suggest that the abrupt reversion of polarity leads to an increased depolarisation of α_1 motoneurons. Knowing that the human muscles are heterogenic [2]and that the compensated phase delivers a relative apolarity, a biphasic current type should be preferred. In conclusion :depending on the goal of FES STF should been stimulated with a biphasic, low frequent current, with a pulse duration of 400μs and a frequency of 33 Hz, and FTF with a biphasic, low frequent current, with a pulse duration of 400μs and a frequency of 100 Hz .

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DEVELOPMENT OF A BICYCLE FOR FES LEG EXERCISING

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INTRODUCTION

A spinal cord injury has a severe impact on human life in general as well as on the physical status and condition (osteoporosis, muscle atrophy, spasticity,...).
One possible method to improve general health and physical fitness, is the use of Functional Electrical Stimulation (FES) in combination with a bicycle or ergometer [1]. The effects of leg-cycle exercising with FES are an increased strength and endurance and an increased thigh girth (see e.g. [2]). No adverse effects are reported and this training method is judged to be safe.
The reason for starting the present study was twofold. First, there was a need for an exercise system to recondition comfortably the Quadriceps as well as the Hamstrings of paraplegics in order to train them for walking with a hybrid orthosis (ARGO + FES). Secondly, a follow-up study on the ARGO indicated mainly therapeutic reasons for using it. Paraplegics seem to be more aware of the fact that they need some physical exercise and FES-cycling could be a method for achieving this. However, only few paraplegics are currently using a leg exerciser due to the limited number of commercially available systems and their rather high cost.

DESCRIPTION

To get a better view on the parameters that are important for the development of a FES-leg exercising system, a versatile prototype is constructed. This prototype can be divided in two major parts: the mechanical part (the leg cycling unit) and the electronics part for control of the cycling movement.

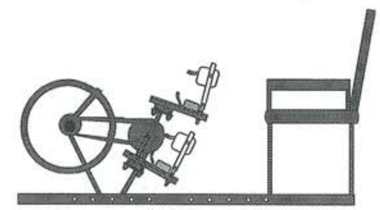

Figure 1 Leg cycling unit

Bicycle mechanics. The mechanical part consists of a chair and a cycling unit. The chair itself is kept as simple as possible because it will be replaced later on by the wheelchair. A normal chair is used with the same height of a wheelchair and with removable armrests.
The cycling unit is composed of a crank with pedals linked by a chain to a flywheel with a brake. The chair and cycling unit are connected by a bar allowing horizontal positioning. The vertical position of the crank can also be changed. In this way the pedals can be positioned according to the leg length and the patient's preference.
The pedal radius is variable to allow different degrees of leg movement. Furthermore, the feet are strapped on wooden plates that are attached to the pedals so that the positioning of the feet

relative to the pedal can be changed. The legs are kept in the sagittal plane by stabilising bars with straps.

Electronics and control. The bicycle is instrumented with an angle encoder on the crank, by which the absolute crank angle can be measured. In this way the position of the legs at each moment is clearly defined. This data is acquired by a PC and is used as an input for the control of the Functional Electrical Stimulation.

The control program itself is written in C++ for Windows to make it user-friendly. For each individual patient a stimulation table is generated first: it links the crank angle with the corresponding stimulation parameters. This look-up table is stored in the PC for consequent use in the control of the cycling exercise.

A commercially available 2 channel stimulator, the EMPI Respond Select, is used for FES. The control program switches the stimulation channels on and off and multiplexes the output of the stimulator to allow simultaneous stimulation of the Quadriceps of one leg and the Hamstrings of the other leg and vice versa.

RESULTS AND DISCUSSION

The developed prototype is tested with several para- and tetraplegics. Most of them were very enthusiastic about it. The goal of these tests was to improve the bicycle and to optimise the stimulation pattern rather than to achieve a real training.

The most important findings of these tests were the following:

- A clear asset is the fact that patients have the impression that after a 10 minutes cycling session their legs are better mobilised than after a session of physiotherapy and that the effect lasts longer. This could be due to the greater intensity of this type of exercising: at a speed of 50 rounds a minute, each leg is flexed and extended 500 times.
- A smaller length of the crank makes the cycling movement easier for non-trained persons. It has also the advantage that there is less flexion at the hip and the knee, which avoids eventual discomfort and possible stretch reflexes of the Quadriceps. At the other hand, the effect of mobilising the legs is less explicit if the cycling radius is smaller.
- The optimisation of the stimulation pattern for each individual is not easy. Starting from a standard pattern, the stimulation parameters are adjusted for the different patients based on the patient's sensation of cycling comfort and on visual examination of the muscle contractions. Objective data could greatly facilitate this optimisation process. Therefor we will mount force transducers on the pedals to measure the exerted forces at each moment.

As this system works properly and the test results are very promising, it will be further improved and a new prototype will be developed. It will include a possible direct connection to the wheelchair, which will make it smaller and even more user-friendly.

ACKNOWLEDGEMENT

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Prototype Hip-Knee-Ankle-Foot-Orthosis. The Influence of Knee Flexion in the Swing Phase on Energy Cost: A Case Study

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Introduction

Reduction of energy cost and upper body load during walking is acknowledged as important issue in the development of new walking orthoses for paraplegic individuals. Components in the mechanical brace which may contribute to a more efficient gait pattern are alignment in frontal plane, lateral stiffness of the leg braces and knee flexion during the swing phase. The currently available orthoses all provide ambulation with extended knees in the swing phase. Patients who use such an orthosis clear the foot during swing phase by adopting compensatory movements at other joints [Kaufman, 1996]. These compensations result in increased muscular effort and increased vertical displacement of the body's centre of mass. Knee flexion will effectively shorten the swing leg during swing phase, thereby reducing the centre of gravity amplitude [Lehman, 1978]. A first prototype Hip-Knee-Ankle-Foot-Orthosis is jointly developed by the University of Twente and Roessingh Research and Development (the UTR orthosis). This orthosis provides knee flexion during swing phase. In this study we compared the physiological requirements of one subject in three different systems: the UTR with locked knees (UTR_{locked}), the UTR with unlocked knees (UTR_{swing}) and a reference system (Advanced Reciprocating Gait Orthosis or ARGO).

Methods

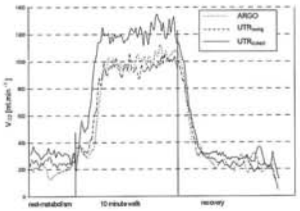
The subject in this study was a man (35 years, 66 kg) and had a complete thoracic lesion (level: T12). Once he was fitted in the UTR he underwent gait training for approximately 2 months in order to be able to control the knee flexion mechanism. Oxygen uptake measurements were performed at a self selected walking speed, during a 10 minute walk. V_{O2}, V_{CO2}, HR, RER, V_e and walking speed (v) were determined during steady state. Oxygen cost was calculated according to: E_{O2} [$ml.m^{-1}.kg^{-1}$] = V_{O2} / v.

A second set of oxygen uptake measurements was performed on a treadmill in order to be able to compare V_{O2} and E_{O2} at equal walking speeds. Walking speed was increased gradually from 0.2 to 0.6 $m.s^{-1}$ during the treadmill test. Each phase took approximately 5 minutes and the actual walking speed in each phase was measured during the treadmill test.

Results

Steady state oxygen uptake was about 20 % higher in the UTR_{locked} with respect to UTR_{swing} as well as ARGO (figure 1). Unlocking the knee in the UTR reduces oxygen uptake compared to ARGO, but walking speed was reduced as well. Consequently, efficiency (E_{O2}) of walking did not improve in the UTR_{swing}.

Differences in walking speed between the orthoses hamper a reliable interpretation of the findings. Comparison of E_{O2} at unequal walking speeds is doubtful since E_{O2} is very much depending on walking speed [Hirokawa, 1990]. Since walking speed between orthoses was different, a treadmill experiment was conducted in order to standardise walking speed. Walking speeds higher than 0.6 $m.s^{-1}$ could not be attained due to the high cadence required for these walking speeds. Steady state V_{O2} was determined during the last 2 minutes of each phase. A plot of V_{O2} and E_{O2} against walking speed is presented in figure 3. Both V_{O2} and E_{O2} were approximately 10-15 % higher in the UTR_{swing} compared to ARGO.



	ARGO	UTR_{swing}	UTR_{locked}
Heartrate [$b.m^{-1}$]	139.5	140.5	151.9
V_e [$l.min^{-1}$]	30.4	32.4	36.6
V_{O2} [$ml.min^{-1}.kg^{-1}$]	19.8	19.0	24.8
V_{CO2} [$ml.min^{-1}.kg^{-1}$]	19.9	19.9	24.7
RER	0.99	1.05	1.00
E_{O2} [$ml.m^{-1}.kg^{-1}$]	0.78	0.98	1.00
v [$m.s^{-1}$]	0.43	0.32	0.41

table 1 Steady state comparison of the ARGO, UTR_{swing} and UTR_{locked}. Ve: expiratory volume, TV: tidal volume, Fr: breathing frequency, RER: respiratory exchange ratio

figure 1. Oxygen uptake in ARGO, UTR_{swing} and UTR_{locked} during the 10 minute walking test. V_{O2} is rapidly increasing up to 1200 $ml.min^{-1}$ for the UTR_{locked} and 1000 $ml.min^{-1}$ for UTR_{swing} as well as ARGO.

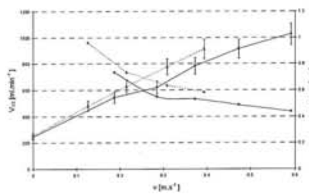


figure 2 (left) Treadmill experiments with the ARGO (solid line) and UTR_{swing} (dashed line). The actual walking speed is displayed (right y-axis). V_{O2} of UTR_{swing} is equal to ARGO, but walking speed differed approximately 0.08 $m.s^{-1}$. Our subject was not able to walk faster than 0.4 $m.s^{-1}$ in the UTR_{swing}.
figure 3 (right) Plot of V_{O2} (left) and E_{O2} (right y-axis) against walking speed. Mean values during the last two minutes of each phase. Standard deviation is presented for V_{O2}. Solid line represents ARGO.

Discussion

Our case report is one of the first studies which has been published on knee flexion during the swing phase in Hip-Knee-Ankle-Foot-Orthosis. The findings indicate that unlocking the knee was not beneficial with respect to energy expenditure. V_{O2} and E_{O2} were even higher (10-15%) in UTR_{swing} with respect to ARGO.Lehman (1978) studied a Knee-Ankle-Foot-Orthosis with an unlocking mechanism in two paraplegic patients. He concluded that knee flexion in the swing phase would only be beneficial with respect to energy expenditure at walking speeds at or above 73 $m.min^{-1}$ (1.2 $m.s^{-1}$). Only able bodied subjects could achieve these ambulation rates [Lehman, 1978]. Knee flexion in the UTR is a passive mechanism due to hip flexion and inertia of the lower leg A sufficient hip flexion velocity yield more knee flexion. However, hip flexion moment can only be generated by means of the upper body. Lehman concluded that paraplegics would not benefit from a knee flexion mechanism because of weakness of hip flexors. Addition of FNS in order to generate an hip flexion moment during swing phase may improve the results.

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CHAPTER 8

LOW BACK / ERGONOMY

EMG BASED FORCE INSENSITIVE PARAMETERS REFLECT MUSCULAR IMBALANCES IN LOW BACK PAIN PATIENTS.

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INTRODUCTION

Most tests of back muscle function are based on the individual's interpretation of what his/her maximum performance of strength, speed or range of motion is (e.g. Thorstensson & Nilsson 1982, Marras et al. 1993, Roy et al 1989). Therefore, patients with low pain tolerance in combination with fear of re-injury may have difficulties performing such a task. Thus, results from such tests will inevitably be contaminated by the individual's conscious and/or unconscious motivation to perform to the full extent of their physical capability. Consequently, any assessment test based on the concept of a maximum performance, be it muscular strength, endurance, range of motion or certain kinematic parameters of voluntary movements, share the common flaw that they are cognitively perceived by the subject and thus can be voluntarily regulated in a manner that may influence the outcome of the test. This makes these tests unsuitable for subjects who are in acute pain or just being unwilling or unable to cooperate during the test. The aim of this study was to develop a set of force-insensitive electromyographic (EMG) parameters that can be used for classification purposes in populations of subjects who develop low trunk extension forces due to pain, fear of injury or simply unwillingness to cooperate in the test.

METHODS

A group of ten healthy males were tested in the Back Analysis System (BAS) at different levels (20-80%) of their maximum voluntary contraction (MVC). The BAS consists of a postural restraint apparatus which stabilizes the pelvis and lower limbs during the test. The subject then exerts a sustained isometric trunk extension force for 30 s against a harness attached to two force transducers (cf. Roy et al. 1989). Lumbar EMG activity was recorded with six active electrodes placed bilaterally over sites at L1, L2 and L5 levels corresponding to the longissimus thoraces, iliocostales lumborum and multifidus muscles. EMG signals were processed in real time to assess spectral information (median frequency - MF, and amplitude - RMS). A post-hoc analysis was performed in a group of fourteen chronic low back pain patients that had previously been tested at 80% of MVC. The patients were specifically selected for having pain in the lumbar back during the BAS test as indicated by their score on a visual analog scale. A set of parameters were calculated to monitor spectral imbalances between muscles of the lumbar back. These imbalance parameters were derived by taking a sample-by-sample ratio between contralateral (right/left L1, L2 and L5) lumbar levels of the MF and RMS signals of interest between 3-30 s of the contraction. The ratios were transformed into a set of corrected ratios which were centered around 0 where a negative ratio indicated that the MF/RMS of the left side was larger than the one for the corresponding right side. A direct mean of the three contralateral imbalances as well as a mean of the absolute value of each ratio was calculated for all subjects.

RESULTS

In the group of control subjects, all three contralateral MF and RMS imbalance parameters remained constant over the whole force range. No significant differences were found between force levels for any of the parameters. The behavior of these parameters was also consistent within the individual subjects. The overall contralateral imbalance behavior for the two groups, as indicated by the direct mean and the mean of the absolute values, is shown in Fig. 1. Segmental MF and RMS imbalances were present in both groups of subjects as indicated by the mean of the absolute values (Fig. 1). However, these segmental differences were canceled out in the healthy subjects when the direction (+ or - of ratio) of the imbalance was taken into consideration (Fig. 1), whereas they remained in the LBP patients.

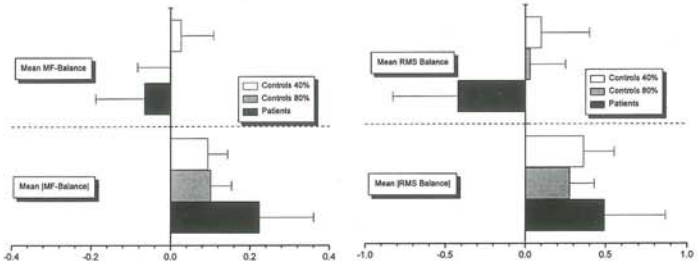

Figure 1. Mean and Mean of Absolute values of overall contralateral MF and RMS for patients and control subjects. Note that segmental imbalances are present both for the control subjects and the patients as indicated by the absolute values. However, the mean values indicate that these imbalances are canceled out for the control subjects but not for the patients.

DISCUSSION

A set of EMG based imbalance parameters, defined as ratios between pairs of contralateral MF and RMS signals, were introduced. These parameters were found to be insensitive to the level of force exerted by the subject which makes them attractive for testing subjects that are in acute pain and/or are unwilling to exert large forces. In addition, these parameters appear to reflect compensatory mechanisms for segmental imbalances that are present in healthy subjects but not in LBP patients. Such information could help the therapist to improve the specificity and quality of the rehabilitation process for the individual patient.

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AMBULATORY BACK LOAD ESTIMATION - VALIDATION IN LIFTING.

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INTRODUCTION

Many people in western society today experience in their lives a-specific low back pain. Many of them to a degree that their daily and work life are severely distorted. The severity and magnitude of these problems has focused the attention of government and industry towards prevention and the attention of the rehabilitation field towards re-education programs and environmental adaptation methodology. This has increased the need for portable, non-interfering instruments to reliably monitor the functional load on the low back during actual work and natural daily activities [3,7]. An attempt was made to develop such an instrument combining inertial sensing, surface EMG recording techniques, a portable recording system, a simple biomechanical cantilever-model and artificial neural network technology [1,2]. Output of the system is formed by patterns and statistics of back posture and movement, back extensor muscle activation, net back extensor moment and compressive force CF the intervertebral body at L5-S1. CF was chosen as operationalisation of back load. This paper discusses and validates the method for net extensor moment assessment, comparing it against a laboratory based assessment method using a 3D kinematic video system, force plates and a 3D linked segment model.

METHODS

The compressive force is estimated by an artificial neural network that has learned the relation between input parameters back inclination, back angular velocity, back angular acceleration and back extensor muscle Smoothed Rectified EMG (SRE) and output parameter compressive force (CF). The neural network was trained in a supervised mode using data sets sampled from a set of random back flexion-extension movements with different known external loads. For each set of input values for the kinematic variables an estimate for CF was calculated using a biomechanical cantilever model. The kinematic variable values were offered together with the SRE signals as inputs to the artificial neural network with the calculated CF value as desired output value. For each subject 160 sec of calibration movements were used. The biomechanical model was of the cantilever type proposed by Dolan and Adams [4] modified to full dynamic back flexion/extension [1]. Inertial sensing, applying miniature rate gyroscopes and accelerometers, was used to assess back movement kinematics [IEEE 96]. EMG was derived from 4 sets of bipolar electrodes placed symmetrically left and right form the spine at levels L3 and T10. Data recording was performed using a custom designed portable data acquisition system, specially adopted to accommodate inertial sensors plus EMG sensors and record long periods handling large amounts of data. Kinematic data were derived by combining the inertial sensor signals in order to avoid temperature and integration drift distorting the data. As a reference ('golden standard') simultaneous recorded data using a 3D video based movement analysis system (VICON) with a customized marker pad set [5] plus an extra marker on the skin above the spine at the L1 level was assessed. 9 subjects performed 32 sets of lifting experiments comprising 2 lifting techniques (back lift and leg lift), 2 external loads, symmetrical and asymmetrical lifts and 2 different speeds.

RESULTS

Kinematic data could be derived with great accuracy and reliability [IEEE 96]. So no differences between estimate and reference extension force can be attributed to inaccuracy in assessment of kinetic data. Typical single lift traces are shown in Fig. 1 (back lift) and Fig. 2 (leg lift). The dark trace is the estimated back extension force and the light trace the force estimated using the reference method. The deviation in the begining of the trace in Fig. 1 appeared to be caused by the subject leaning on the load box which is not detected by the current method. The steep rise reflects the stepwise inclusion of the load box upon detection by a switch under the box. The lift itself however shows a striking resemblance with the reference trace. Fig. 2 shows in the beginning of the lift a clear underestimation. Close inspection of intermediate results showed that this was a direct result of neglecting linear accelerations.

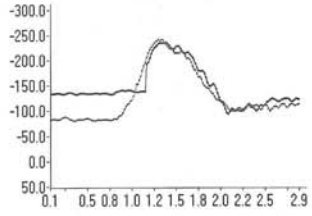

Fig. 1: Net extensor moment single back lift, 6,7 kg. Dark: ambulatory system, gray: reference system.

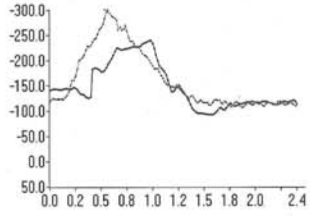

Fig. 2: Net extensor moment single leg lift, 6,7 kg. Dark: ambulatory system, gray: reference system.

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CONTROL STRATEGY PARAMETERS IMPROVE DISCRIMINATION BETWEEN LBP PATIENTS AND NORMALS USING SURFACE EMG

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INTRODUCTION

It was shown before that discrimination was possible between LBP patients and normals by analyzing the slope and intercept values of EMG median frequency (MF) vs. time traces from a set of back extensor muscles [6]. It was also shown that MF vs. time traces were influenced by two main underlying electrophysiological mechanisms [1,3]. One being the peripherally located changes linked to muscle fiber conduction velocity decrease and one being the more centrally located changes in the muscle force generation control strategy. Because the effects of these two mechanisms on the MF traces appeared to be not correlated over different subjects, they will generate a certain amount of unexplained variance which will diminish the discriminative power of the MF trace parameters. Therefore it may be expected that adding a second set of parameters derived from the same EMG recordings, but only linked to one of the two mechanisms, may bring extra discriminative power.
This paper examines whether adding estimations of the average firing rate (AFR) trace parameters increases the discriminative power of the traditionally used set MF trace parameters.

METHODS

The measurement set up was based on a set-up proposed by Roy [6]. Subjects were positioned in a frame to stabilize pelvis and lower limbs. such that the generated back extension force could only be generated by the lumbar extensor muscles. Isometric back extension force was recorded in upright trunk position by a double bridge strain gauge device hooked to a small shoulder harness. 6 EMG electrodes were placed on M. Iliocostalis Lumborum, M. Latisiimus and M. Multifidus. EMG and exerted force were recorded simultaneously with a 1024 Hz sample rate. The exerted force was fed back visually to the subject, to allow for subject force level control. Every subject was asked to exert a back extension contraction at an 80% MVC force level, where the 100% force level was derived from 3 test contractions defined as the maximum force level which could be sustained for 2 seconds. For every recording for all six muscles MF and AFR values were derived over 2 second intervals and stored as a function of time. The slopes and intercepts of these functions were estimated using linear regression.
The algorithm used for estimating AFR [2] 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) [5]. 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]).
14 normals and 23 LBP patients participated in the experiments. The hypothesis was that it would be possible to generate a discrimination,function with a lower misclassification error when using not only the 6 MF slope and 6 MF intercept parameter values but additionally the 6 AFR slope and 6 AFR intercept parameter values.

The discriminant analysis was performed using a Jackknife design. This technique allows a legitimate validation of a discriminant function of n-1 cases (n is the total number of subjects).

RESULTS AND DISCUSSION

Fig 1 and 2 show for two sets of parameters the discriminant function value histograms for all normals (white) and all LBP patients (gray)). In both figures vertical lines indicate the median discriminant function values for both groups. Subjects with a discriminant function value to the left side of the border value (large tick on x-axis) were classified 'normal' and patients with a value right of this line were classified 'LBP patient'. To determine the discriminant function of Fig. 1 only the MF slope and intercept values were used, while for Fig. 2 both the MF and AFR slope and intercept values were used. Fig . 1 and Fig 2. show that using the MF/AFR-set a stronger distinction between both subject groups emerges resulting in more correct classifications. This is also indicated by the increased difference in median scores while for both groups (from 3 to 5.6) indicated by the vertical lines in Fig. 1 and Fig. 2. In this study the percentage of correct classification were 92 % for the MF set and 100 % of the MF/AFR set. Obviously by adding the AFR analysis extra information distinguishing LBP patients from normals is obtained.

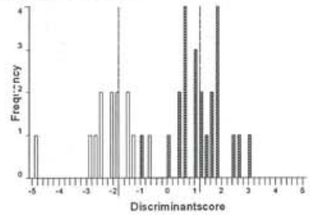

Fig. 1: Histograms of discriminant function values for normals (white) and LBP patients (gray). The discriminant function is based on the MF-set. Vertical lines indicate group medians.

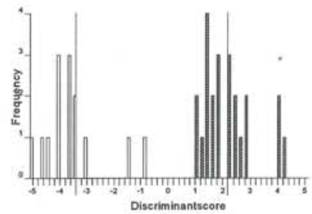

Fig. 2: Histograms of discriminant function values for normals (white) and LBP patients (gray). The discriminant function is based on the MF/AFR- set. Vertical lines indicate group medians.

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LUMBAR PARASPINAL MOTOR EVOKED POTENTIALS IN HEALTHY HUMANS AND IN PATIENTS WITH LOW BACK PAIN.

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INTRODUCTION

Back pain disability increased enormously over the last decades, being currently the leading cause of disability under the age of 45 [9].
In the majority of cases low back pain is not caused by disc herniation or nerve root compression, but is most probably due to so-called mechanical factors, such as structural changes in the lumbar spine, pelvis, pubic symphysis, hips, sacroiliac joints and zygapophyseal or facet joints, and their consequences on the lumbar muscle tone, connective tissues,ligaments, tendons and joint capsules[9]. So far, there has been a lack of objective techniques to quantify muscle spasms in those cases of mechanical low back pain without neurological deficits or severe radiological anomalies. Since 1985, the function and integrity of the corticospinal tracts can be investigated through magnetic transcranial motor cortex stimulation to obtain motor evoked potentials (MEPs) of the recorded muscles[1]. Up to now, magnetic transcranial stimulation (TCS) has already several clinical applications[8], but has not been much used to study the low back muscles. The influence of body position on the MEPs has never been evaluated, neither in healthy humans nor in patients with low back pain. The present study is designed to describe the MEP characteristics of the lumbar low back muscles and the influence of body position on neurocontrol, both in healthy humans and in patients with low back pain.

MATERIAL AND METHODS

We investigated the lumbar paravertebral muscles at the L5 level, using magnetic transcranial brain stimulation during three different body positions, as well in 10 normal humans (ranging in age from 20 to 40 years) as in 10 patients with low back pain. None of the patients had irradiating pain in the lower limbs; they all had a normal clinical neurological examination, and showed no evidence of intervertebral disc herniation.

Magnetic TCS was performed with a 19 mm-diameter coil positioned at the vertex (Cz according to the international 10-20 EEG system). Recording electrodes with a diameter of 0.5 cm were placed on the lumbar paravertebral muscles at the L5 level. In this study the recording electrodes were placed with the anode or active electrode at the needle EMG localization for the multifidus muscle at the fifth lumbar level[5]. Supramaximal stimulation was applied with the subject in three different body positions: in relaxed dorsal recumbency on an examination table, during unsupported relaxed sitting, and finally during unsupported relaxed standing. For each body position at least 3 stimuli were applied and the recorded MEPs were superimposed. The shortest latency time and largest amplitude were taken into account. The latency time was measured at the first negative deflection of the MEP, and the amplitude was measured from peak to peak. For magnetic stimulation we used a Magstim 200 magnetic stimulator (Magstim company, UK). The MEPs were recorded on a Medelec/Teca MS 20 Mystro or Sapphire Premiere electromyograph, with filter settings from 10 Hz to 5 KHz. The sweep speed was set at 100 msec.

RESULTS

The mean MEP latency time in healthy subjects was found to be 16.11 ± 0.93 msec during lying, 14.47 ± 1.0 msec during sitting, and 13.26 ± 1.26 msec during standing. The mean peak-to-peak amplitude of the lumbar paraspinal MEPs was 121.25 µV during lying, 185.0 µV during sitting, and 263.0 µV during standing. There was a decreasing latency time and increasing amplitude from lying to standing. Definite and reproducible late responses werd present in all examined healthy subjects during sitting and standing with latency times of 63.87 ± 4.17 msec and 63.14 ± 2.55 msec respectively. The patients revealing clear lumbar paraspinal muscle spasms showed MEP latency times of 14.4 msec during lying, 15.6 msec during sitting, and 16.9 msec during standing. MEP peak-to-peak-amplitudes werd found to be maximally 1100 µV during lying, 430 µV during sitting, and 260 µV during standing respectively. Moreover, they also had large and clearly reproducible late responses during the lying position with shortest latency time of 61.1 msec.

In other words, patients with low back pain and muscle spasms, showed the opposite neurocontrol pattern of their paraspinal muscles as compared to normal humans as they revealed the shortest latency time, greatest amplitude and most pronounced late responses during lying.

DISCUSSION

The lumbar paraspinal MEP latency times (13 - 16 msec) are comparable to the findings described previously by other authors[4,10] who found latency times of about 13 msec at the thoracolumbar and L3 levels. Our results confirm a direct corticospinal projection to the lumbar paravertebral muscles, most likely through fast conducting fibers in a mono- or oligo-synaptic pathway, similarly as was demonstrated for the corticospinal pathways to the limb muscles. As in healthy humans the MEP latency time shortens and the MEP amplitude increases from the lying to the standing position, it can be concluded that an inhibition occurs during lying, and a facilitation during sitting and most pronounced during standing, since the erector spinae muscles are slightly contracted during sitting and standing, and relaxed during lying.

In patients with low back pain and paravertebral muscle spasms however, an inverse neurocontrol of the low back muscles was seen, as they showed MEPs with largest amplitude and shortest latency time during lying, and low amplitude MEPs with longer latency time during standing. These changes in facilitatory and inhibitory mechanisms might reflect either an altered cortical or spinal excitability, or a different proprioceptive inflow, as their low back muscles are in a prolonged state of contraction or muscle spasm. Possibly even more important is the presence of late responses, that were found in all examined healthy subjects during standing and less pronounced during sitting, though mostly absent during lying, whereas in patients with low back pain and paravertebral muscle spasms clearly reproducible late responses were also seen during lying. The latency times of the late responses we found were within the same range of the late responses other authors described in the limbs[2,3,6] and erector spinae muscles[4]. We suggest that in case of low back muscle spasms the proprio-ceptive input to the central nervous system is altered, resulting in a changed spinal and/or cortical excitability, and either a consequent reactivation of motor units, or a slower conduction or central delay as a result of activation of alternative polysynaptic pathways or structures, such as cortico-bulbo-spinal fibers terminating on the spinal interneurons in the intermediate zone of the spinal grey matter[7].

This technique of magnetic transcranial brain stimulation of the low back muscles during different body positions therefore can offer interesting additional information to document the complaints of low back pain patients with negative radiology and without other neurological deficits, moreover since the alterations in MEP latency and amplitude of the early MEPs and especially the changes of the late responses cannot be simulated in a constantly reproducible way. This easily to perform technique thus can objectivate quantitatively low back muscle spasms, and thus rule out less objective complaints. Consequently, it will be also useful to direct therapeutical measures.

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MUSCLE CONTROL AND MOVEMENT PATTERNS IN BACK PAIN

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Etiology in most cases of back pain remains undiscovered. Although there is no doubt that muscles play in the pathogenesis of back pain an important and often even a decisive role, in clinical praxis the protective role of muscles on joints is considered almost only from the viewpoint of their strength. This narrow aspect is indeed one of the reasons of unsatisfactory results in treatment and prevention, in particular in regard with the chronic pain syndromes.
In this presentation other important but ignored factors necessary for protection of joints are stressed. In particular it is a) the recruitment of motor units (motor units firing order) as a presumption for a reasonable muscle contraction speed, b) the quality of movement patterns and a fine muscle coordination in relation to the sequencing of activation of muscles during various movements and their importance in producing pain syndromes in even remote areas. c) feedback EMG control to achieve a muscle function control which at our present knowledge is considered as the most desirable.
Surface electromyography is the only method by means of which these functions can be examined and evaluated. Kinesiological EMG thus contributes to a better understanding of the problem, can substantially improve the quality of treatment and the observational skill of the clinician.

LUMBAR PARAVERTEBRAL MUSCLE PERFORMANCE IN LIFTING EXERCISES BY MEANS OF ELECTRONICALLY PROCESSED EMG

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INTRODUCTION

Recent literature reports have focused on the possible role of paravertebral muscle function in the development and persistence of lumbar complaints (1, 2, 3, 4, 5).
The present study is designed to assess lumbar paravertebral muscle performance by means of electronically processed EMG.

METHODS

During two lifting exercises the activity of the m. Multifidus (MF) and the m. Iliocostalis lumborum pars thoracis (ICLT) are registrated.
A sample was selected, and during this period the AEMG and average ZCR were calculated by the computer.
The product of AEMG x ZCR has been selected as a parameter to study muscle performance during the following exercises :

1. lifting of an increasing amount of weight in the sagittal plane with a round back
2. asymmetric lifting with a round back

Thirteen male and 6 female volunteers in good health were included in the study.

RESULTS AND DISCUSSION

Lifting of an increasing amount of weight in the sagittal plane with a round back (fig. 1)

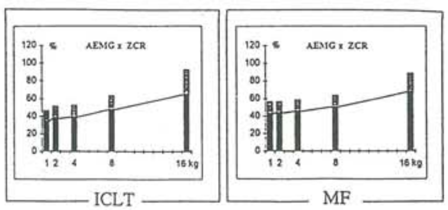

Fig 1 : Lifting in a sagittal plane with a round back : comparison of the increase (% of maximal voluntary isometric contraction) in increasing loading contractions

For the ICLT the correlation between the considered parameters and the increase in weight is strictly linear. For the MF the increase is significantly more than linear. One of the possible explanations is that the MF has a different function than the ICLT.

Asymmetric lifting with a round back (fig. 2)

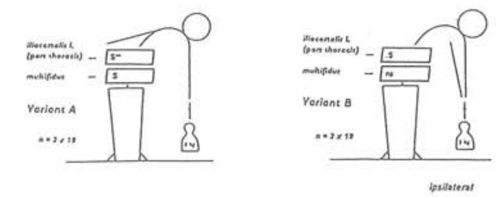

Fig 2 : Asymmetric (torsional) lifting with a round back : Graphic representation of the differences between ICLT and MF during variant A and B. The significances are calculated by the Wilcoxon-test. $p > 0.05$: not significant (ns), $0.05 > p > 0.01$: probably significant (S), $0.01 > p > 0.005$ significant (S*), $0.005 > p$: highly significant (S**).

It appears that during exercise A the contralateral ICLT develops a highly significant peak work rate (S**) in comparison with the ipsilateral ICLT. This difference becomes less pronounced (S) during bimanual grasping (exercise B).
Consequently, the ICLT follows the load of the contralateral shoulder.
Conversely, during exercise A the peak power of the MF shows an only minor difference (S) between the ipsilateral and contralateral side, which disappears completely during exercise B (ns). Consequently, the MF follows only partially the load on the contralateral shoulder during these markedly asymmetric lifting exercises. From the asymmetric action of both ICLT during exercise A it can be concluded that these muscles mainly contribute to the moment of extension. The contralateral ICLT acts as a levator. The more symmetric action of both MF during this highly asymmetric exercise indicates that these muscles mainly aid in the stabilization of the lumbar spine.

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BETWEEN-SUBJECT VARIANCE OF BACK MUSCLE ACTIVATION PATTERNS IN MECHANICALLY COMPLEX TASKS

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INTRODUCTION

In submaximal motor tasks, the trunk is a highly indeterminate mechanical system, i.e. net torques around the lumbar spine may distributed in an infinite number of ways among torque producing structures. To determine loads acting on the lumbar spine during everyday activities, it is nevertheless necessary to solve the indeterminate set of equations describing this system. Some authors have proposed the use of optimization criteria [e.g. 1], others have used empirical relationships between motor tasks and muscle activity patterns [e.g. 2], to obtain such solutions. Both approaches presuppose that the distribution of the torque among muscles and passive structures is based on a control strategy which is generalizable across subjects. Neglecting between subject variation in motor control may limit the validity and applicability of models of the trunk [e.g. 3]. This study assessed between-subject variance in the activation of the erector spinae muscle, in order to evaluate whether this might prohibit the implementation of a general control principle in a distribution model of the trunk. It was assumed that between subject variance might be related to task complexity, therefore trunk activity at varying degrees of asymmetry in trunk posture was studied.

METHODS

12 male subjects performed isometric contractions at 20, 40, 60 and 100% of maximum in a neutral and 15°, 30° and 45° twisted posture. Surface-EMG of 6 bilateral back muscles (multifudus, iliocostalis lumborum, longissimus thoracis) was recorded during these 5-s contractions. Muscle activation was expressed by the rectified and averaged EMG of a 3-s plateau in these contractions. Rectified and averaged EMG values were normalized to the value obtained at 100% and 0° (nRAEMG). ANOVA was performed to test the effects of angle of twist, relative torque and muscle on nRAEMG. Subsequently, linear regression analyses were performed on individual and pooled data of each muscle with angle of twist and relative torque as the independent variables and nRAEMG as the dependent variable.

RESULTS

The nRAEMG values were significantly affected by twisting and torque (both $p < 0.001$). In addition, the effect of muscle ($p < 0.001$) and the interaction effects of muscle with torque and angle respectively were significant (both $p < 0.001$). In all cases a monotonic relationship between nRAEMG and torque was present. Furthermore, the nRAEMG was higher in the muscles on the side contralateral to the direction of the rotation (i.e. the left side). The difference between the left and right sides increased with the angle of twist and the torque.

The regression analyses yielded significant fits for 66 out of 72 (12 subjects x 6 muscles) sets of data. For these 66 data sets the percentage of variance accounted for by the model averaged 75.1 (SD 16.6)%. Figure 1 gives the regression lines for the right longissimus muscle, which was the muscle showing the largest between-subject variance. The variance clearly increased with an increasing angle of twist. As was seen in the other muscles, the data in 10 out of 12subjects clearly follow a similar pattern, whereas the data in 2 subjects diverged from the general trend. Excluding these 2 subjects from the regression analysis on pooled data yielded models accounting for in between 57 and 77% of the variance.

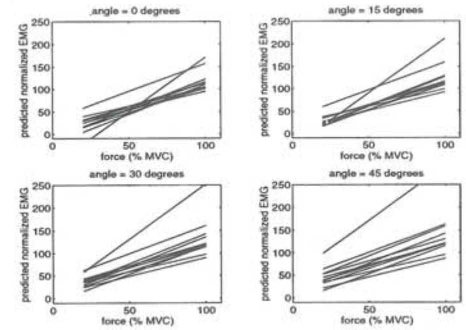

Figure 1. The predicted nRAEMG of the left iliocostalis muscle as a function of relative torque in the four angle conditions based on the regression analysis on individual data sets.

DISCUSSION

Asymmetric activity of the erector spinae in twisted postures has been shown previously in similar tasks [4]. The occurrence of this asymmetry can be understood from the fact that an attempted movement perpendicular to the trunk's frontal plane requires a torque around the flexion extension axis as well as around the lateral flexion axis, as defined with respect to the pelvis. With an increase of the angle of twist the lateral flexing component of the total torque increases. This component can be produced by the contralateral erector spinae muscle.

The fact that a model could be found, which fitted the pooled data fairly well, indicates that a generalizable pattern underlies the activation of the erector spinae muscle group in most subjects in the present set of tasks. This shows that using a general control principle to predict muscle activity in distribution models of the trunk is a viable approach. However, 2 subjects showed a pattern of activity which was clearly different from the general pattern. Prediction of activity levels in these subjects on the basis of the general model may be wrong by a factor 2. It might well be that exactly this variation gives a clue as to why some subjects develop low back pain and others do not. To obtain data on such between-subject variation, EMG driven models are needed, while models driven by deterministic control principles have the advantage of being applicable in simulations and of being open to validation by means of EMG [2]. It, therefore, seems justified to conclude that both type of models should be considered complementary.

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FOREARM FLEXOR AND EXTENSOR MUSCLE EXERTION DURING GRIPPING - A REVIEW

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Introduction
Gripping is a frequently used function in all kinds of manual activities. The basic function in gripping is the finger flexion activated by the finger flexors in the forearm. Hence, electromyographic (EMG) evaluation of gripping work has been focused on the forearm finger flexors. There are only sparse comments in basic textbooks that other forearm muscles are activated as well. A number of recent laboratory experiments as well as field investigations have shed light on these matters and the unanimous results are remarkable with implications for the general comprehension of muscular exertion during gripping.

Forearm muscle acrivation dusring gripping
A fundamental biomechanical circumstance is that the extrinsic finger flexion tendons when transferring a force to the fingers also generate a flexion torque over the wrist joint (Snijders, et al. 1987). In order to preserve a neutral wrist angle, a counteracting torque has to be generated by the wrist extensors. Evidence of wrist extensor activity (and fatigue) during gripping has been reported by several investigators (Byström and Kilbom 1990, Byström, et al. 1991, Jonsson and Nilsson 1979, Snijders, et al. 1987). In a recently reported laboratory study of intermittent gripping with a freely hanging handgrip, Hägg and Milerad were able to demonstrate more pronounced EMG fatigue signs in the extensor muscles compared to the muscles on the flexor side (Hägg and Milerad 1996). Similar results were obtained by Roy et al. studying gripping work with and without a pressurised space glove and by Kilbom et al. studying work with plate-shears (Kilbom, et al. 1993, Roy, et al. 1991). In a study of forearm load patterns during automotive assembly work, Hägg et al. showed a more static load pattern on the extensor side compared to the flexor side where load peaks were higher but pauses were more frequent (Hägg, et al. 1996). This finding further highlights the unfavourable working conditions of the forearm extensor muscles.

Wrist stabilisation
The study by Roy et al. is specially interesting since they found a *reduced* extensor fatigue when performing the same mechanical output with a pressurised glove on the hand. The glove stabilises the wrist and active stabilisation by the extensors is not needed. In a study of forearm fatigue during one handed carrying tasks Kilbom et al. found more pronounced EMG fatigue on the *flexor* side (Kilbom, et al. 1992). In this case, however, the wrist is passively stabilised by the force of gravity from a heavy burden.

The role of wrist stabilisation in gripping has been overlooked. In laboratory experiments the handgrip/forcetransducer arrangements are crucial for the forearm load. A fixed handgrip is likely to relive the load on the forearm extensor muscles. A passive wrist stabilisation during practical work probably has the same effect.

Conclusions
In conclusion, during unsupported gripping, the forearm extensor muscle group exhibit the highest degree of muscular exertion which might be an important clue to the understanding of why tendon disorders are much more frequent on the lateral side than on the medial side (e. g. "tennis elbow").

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EVALUATION OF SURFACE EMG RECORDED FROM THE UPPER TRAPEZIUS MUSCLE IN A GROUP OF FEMALE SUPERMARKET CASHIERS

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INTRODUCTION

In order to identify occupations or work situations with high risk for development of trapezius myalgia, it is, according to the "Cinderella" hypothesis (1), important to evaluate, not only the amount of load on the muscle, but also the time distribution of the load. In this study a set of simple EMG-parameters has been applied to evaluate both load and temporal aspects of the muscle activity in the upper trapezius in a group of female supermarket cashiers.

METHODS

EMG signals were recorded from 33 female cashiers during 2 hours of normal work. Surface electrodes were placed bilaterally over the descending part of the trapezius muscle (midway between C7 and the acromion) and connected to the MyoGuard (Synectics Medical AB), a signal processor based ambulatory monitoring device which calculates and stores muscle activity as an averaged rectified value (ARV) every second (2). In order to obtain a reference value and to assess the subject's ability to relax, two reference contractions and two relaxations were performed just before the start of the measurement. During the reference contractions the subject held her arms straight forward with straight elbows (90° forward flexion) for 20 seconds. The relaxations were performed during 20 seconds between and after the two reference contractions, while the subject was seated comfortably and asked to relax with her hands in the lap. No feedback was provided. The EMG-signals were then recorded during 2 hours of regular work. After the measurement the recorded data were transferred to a PC for further processing including the following steps, **i)** a reference voluntary electrical activation (RVE) value was determined as the mean of the ARV data recorded during the reference contraction with the most stable output (i.e. the lowest variance). **ii)** The recorded data were normalised with respect to RVE, producing data expressed in per cent of RVE (%RVE). **iii)** The following set of parameters was calculated: *MeanARV*, i.e. the mean of the recorded ARV data; *Work/RestThreshold*, based on the mean value of the best relaxation (i.e. the one showing the lowest activity). To this best relaxation value 10 %RVE was added; *RestTime*, the percentage of the total measurement time when ARV-data are below the Work/Rest threshold; *NrOfRestPeriods*, the number of uninterrupted sequences when ARV-data are below the Work/Rest-threshold; *MeanWorkLth*, the mean length of the uninterrupted sequences when ARV-data are above the Work/Rest threshold; *MeanRestLth*, the mean length of the uninterrupted sequences when ARV-data are below the Work/Rest threshold.

RESULTS

The work carried out during the two hours was quite uniform with each cashier serving about 25 customers/h. The mean number of items registered was 620/h and the mean total weight of the items handled was 330 kg/h. This work resulted in a mean EMG-activity in the same order of magnitude as the reference contraction. The parameters concerning the temporal aspects of the load show that there were about 2 resting periods every minute and that the average length of such a period was about 5 seconds. Since the recordings from the left trapezius in some cases were disturbed by ECG activity, the analysis was performed using data from the right

hand side only. The following table shows the outcome of the analysis based on the 33 subjects:

	MeanARV [%RVE]	*Work/Rest Threshold [%RVE]*	*RestTime [%]*	*NrOfRest Periods [/h]*	*MeanRest Lth [s]*	*MeanWork Lth [s]*
mean	91	38	24	140	5.5	54
stddev	60	27	20	83	2.1	110
min	38	16	1	9	2.6	4
median	74	35	17	134	5.3	23
max	314	83	78	334	13.2	603

DISCUSSION

We conclude that the work of the supermarket cashier is quite strenuous. A mean muscle activity (*MeanARV*) close to 100 %RVE must be considered high, especially since the load is present for hours and in some cases even full working days. As is indicated by the difference between the mean and median values of *MeanARV*, some of the subjects showed mean EMG-activity about 2-3 times the reference contraction. A possible explanation could be that some subjects have managed to perform the reference contraction using other shoulder muscles than the upper trapezius, thus producing less output to the electrodes. The *Work/RestThreshold* was in some cases surprisingly high, thereby affecting other parameters depending on this value. Since the threshold value is based on activity recorded during attempted relaxation, these sometimes high values reflect **i)** the subjects ability or disability to relax; **ii)** that noise or disturbances impaired the recording; **iii)** that the electrical output to the electrodes were poor during the reference contraction as a result of the chosen posture and/or the placing of the electrodes which could correspond to the innervation zone of the muscle. These problems could possibly be avoided by altering the reference contraction and placing the electrodes 2 cm lateral to the midpoint of a straight line between C7 and the acromion, as recommended in (3). On the other hand, too high values of the *Work/RestThreshold* also tend to overestimate the outcome of the other parameters concerning relaxation possibilities during the performed work. With this in mind, the short mean duration of *MeanRestLth* and the minimum *RestTime* value of only 1 per cent is alarming. It is obvious that these temporal aspects makes the analysis of the studied work more complete and, given the "Cinderella" hypothesis, tells us whether the low threshold motor units ever get a chance to relax or not.

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GENDER DIFFERENCES IN JOINT COORDINATION OF SQUAT LIFTING

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INTRODUCTION

Work technique is a somewhat unclear concept but the term is often used as well in occupational biomechanics as in sport biomechanics. The overall goal in occupational biomechanics is to recognize techniques that reduce the risks of accidents and stresses on the lower back. In sport biomechanics the main purpose for developing and introducing new techniques is to maximize the performance of competitors.

Work technique has an obvious connection to performance and movement coordination. There are reasons to believe that gender differences exist in the performance of certain tasks, for instance because of different anthropometric characteristics. Gender differences in the performance of an incremental lifting machine test were observed in terms of timing, displacement, velocity, acceleration, force and power (Stevenson et al., 1996). The purpose of the present study is to investigate whether differences in hip-knee coordination can be observed by means of phase-plane analyses of two squat lift experiments; one recent study on twelve female subjects and a previous study on ten male subjects (Lindbeck and Arborelius, 1991).

METHODS

Ten male subjects participated in the previous study. Ages ranged from 28-45 years with a mean of 37.0 years. Means of statures and weights were 176.8 cm and 72.2 kg, respectively. Corresponding values for the women were: ages ranged from 22-60 years, (mean = 39 years), mean stature = 166.9 cm, and mean weight = 63.9 kg.

The subjects performed squat lifts of a box using two velocities, namely a fast lift (FL) of approximately 1 s and a slow lift (SL) of 2 s. The lifting time was defined as the time during which the box was moving. Five trials of each lift type were measured. Only the third was analyzed. The box was 25 cm long in the sagittal plane, provided with handles 25 cm above its base and 40 cm apart. The box was lifted from the level of the force plate to a table adjusted to navel height. The weight of the box was for the men 12.8 kg and for the women 8.7 kg. The difference in load was assumed to correspond to differences in physical capacity.

The movements of markers attached to the ankle, knee, hip, shoulder, elbow, and wrist joints, respectively, were sampled at 50 Hz by means of optoelectronic motion analysis systems (Selspot II in the first study and Mac Reflex in the second). Angular motions were then calculated from the recorded marker data.

RESULTS

The interjoint coordination was quantified as a relative phase angle between the knee joint and the hip joint, respectively, as suggested by Burgess-Limerick et al (1993). Angles and angular velocities for the hip and knee joints were normalized to the interval [-1,1]. Plots of normalized hip angles vs. knee angles (Figure 1a.), phase plane plots, i.e. joint angles vs. joint angular velocities, for the knee and hip joints, respectively, and the corresponding phase angles (Figure 1b.) were produced for all subjects. Relative phase angles, i.e. the knee joint phase angle subtracted from the hip joint angle, were determined (Figure 1c.) and finally max and min values of the relative phase angles were calculated for all subjects (Figure 1d.).

The angle-angle diagrams indicated a better interjoint coordination for women than for men and the curves for slow lifts were in general similar to those for the fast lifts. For the men, however, the coordination pattern appeared to differ depending on whether the lift was a slow or fast lift.

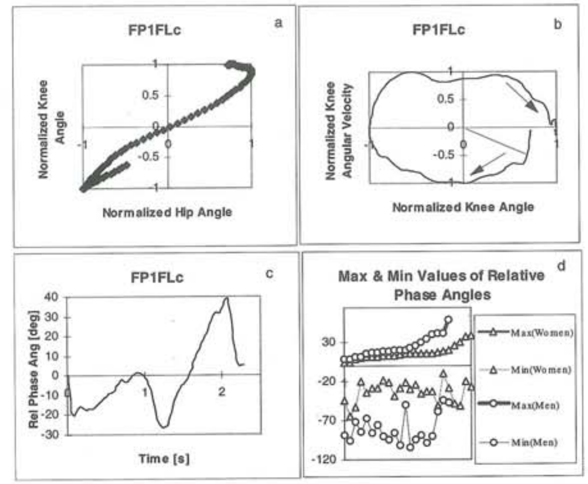

Figure 1. Normalized kinematic data for a fast leg lift. a) Hip angle vs. knee angle, b) knee angle vs. knee angular velocity, c) hip-knee relative phase as a function of time, d) sorted max and min values of the relative phases for all analyzed trials.

DISCUSSION

The analyses indicated that the hip-knee coordination was better in-phase for the women than for the men. The reason for that is difficult to explain. Burgess-Limerick et al. (1995) reported that increasing load mass increased the deviation from perfectly in-phase-coordination. The men in our study lifted a heavier weight and it might be that it did not correspond to a greater physical capacity. The possibility of existing gender differences in motion patterns has not been much considered but is probably an important factor in studies of work technique and performance.

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Muscle Fatigue Estimation *in vivo* - seated on the vehicle seat -

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I. INTRODUCTION

Quantifying muscle fatigue, including lower back and shoulder pain, is vital to the suppression or reduction of pain. It may be helpful to measure the effects of muscle training to prevent overload. The purpose of this study is to measure and estimate muscle fatigue utilizing EMG and muscle viscoelasticity. Muscle fatigue, which has been caused by sitting on a standard/lesser-quality vehicle seat for a long period time, is measured. The authors propose an analytical method using PCA (Principal Component Analysis) with EMG and viscoelasticity. The final goal of this study is to propose an objective muscle fatigue index based on PCA. It corresponds to the subjectivity of the fatigue, and can define clearly the muscle fatigue for each seats.

II. EMG AND VISCOELASTICITY

A. *Vehicle Seat System*

When an estimate of muscle fatigue in the lower back during a long period driving can be established, it will be very useful for reducing muscle fatigue and designing improved vehicle seats. Since this measurement is performed while a subject keeps sitting on a vehicle seat, the subject's posture does not change. The seat system used in our experiment consists of a vehicle seat, a steering wheel, a foot pedal and a personal computer. The sitting posture of the subject strongly influences the occurrence of muscle fatigue. A typical arrangement is shown in Fig.1. In this study, muscle fatigue in the lower back is evaluated for two types of seats. "the standard seat" which we usually use in the vehicle, includes four holes (ϕ25mm) to measure muscle viscoelasticity. These holes do not influence muscle fatigue. "the worse seat" is remodeled to induce muscle fatigue of the lower back by removing a sponge rubber and two "S" springs for lumbar support. This seat forces the subject to sit with a stoop.

B. *EMG and Muscle Viscoelasticity*

In order to create the driving situation, the subject grasps a steering wheel with his left hand and chases a moving object, projected on the monitor, by manipulating a track ball with his right hand. Measuring points should be selected for areas in which a change in muscle stiffness is likely to occur. The four measuring points for viscoelasticity are each 40mm away from the right and left side of the spinal column between the 3rd and the 4th lumbar vertebra, and between the 4th and the 5th lumbar vertebra as shown in Fig.2. The random vibration (30 - 1000Hz) is applied on the lower back and the biomechanical impedance is obtained[1]. The viscosity and elasticity are calculated from the impedance spectrum[2]. EMG is also measured to estimate muscle fatigue. EMG electrodes are attached in each channel at the right and left side of the lower back as shown in the figure. After a subject is seated, viscoelasticity and EMG are measured every 30 minutes (1 set of measurements), and this set is practiced for 3 hours. The subjects are six men who are about 22 years old and frequently drive a car. The subject rates subjective pain in his lower back using a 6-grade classification system from 0 (right after sitting) to 5 (a pain which the subject cannot endure). In order to increase the EMG discharge volume in the seat-sitting, the subject pulls the dynamometer, which is fixed to the wheel, by stretching both arms shown in Fig.1. The pulling force of the dynamometer is 15kgf and muscle fatigue is not influenced pulling the dynamometer.

III. EXPERIMENTAL RESULTS

The changes in elasticity μ_1 and viscosity μ_2 between the standard and worse seat are shown in Fig.3(a) and (b) respectively. The vertical axis is normalized by the value at 0 hour (right after sitting). μ_1 increases at a similar rate for both seats. A difference between the two seats

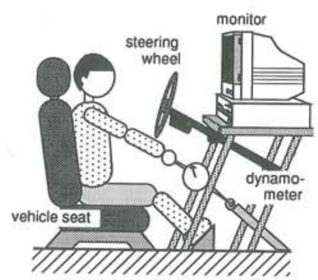


Fig.1 Arrangement of vehicle seat system.

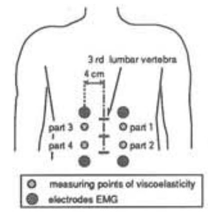
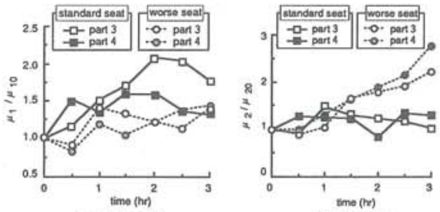


Fig.2 Measuring points.



(a) elasticity (b) viscosity

Fig.3 Changes in elasticity and viscosity.

is found in μ_2. In the worse seat, μ_2 increases steadily, but in the standard seat, μ_2 almost never changes. In the case of the worse seat, the EMG spectrum gradually sharpens and its amplitude increases, shown in Fig.4. Then MNF decreases and IEMG increases.

IV. DISCUSSION

The elasticity μ_1 increases at a similar rate for both seats, while the viscosity μ_2 shows a different tendency. The result of PCA are shown in Fig.5. The figure shows that the 1st component expresses the change in muscle stiffness and the 2nd component expresses objective fatigue, based on the former results. The authors construct a perpendicular to a line from each measurement point to calculate the new index for estimating muscle fatigue utilizing the synthesis of two components. The distance between the foot of the starting point perpendicular (right after sitting) and that of each measurement point is obtained. The distance displays objective muscle fatigue. The relation between objectively-measured muscle fatigue and subjectively-rated fatigue is shown in Fig.6. In the worse seat, objective muscle fatigue corresponds to subjective fatigue and it appears that the muscle becomes fatigued. In the standard seat, such a change is not found and the dispersion of the measurement points is narrow. The muscle is supposed to become less fatigued than in the worse seat. Then subjective fatigue can be estimated using the new objective muscle fatigue index proposed by the authors.

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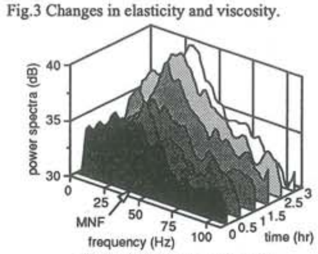

Fig.4 EMG spectrum change.

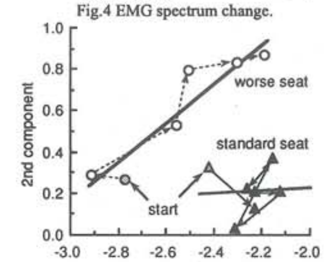

Fig.5 1st and 2nd component of PCA.

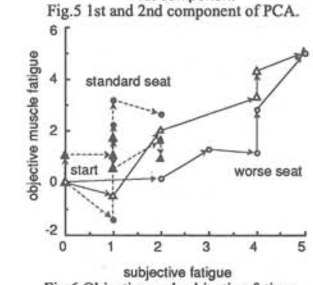

Fig.6 Objective and subjective fatigue.

AN EMG-STUDY OF THE INTERNAL EXPOSURE LEVEL ON LUMBAR SPINE AND SHOULDERS DURING LIFTING.

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INTRODUCTION

Manual material handling still take place in many industries. Contemporary numerous people suffers from musculoskeletal problems (e.g. Biering-Sørensen 1983). The causes are multiple and can be due to both individually-related risk factors, physiological as well as psychological, and work-related risk factors. Among the latters are asymmetric and repetitive lifting (e.g. Riihimäki 1991). The musculoskeletal disorders are in particular localized to the lower back and shoulder neck region.
The aim of the present study was to investigate, by surface EMG technique, differences in the internal exposure level on the lower back and shoulders during repetitive lifting of mail transport-boxes using different lifting heights and frequencies.

METHODS

11 experienced healthy male post-workers 25(21-29) yrs, 72.3(56.0-88.5) kg and 180.0(173.1-191.9) cm gave their informed consent to volunteer in the study. A mock-up was designed in the laboratory. Mail-transportboxes (56 x 33 x 19.8 cm), 10 kg each including mail, were stacked in 1.60 m colums consisting of 8 boxes. The delivery height was 0.80 m above floor.
Independent variables: A total of 9 interventions - 3 different lifting height-categories (low: 36.3 cm and 54.4 cm, medium: 72.5 cm, 90.6 cm, 108.7 cm and 126.8 cm, high: 144.9 cm and 163.0 cm) and 3 different lifting frequencies (6, 12 and 18 lifts per minute) was applied. Each lifting session endured 8 minutes, and was accomplished in a balanced way.
Dependent variables: EMG was recorded bisymmetrically from m. erector spinae and m. trapezius. The myoelectric activity was picked-up by bipolar diaposable Ag/AgCl-surface electrodes (E-10-VS, Medicotest, Denmark) with an interelectrode distance of 2 cm. Electrode-positions on the m. erector spinae were at L3-level 3 cm laterally from midline of the back, and on the m. trapezius midbetween processus prominens and articulatio acromio-clavicularis. The signals were recorded on an on-line system (Physiometer, Phy-400, Premed, Norway).
The root-mean-square (RMS) of the EMG-signals were evaluated by computing the Amplitude Probability Distribution Function (APDF) (e.g. Jonsson 1982). The APDF of the myoelectric signals offers a simple profile of muscular load during a work period, exposing e.g. the static(p 0.1), the median(p 0.5) and the maximum load level(p 0.9). The EMG activity are presented as mean values in percent of the EMG at MVC (%MVE). Non-parametric statistics (Friedman test) were used to test the signifiance of difference. Significance-level was set at the $p<0.05$.

RESULTS

Since no significant side differences were present only results from the right muscles are presented. The mean internal exposure level(%MVE) during the 9 interventions are shown in the table. The signs * and # indicate significant differences, as follows:
* significantly larger %MVE than at the preceding lower frequency on the same height-category, # at the same frequency the %MVE is significantly smaller compared to the low height-category(m.erector spinae) and high height-category(m.trapezius) respectively.

height		low			medium			high	
frequency	6	12	18	6	12	18	6	12	18
p 0.1:									
m. erector spinae	1.4	1.7*	4.2*	0.9#	1.4*	1.8*#	0.9#	1.3*	2.0*#
m. trapezius	0.7	0.6#	0.7*#	0.6#	0.7#	1.1*	0.9	1.9*	2.2
p 0.5:									
m. erector spinae	5.6	10.9*	17.1*	3.8#	6.6*#	9.5*#	3.5#	6.5*#	9.2*#
m. trapezius	2.2#	2.8*#	3.3#	2.6#	3.5*#	5.6*	4.2	7.6	9.5
p 0.9:									
m. erector spinae	24.5	31.5*	35.3	15.4#	20.4*#	23.5*#	16.2#	22.4*#	27.0*#
m. trapezius	7.7#	11.0*#	14.3#	9.2#	12.1*#	17.5*	16.6	20.5	24.2

DISCUSSION

The internal exposure level was largest in the lumbar muscles when lifting from the low lifting height-category and in the shoulder muscles when lifting from the high lifting height-category. This is in concert with Habes et al. (1985). The internal exposure level (evaluated from p 0.9) seem to be dependent on the the lifting frequency. Jørgensen et al. (1985) found the same trend. This is probably not caused by the time ratio between the lifts and the pauses, since EMG indicate that %MVE at each lift was larger when the frequency became higher, probably due to a faster lifting speed.
The muscular loads did on average not reach the critical levels (p 0.1: 2-5%, p 0.5: 10-14%, p 0.9: 50-70%) as suggested by Jonsson (1982). In contrast to this the low-back exposure reach critical levels at p 0.9 when the items lifted are 10 to 15 kg heavier, e.g. during forestry work (handling logs in the forestry) and loading/unloading of air crafts (Jørgensen et al. 1985, Jørgensen et al. 1987).

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CHAPTER 9

POSTURE

CORRELATION BETWEEN BODY SWAY AND BREATHING

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INTRODUCTION

According to Gurfinkel (2) a direct correlation between the postural sways and the respiratory movements during quiet breathing and standing could not be found in normal subjects, because of a fine tuned active compensation ('respiratory synergy'). During inspiration the upper and middle parts of the torso shift backward and this movement of a large body mass must be compensated in order to ensure postural steadiness. This compensation belongs to the feed-forward control mechanism and results from a fixed and reproducible interaction of the joints, mainly of the hip. This assumption has recently been partly questioned (1,4), according to the results of the Time Locked Averaging technique (**TLA**). In a number of neurologic diseases there is evidence of a loss or impairment of this compensation (3) with an increase of postural oscillations related to respiratory activity. Therefore, in some neurological diseases - i.e. Multiple Sclerosis (M.S.) and Head Injury sequelae (HI) - the ventilatory chest movements could be a factor impairing balance and steadiness. The aim of this study was to evaluate the relationship between ventilatory behaviour and postural sway in pathological conditions.

MATERIAL AND METHODS

13 subject without any known respiratory or neurological disease, 3 MS patients and 3 HI patients, all of them able to walk without any walking aid, have been evaluated by means of a dinamometric platform for the Centre of Feet Pressure (CFP) parameters, and a thermic transducer for the pneumograms. Each experiments was lasting 180 seconds and the subject stood upon the platform in a confortable position. Two conditions were evaluated: Quiet Breathing (QB) and Deep Breathing (DB). Frequency and amplitude of respiratory signal were acquired and processed together with statokinesigraphic parameters, namely Sway Path (**SP**), Sway Area (**SA**), fore-aft displacement of CFP (**AP**) and maximal range antero-posterior displacement (**RAP**). The AP parameters were further processed by means of TLA in order to extract the repiratory related components.

RESULTS

Normal subjects during DB condition showed CFP parameters larger than during QB condition (+ 8-24%), except 3 subjects. In the neurological subjects the CFP parameters were higher both in QB and in DB conditions, and MS patients showed the highest values. The AP component connected with respiration in QB and in DB was respectly 6% and 13% in the control group, while was larger in the pathologic group but with no significant difference between the two conditions. In normal subjects the TLA showed in QB a backward movement of CFP during inspiration and a forward movement during expiration. On the contrary, the neurological subjects show a forward shift of CFP during inspiration and backward shift of CFP during expiration. In DB condition the AP components of CFP behave in different manner in both the groups and more complex patterns appear.

DISCUSSION

In normal subjects the slight shift backward of the CFP during inspiration and forward during expiration in QB can confirm that the respiratory synergy, mainly the hip antiphase movement tends to minimise the ventilatory related displacements of the body's centre of gravity, but does not cancel them,. In pathological subjects, the observed inversion of the pattern can reveal a different strategy for postural steadiness preservation, in which the hip might be not involved.

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INTERPRETATION OF STANDING PERFORMANCE PARAMETERS OBTAINED USING A TRIAXIAL ACCELEROMETER

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INTRODUCTION

A large number of tests to evaluate stability during standing have been developed [1]. They depend on expensive force plates and can be difficult to interpret. The project objective was to develop a system for clinical application to perform standing stability tests using a triaxial accelerometer. This paper focuses on the interpretation of the results from this test.

METHODS

The measurement method and formulas for calculations of the performance parameters are detailed elsewhere [2]. In this study, five parameters are calculated: mean speed, mean radius, mean frequency, displacement in the anterior-posterior (A/P) and in the medial-lateral (M/L) directions. A low sway speed is associated with greater stability since it indicates that smaller, better controlled adjusting movements are being made. The mean radius is interpreted as inversely proportional to the stability. If the mean frequency is of a similar value under different conditions, it is assumed that the person is able to compensate for these adverse conditions. The A/P and M/L displacements indicate the limits of the area of the travel during standing. These are useful when comparing results of changing feet position.

Dual tasks [3] are added to tax the balancing system and help to distinguish between truly normal and apparently normal standing stability looking at significant increases in the performance parameters (or decrease in the case of mean frequency).

RESULTS

Experimental measurements were made with eight neurologically intact young men under four single task (CPEO: feet in a comfortable position with eyes open; CPEC: feet in comfortable position, eyes closed; FTEO: feet together, eyes open; FTEC: feet together, eyes closed) and three dual task (MASW: mathematics answering with switches; MASP: mathematics answering by speaking; HLSW: auditory Stroop answering with switches; HLSP: auditory Stroop answering by speaking and MESP: memory answering by speaking) conditions. The average of the resulting parameters and their standard deviations (error bars) are in figure 1.

DISCUSSION

The increases in mean speed and A/P displacement (fig 1) for the single task tests, follow the same pattern. Based on these parameters, the feet in comfortable position with eyes open are the most stable, then all dual tasks, then the feet together eyes open test, followed by the comfortable position eyes closed test, and the feet together eyes closed test the least stable. However, looking at the M/L displacement (and maybe also the mean radius) the order is slightly different: the comfortable position eyes open test and dual task tests are the most stable followed by comfortable position with eyes closed, then feet together with eyes open and feet together with eyes closed the least stable. Since the mean radius and M/L displacement parameters are linked to the mechanical stability of the system, they are particularly affected by the reduction in the bàse of support, while the mean speed is linked to the balancing mechanisms and is more sensitive to deprivation of vision; one of the inputs needed to control balance. In normal subjects no changes are expected for dual task tests [3] compared to the same feet position and vision conditions, which in this case correspond to comfortable feet position with eyes open.

All the parameters must be considered together for the evaluation of a subject's stability. This test should be interpreted according to whether the expected changes in the parameters under the different conditions are present or not and on the relative size of their increases and decreases. Clinical relevance would be best evaluated comparing two or more tests done at different stages of a disease or therapeutic intervention. The answers during the dual tasks can be given by either speaking or using hand switches, according to the abilities or preferences of the subject.

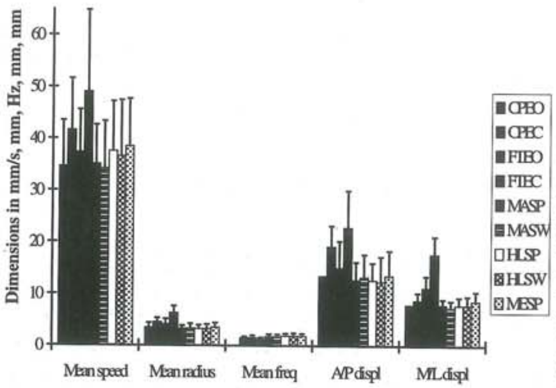

Figure 1. Accelerometer Performance Parameters: eight normal subjects averages and standard deviations. Freq: frequency; displ: displacement.

The triaxial accelerometer based system is: very easy to use, unobtrusive to movement of the subject and inexpensive. By collecting the data on a portable data logger, it can be used in any setting. The results are easy to interpret. Dual task tests can help to discern between subjects who are apparently stable and those who are truly stable. A test based on the accelerometer is being implemented for routine clinical use.

ACKNOWLEDGMENTS

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CORRELATION BETWEEN BODY PARTS' MOVEMENTS DURING GRASPING A DISTANT OBJECT

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INTRODUCTION

Grasping distant objects out of the arm reaching zone is associated with trunk flexion and the corresponding movements at hip and knee joints (Gantchev et al., 1996). The kinematic studies showed backward displacement of the lower part of the body during trunk flexion (Crenna et al., 1987; Oddsson and Thorstensson, 1986), thus stabilizing the position of body center of gravity. The pattern of body parts' displacements during complex movement involving dynamic postural adjustments is thought to depend on the motor task and it reflects to some extent the interaction between the mechanisms of posture and movement control. To study this interrelation we estimated the correlation between distances moved by the various body parts during grasping a distant object.

METHODS

Six healthy male subjects, aged 25-44, participated in the experiments. The standing subject, with his arms along the body, was required to grasp as fast as possible a tennis ball hanging in front of him in one of two positions: (1) at the shoulder level out of the arm reaching zone - about 10 cm away from the elevated and outstretched arm (ORZ), and (2) the ball was situated below the palm of the subject's elevated arm, at knee level (KL). The subjects performed the motor task 10 times for each ball position at time intervals of 5-10 s in a self-paced mode.

Subjects were videotaped in whole size at the sagittal plane by SONY video-camera with frame speed 50/s. For studying arm and body movements, plastic markers were placed on seven reference points on the right side of the subject: on the head, shoulder, the anterior superior iliac crest, knee, ankle, elbow and wrist joints. The kinematic analysis was done by means of software designed for this purpose. The interrelationship of different body parts was evaluated by testing the correlation among the distances moved by the markers for the whole movement time - from the first displacement of the wrist (for the arm movement) and the shoulder (for the body) markers to the time of contact between the palm and the ball. The correlation coefficients were estimated for $n=60$ and $\alpha=0.05$.

RESULTS

High positive correlation between displacements of wrist and shoulder (fig. 1A) and between head and shoulder (fig. 1B) in both task conditions was observed. No significant differences in the coefficients of correlation between the tested experimental conditions was found. As it was expected, no or a low level of correlation was found for the wrist and the rest of body parts (hip, knee and head) in both conditions (fig. 1A). During grasping at knee level, however, the relationships between wrist and the lower body parts became positive.

The data showed lack of correlation between shoulder, hip and knee displacements when the subjects performed grasping out of the reaching zone (fig. 1B), whereas these coefficients increased when the grasping was accomplished through trunk flexion - KL condition ($p<0.05$). Evidently, the relationship between trunk and leg movements during grasping became stronger when the performance of the motor task was associated with more complex postural conditions.

DISCUSSION

The data showed that wrist and head displacements are closely positively related with the shoulder movement, which has been expected as the shoulder represents the support element of the arm and the head. These results suggested that planning and establishing the motor program for arm, head and shoulder displacements involves high degree of synchronization among them.

The stabilization of body center of gravity when the ball had to be grasped at shoulder level out of the arm reaching zone would require small enough displacements in the hip and

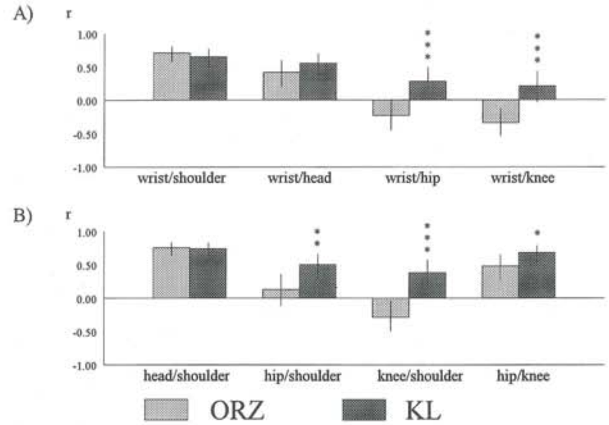

Fig. 1. Correlation coefficients (r and 95% CI, n=60) among displacements of body parts during grasping a distant object associated with dynamic postural adjustments; * - p<0.1, ** - p<0.05, *** - p<0.01.

knee joints. The lack of correlation between shoulder and hip, and between shoulder and knee displacements during grasping the ball in ORZ condition suggested that independent modes of control for arm movement and postural activity are operating.

The increased correlation among the displacements of body parts other than those participating in the focal movement when grasping the ball at knee level might be considered as an adaptive mechanism for posture stability and equilibrium control. The newly established geometrical configuration of body segments aimed to preserve the center of gravity inside the support area and it was determined by the biomechanical constraints for its stabilization (Crenna et al., 1987; Oddsson and Thorstensson, 1986). The stronger relationship between trunk flexion and hip and knee movements during grasping at knee level might be related to the enhanced role of the kinesthetic input in establishing the reference set for posture stabilization and equilibrium control.

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The Co-ordination of Posture and Movement: What are the Associated Postural Adjustments Which Accompany a Simple Lifting Task? **E.M. Earl‡ and J.S. Frank+**
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INTRODUCTION The use of anticipatory postural adjustments to stabilize displacement of the centre of mass (CM) has been illustrated in a variety of voluntary limb and trunk movements[1]. Ankle and hip muscle synergies which serve as fundamental strategies to stabilize posture have been observed during a variety of external perturbations, and during a number of voluntary movements. The purpose of this study was to investigate the nature of the associated postural adjustments (APA's) which accompany a routine daily lifting activity.

METHOD Three healthy adults were asked to stand in front of a table, lift a tray of glasses from the table and place it on a shelf in front of them. The height of the table was standardized such that the top of the table was at waist height for each subject; the endpoint position of the tray on the shelf , the number and location of the glasses on the tray, plus the total mass of the tray were standardized for all subjects. Subjects were asked to perform the task as quickly and accurately as possible, in a simple reaction time paradigm. Six trials were performed following a series of practice trials. Reflective markers were positioned on 10 joint landmarks on the right side of the body. Vertical and horizontal co-ordinates in the sagittal plane, were determined for each marker using the Peak Performance video motion analysis system. The whole body centre of mass (CM) was calculated using this data in conjunction with anthropometric data. CM and joint marker horizontal displacements were quantified using discrete measures in the temporal and spatial domains. All latencies were measured relative to the onset of the vertical hand acceleration which corresponds to the onset of the perturbation by convention (Time = 0.0 s).

RESULTS The magnitude of the horizontal hand displacement was constrained by the task to 30 cm. However, Table 1 illustrates that the three subjects differed with respect to the nature of the primary component of the task, as portrayed by the hand movement. Subject II (S_{II}) exhibited long reaction times (RT) and movement times (MT) in comparison to the other two subjects. Subject III (S_{III}) utilized the greatest hand acceleration in both the vertical and horizontal directions, related to the lift and place phases of the task, respectively.

Table 1: Descriptive Statistics of the Primary Movement (Mean ± 1 Standard Deviation)

Subject	Reaction Time (s)	Movement Time (s)	Peak Hand Horizontal Acceleration (m.s^{-2})	Peak Hand Vertical Acceleration (m.s^{-2})
I	0.189 ± 0.054	3.416 ± 0.260	0.963 ± 0.049	1.373 ± 0.206
II	0.369 ± 0.083	4.142 ± 0.236	0.897 ± 0.135	1.053 ± 0.220
III	0.278 ± 0.039	3.664 ± 0.165	1.527 ± 0.348	1.653 ± 0.259

The mean temporal and magnitude measures of the postural and primary components of the task are presented in Figures 1 and 2. All subjects displayed a posterior displacement of the CM, initiated within 0.155 s of the onset of the perturbation, followed by an anterior displacement of the CM as the tray was placed on the shelf; the magnitudes of the CM excursion were greatest for S_I, and smallest for S_{III}. Each subject utilized anticipatory APA's as depicted by the initiation of either lower extremity (LE) or trunk motion prior to the onset of the hand movement. However, the temporal sequencing and the direction of the movements of the LE, trunk and head segments differed between subjects. S_I and S_{II} exhibited displacements of the LE, trunk and head, first in the posterior, then in the anterior directions. Hip motion was the first evidence of the APA's utilized by S_I, while the onset latency of the ear marker was the shortest for S_{II}. Coincident posterior hip and anterior shoulder displacements indicate that S_{III} utilized a hip strategy during the lift phase of the task; posterior displacement and subsequent anterior displacement of all segments, initiated by the trunk segment, occurred as the tray was placed on the shelf.

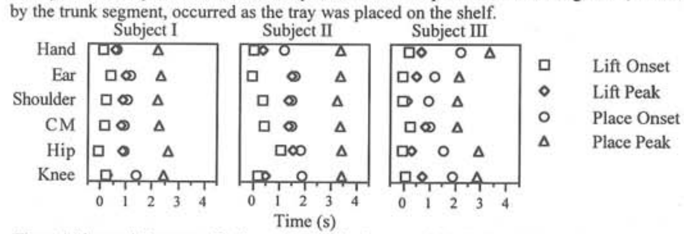

Figure 1: Temporal Measures of Horizontal Marker Displacements for Individual Subjects. Symbols represent the mean latency (n = 6 trials). Negative values indicate events which occurred prior to the onset of the vertical hand acceleration (t = 0).

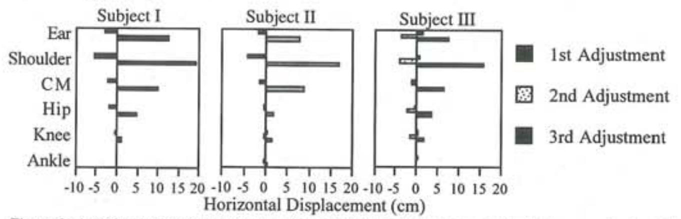

Figure 2: Associated Postural Adjustments for Individual Subjects. Bars illustrate the mean horizontal displacement of the markers in the sagittal plane (n = 6) for each subject. Positive and negative values represent anterior and posterior displacements respectively.

DISCUSSION These young, healthy individuals utilized different strategies to co-ordinate posture and movement during this lifting task. A trade off between the postural and primary components of the task was apparent from S_I who exhibited the fastest RT and MT, and the largest CM excursions. This pattern appears to emphasize the primary component over the postural component of this movement. In contrast, the smallest CM excursions were observed from S_{III} who demonstrated the greatest hand accelerations in conjunction with a hip strategy; this strategy appears to optimize both components of the movement. The amplitude[2] and the speed of movement[3] contribute to the choice of APA which is used during a lifting task. This study provides preliminary evidence that the perceived importance of the postural component relative to that of the primary aim of the task, is another factor which influences the nature of the APA's which accompany voluntary movements.

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ANTICIPATORY POSTURAL ADJUSTMENTS ASSOCIATED WITH VERTICAL SINGLE JUMP MOVEMENTS

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INTRODUCTION

Since Belenkii et al. (1) pioneering work, numerous studies have shown that the voluntary movement is usually associated with postural adjustments that occur simultaneously with or prior to its onset. These postural changes are called anticipatory postural adjustments (APA) and are obviously on feedforward control. The main goal of APA is most often to counterbalance the perturbation caused by the movement (2). At the opposite, in the process of gait initiation, their main role is to disrupt the equilibrium: APA create the necessary conditions for progression, by initiating a forward fall (3). We investigated here if this last functional role of APA also applied for voluntary vertical jumps which, like gait initiation necessitate an initial postural disturbance, but are not forward but upward oriented. Lipshits et al. (4) have shown that there are no APA associated with tip-toe rising movements if subjects are asked to rise on tiptoe and to return immediatly to their initial postition without to maintain the erected posture on tip-toe. Thus, the first question to address was the existence of APA associated to jump. If any, one could expect that the APA characteristics might depend on biomechanical characteristics of movement and of final equilibrium.

MATERIALS and METHODS

The subjects stood barefoot on two force plateforms which were used to record the reaction forces under each foot in three othogonal directions: vertical (Fz), parallel to the ground in the sagittal plane (Fy), and in the frontal plane (Fx). The corresponding moments were also recorded. Vertical acceleration of the head was recorded by a miniature unidirectional accelerometer that was taped to the subject's forehead. The vertical amplitude of the jump was estimated from the acceleration signal by double integration. Surface EMG activities of muscles soleus, tibialis anterior, biceps femoris, vastus lateralis were recorded from both sides of the body. EMG signals were amplified and band-pass filtered (10-500 Hz). All the signals was sampled at 1000 Hz. The experiments were performed on 5 male subjects aged from 22 to 29 years. They were asked to perform 36 single vertical jumps, separated by a of rest. In each series, the subjects had to jump at different heights under the instructions of the experimenter, keeping head and trunk as vertical as possible, and landing at about the same place on the plateform. Before each movement or hopping task, the subjects were asked to stand with their hands clasped behind their back, the trunk and head vertical, and their knees flexed at 30° with respect to full extension. This initial squat position was choosen to avoid countermovements which required particular control of muscle stiffness (5). In this initial posture, subjects were looking at a led placed about 2 m in front of their head and were asked to perform the required movements as soon as the led was switched on.
Time of onset of the soleus EMG burst, which preceded heel-off, was taken as the onset of voluntary movement (t_0). The integrated EMG activity of muscle tibialis anterior was calculated over the period of time between the onset of EMG burst and t_0. The instantaneous coordinate of the center of pressure (Cp) under each foot was calculated from the plateforms signals. Jump impulsion in the vertical direction was obtain by integration of Fz over the duration of the vertical acceleration phase of the body.

RESULTS

A consistent backward shift of the center of pressure demonstrated apparent anticipatory postural adjustments. The maximum Cp shift ranged from 3.0 to 3.7 cm, depending on the subject and for a given subject, on the height of the jump. Onset of Cp displacement occured 100-500 ms before t_0. Cp shift was systematically preceded and accompanied by a tibialis anterior excitation which followed a more or less deep soleus inhibition. The latency of soleus inhibition and that of tibialis excitation (with respect to t_0) covaried linearly. There was also a linear correlation between these delays and that of the onset of Cp shift. Excitation of biceps femoris and vastus medialis occured simultaneously or after t_0. Since subjects were instructed to jump at a given height, and thus to rise their center of gravity of a given height above the floor, one could expect that this parameter could be programmed as it is assumed for other tasks (6). In the present task, the height of the center of gravity might be a function of the vertical impulsion. As a matter of facts, there was a direct proportionality between these two parameters. The onsets of the APA phenomena showed a tendancy to occur with a greater latency when impulsion was higher. In the same way, tibialis integrated EMG and amplitude of Cp shift tended to be greater for greater impulsions. However this trends were or were not statistically significant depending on the subject.

DISCUSSION

The present experiments demonstrated that voluntary vertical jump is associated with anticipatory postural adjustments similar to those described for numerous forward oriented movements of usual life: initiation of gait, forward head and trunk bending, rising on tip-toes, standing up, fast rising of the arms (7). Vertical jump APA are based on a synergy between soleus and tibialis anterior which results in a backward shift of Cp. The characteristics of these APA seem to depend on the vertical impulsion of the following jump. Nevertheless, since the relationships between impulsion and APA characteristics were not always statistically significant, it is likely that APA also depend on other parameters of the task..

In conclusion it seems probable that, as suggested by Crenna and Frigo (1991), when the goal of the movement is a perturbation of the postural equilibrium, the aim of the APA is to adjust the configuration of the forces external to the body, allowing an optimization of the action of the prime mover muscles. For this purpose, APA programmes use a simple preset ankle synergy.

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SPECIFICITY OF ARM MOVEMENTS FOLLOWING EXTERNAL PERTURBATION OF UPRIGHT STANCE

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INTRODUCTION

Rapid movements of the arms seem to be an integral part of the postural response following a perturbation of the support surface (McIlroy and Maki 1995, Oddsson et al. 1995). They appear at short latencies consistent with automatic postural responses (McIlroy and Maki 1995) and they are present even if the subject is performing a lifting task (Oddsson et al. 1995). It can be argued that movements of the arms could act to secure posture in different ways. Grasping a firm object in the environment would immediately increase stability. However, rapid movements of the arms could also serve to transfer momentum to the body thus decreasing the effect of an external perturbation. In addition, movements of the arms allows shift of body mass in relation to the support area which in some situations may decrease the risk of a fall. In addition, in case of a fall the arms could be used to attenuate the impact against the ground. However, the exact function of these protective responses and their relation to postural stability remains to be determined. The aim of this study was to further investigate postural responses of the arms following external perturbations of upright stance. It was hypothesized that the postural responses of the arms would be scaled to the magnitude of the perturbation and display directional specificity.

METHODS

Nine healthy, physically active males (age 22-40 yrs, height 1.75-1.95 m, weight 72-83 kg) participated as subjects in the study. They were perturbed at the feet in eight different directions (straight forward and every 45[th] deg covering 360 deg) at three different magnitudes (0.05, 0.07 and 0.11 m at 9.81 m/s^2) while standing on a balance platform (BALDER, BALance DisturbER) instrumented with a force plate (AMTI). Electromyographic (EMG) activity was recorded bilaterally from the anterior and posterior portions of the deltoideus muscle (lAD, lPD, rAD, rPD), soleus and erector spinae. Only shoulder muscle responses will be reported here. Onset of muscle activity was measured in relation to start of platform acceleration. Magnitude of EMG activity was estimated from the RMS amplitude. Subjects were standing quietly and relaxed looking straight ahead prior to each perturbation. They were instructed to avoid taking a step if possible. There were no objects in the vicinity of the platform that could be grasped for external support.

RESULTS AND DISCUSSION

The arm responses displayed early onsets (80-120 ms), consistent with automatic postural responses (McIlroy & Maki, 1995). All subjects displayed some form of directional specificity in their arm responses. Occasionally, subjects displayed a general preparatory response which was less directional specific. The directional specific responses appeared to vary with the magnitude and/or individual perception of a certain perturbation level, i.e. a certain arm response triggered in one subject may have been triggered at a different level of perturbation in another subject. For backward perturbations, seven out of the nine subjects displayed clear magnitude scaled directional specific responses, in either both (2 subjects), the anterior (3 subjects) or the posterior (2 subjects) deltoid muscle. Figure 1 shows an example of EMG responses from the rPD and lPD displaying this type of behavior for one subject. A directional scaling of the response was seen most clearly for directions involving a backward displacement. For these directions the lPD showed highest activity for backward left

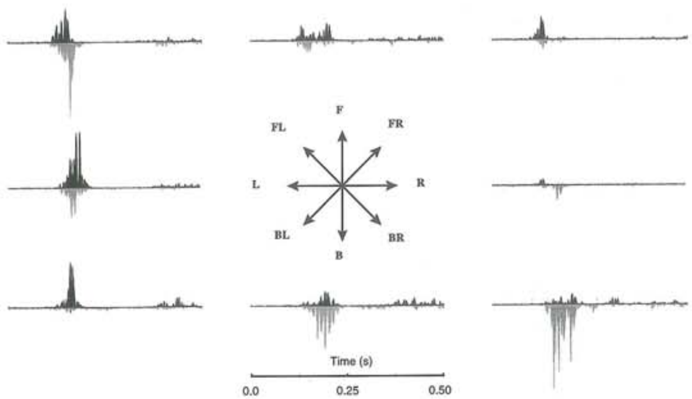

Figure 1. Rectified EMG responses from the left and right posterior deltoid (lPD - upward and rPD - downward deflection) for one subject perturbed at 0.07 m magnitude on the BALDER platform. Directions of perturbation are indicated with arrows (F-forward, B-backward, R-right, L-left). Traces start at onset of platform acceleration and continue for 500 ms.

(BL) perturbations, whereas intermediate levels were seen for backward (B) and low levels for backward right (BR) perturbation. The opposite relationship was seen in the rPD (Fig. 1). The anterior deltoideus muscles displayed a pattern opposite to their posterior counterparts. This suggests that for a BR perturbation the arms would swing in a diagonal pattern with the right arm moving in a backward direction and the left arm forwards, whereas for a BL perturbation the left arm would swing backwards and the right arm forwards. Currently, our results support the presence of varying forms of arm responses related to magnitude and direction of perturbation. In addition, subjects appeared to display blends of responses, i.e. a graded combination of responses with directional specificity and a general preparatory response.

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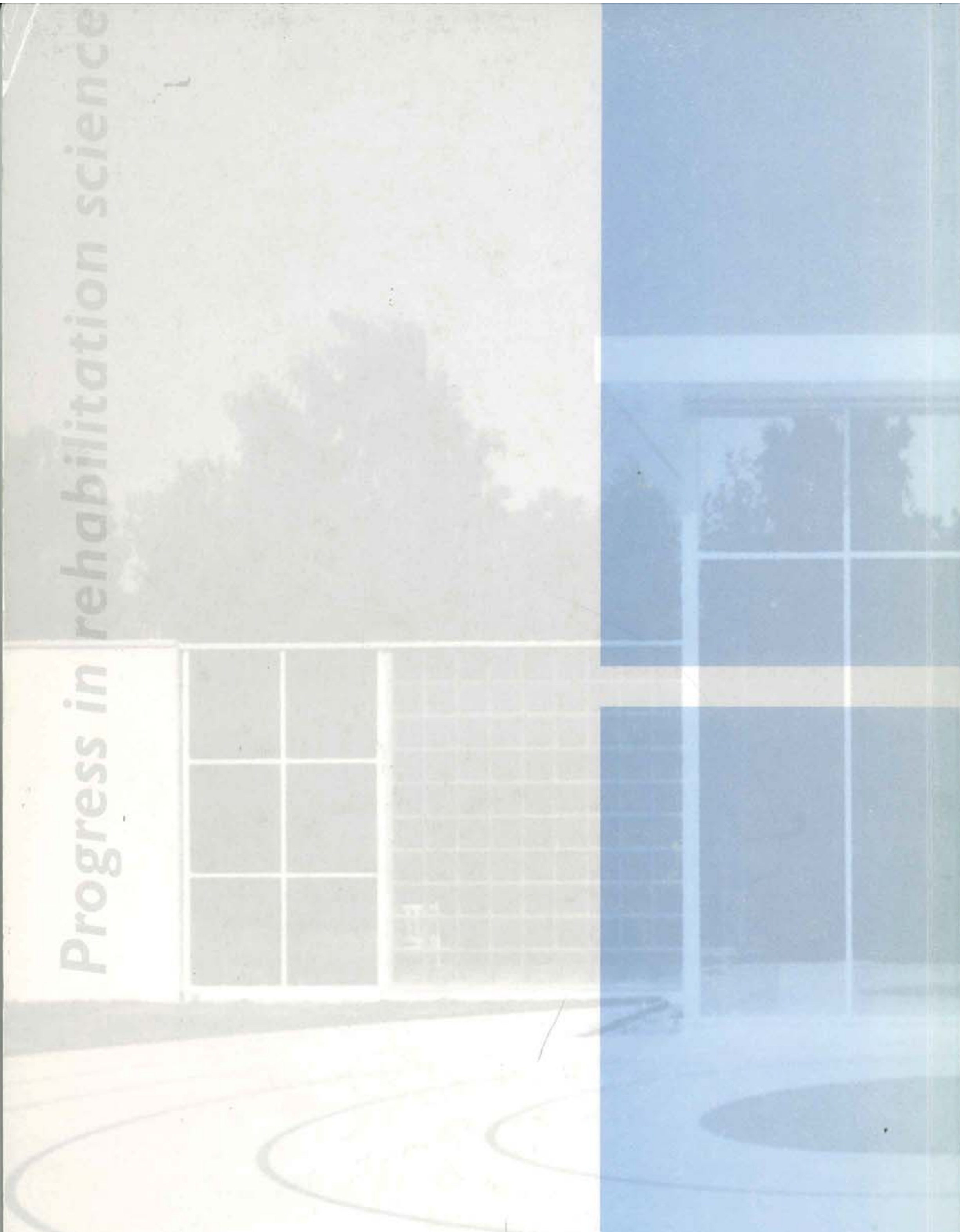

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