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FROM CELL TO MOVEMENT: TO WHAT ANSWERS DOES EMG REALLY CONTRIBUTE?

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INTRODUCTION

The electrophysiological phenomena reflect the active state of living cells. In this sense, the electromyogram is connected to the complex activation of skeletal muscles which results in static and dynamic active force exertion and movement control. Today with refined electrode and amplifier technologies it is easily possible to obtain signals from a muscle. But it still is a big challenge to utilize those signals in order to obtain detailed, repeatable, reliable – and meaningful results. In the following some issues will be considered on various levels of the muscle with reference to the corresponding electrode and recording techniques as indicated by figure 1. Selected implications will be discussed.

Considerable progress has been achieved on different levels during the past years. The biomechanics of muscle cell contraction has been studied regarding the subcellular and cellular phenomena yielding new and deep insight into the biochemical and mechanical mechanisms involved. The electrophysiological excitation at the cell membrane had been investigated by development of voltage clamp techniques, using microelectrodes. With the patch clamp technique it became possible to observe individual electrophysiological channel dynamics under controlled conditions. Recently, the molecular structures of different channel types have been identified and analysed supported by complex computer modelling.

With larger electrodes the activity of muscle cell groups and of complete muscles can be observed. Non-invasive electrode systems pick up the superimposed potential generated by all muscle cells within the volume conducting body. Also here, new developments in measurement methodology, electrode arrangements, signal processing and modelling are in progress to improve the interpretation of the EMG.

However, many issues are unsolved, and one question is: can we make use of all that knowledge to better understand the EMG, or what properties and parameters are represented in the EMG signals? What kind of restrictions may cause limits in the interpretation of the EMG signals?

MUSCLE CELL ACTIVITY

The motor unit (MU) is the functional element in voluntary contraction of skeletal muscles. The access to single muscle fibres and their electrophysiological excitation is possible by isolating the cell in a preparation or by inserting highly selective electrodes. Both are invasive procedures which will not be considered here.

Indwelling needle and wire electrodes can be inserted into the muscle tissue. If the active electrode area is small the electrical activity of single fibres or small motor units may be picked up. Needle and wire electrodes have in this sense a high selectivity with little interference signals from adjacent cells. They are adequate to e.g. study the firing rate of a motor unit. One of the drawbacks is the relative movement between the electrode and the tissue caused by contraction. This implies a number of restrictions for the interpretation of the signals. Also, the method is invasive, poorly repeatable and painful. In small or deeper located muscles or parts of a muscle it is an extremely valuable method.

If we wish to consider activities of a single motor unit or a population of motor units we have to use multiple electrode systems. Insertion of such systems show even more problems of the kind mentioned above. Therefore, non-invasive electrodes may be used to obtain the superimposed action potentials, combined with adequate signal processing including decomposition features.

INVESTIGATION OF MOTOR UNIT PROPERTIES

Studies on the level of MUs enables valuable estimates on different muscular properties and neuromuscular control. For instance, the estimation of single motor unit conduction velocity (CV) may provide information about fiber type and muscle fatigability, while the investigation of recruitment pattern and fire rate gives inside into CNS control strategies. Thus investigation of the MU properties can contribute to the basic understanding of muscle physiology as well as to the diagnosis of neuromuscular disorders.

A proper recording technique for the analysis of MU properties requires the recording of the overall potential, resulting from the potentials of its muscle fibers. Since the cells of one individual MU are spread over a wide cross-section of the muscle, invasive EMG techniques such as needle or wire EMG come to their limits here. The application of specialized surface EMG (SEMG) techniques helps to overcome these problems.
SEMG techniques with a high spatial resolution have been developed within the last decades, making use of one or two-dimensional electrode arrays. The spatial arrangement of little bar or pin electrodes with small inter electrode distances (in the order of mm) allows the enhancement of selectivity by means of spatial filters. Using appropriate filter techniques, it is now possible to detect single MU potentials (MUPs) even at high force levels.

Still it is necessary to decompose such recorded signals in order to track single MUs during an extended time period. Here the combination of invasive and non-invasive methods has shown good results, using the highly selective invasively recorded EMG to trigger the SEMG recordings on single MUPs. With the development of higher order spatial filters it also became possible to track MUs non-invasively from SEMG using advanced decomposition programs. However, the problem of EMG decomposition is not yet solved satisfactorily.

While the timing of the MUPs provides information about CNS control strategies, conclusions on the muscle physiology may be drawn from the MUP shape. While the use of spatial filters enables the identification of MUPs, it also leads to serious deformations on the signal shapes. Therefore it is meaningful to also study monopolar signals in addition. Monopolar signals are superimposed by spuriouse signal components which can be suppressed by averaging over several MUPs. It has been shown that this can be done non-invasively by triggering on the spatially filtered data (Fig. 2). The monopolar signal clearly shows non-travelling components which are suppressed by differentiating electrode systems including bipolar leads. This opens the detection of further muscle fiber properties as e.g. fiber length. This knowledge may prove to be important specially in the field of dynamic EMG, that is EMG acquired during non-isometric and non-isotonic contractions.

Utilizing a one or two dimensional electrode-array, it becomes possible to simultaneously obtain potentials from different acquisition sites. This gives the possibility to estimate for instance CV of MUs from the comparison of signals recorded at different locations. Different methods have been developed in this field, trying to minimize the variance of the estimate. However, due to the non ideal conditions of distorted signal shapes, CV estimates depend on the method adopted. Modelling and recent experimental studies indicate that also the choice

of the spatial filter influences the estimates of CV strongly. This has to be kept in mind when comparing studies which report CV estimates with different types of recording systems.

Finally, when discussing the number of interesting parameters concluded from high spatial resolution EMG techniques, of which not all could be reported here, we have to keep in mind that the results obtained are highly affected by physiological details, such as muscle temperature, relative muscle fiber diameter or inhomogenities in the connective tissue forming part of the volume conductor around the muscle.

GLOBAL MUSCULAR ACTIVITY

In contrast to investigations on cellular or motor unit level, which need very specialised detection techniques, the detection of the compound muscle activity on a more global scale seems to be less difficult. Bipolar electrode arrangements, consisting of relatively large electrodes with interelectrode distances of more than 10 mm, are commonly used in applications in biomechanics, ergonomics or rehabilitation. In these applications mostly the activation pattern (timing) of different muscles or muscle groups or the intensity of muscle activation are of interest. Due to the relatively simple recording technique the detection of the global muscle activity is popular. However, the difficulties with the detected surface-EMG signals appear, when the signals have to be interpreted. The signal amplitude, its power spectrum or the onset of muscular activity are examples for parameters used for information extraction in different applications. However, even for these very established parameters it is often not clear on what underlying effects they depend. Some examples will be shown here.

The mean frequency of the power spectrum is often used to quantify muscular fatigue. A shift of the mean frequency towards lower frequencies is often interpreted as a decrease of the conduction velocity. However, up to now it is not clear whether this effect can be attributed to a decrease of the conduction velocity on the muscle fibre level or to a change in the recruitment toward slower motor units. Further basic investigations are needed to support the interpretation of the frequency shift during fatiguing measurements.

The SEMG amplitude, which is often used to quantify the level of muscular contraction, is known to depend on several different effects like distance between muscle and recording electrodes, localisation of the electrodes relative to the anatomical structures of the muscle or physical parameters like electrode size or interelectrode distance. The European initiative SENIAM gives recommendations for a proper electrode localisation and signal detection in order to minimised the uncertainty in the signal and to make the results more comparable.

The detection of the muscular coordination pattern of different muscles or muscle groups by specifying the on- and the off-times of each muscle is frequently used in clinical applications. However, how to detect of the onset of muscular activity correctly is still discussed and not solved. Additionally, the problem of cross-talk between the different muscles lead to a some degree of uncertainty in the detected coordination pattern. Some of these problems are less when the signals are interpreted by highly experienced persons. To make their experience available for a lager number of users, expert-systems have been designed which support the physician in the interpretation of the surface-EMG signals. Figure 2 shows one attempt based on the fuzzy-inference methodology to build such an expert-system for the supported interpretation of SEMG signals detected synchronously at different muscles.

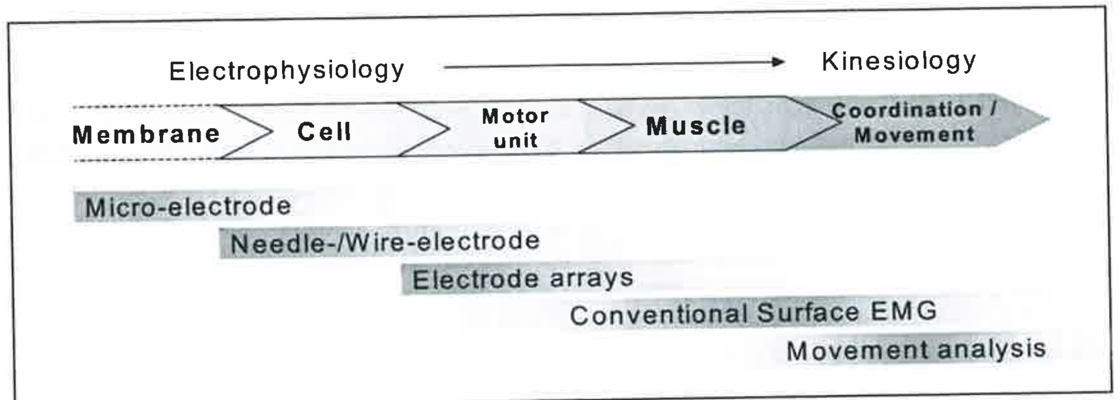

Figure 1: Overview muscle components and recording techniques

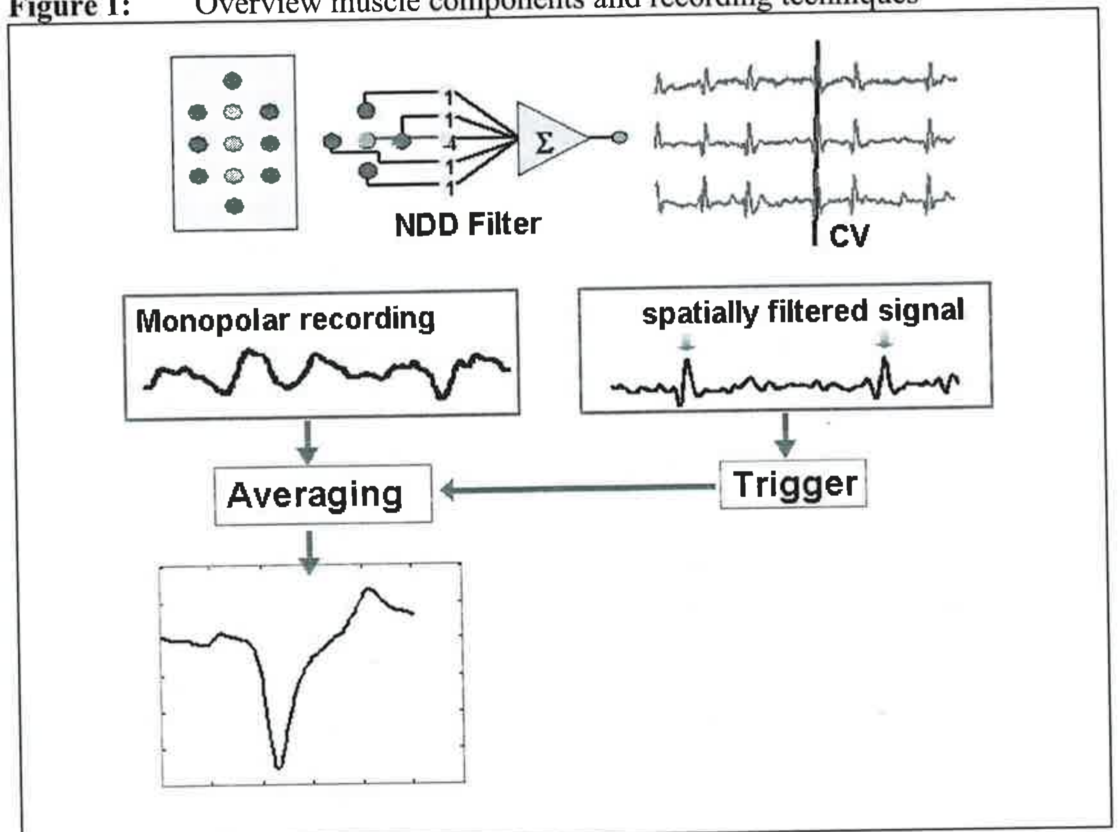

Figure 2: Spatial filtering of EMG signals recorded with an electrode array. Simultaneously recorded signals from different recording sites can be utilized to estimate CV, while information on the monopolar signal shape can be extracted non-invasively by averaging, using the filtered signals for triggering on single MUPs.

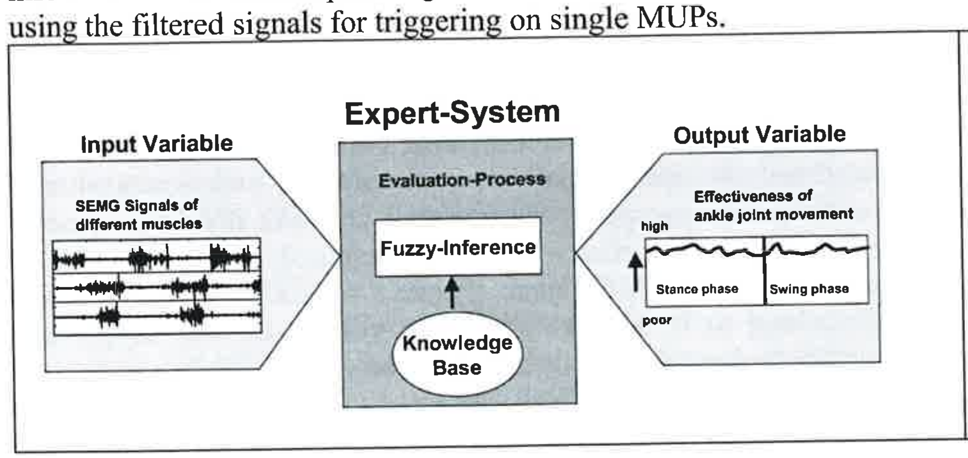

Figure 3: Principle of an expert-System to support the interpretation of the muscular coordination pattern detected by surface EMG.

THE LINEAR ELECTRODE ARRAY: A TOOL WITH MANY APPLICATIONS

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INTRODUCTION.

Electrode arrays provide a very powerful tool for non invasive investigation of muscles through surface EMG. They have been used in research labs for over twenty years, since the pioneer work of Hayes et al., Hilfiker and Meyer, Rau and Reucher, Masuda et al (9, 10, 12, 16). It is the purpose of this lecture to provide an overview of their applications and of their present and future clinical relevance.

An electrode array samples the EMG bidimensional surface potential distribution in a number of points (or surfaces) detecting the evolution of this potential distribution in space and in time.

Except for the case of monopolar pick-up, a spatial filter is always introduced by the detection system. This filter enhances some signal features and attenuates others: it always implies loss of (often undesired) information. Linear arrays sample a section of the "image" represented by the monopolar surface potential distribution and provide linear combinations of the samples. Two dimensional electrode structures, such as the Laplacian or the concentric ring system, process the same "image" in two dimensions. When aligned with the fiber direction of a motor unit (MU), a linear array detects the generation, propagation and extinction of the motor unit action potentials (MUAP) and provides information about the geometry of the MU and MUAP conduction velocity (CV). Full information is provided by the monopolar detection. Progressively reduced information is provided by linear combinations of monopolar voltages which, however, allow to focus on features of interest and to eliminate confounding components. As a first approximation, this process of "spatial filtering" may be seen as a rough form of de-blurring of the signals blurred by the volume conductor interposed between the sources and the detection surfaces or, in certain cases, as a way to reduce undesired components such as the "fiber-end effect". Important applications concern a) the identification of anatomical/geometrical properties of the MUs (in particular, the location of the innervation zone) and consequent identification of the optimal location of a single electrode pair, b) high resolution estimation of CV of single MUs, c) surface EMG decomposition into the constituent action potential trains, d) volume conduction studies.

IDENTIFICATION OF ANATOMICAL/GEOMETRICAL PROPERTIES OF MOTOR UNITS.

Fig.1 depicts a short segment of a set of single differential signals obtained with a linear electrode array applied to the biceps brachii. A number of MUAPs can be visually identified. They can be extracted and repeated MUAPs may be classified as generated by the same MU (13). The innervation and termination zones (IZ and TZ) can be clearly identified. A symmetric pattern indicates proper alignment between the fibers and the electrode array. Fig. 1 also shows that two electrodes placed symmetrically with respect to the IZ would detect a small signal and that the best position of an electrode pair, suitable for estimating EMG variables, is on one side of the IZ, as indicated in Fig. 2 (7). Combinations of linear arrays placed along the fiber directions and perpendicular to it allow estimation of MU location, as demonstrated by Roeleveld et al (17, 18, 19) and depicted in Fig. 3.

HIGH RESOLUTION ESTIMATION OF CV OF SINGLE MOTOR UNITS.

CV estimation is traditionally based on the ratio between the distance separating two electrode pairs and the estimated delay between the corresponding signals. If many signal pairs are available (on the same side of the IZ, as indicated in Fig. 1) the estimate of delay may be obtained with greater accuracy using the beamforming or the maximum likelihood algorithms, as demonstrated by Farina et al. in 2001 (5). If the multiple channel delay algorithms are applied to many individual MUAPs generated by different MUs, the CV distribution of MU firings may be obtained. The CV of a single MU may be estimated with a resolution of 2% to 5% and small changes over time can be detected (5).

SURFACE EMG DECOMPOSITION INTO THE CONSTITUENT MUAP TRAINS.

After 20 years of research, decomposition of a needle or thin wire interferential signals is now ready for clinical applications. Decomposition of surface EMG is still in its infancy and the resolution of superpositions is not yet satisfactory. Classification techniques based on neural networks and template matching are acceptable only when superpositions are limited and MUAP features reasonably different (14).

VOLUME CONDUCTION STUDIES.

Electrode arrays allowed to demonstrate that a major source of crosstalk is the fiber-end effect. Indeed the extinction of the MUAP at the tendon junctions in a muscle can often be clearly detected on another muscle (4, 8). This observation would not be possible with single electrode pairs and opens up a totally new approach to this problem that plagues surface EMG. Fig. 4 shows an example. This is a first step towards the solution of the problem and a challenging research area for the future.

CLINICAL APPLICATIONS AND FUTURE PERSPECTIVES

Clinical applications of surface EMG, at this time, are mostly limited to nerve conduction velocity studies, biofeedback, movement analysis and identification of muscle activation intervals. However, as anticipated by Zwarts et al (21), an increasing number of neurophysiological studies show the potentials of the techniques for estimation of MU size and number (1, 19), studies of firing rate modulation induced by magnetic cortical stimulation (11), studies of specific neuromuscular diseases (3, 15), end-plate disorders, quantification of pathological muscle CV and fatigue (2), etc. The identification of individual MUs and of their innervation zones should lead to optimization of the use of botulinum toxin, by minimizing the injected quantity and allowing monitoring of the denervation-reinnervation process. In conjunction with electrical stimulation and with mechanomyography, array EMG detection is expected to allow studies of recruitment and advances in non-invasive assessment of muscle fiber type constituency by joint estimation of electrophysiological and mechanical properties of individual MUs. Other application areas concern monitoring of muscle reinnervation following peripheral lesions or transplanted limbs, MU topography, studies of innervation of the anal and urethral sphincters to provide guidelines during surgical procedures.

The expected progress in surface EMG decomposition should soon allow reliable estimation of firing rates and recruitment strategies thereby providing more comprehensive information on central mechanisms as well as on peripheral properties of the system.

International research agencies, such the European Community and the European Apace Agency, have recently sponsored research in this field by supporting concerted actions and shared cost projects in the areas of standardization, applications to ergonomics and occupational health, sphincter electrophysiology in relation to incontinence, EMG monitoring of electrical stimulation effectiveness as a countermeasure in microgravity environment. Applications in sport medicine for assessing effectiveness of training strategies is also attracting attention.

Academic training of experts in the field needs specific attention. EMG in general, and surface EMG in particular, are extremely marginal in the education curriculum offered by

most Schools of Neurology, Rehabilitation, Sport and Occupational Medicine as well as in the recently established Schools of Movement Sciences and Exercise Physiology. International efforts and guidelines in this direction are urgently needed to insure proper training of future experts.

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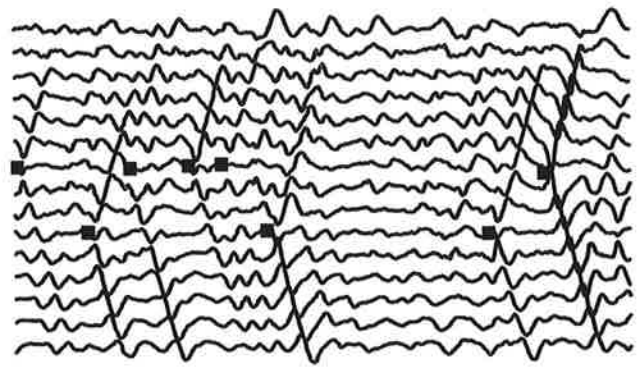

Fig. 1. Voluntary contraction of the biceps brachii at 50% MVC. Single differential signals from a 16 contact linear array with interelectrode distance of 10 mm. 8-10 motor units innervated in two locations can be indentified. Channels located between the innervation zones (%) show bidirectional propagation and are not suitable for global estimation of conduction velocity. Muscle-tendon junctions are visible for some MUs.
From: R. Merletti et al. 1999 (13), with permission.

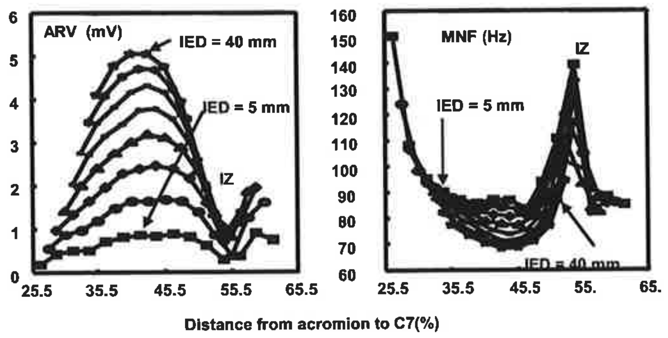

Fig. 2. Average Rectified Value (ARV) and Mean Frequency (MNF) of single differential EMG signals detected in different locations and for different inter-electrode distances (IED) above the trapezius muscle. IED is incremented in steps of 5 mm. A well defined innervation zone (IZ) is evident at 55% of the distance between acromion and C7, where EMG amplitude is minimal while MNF is maximal. From: D. Farina et al. (7) with permission.

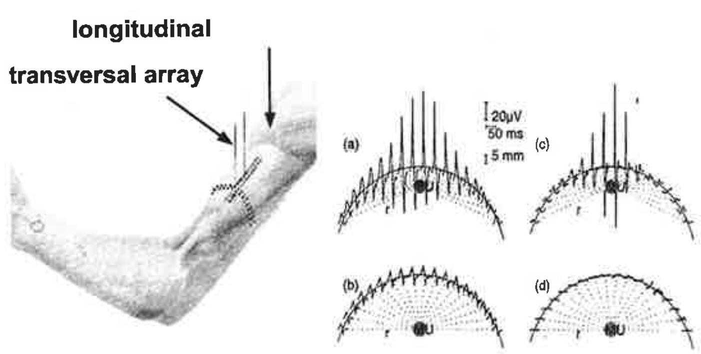

Fig. 3. Left: detection of surface EMG using two perpendicular arrays placed on a biceps brachii. Right: a) monopolar detection of a superficial MU, b) monopolar detection of a deep MU, c) differential detection of a superficial MU, d) differential detection of a deep MU.
From Roeleveld et al. (17), with permission.

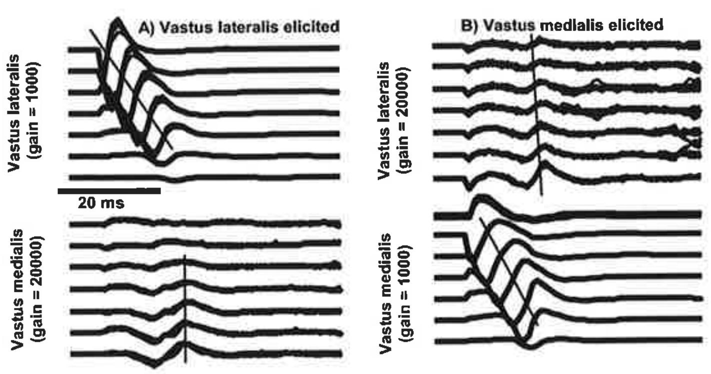

Fig. 4. Electrical stimulation of the vastus lateralis (A) or vastus medialis (B) and single differential detection of surface EMG with an eight electrode array on each muscle. Observe the propagating M-wave on the stimulated muscle and the non- propagating signal synchronous with the M-wave extinction on the non stimulated muscle presumably due to volume conduction of the fiber-end potential generated by the stimulated muscle. From Farina et al. (8), with permission.

MOTOR MECHANISMS OF BALANCE DURING QUIET STANDING

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MEASUREMENT CHALLENGES

Quiet standing as a human "movement" has been confounded at the measurement level for two major reasons. First, the vast majority of studies have been limited to force plate studies using only one force platform. Second, because the displacements, velocities and accelerations of the individual segments and joints are so small that the vast majority of optical systems are not precise enough to yield meaningful kinematic data. The major kinetic variable measured from a force platform is the centre of pressure (COP) and for a single platform produces a two dimensional "spaghetti" diagram that was the spatial/temporal summation of the A/P COP and M/L COP. The control of both these COPs was considered to be the ankle muscles. Not until two platforms were used to separate the contributions of the individual limbs was it possible to identify the fact that the "spaghetti" plot was nothing more than the spatial/temporal summation of two completely separate mechanisms (Winter, et al. 1996). In side-by-side standing, for example, the A/P COP was under the control of the ankle plantarflexors and dorsiflexors while the M/L COP was identified as being under the control of a load/unload mechanism controlled by the hip abd/adductors. The major kinematic variable reported was the body centre of mass (COM) and this has varied from crude single point one-dimensional estimates (Horak, et al. 1992) to a 3D, 14 segment bilateral estimate (Winter, et al. 1998). Standard TV based optical systems lack the precision; however, IRED based systems, such as OPTOTRAC, have a precision of 0.01 mm which is sufficient to record displacements of the leg ($\approx$ 0.2 to 0.5 mm) to the head ($\approx$ 2 to 4 mm). Thus estimates of the total body COM are now possible ($\approx$ 3 mm in the A/P direction and 1.5 mm in the M/L direction).

MODELS AND MECHANISMS OF BALANCE

With precise and accurate measures of COP and COM it is possible to develop a valid biomechanical model of quiet standing: the inverted pendulum. In both A/P and M/L directions a simple inverted pendulum model (Winter, et al. 1998) showed that:

$$COP - COM = -K \bullet \ddot{COM} \qquad \text{Eqtn1}$$

where: K = I/Wh, an individual anthropometric variable; I is the total body moment of inertia about the ankles, W is body weight and h is height of COM above the ankles.
$\ddot{COM}$ is the horizontal acceleration of the COM in either the A/P or M/L directions.
A regression of COP-COM vs COM for 10 subjects yielded an average r of 0.91 in the A/P direction and 0.78 in the M/L direction. The lower M/L score was due to the fact that COP - COM was only twice the noise level while the A/P measures were five times the noise level (Winter, et al. 1998). Eqtn.1 gives us the framework to understand the horizontal accelerations of the COM, those acceleration which could be either destabilizing or destabilizing. COM is the passive controlled variable while COP if the active controlling variable. In quiet standing COP oscillates either side of COM to keep COM in a safe and fairly constant position between the two

feet. If COP is ahead of COM it accelerates COM posteriorly; if COP is to the right of COM it accelerates it to the left, etc.
Detailed segment and joint kinematics of quiet standing have not yet been reported. COM measures are a weighted summation of the trajectories of all body segments in both A/P and M/L planes but do not necessarily reflect the fact that all segments and joints are moving in unison. A recent study (Gage et al. 2002) using a 14 segment bilateral model revealed that all segments move in synchronization with the total body COM and that the largest movement is at the ankle and its angular displacement is highly correlated with the total body COM. A plot of all the rms trajectories vs height of each segment above the ankle yielded a straight line passing though the ankle at (0,0). For 11 subjects standing quietly for two minutes the average R^2 for this regression averaged 0.965 in the A/P direction and 0.956 in the M/L direction. The correlation between the individual segment and the total body COM trajectories ranged from 0.923 to 0.999 and averaged 0.972. Thus it is quite evident that quiet standing involves all segments from the legs up to the head moving in unison. Also, a correlation between the individual ankle joint angular displacements and the total body COM gave an average r^2 of 0.876 for the left ankle, 0.891 for the right ankle and 0.896 for the average left and right ankles. Thus the simplified inverted pendulum model (Eqtn.1) is a valid representation of all segments and reinforces the assumption that COM is the single controlled variable.

The debate that now arises as to whether COP is under active reactive control or passive stiffness control. If reactive control were involved some form of sensory input must be available at the joint, muscle or vestibular level. Our measures of joint angles (0.1° in M/L, 0.25° in A/P), angular velocities (0.05°/s in M/L, 0.16°/s in A/P) and head accelerations (1.1 cm/s^2 in M/L, 1.9 cm/s^2 in A/P) are all at or below reported sensory thresholds (Winter,et al. 1998). Also, if reactive control of some sensory system were present that tracked the COM then there would be afferent and efferent latencies ($\approx$ 50 to 70 ms) plus delays in build up of muscle tension ($\approx$ 70 to 100 ms) as predicted by the newly recruited muscle twitches. Thus A/P COP which is controlled by the plantarflexors would be predicted to lag the COM by 120 to 170 ms. However our measures of COM show that the COP lags by only 5 ms (Winter, et al. 1998). Such a close relationship suggests a near 0th order control, i.e. a simple passive spring system with a small amount of friction. As the body sways forward the ankle spring increases it tension causing the COP to increase and move ahead of the COM, thus causing a posterior acceleration of the COM. Then as the body swing backwards the spring tension decreases causing the COP to decrease and move behind the COM and decelerate its backward movement. The theoretical relationship between M/L sway and stiffness were closely matched by experimental results (Winter, et al. 1998).

More recent evidence is now available as to why the ankle A/P stiffness is always higher than gravitational load (which varies linearly with sway angle from the vertical) is suggested by the non-linear series elastic plantarflexor characteristics reported by Winters and Stark (1988). They report a non-linearity that is close to a square law, thus the tension in the plantarflexors would increase as the square of the ankle angle. Thus at the operating point of the system (the intersection of the gravitational load line and the non-linear series elastic curve) the plantarflexor ankle moment would increase twice as fast as the gravitational load. The operating point is decided by the plantarflexor muscle tone which could drift over time causing the operating point to also drift but the changes in the plantarflexion moment about that changing operating point would always be twice that of the gravitational load. A recent model (Winter et al. 2002) assumes this non-linearity in the plantarflexors combined with a "noisy" COM which is perturbed internally by respiratory and cardiac mass shifts (Hunter and Kearney, 1981). With the COM trajectory as an input the sway angle, 2_{sw}, is predicted from:

$$2_{sw} = COM/h \qquad \text{Eqtn.2}$$

The ankle moment, M_a, is predicted by the non-linear relationship:

$$M_a = K2_{sw}^{\,n} \qquad \text{Eqtn.3}$$

where: "n" must be greater than 1 for COP to increase more rapidly than COM.

M_a is independently calculated from the COP:

$$M_a = COP \bullet R - m_f x_f \qquad \text{Eqtn.4}$$

where: R is the ground reaction force, COP is the distance of COP anterior of the ankle, m_f is the mass of the foot and x_f is horizontal distance of the foot COM to the ankle.

For 11 subjects standing quietly for two minutes n was varied until the predicted M_a (Eqtn.3) most closely agreed with the independently calculated M_a (Eqtn.4). A perfect agreement between the predicted and measured M_a would result in an rms difference of 0 N.m and a regression of the predicted vs calculated M_a would yield an $r^2 = 1$ with a slope of 1. The results from our model gave an average rms difference of 0.70 N.m, an average r^2 of 0.965 with an average slope of 0.985. The non-linearity power "n" ranged from 1.54 to 2.10 with an average of 1.66. Thus this simple non-linear passive model with nothing more than 2_{sw} as input yields an extremely close prediction of M_a. The inherent simplicity of such a balance control system is that the sensory system is not involved any more than setting the muscle tone and even this can be somewhat "sloppy" as changes in muscle tone only cause the operating point to drift forwards and backwards along the gravitational load line (these drifts have been documented experimentally in all subjects).

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CORTICAL REORGANIZATION IN TRAINING

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INTRODUCTION

The adult human cortex has the ability to reorganize and adapt to compensate for environmental changes or lesions. [1] The term “plasticity” defined as “any enduring change in cortical properties, as for example, in the strength of internal connections, representation patterns, or neuronal properties either morphological or functional” [2] refers to this reorganization. [3,4,5,6,7] Transcranial magnetic stimulation (TMS) has been used as a non-invasive method for evaluation of neural plasticity of central nervous system. We evaluate the motor leaning by using TMS. TMS has the characteristics in stimulation the more interneurones in the superficial layer of motor cortex, compared with the transcranial electrical stimulation, which stimulates the more Betz’s neurons, corticospinal neurones. The direction of stimulation by TMS is in parallel with skull.

METHODS

TMS [4,8,9,10,11,12] with twin coil is given to the brain of normal subjects and patients with Parkinson’s disease, by recording the TMS induced movement [3] of thumb with electric goniometer, and the surface EMG. The subject is asked to move the thumb the opposite direction actively, repetitively to the TMS induced movement in 2-3Hz, without rhythmic sound or with rhythmic sound.

RESULTS

In normals after repetitive movement of thumb for 1-2 minutes the direction of TMS induced movement is changed to the direction of active movement. This change is not observed after 5 - 10 times of active movements of thumb. The change of the direction of TMS induced movement after active movements of thumb for 30 minutes, has remained more than 60 minutes after stopping the active movement as memory trace. In imagery there is no change in direction of TMS induced movement as memory trace in primary motor cortex or premotor cortex.

In patients with Parkinson’s disease, the change of the direction of TMS induced movement by active movements is not appeared smoothly after active movement. It takes more time compared with normal subjects. When active movement is done with rhythmic sound, the change of the direction of TMS induced movement is caused easily than it without rhythmic sound(figure1). And it lasts even after stopping of active movement.

DISCUSSION

The motor memory after training is different from the memory of knowledge in the place of store in the brain. These studies suggest that the motor cortex is able to keep a memory trace of the most recently practiced movements, which make the beneficial effects of pre-performance practice in musician and athletes and be the first step in skill acquisition. [1,3,4] From this study the motor learning is stored at least in the primary motor cortex or premotor cortex. The

information of the movement with internal rhythm comes from limbic cortex to the basal ganglia, and to the primary motor cortex (internal voluntary movement circuit). The information of the movement with sensory cue (rhythm) comes from sensory system to the cerebellum, and to the premotor cortex and to the primary motor cortex (external voluntary movement circuit). [13,14,15] In Parkinson’s disease, the connection between the basal ganglia and the motor cortex is not sufficient well. the external voluntary movement circuit might compensate the insufficient internal voluntary movement circuit in Parkinson’s disease from the point of motor learning in this study.

In our another study, we ask the patients with Parkinson disease to use handy auditory cue apparatus and analyze the parameter of gait and the pattern of muscle discharges. By using the handy apparatus, they can walk more rapid and regularly and the pattern of muscle discharges become normal pattern. From these studies this has been used for training apparatus.

It is interesting when the neural plasticity play a beneficial effect functionally [1] with a compensatory mechanism. But it is possible that the plasticity may become undesirable phenomenon. The enhanced method to the direction of beneficial effect might be training. The neural reorganization is ofen related to the usage dependent phenomenon. [4,16,17,18]

The active repetitive finger usage induces memory trace in the motor cortex, with some duration from 30 minutes to several hours or a few days.

The Olympic athetic players show the increased motor mapping area and changed motor mapping place in motor cortex, which is the changes with long duration, and related with long term training.

Voluntary Movement Circuits

Internal Circuit	External Circuit
↓	↓
limbic system	sensory information
↓	↓
basal ganglia	cerebellum
↓	↓
supplementary motor cortex	premotor cortex

primary motor cortex

muscle contraction

movement

Figure1 : In Parkinson’s disease the internal voluntary movement circuit is impaired, the external circuit compensate the decreased function of it.

CONCLUSION

The motor memory is stored in the motor cortex, and the external voluntary movement circuit might compensate the state of insufficient internal voluntary movement circuit.

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THE DECLINE IN STEADINESS OF SUBMAXIMAL CONTRACTIONS WITH ADVANCING AGE

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When an individual performs a steady contraction with hand, arm, or leg muscles, the force exerted by the limb is not constant but rather it fluctuates about an average value (Christou et al 2002). The variability of the force about the mean, which is quantified in absolute terms as the standard deviation or in relative terms as the coefficient of variation, often increases with the average force exerted by the involved muscles. Based on protocols that have involved low-to-moderate forces during isometric and anisometric contractions, a number of studies have found that there can be differences between young and old adults in the amplitude of the force fluctuations. The purpose of the presentation is to describe the functional consequences of greater force fluctuations, to characterize the changes that occur with advancing age, and to examine the mechanisms responsible for this effect.

Functional Consequences

The presence of force fluctuations in the motor output influences the capacity of an individual to achieve a desired force and to produce an intended trajectory. For example, the force exerted by the thumb and index finger when performing the pinch grip must be greater than the difference between the minimum in the force fluctuations and the force required to prevent slipping (Cole, 1991; Kinoshita & Francis, 1996). Similarly, the rapid performance of a simple aiming movement requires large control signals, which increases the variability of the trajectory and reduces the accuracy of the final position (Fitts, 1954; Kim et al 1999; van Galen & Huygevoort, 2000). Furthermore, the fluctuations in muscle force during a voluntary contraction cause variability in motor output across trials, which impair the ability of an individual to achieve a desired force, trajectory, or force-time integral (Carlton & Newell, 1993; Christou & Carlton, 2001; Christou et al 2001). The motor output sent form the nervous system to muscles, therefore, must accommodate the force fluctuations for the successful completion of goal-directed movements.

Force Fluctuations in Older Adults

The fluctuations in muscle force during a voluntary contraction have been quantified as the standard deviation of force during isometric contractions and as the standard deviation of acceleration during anisometric contractions in hand, arm, and leg muscles. Although the amplitude of the fluctuations has been found to differ between young and old adults, this is not a consistent finding. The factors that influence this relation include the muscle group performing the task, the type of muscle contraction, the intensity of the muscle contraction, and the physical activity status of the individual.

1. *Muscle Group*

The coefficient of variation for force is greater for old adults compared with young adults when performing low-force isometric contractions with the first dorsal interosseus (hand) and knee extensor muscles, but not for the elbow flexor muscles (Burnett et al 2000; Galganski et al 1993; Graves et al 2000; Hortobágyi et al 2001; Laidlaw et al 2000; Tracy & Enoka, 2002). Furthermore, the force fluctuations during submaximal isometric contractions (2.5 to 50% MVC) were moderately correlated ($r = 0.5$ to 0.7) within the same individuals for the elbow flexor and knee extensor muscles, but not between the first dorsal interosseus and the other two muscles (Tracy et al 2002). Mechanisms related to the lesser upper limit of motor unit recruitment in intrinsic hand muscles may contribute to this difference between muscles.

2. *Contraction Type*
The standard deviation of acceleration for the index finger is much greater when young and old adults use the first dorsal interosseus muscle to lift a light load compared with supporting the load (Laidlaw et al 2001). Although the standard deviation of acceleration is similar for slow concentric and eccentric contractions (Christou et al 2002), the fluctuations are greater for eccentric contractions at moderate-to-fast speeds (Christou et al 2001) and the trial-to-trial variability in motor output is greater for rapid eccentric contractions (Christou & Carlton, 2002; Christou et al 2001). Furthermore, the standard deviation of acceleration is greater during eccentric contractions compared with concentric contractions for old adults when they lift light loads with the index finger (Burnett et al 2000; Laidlaw et al 2000).

3. *Contraction Intensity*
The standard deviation of force increased with target force for all muscles examined in both young and old adults. The coefficient of variation for force, however, declined with low target forces for the first dorsal interosseus (Burnett et al 2000) and knee extensor muscles (Tracy & Enoka, 2002), whereas it increased for the elbow flexor muscles (Graves et al 2000). There was no difference in the coefficients of variation between young and old adults for the elbow flexor muscles, whereas the coefficient of variation was greater for the old adults at low forces with the other two muscles. Similar to the standard deviation of force, the standard deviation of acceleration increased with average muscle power for both concentric and eccentric contractions performed with the first dorsal interosseus muscle (Christou et al 2001). Although the young adults had a greater average standard deviation of acceleration than old adults at a given movement speed, there was no difference between the two groups of subjects in movement accuracy. In contrast, the old adults were less accurate with the movement for the same level of fluctuations in acceleration, had more variable levels of acceleration across trials, and were more variable with eccentric contractions.

4. *Physical Activity Status*
The greater coefficient of variation for force exhibited by old adults at low forces with the first dorsal interosseus muscle disappears after a few weeks of strength training (Bilodeau et al 2000; Keen et al 1994). Furthermore, this effect occurrs for both isometric and slow anisometric contractions whether the subjects train with light or heavy loads (Laidlaw et al 1999). In contrast, there are mixed results on the effect of strength training on force fluctuations during isometric and anisometric contractions with the knee extensor muscles (Hortobágyi et al 2001; Tracy et al 2002).

Mechanisms That Influence Force Fluctuations
The unitary functional element of the neuromuscular system is the motor unit. The net force exerted by a muscle when many motor units are activated results in a force of varying amplitude, the fluctuations of which depend on the contractile and discharge characteristics of the most recently recruited motor units (Allum et al 1976; Christakos, 1982; Erim et al 1999) and are, according to the Size Principle, the largest motor units that have been activated for the task. Based on this rationale, it is evident that the fluctuations in the force exerted by a single muscle can be influenced by the properties of the activated motor units. These include factors that influence the amplitude of the motor unit twitch and those that determine the rate at which the motor neuron discharges action potentials:

1. *Motor Unit Force*
Apoptosis of spinal motor neurons results in a decline in the number of motor units in the muscles of older adults but an increase in the innervation number of surviving motor units (Campbell et al 1972, Tomlinson & Irving, 1977). Accordingly, the spike-triggered average forces of motor units in the first dorsal interosseus muscle are larger in older adults (Galganski et al 1993). Thus, the fluctuations in motor unit force when the unit is first recruited are greater in older adults, which could contribute to the difference in the force

fluctuations at low forces between young and old adults. However, several weeks of strengthening exercises performed with the first dorsal interosseus muscle reduced the force fluctuations in older adults but did not change the distribution of motor unit forces (Keen et al 1994), an effect that was corroborated with computer simulations (Taylor et al 2000; Yao et al 2000). Hence, the larger motor units in the first dorsal interosseus muscle of older adults do not appear to contribute to the difference in the force fluctuations between young and old adults.

2. *Motor Unit Synchronization*
When motor neurons discharge action potentials at or near the same time, which is quantified as motor unit synchronization (Semmler, 2002), there is an increase in motor unit force and the force fluctuations during submaximal isometric contractions (Taylor et al in press, Yao et al 2000). Experimental measurements indicate that older adults do not exhibit greater levels of motor unit synchronization between pairs of motor units during low-force isometric contractions, even though the force fluctuations are greater (Semmler et al 2000). However, we have recently found that motor unit synchronization and the fluctuations in acceleration are greater during eccentric contractions compared with concentric contractions performed with the first dorsal interosseus muscle by young adults (Semmler et al 2001).

3. *Motor Unit Discharge Rate*
The average discharge rate of motor units in the first dorsal interosseus muscle when subjects exert low forces and lift light loads is not different between young and old adults (Laidlaw et al 2000; Semmler et al 2000). However, the discharge rate during these tasks can be more variable and be associated with increased fluctuations in force and acceleration for older adults (Laidlaw et al 2000). Furthermore, discharge rate was more variable and the force fluctuations were greater for eccentric contractions compared with concentric contractions for old adults. Computer simulations, moreover, indicate that an increase in discharge rate variability can cause an increase in force fluctuations during isometric contractions (Taylor et al 2000).

In addition to motor unit properties in a single agonist muscle, a fourth mechanism that could influence the force fluctuations is the pattern of activation of the agonist and antagonist muscles. Because the net force depends on the difference in the force exerted by the agonist and antagonist muscles, alternating activation of the agonist and antagonist muscles can cause fluctuations in force and acceleration. Vallbo & Wessberg (1993), for example, found that young adults used reciprocal activation of agonist and antagonist muscles at 8-10 Hz when performing slow finger movements. Although older adults coactivate the antagonist muscle more often than young adults (Burnett et al 2000: Spiegel et al 1996), we have found no consistent pattern of alternating coactivation or peaks in the power density spectra for acceleration to indicate that the steadiness of isometric and slow anisometric contractions was due to the amplitude of 8-10 Hz cycles. These results suggest that for a simple motor system involving single agonist and antagonist muscles, the force fluctuations depend primarily on the discharge characteristics (variability and perhaps synchronization) of the active motor units.

Most motor systems in the human body, however, comprise multiple agonist and antagonist muscles. Given the exponential decline in the relative contribution of newly recruited motor units to the net force (Fuglevand et al 1993), it seems likely that fluctuations in the force exerted by multiple muscle systems is minimally influenced by the behavior of individual motor units. Alternatively, the fluctuations might be caused by variation in the distribution of activation among the involved muscles (Graves et al 2000) or by rhythmicities that are imposed by the nervous system (Brown, 2000; Halliday et al 1999; Wessberg & Kakuda, 1999).

Despite the functional significance of fluctuations in the motor output for the accuracy of goal-directed movements and its deterioration with advancing age, there is not yet a physiological explanation of the mechanisms that are responsible for this phenomenon.

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INTERPRETATION OF EMG CHANGES WITH FATIGUE: FACTS, PITFALLS, AND FALLACIES

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INTRODUCTION

Failure to maintain the required or expected force, defined as muscle fatigue, is accompanied by changes in muscle electrical activity. Piper (16) was the first who observed a reduction in frequency of the surface EMG (Piper rhythm) when a contraction was sustained. Besides such a frequency shift, Cobb and Forbes (1) found a consistent increase in the amplitude of surface recorded EMG. Since then, the studies can be divided into investigations directed at discovering signs of fatigue (5,6,9-19) and/or reasons for fatigue (2,5,6,13,14,16,19). Opinions agree that muscle fiber propagation velocity (***v***) decreases with fatigue and that EMG power spectrum shifts during fatigue mainly owing to slowing down of ***v*** (5,6,9-16, 18,19). The fact that spectral characteristics can drop even without any changes in ***v*** or in proportion exceeding the ***v*** changes has prompted many authors to conclude that there must be other factors contributing to the observed shift in the power spectrum too. These factors and the way they could affect EMG spectral characteristics have been so far unknown. Voluntary and electrically elicited EMG signals have a similar spectral shift during fatigue (8,15). Simulations performed by (11) showed the power spectra of interference EMG signals and of motor unit (MU) potential (MUP) as essentially similar and dispersion in the action potentials produced by individual fibers within MU as leading to dramatic changes in the MUP power spectrum. Thus, an analysis of factors affecting the spectral characteristics of MUPs or M-waves can help understand the reasons for EMG power spectrum shift.
Changes in amplitude seem to be more complicated and contradictory, since literature provides data on increased, almost unchanged and decreased amplitude characteristics of EMG, M-wave or MUP during fatigue. Moreover, simultaneous decrease and increase in amplitude of MUP and M-wave detected with indwelling and surface electrodes was referred to as paradoxical. Analysis of changes in MUP or M-wave size and shape with fatigue makes it possible to reveal the peripheral factors that contribute (along with the central factors) to change in amplitude and spectral characteristics of EMG signals. The aim of the present report is 1) to analyze possible reasons for EMG changes with fatigue; 2) to point out pitfalls and possible fallacies in the interpretation of amplitude and spectral characteristics of experimental results; 3) to discuss ways for increasing sensitivity of different EMG characteristics to changes in individual muscle fibre parameters.

METHODS

It is known that besides ***v***, other parameters such as duration (T_{in}) of intracellular action potential (IAP), amplitudes of IAP and after-potential, desynchronization of fibre activation, etc. change, or could change during fatigue. Moreover, duration of different phases of IAP alter non-simultaneously and in different proportion. Since individual parameters cannot be affected independently by an explorer, and, as a rule, the source-electrode distance is unknown in experiments, in order to analyze the processes, we have resorted mainly to mathematical modelling of muscle potentials, which does not suffer from these disadvantages. MUPs and M-waves were calculated with the help of a convolution model (3) that included source morphology, finite length of the fibres, and a way to simulate IAP shape without simplification. The volume conductor was assumed to be infinite or semi-infinite, homogeneous and anisotropic (K_{an}=5). We did not take into account the more complicated restriction of the volume conductor (that increased MUP and M-wave size) nor the presence of layers of weaker conductivity (skin and fat) whose effect roughly corresponded to the increase of equivalent source-electrode distances. These parameters and their effects did not change during fatigue but they considerably complicated description and solution of the problem. IAP amplitude, duration, and shape during different stages of fatigue were varied mainly according to the parameters obtained in *in vitro* experiments, and the result from *in situ* experiment was also taken under consideration. Data on ***v*** changes were taken from single fibre experiments. Experimental data on IAP change with fatigue were taken from literature.

RESULTS

Spectral characteristics. Presentation of MUP or M-wave through convolution in the temporal domain is transformed into product in the frequency domain. Thus MUP or M-wave power spectrum (PS) can be represented as a product of PS of the input signal (IAP first temporal derivative) and of PS of the corresponding impulse response (IR). IR depends on the distances between electrode and activated fibres, arrangement and longitudinal position of the electrode with respect to the end-plate area, volume conductor properties, location and scattering of the end plates and fibre ends, desynchronization of fibre activation, diameters of the fibres, and propagation velocity along individual fibres.

The product of the two spectra is distance-dependent because distances affect strongly the position of PS of IR on the frequency axis, while PS of the input signal (IS) is distance-independent. As a result, the effect of any parameter affecting IS or IR on the MUP or M-wave PS is also distance-dependent. Although PS of IS does not depend on ***v***, its effect on the resultant MUP or M-wave PS makes insignificant ***v*** effect under intramuscular recordings. On the other hand, position and arrangement of the electrode affect IR and thus its PS, which results in different sensitivity of MUP or M-wave PS to changes in IAP duration and shape, and especially to changes in negative after-potentials.

MUP or M-wave amplitudes. Besides in convolution form, MUP or M-wave can also be represented as a linear summation of temporally and spatially dispersed single muscle fibre potentials (SMFP). SMFPs have phases reflecting the excitation arising at the end plates (leading edge) and its extinction at the fibres' ends (terminal phases) as well as phases that reflect propagation of depolarized zones from the end plates to the fibre ends. Since formation of the electric field is essentially a spatial problem, size of individual SMFP is determined by the IAP profile whose effect is distance-dependent. So are the MUP or M-wave amplitudes.

A ratio (**R**) between distance (**y**) and length of the depolarized zone (**LDZ**, i.e. product of T_{in} and ***v***) defines an approximate division of distances **y** in an equivalent isotropic volume conductor into small (**R**<<1), medium (**R** $\approx$ 1) or large (**R** > 2) ones. Depending on **R**, the effect of IAP asymmetry (ratio of IAP falling and IAP rising phases), IAP profile length, or/and IAP negative after-potential could predominate at small, medium and large distances, respectively. Following **LDZ** lengthening, the SMFP and MUP or M-wave amplitudes first increase, and then decrease. The larger the source-electrode distance is the greater is **LDZ** at which the corresponding amplitude is maximal. IAP negative after-potential could increase amplitude of SMFP, MUP or M-wave detected far away from the active fibres, but its effect would depend on the electrode longitudinal position. Desynchronization in activation of fibres constituting MU decreases the MUP amplitude at small distances stronger than at large ones.

Changes in the duration of MUP negative phase are determined by alterations in duration of IAP or/and propagation velocity. The effects are also distance dependent. In general, at small and medium distances, changes in T_{in} mainly affect the duration of MUP negative phase. At large distances, changes in propagation velocity and negative after-potentials are the main factors affecting MUP or M-wave duration. However, there are differences between the effects depending on the position of the detecting probe: midway between the end plate area

and tendons, or above the end plate region. In the later case, T_{in} affects negative MUP duration even at large distances.

Amplitude characteristics such as MUP or M-wave area or *rms* combine the effects of changes in MUP or M-wave amplitude and duration that are not directly correlated. As a result, amplitude characteristics can alter in a way different from that of amplitude changes. Nevertheless, the character of changes is distance dependent and MUP area or *rms* could also increase, remain almost unchanged, or decrease during fatigue.

Terminal phases of SMFPs repeat the shape and duration of IAP and do not depend on ***v***; their polarity is positive above the fibres and negative behind the fibre ends. In MUPs or M-waves, their shape represents the sum of IAPs desynchronized according to the arrival times at the fibre ends, which depend on scattering of the end plates and fibre ends and on desynchronization in fibre activation and ***v*** along individual fibres. Duration of terminal phase does not change with the distance from the source, while its relative weight in the detected potentials increases with the increase of source-electrode distance. If a positive after-potential follows the IAP spike, an additional phase should be seen in SMFP, MUP or M-wave, i.e. the terminal phases would become positive-negative above the fibre length and negative-positive behind the fibre ends. On the other hand, if the corner frequency of the high pass filter is 20 Hz, as is generally recommended, a similar additional terminal phase could be seen in MUP or M-wave even when IAP has a negative after-potential.

DISCUSSION

It follows from representing MUP or M-wave through convolution between IS and IR that the properties of potential power spectra should differ from those of IR power spectra (4). Krakau (7) was the first who represented the nerve fibre action potentials through convolution. However, following the expression for the transfer function only, the author made a conclusion about the effect of velocity on the frequency maximum of an action potential. Later, Lindström (9) was interested in the signal filtering in the muscle as a volume conductor and therefore related all expressions to the action potential just outside the fibre. He noted that all formulas, except for the 'correlation factor', contained the factor $2\pi f/v$, which indicated that any change in ***v*** was accompanied by translation along the frequency axis. Following (10), Stulen and Deluca (18) and (11) concluded that ***v*** scaled inversely the frequency components of EMG power spectrum. Along these lines, many authors assumed that the theory predicted unconditional proportional change of characteristic frequencies with ***v***. The results represented in this report demonstrate the effect of IAP and after-potentials that can also affect considerably PS. Data on differences in character of EMG amplitude changes have prompted many investigators conclude that amplitude characteristics are not reliable (19, 5). However, knowledge of differences in changes of amplitude and duration depending on the distance from the source and end plate area could be used to distinguish between the effects of different parameters, e.g. changes in after-potentials or IAP duration. A considerable increase of M-wave amplitude at the beginning of a fatiguing contraction (2) when the Na^+-K^+ pump has hardly been activated could be explained by IAP profile lengthening.

On the other hand, not only fatigue, but the level of muscle force also affects the EMG amplitude and frequency characteristics. To overcome this problem, (12) and (6) have proposed the Joint Analysis of EMG Spectrum and Amplitude (JASA) method. But in accordance with this method muscle fatigue could be misclassified as 'force decrease' in smaller or more superficial muscles, as well as in the final stages of fatigue, when EMG amplitude decreases or during maximal volume contraction (16). On the basis of the above findings we propose an index that mainly reflects the increase of negative after-potentials and does not depend on the source-electrode distance, volume conductor properties, or level of muscle force.

CONCLUSIONS

1) Surface detected EMG amplitude characteristics can increase owing to IAP profile lengthening during fatigue even when IAP amplitude considerably decreases. 2) MUP or M-wave power spectrum is determined by a product of two spectra whose interrelations vary with the source-electrode distance that affect the sensitivity of MUP or M-wave power spectrum to changes in different parameters (IAP duration, propagation velocity, and after-potentials). 3) Appropriate electrode arrangement and position can intensify or decrease sensitivity of amplitude and frequency characteristics to changes in individual parameters.

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SURFACE EMG RECORDING OF SINGLE MOTOR UNIT ACTION POTENTIALS FROM THE EXTERNAL ANAL SPHINCTER

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INTRODUCTION

Electrode arrays have been used for surface EMG recording of single motor unit action potentials (MUAP) propagating from the innervation zone (IZ) to the tendons in muscles such as the biceps brachii, upper trapezius and tibialis anterior. Application of this technique to the circular anal sphincter muscle allows to address the question of the relevance of innervation and asymmetry in fecal incontinence. In addition, knowledge of the innervation zones of the two hemisphyncters would be useful to minimize surgery related muscle damage. It was the objective of this work to determine if innervation zones and other features of the anal sphincter could be detected using electrode arrays placed on an anal probe.

METHODS

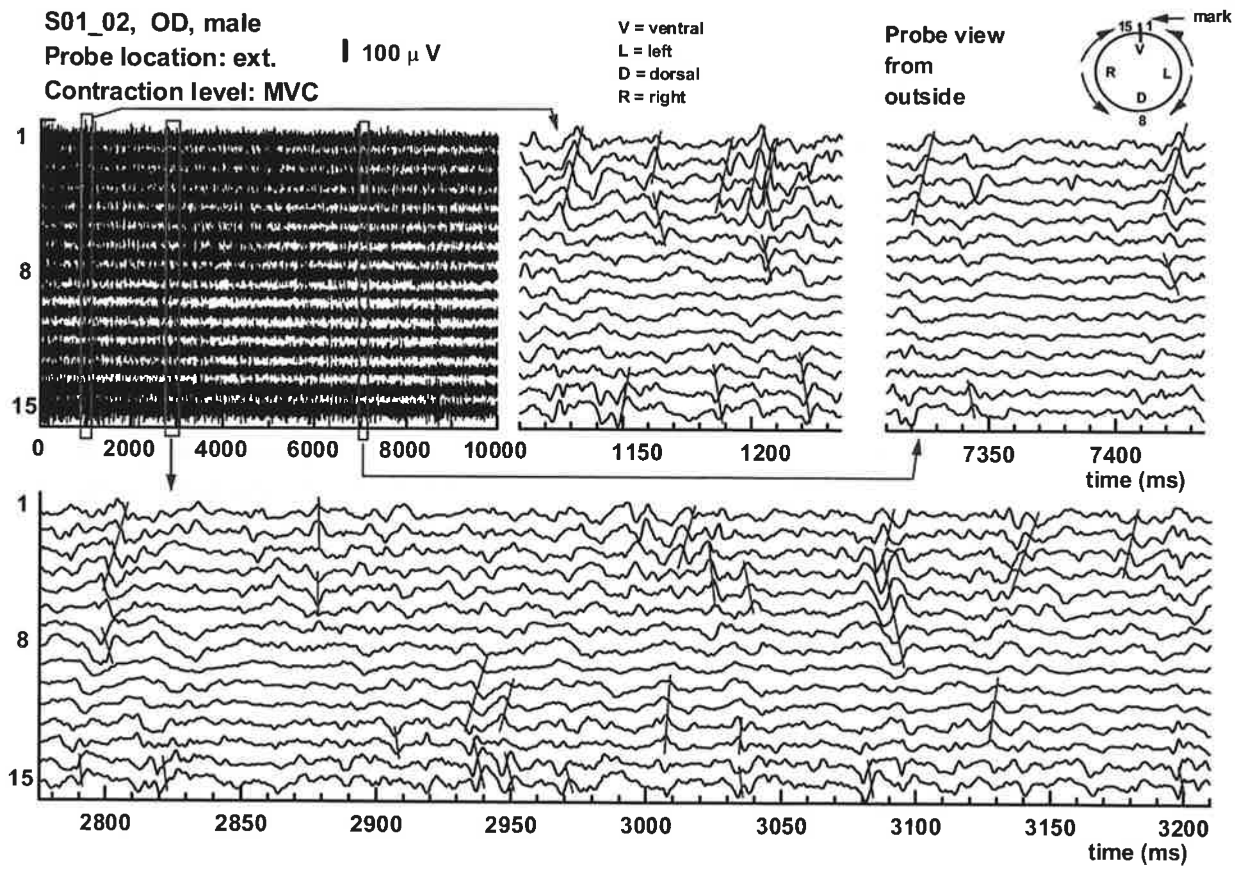
Cylindrical probes, with 12 mm diameter, carrying 16 silver bars (1 mm diameter, 10 mm length, 2.4 mm apart) to record MUAPs circumferentially along the muscle fibers were used to detect EMG from the external anal sphincter muscle near the anal orifice and 1 and 2 cm proximal in 8 healthy volunteers (6 males, 2 females, age 22-41 years.). Each contraction was sustained for 10 s. Signals were recorded differentially between adjacent electrodes during rest and during moderate and maximal voluntary contractions in each of the three locations, and were evaluated to detect the left and right IZs (expressed using a 12-h-clock representations). The EMG signals were amplified (gain of 1000 or 5000, 3dB bandwidth 10Hz-500Hz, 40 dB/decade), sampled at a rate of 2048 Hz and stored on a PC after 12 bit A/D conversion.

RESULTS

Single MUAPs were reliably observed during maximum and moderate contractions and occasionally in relaxed conditions. Two major innervation patterns were noted and described according to a 12 hour clock face: a) ventrally between 11 and 1h in all subjects, mostly bilateral, travelling dorsally at either side; b) in 6/8 subjects dorsolateral at 4-5 h and 7-8 h, mostly bileratal, travelling both ventrally and dorsally. In 6/8 cases, the middle segment of the anal canal showed activity smaller than the distal and proximal end. The averagedistance travelled by a MUAP was 3,5 hours, but could range between 2 and 6h. About 50 % of the MUAPs crossed the midline at either 12 of 6 h. At least half of the subjects showed non-travelling MUAP presumably due to longitudinal motor units or crosstalk from nearby active muscles. MUAPs with apparently different propagation velocities where observed possibly due to different radial distances from the surface of the probe. The figure depicts signals from a contraction showing larger signals on the left side and different MU structures

DISCUSSION AND CONCLUSIONS

Circular electrode arrays reveal individual MUAPs and patterns of functional innervation of the external anal sphincter and allow the identification and characterisation of individual MUs in terms of length, IZ, MUAP morphology and, possibly, conduction velocity and firing rate. The circular EMG electrode array is a useful tool for electrophysiological studies of the anal sphincter for detecting abnormalities of this muscle. Similar conclusions have been reached about the urethral sphincter using a probe with 5 mm diameter and 8 equally spaced electrode contacts.



Results from a maximal voluntary contraction showing larger signals on ch 14-15 and 1 to 8 and smaller signals on ch 9 to 13 presumably due to asymmetry in sphincter structure or control. Innervation zones can be detected under ch. 4, 5 and 6 and under ch. 11, 12 and 13. Non propagating potentials, possibly due to more distant motor units, are visible at t = 2880 ms.

Ch. 1 and 15 are located ventrally, ch. 8 dorsally. Ch 1 to 8 are on the left of the subject and ch 9 to 15 on the right. The subject was asked to contract naximally. Peak to peak amplitude is about 100 μV.

ACKNOWLEDGMENT.

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EMG-BASED APPROACH TO IDENTIFYING FUNCTIONAL MOTOR ACTIVITIES

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INTRODUCTION

In this study we describe the feasibility of a surface EMG-based monitoring system for functional motor activities encountered during daily living.

METHODS

Eleven functional motor activities, derived from the Functional Independence Measure (FIM) scale, were selected for testing; they included eating, grooming, dressing, transfer, ambulation, and toileting activities. Misclassification of tasks not belonging to these 11 FIM activities was evaluated by including an additional set of 11 motor tasks (Non-Identification FMAs) comprised of fine- or gross-motor activities utilizing similar muscle groups and movements as those of the 11 FIM tasks. A procedure that combines feature extraction algorithms, a multi-layer feedforward neural network, and an adaptive neuro-fuzzy inference system was implemented for task identification. Different combinations and numbers of EMG channels (maximum of 8 muscle sites) and different neural network topologies were compared to optimize performance. Performance results from data collected on six healthy volunteers with no known neuromuscular disorders were based on the operating characteristics (sensitivity vs. specificity and sensitivity vs. misclassification) of the algorithms.

RESULTS

For a misclassification value equal to 10%, combinations of 7 and 8 channels lead to sensitivity values of approximately 90% and specificity values of approximately 99%. For combinations of 4 to 6 channels, a 10% misclassification corresponded to sensitivity of approximately 82%, and specificity of approximately 99%.

CONCLUSION

These results indicate that EMG technology can be used to identify specific motor activities. As few as 4 EMG channels may be sufficient for practical applications.

EVALUATION OF CHANGES IN THE JOINT ANGLE ON BASIS OF NON-INVASIVELY EXTRACTED MOTOR UNIT ACTION POTENTIALS

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INTRODUCTION

Changes in muscle fiber length and electrode position affect the detected signals as well as their amplitude and frequency characteristics (1) used to estimate muscle functional state. Studies of motor unit potentials (MUPs) under extremely weak muscle activity (2) or M-responses under supramaximal nerve activation (3) have shown that changes in joint angle are correlated with the length of active muscle fibers and with latencies of certain phases of muscle potentials. The time between the maxima of negative phase and positive terminal phase of MUPs was measured in (2), while latency of the negative phase of M-responses detected monopolarly behind the *m. biceps brachii* tendon was measured in (3). Actual muscle activity is also accompanied by changes in joint angles; it is, however, neither extremely weak nor maximally synchronized. To interpret actual SEMG properly, one should know the actual joint angle or fiber length. It is practically impossible to detect activity of individual MUs under voluntary contraction. Using M-responses is also not appropriate. The aim of the present study was to investigate the possibility of extracting information on the active fiber length from SEMG signals (obtained under voluntary contractions) so as to correlate fiber length and joint angle.

METHODS

Theoretical basis of the method: To estimate muscle fiber (semi-) length, we extracted MUPs from SEMG signals. Then, the delay between onset of excitation at the end plates and its extinction at the fibers' ends was used to relate the length of the active fibers with changes in the joint angle, having regard to the excitation conduction velocity. Monopolar SEMG recordings suffer from a low signal quality. A common technique to obtain the required information from potentials recorded monopolarly, is averaging. This can be done non-invasively using the same data set for triggering, after spatial selectivity is enhanced by using spatial filters (4). To do so, we used the two dimensional Laplacian, NDD filter. From the spatially filtered signals, the MUPs were assigned to the single motor units (sMUs), and tracked with time. Then, the monopolar signals were averaged over several MUPs of the same MU. *Experimental protocol:* High spatial resolution EMG (HSR-EMG) was utilized to detect muscular activity from *m. biceps brachii* non-invasively using a multi-electrode array that allows the monopolar detection of the SEMG. To record the exact timing of the MUP genesis at the end plates, the array was located above the distal innervation zone. Measurements were performed on three healthy subjects. While the joint angle in the shoulder was fixed in 90° abduction, the elbow angle was varied over the range 70° to 160° in 10° step, 180° indicating full extension of the elbow. The succession of angles was random. To avoid fatiguing effects, the subjects were given breaks of at least two minutes between the single measurements. A moderate muscle activity was performed at each joint position during 10 sec.

RESULTS

Using the information contained in monopolarly recorded single MUPs, relative changes in the muscle fiber length and thus in the joint angle were extracted from the SEMG data [Fig.1]. Initial estimates demonstrated the correlation between the elbow angle and fiber length. No dependence between conduction velocity and joint angle was found.

DISCUSSION

The method described presumes a monopolar recording technique. It does not reduce the end-of-fiber effects and provides the possibility to follow sMUs over time. The sMUs can be obtained directly from spatially filtered HSR-EMG data that can be also used to trigger averaging necessary to extract monopolar MUPs. A strong dependence of the conduction velocity on the joint angle described in literature by some authors also in case of the human *m. biceps brachii*, was not found in the actual data. Consequently, the change in the latency between the excitation genesis and its run out at the fibers' ends was caused by the change in the fiber semi-length only. The fact, that the relative change in the estimated fiber length does not go linear with the change in the joint angle might be due to the anatomical properties of this specific muscle or to the fact that not the same MU was always extracted under different joint angles.

CONCLUSION

Although the need to average the MUPs in order to increase the signal quality of the monopolarly recorded signals limits the use of the methodology described, the HSR-EMG method might prove to be a useful tool to extract information on the active fiber length from surface detected EMG data in static, slowly changing or repetitive movements. Moreover, the methodology has shown the basic advantage that it provides information on the muscle fibers semi-length that is defined on the very same single MUs investigated by the HSR-EMG on whatever other parameters are of interest.

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ACKNOWLEDGEMENT

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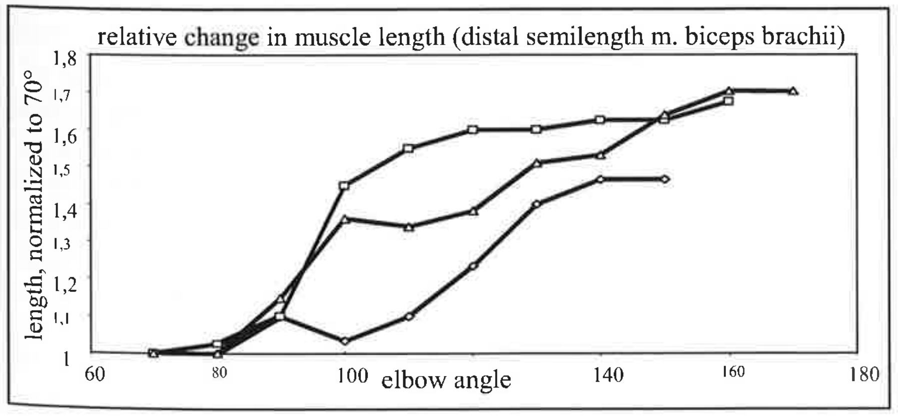

Fig.1. Normalized estimate of *m. biceps brachii* distal semi-length obtained for different elbow angles. Different curves represent data for three subjects studied.

TIME – FREQUENCY ANALYSIS OF MYOELECTRIC SIGNALS COLLECTED DURING CLENCHING

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INTRODUCTION

The role of muscle contraction in tension-type headache is still controversial. In a previous study we demonstrated that the electrical manifestations of muscle fatigue during isometric and constant-force contractions were not different in healthy and pathological subjects. Based on this result, we decided to extend our study to myoelectric signals collected in dynamic conditions during repeated clenching. This contribution reports the results of this study, that is the first application of time-frequency analysis to the study of the role of muscle disorders in tension-type headache.

MATERIALS AND METHODS

The sample population involved in this study consisted of 11 subjects, 2 males and 9 females, aged between 20 and 40. All the subjects suffered from tension-type headache, defined according to the International Headache Society classification. From the point of view of disorders of the pericranial muscles, one subject was had no symptoms, four subjects reported pain on the left side of the face, one on the right side, and the remaining four subjects on both sides. Pain was assessed through a visual analog scale. Each patient underwent clinical examination and a muscle fatigue test, which consisted of two series of continual contractions of jaw muscles lasting 1s each and repeated for 60s.

Myoelectric signals were acquired by means of two single differential probes connected to a myoelectric signal amplifier (Bagnoli2, Delsys, USA) and positioned on the right and left masseter muscles. The acquired signals were segmented into bursts of activity corresponding to the contractions of the masseter muscles by means of a double–threshold statistical detector [1], to isolate the electrical muscle activity corresponding to each contraction. Then, each signal burst was processed to obtain its time–frequency representation. To this purpose, we adopted the Choi–Williams time–frequency transform, which belongs to the Cohen's class of time–frequency bilinear distribution. These distributions have already proven useful in previous studies for the detection of the spectral changes affecting the surface recorded EMG signal during dynamic contractions [2,3]. Moreover, we used an original cross–time frequency based approach for estimating the instantaneous mean frequency (IMNF) of the signal bursts.

Plotting all the IMNF curves corresponding to either the left or the right masseter muscle of each single subject normalized over one thousand samples, it was evident that the IMNF time courses within each contraction were sufficiently similar, and hence we decided to represent them by their mean curve. Two different kinds of shapes were clearly recognized. In the first case we observed no trend from the beginning to the end of the burst, while in the second case we observed an increment of IMNF followed by a decrement. The mean curve was than used to classify the subjects.

RESULTS AND DISCUSSION

The time course of the mean IMNF curves observed on our sample population could be classified according to three different types of behavior:

1. Left and right masseter muscles showed an IMNF behavior clearly different, and this was seen in 3 out of 11 subjects;
2. Left and right masseter muscles showed an IMNF behavior that was similar and consisted of an increment of IMNF followed by a decrement, and this was seen in 4 out of 11 subjects;
3. Left and right masseter muscles showed an IMNF behavior substantially constant, and this was seen in 4 out of 11 subjects.

Although the number of subjects considered in this study was rather small, and hence results may not be considered as statistically significant, there was a clear match between the type of behavior of the IMNF time course of left and right masseter muscles and the clinical conditions of patients:

a) complete clinical examination showed that the subjects belonging to group 3 did not experience pain related to the masseter muscles.
b) On the contrary, subjects belonging to group 1 and 2 reported pain on either one or both masseter muscles and could further be differentiated according to their clinical situation as shown by the following table:

	Group 1	Group2
Time passed from the first manifestation of pain	More than 2 years	Less than 2 years
Pain intensity (VAS)	From 18 to 48	From 59 to 100
Factors increasing pain	Stress and cold	Muscle activation

A possible interpretation is that subjects suffering from more intense and long lasting pain developed different strategies of activation of the masseter muscles trying to reduce their discomfort. On this point further studies are needed.

Notwithstanding the small sample population size, we believe that this test is very reliable to separate subjects in which masseters were, at least to some extents, responsible for tension type headache from those in which masseter muscles were not a cause of pain.

CONCLUSIONS

This study reports some preliminary results obtained by applying muscle fatigue evaluation in subjects suffering from tension type headache. It was demonstrated that the proposed technique is able to separate subjects in different groups according to their clinical situations and symptoms. Although the sample population we considered is rather small, we believe that the results we obtained must be considered as encouraging and justify the extension of this study to a larger population.

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MOTOR UNIT RECRUITMENT AND PROPRIOCEPTIVE FEEDBACK DECREASE THE COMMON DRIVE

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INTRODUCTION

During voluntary contractions the firing rates of motor units are cross-correlated indicating that the motor units have a Common Drive. We explored the relationship between the recruitment of a motor unit and the fluctuation of the Common Drive in three muscles having varying amounts of muscle spindles.

METHODS

In this study we investigated the common drive during isometric constant-force contractions and isometric ramp contractions in the First Dorsal Interosseous muscle, the Trapezius muscle and the Tibialis Anterior muscle. A time-frequency analysis technique was used to calculate the time course of the common drive throughout the duration of the contraction.

RESULTS

We found that in most contractions the Common Drive was not constant throughout the contraction. A decrease in the common drive was noted when a motor unit was recruited during the contraction. The amount of decrease in the Common Drive across muscles was found to be proportional to the density of muscle spindles in the muscles and to the number of motor units recruited during the contraction.

DISCUSSION

It is proposed that when a motor unit is recruited, the unfused muscle fibers slacken nearby spindles, the Ia firings decrease thus decreasing the excitation to the homonymous motoneuron pool. The contracting fibers will also excite the Golgi tendon organs, which will increase the discharge of the Ib fibers, thus increasing the inhibition to the homonymous motoneuron pool. Both effects will increase the discord of the firings of the motor units resulting in a decrease in the Common Drive. This account explains why the common drive has been reported to be lower: in muscles with a lesser spindle density, in contractions where motor units are recruited, in contractions of the elderly where the force output is less steady, in contractions of the non-dominant side where the force output is less steady, and in contractions where the proprioceptive input is altered.

DURING FATIGUING CONTRACTIONS VASTUS LATERALIS MOTOR UNITS SHOW COMMON FIRING ADAPTATIONS

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INTRODUCTION

The force level of a voluntary muscle contraction is controlled through two mechanisms: recruitment of motor units and modulation of the firing rate of active motor units. The general belief is that a motor unit, once recruited, tends to remain active during a constant force contraction, although alterations in the recruitment of motor units during sustained contractions have been the subject of discussion for decades. This project was designed to systematically investigate changes to motor unit firing behavior during long-duration muscle activity.

METHODS

Motor unit firing patterns were studied in the vastus lateralis (VL) muscle during a series of isometric knee extensions. Five healthy, young men, age 20 – 22years, participated in the study. The fatigue protocol required subjects to control knee extension torque by following a target trajectory on a computer screen. Each ramp and hold contraction in the series lasted one minute followed by a 6-second rest period. The contraction level of the hold phase was set to 20% of the maximal voluntary contraction (MVC) torque of each subject, measured at the beginning of the experimental session. A quadrifilar fine wire electrode was inserted into the muscle to record electromyographic (EMG) signals as subjects repeatedly contracted the muscle until exhaustion. The Precision Decomposition technique, developed at the NeuroMuscular Research Center, was used to identify the firing times of concurrently active motor units from the intramuscular EMG signals.

RESULTS

The threshold forces for recruitment, normalized to the initial MVC value and the firing rates of the same set of motor units were tracked at regular intervals during the fatiguing, intermittent contraction. Preliminary analysis showed a steady decrease in the normalized recruitment threshold of all active motor units and the progressive recruitment of new units, without a change in recruitment order. Earlier recruited, low threshold motor units always exhibited higher mean firing rates than higher threshold units, but all units displayed similar time-dependent firing rate adaptations. While the knee extension torque was held constant, motor unit firing rates changed from an initial slow decrease over time to a slow increase over time.

DISCUSSION

These results indicate that in the VL, the ordered recruitment and firing rate behavior commonly observed in skeletal muscle is also maintained during fatigue. In contrast to previous reports on other skeletal muscles, no evidence was found for a divergent adaptation of motor unit sub-populations based on the initial recruitment threshold. In light of the observed collective changes in firing behavior, the adaptations in the motor unit activation might simply reflect changes in the drive to the motor unit pool. Thus, in order to achieve the imposed motor task of a constant torque output, the central drive was modulated in such a way as to accommodate changes in the mechanical output of the motor units in the fatiguing muscle.

ACKNOWLEDGEMENTS

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CRITICAL EVALUATION OF CONTINUOUSLY ACTIVE MOTOR UNITS IN LONG-TERM EMG SIGNALS

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INTRODUCTION

Long-term analysis of neuro-muscular systems, such as the analysis of the development of chronic muscle pain, requires intramuscular electromyogram (EMG) recordings that may last for several hours. It is obvious that the performance of the decomposition technique affects the physiological investigation, both the statistics of the interpulse intervals and the motor unit action potential (MUAP) waveform characteristics. In this paper we present a critical evaluation of decomposition results of long-term EMG signals that were automatically decomposed by our decomposition software (EMG-LODEC) [1].

METHODS

30 minutes computer work was performed. Muscle activity was recorded from the upper trapezius muscle using four fine wires as electrodes, and was analysed by EMG-LODEC [1]. EMG-LODEC describes action potentials by a set of wavelet coefficients that are used for the clustering, supervised classification, and to describe the goodness of fit. To test the decomposition results, the different motor unit (MU) classes are compared to each other using two morphological properties of the MU classes the energy difference ΔE and the shape difference between the MU classes d [2]. ΔE is calculated as $\Delta E = E_i - E_j$, where E_i and E_j are the energies of the MUAP templates (averaged over time) of class i and j, respectively. To calculate d, first the boundaries of the adapted MU class mean signal are determined (confidence interval of the adapted MU template is limited by the 95th percentile of only these MUAPs that characterize a class). d represents the difference in shape between the average templates of each MU regarding the estimated time aligned boundaries. If a template lies completely between the boundaries, d is 100%. The corresponding ΔE and d for all pairs of MU classes were calculated and when ΔE is smaller than 0.2 (20% energy error) and d is larger than 70%, the corresponding MU classes are merged. The threshold were chosen due to observed variations in amplitude of the same MU, which can make up more than 30% of the maximal peak-to-peak amplitude [2]. The information of all three channels were used.

Furthermore, the performance of MUAPs' classification in the context of the extracted wavelet coefficients are described by a certainty function. The certainty $c_{a,r}(\underline{x})$ in classifying a MUAP $\underline{x}$ to a particular class is defined as

$$c_{a,r}(\underline{x}) = c_a(\underline{x})c_r(\underline{x}), \tag{1}$$

$$c_a(\underline{x}) = 1 - \frac{\left\|\underline{x} - \overline{\underline{x}}_{j_0}\right\|^2}{\left\|\overline{\underline{x}}_{j_0}\right\|^2}, \tag{2}$$

$$c_r(\underline{x}) = 1 - \frac{\left\|\underline{x} - \overline{\underline{x}}_{j_0}\right\|^2}{2\left\|\underline{x} - \overline{\underline{x}}_{j_1}\right\|^2}, \tag{3}$$

where c_a is the absolute shape similarity, c_r is the relative shape similarity, $\overline{\underline{x}}_{j_0}$ is the nearest class men vector and $\overline{\underline{x}}_{j_1}$ is the second nearest class mean vector. The bigger $c_{a,r}(\underline{x})$ is the better a MUAP is classified. On the basis of equation (1) four quality criterions are defined:

criterion 1: $c_{a,r}(\underline{x}) \geq 0.9$ (MUAP is correctly classified), (4)

criterion 2: $0.8 \leq c_{a,r}(\underline{x}) > 0.9$ (MUAP is in all likelihood correctly classified), (5)

criterion 3: $0.6 \leq c_{a,r}(\underline{x}) > 0.8$ (MUAP is probably correctly classified), (6)

criterion 4: $c_{a,r}(\underline{x}) < 0.6$ (MUAP is possibly falsely classified (e.g. strong superimposed MUAPs)). (7)

To further determine the accuracy of the decomposition, visual plots (waterfall plots showing the MUAP's waveforms of the respective ten second interval) were used, as shown in Fig. 1. These waterfall plots and the goodness of fit of the wavelet coefficients enabled us to verify possible misclassifications, i.e. if different trains of MUAPs were correctly merged. Besides the visual plots characteristic variables, i.e. peak-to-peak amplitude, shimmer variance and phase stability were calculated.

RESULTS

The decomposition of the 74 intramuscular signals identified 887 different MUs. Thirteen MUs were continuously active throughout the 30 minutes task. In eleven of these the decomposition achieved good accuracy, in the remaining two MUs it was moderate.

DISCUSSION

The goodness of fit based on the wavelet coefficients, the calculated variables and the visual plots enabled us to test the quality of the decomposition results that included several thousands of single action potentials.

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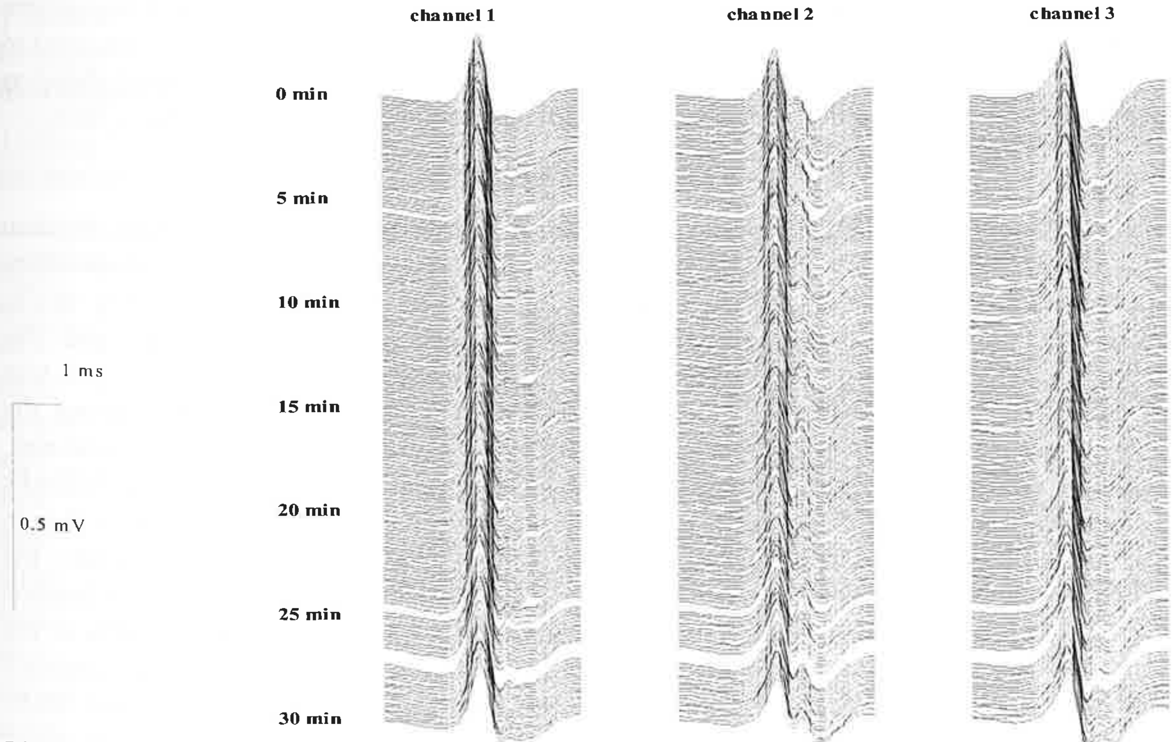

Figure 1: Waterfall plot of MUAP templates for a Cinderella MU that is active throughout the entire computer task. The MUAP templates are updated every 10-seconds. The MUAP templates are plotted for all three channels. (ΔE=80.4% and d=54.8% compared to the most similar separated MU class; 87.7% MUAPs with criterion 1 and 2, 1.9% MUAPs with criterion 4.)

MOTOR UNIT FIRING PATTERN IN THE TRAPEZIUS MUSCLE DURING LONG-TERM COMPUTER WORK

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INTRODUCTION

Muscle pain in the shoulder/neck area and the upper extremities is a common and increasing problem among computer workers (Öberg and Åström, 2000). Hägg (1991) formulated the "Cinderella" hypothesis, proposing that the pain is caused by an overuse of low-threshold motor units (MUs). While this hypothesis is supported by laboratory studies that showed continuous activity of single MUs during e.g. long-term (60-minute) static contractions (Thorn et al., 2002), MU substitution (a shift from previously active MUs to newly recruited ones) has also been observed. It is unknown if the MU behaviour in an experiment more similar to long-term computer work may differ from that during a solely static muscle contraction. The purpose of this study was to investigate MU firing patterns in the trapezius muscle during a long-term computer work task. Specifically, are MUs continuously active during such work?

METHODS

Four subjects participated in the study. Surface and 3-channel intra-muscular fine-wire EMG (IEMG) was recorded from the right trapezius muscle during a 60-minute combined mouse- and keyboard work task, which consisted of editing a text where every 20th word was in boldface. The subject was asked to double click on each boldfaced word, un-bold and retype it. A semi-automatic classification program, EMGLODEC, (Zennaro et al., 2001), was used to decompose the signal into motor unit action potential (MUAP) trains. A MU was defined as active when its low-pass filtered (0.5 Hz) pulse train (of identified firings) was above zero.

RESULTS

Subject 4 was excluded from the study due to poor quality of the IEMG signals. Median surface EMG (SEMG) levels for Subjects 1-3 were 7.1, 14.1 and 3.3% of maximal voluntary electrical (MVE) activity, respectively. The average gap frequencies (SEMG <1% MVE, during ≥1/8 s) and the relative time with SEMG <1% MVE were less than 1 min^{-1} and 1%, respectively, for Subjects 1 and 2, and 13 min^{-1} and 13.4%, respectively, for Subject 3. It was observed from video recordings that Subjects 1 and 2 lifted their shoulders during the keyboard input task, which induced periodical SEMG increases and less gaps and muscle rest. The average classification rate of decomposed IEMG segments into one or several MUAPs was 87% for Subject 2. For Subjects 1 and 3, there were decomposition problems due to external signal disturbances and too large pick-up volumes of the electrodes. Therefore, the classification rate for Subjects 1 and 3 was only 59 and 42%, respectively. Classification results from Subject 2 showed 10 out of 15 identified MUs to be firing during >90% of the working time while only 1 was active during <70% of the working time (Figure 1). Subject 1 showed a somewhat lower degree of long-term MU activity; among the 15 identified MUs, the activity percentage was >90% for 5 MUs and <70% for 4 MUs. For Subject 3, none of the 12 identified MUs showed an activity percentage >90% while 6 were active <70% of the working time. A low classification rate, as for Subjects 1 and 3, induces a negative bias in the estimated firing rate, which should be considered when interpreting the result.

DISCUSSION AND CONCLUSIONS

The existence of long-duration active MUs, in at least one of the studied subjects, supports the Cinderella hypothesis in the context of computer work. Inter-individual differences in MU activity behaviour have previously been reported from more standardised tasks (e.g. Thorn et al., 2002). However, as a caveat, the present study contains results from only 3 test subjects, where 2 subjects showed a low rate of segment classification. Under these circumstances it is not possible to make any population-level conclusions about the MU activity behaviour among healthy computer users.

Changes in MU activities that could possibly be MU substitutions were found in all studied subjects, which may mean that only some of the registered low-threshold motor units follow the concept of the Cinderella hypothesis. These substitutions may also be related to high overall MU activity.

In conclusion, the study showed an existence of long-term active MUs in a computer work task. This result supports the Cinderella hypothesis. Indications of substitution were also found. The determinants for the prevalence of substitution as well as inter-individual differences are still not completely understood. Therefore, the present study will be enlarged in the near future with more participating subjects.

ACKNOWLEDGEMENT

We want to thank Ms. Gunilla Zachau for her invaluable support during the analysis work.

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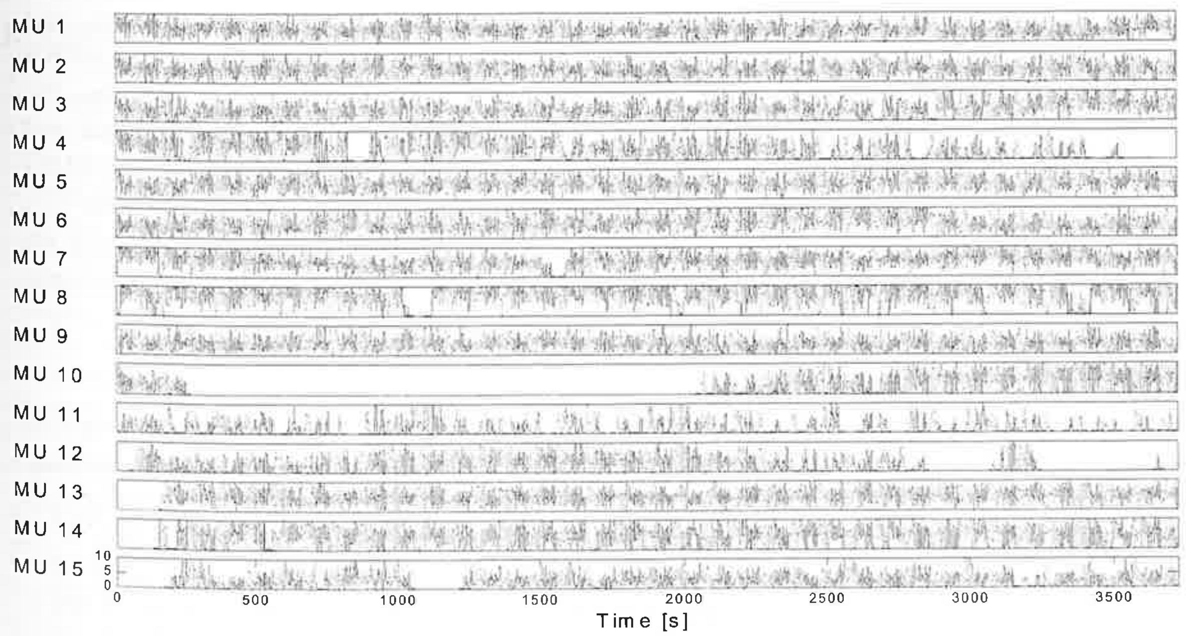

Figure 1: Firing frequencies (Hz) for identified MUs in Subject 2 during the 1-hour task.

EFFECT OF EXPERIMENTAL MUSCLE PAIN ON MOTOR UNIT CONTROL AND CONDUCTION PROPERTIES

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INTRODUCTION

There are a few studies which report inhibition, due to experimental muscle pain, of single motor unit (MU) activity by the use of intramuscular EMG signal decomposition. These findings are in agreement with the pain adaptation model [4] which predicts decrease of muscle activity in presence of muscle pain. Past studies [1][5] addressed the analysis of single MU activity during experimental muscle pain with two basic limitations: 1) only one level of pain (pain off or pain on) was assessed, thus it was not possible to investigate the firing rate modulation by the intensity of pain, 2) only MU control properties were investigated, without any indication on MU conduction properties (peripheral changes in the neuromuscular system). The present study aimed at investigating single MU conduction velocity (CV) and firing rate during stimulation at different levels of nociceptive muscle afferents in constant force isometric contractions.

METHODS

The tibialis anterior muscle of 12 subjects (5 males, 7 females; mean age = 23.2 years, std = 2.2 years) was investigated. The study was conducted in accordance with the Declaration of Helsinki, approved by the Local Ethics Committee, and written informed consent was obtained from all participants prior to inclusion. The subject sat comfortably on a chair with his/her foot fixed in an isometric force brace incorporating one torque transducer. The angle of the ankle was fixed at about 90 degrees while the angle of the knee varied between 110 and 130 degrees depending on the height of the subject (the thigh was always in the horizontal position). Muscle pain was induced by three intramuscular injections of hypertonic saline (5% NaCl, 0.2, 0.5 and 1 ml) separated by 140 seconds each. In this way, three well-defined levels of pain were obtained. Six isometric contractions (two for each pain level) of 20 seconds separated by 50 seconds at 10% MVC were performed after each of the saline injections. Surface EMG (with an array of 16 electrodes, interelectrode distance 5 mm) and intramuscular EMG (by four fine wire electrodes) signals were recorded and processed with a technique (based on spike triggered averaging) we recently proposed [2]. Briefly, the intramuscular EMG signals were decomposed to have indication on single MU firing rate and to average the multi-channel detected surface EMG signals. From the averaged surface EMG signals the CV of single MUs was estimated by low-variance multi-channel methods [3]. During the painful condition the subject was asked to continuously score the pain intensity on a 100 mm visual analogue scale (VAS) of pain with the lower extreme labeled "no pain" and the upper extreme labeled "highest pain imaginable".

RESULTS

The average VAS response from the 12 subjects in reported in the Figure. With the protocol designed it was possible to obtain a clearly increasing level of subjective pain. The force was constant in all 72 recordings. A total of 54 MUs were identified from the intramuscular and surface signals. The mean MU firing frequencies were 10.75 pps, 10.27 pps, 10.03 pps, 9.77 pps, 9.72 pps, and 9.62 pps for the six contractions. A one-way ANOVA of the mean MU firing frequency with the pain level as independent factor was highly significant ($p<<0.001$). The post hoc Student-Newman-Keuls test disclosed pair-wise differences ($p<0.01$) of mean

firing rate between the first three contractions and between these contractions and the last three. The mean firing rates in the last three contractions were not significantly different between each other, although there was a tendency for a decrease. Firing rate was correlated with the level of pain. The mean single MU CVs were 3.91 m/s, 3.88 m/s, 3.89 m/s, 3.83 m/s, 3.87 m/s, and 3.86 m/s in the six contractions. A one-way ANOVA of single MU CV with the pain level as independent factor was not significant and CV was not correlated with the pain level.

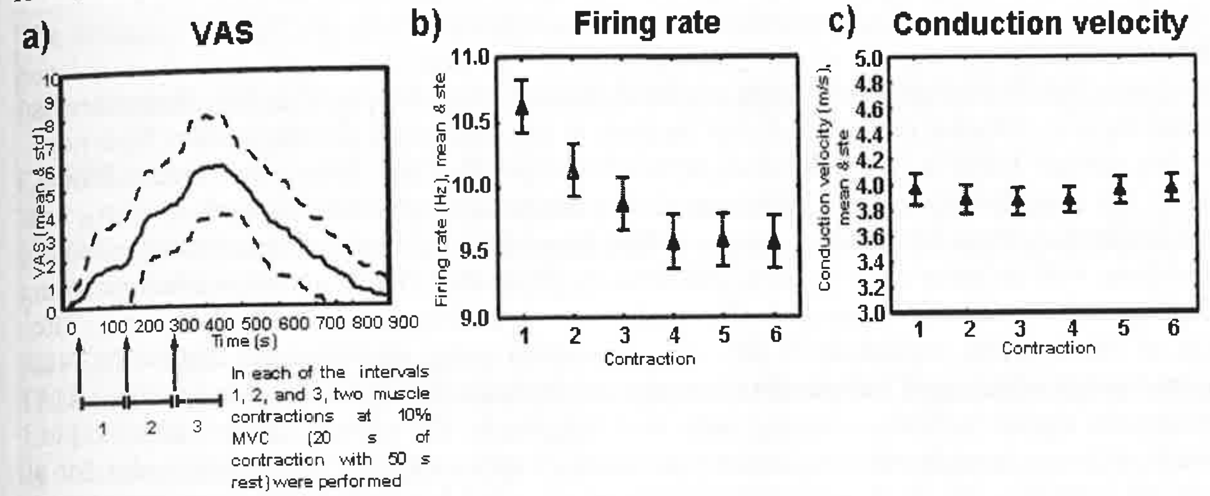

Figure. a) Mean and standard deviation of the VAS provided by the subjects during the painful contractions. The three instants of time corresponding to the injection of the hypertonic saline are indicated with arrows. b) The mean (± standard error of the mean) firing rate of the detected MUs for the six contractions. c) The estimated conduction velocity (mean ± standard error of the mean) of all the detected motor units during the six contractions.

DISCUSSION AND CONCLUSIONS

This study showed that experimental muscle pain induces inhibition of MU firing rate by a reflex activity of the afferents of groups III and IV. In addition, in painful conditions it was found that single MU CV does not significantly change, suggesting that the modulation of firing rate in constant force contractions is not related to changes of peripheral MU conduction properties but rather to a central mechanisms.

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MOTOR UNIT DECOMPOSTION OF SURFACE EMG USING MULTICHANNEL BLIND DECONVOLUTION

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INTRODUCTION

Motor unit (MU) decomposition from needle EMG (1) is widely used today. However, its measurement is invasive and it would not be easy to use. Although decomposition from non-invasive surface EMG is desirable, such signal decomposition has been considered difficult, even if not impossible, because both spatial and temporal resolutions of surface EMG are much lower than those obtained by needle EMG. Nevertheless, a multichannel measurement, easily done with an array of electrodes, can actually allow the extraction of much information including amplitude, waveform and phase of motor unit action potentials (MUAPs). The study of blind signal separation in MU decomposition using multichannel electrodes was reported in (2) which used independent component analysis (ICA). Unfortunately, since ICA decomposes signals basically utilizing only their amplitude, the method cannot, in principle, identify different waveforms or different phase-delays appeared on multiple electrodes for a single MUAP. Thus practical application of ICA needs careful deposition of electrodes placed linearly in the transverse direction to muscle fibers. Considering the fact that waveforms and phase-delays are actually different among channels, deconvolution is needed in processing surface EMGs measured by electrodes array. In this paper, we apply multichannel blind deconvolution method, as an automatic MU decomposition, to surface EMGs measured by square array of electrodes, and also discuss about our results.

METHODS

A 5 by 4 array of Ag/AgCl electrodes (2mm in diameter for each) was attached on the left forearm of a subject. The electrodes were placed separating by 10mm along muscle fibers and by 4mm vertically. 16-channel bipolar surface EMGs were recorded for each pairs of adjacent electrodes along muscle fibers. The measurement was done with weak voluntary isometric contraction of a ring finger's flexion for about 5 minutes. Signals were recorded with 12-bit A/D conversion at 1kHz sampling rate, and were filtered by 250Hz cut-off anti-aliasing filter and by 2.5Hz digital HPF.

The surface EMG can be regarded as a signal which passed through noncausal filter, and its transfer function has nonminimum phase characteristics. Therefore, we apply multichannel blind deconvolution of nonminimum phase systems(3) for MU decomposition. An n-dimensional output vector $\mathbf{y}$ is linearly computed through noncausal filter $\mathbf{W}(z)$ from an n-dimensional input vector $\mathbf{x}$ as

$$\mathbf{y} = \mathbf{W}(z)\mathbf{x} = \mathbf{L}(z)\mathbf{R}(z^{-1})\mathbf{x}, \qquad \text{(eq.1)}$$

where $\mathbf{W}(z) = \sum_{p=-N}^{N} \mathbf{W}_p z^{-p}$, z^{-1} is the delay operator, $\mathbf{W}_p$ is an $n \times n$ -dimensional coefficient matrix at time lag p and $\mathbf{W}(z)$ is decomposed into a cascade form of two FIR filters. $\mathbf{L}(z)$ and $\mathbf{R}(z^{-1})$ are updated by information backpropagation method for each sample.

RESULTS & DISCUSSION

400ms samples of the recorded EMGs and three selected outputs are shown in Fig. 1. The other outputs (not shown here) are visually discarded as noise. Each of these three output signals seems to indicate decomposition into a single MU, and the waveforms like delta-function show successful deconvolution for the entire MUAP transfer function. There are eight pulses from three MUs during the 400 msec, and in the middle of the figure, we can observe that three MUs are firing simultaneously. We detected peaks of each MUAP train, and examined distributions for amplitude and firing period. They are unimodal and it is consistent with physiological knowledge on a single MU. We also calculated the i-th column vectors of $\mathbf{W}^{-1}(z)$, which represent the MUAP waveforms for the i-th output component. The obtained MUAP waveforms corresponding to the selected output components were also consistent if they were assumed to be decomposed into a single MU. Moreover, the innervation zones could be obviously discriminated where the phases of MUAP is reverse.

CONCLUSION

We performed automatic MU decomposition of multichannel surface EMGs by using multichannel blind deconvolution of nonminimum phase systems, for which any specific prior knowledge about EMGs is not necessary.

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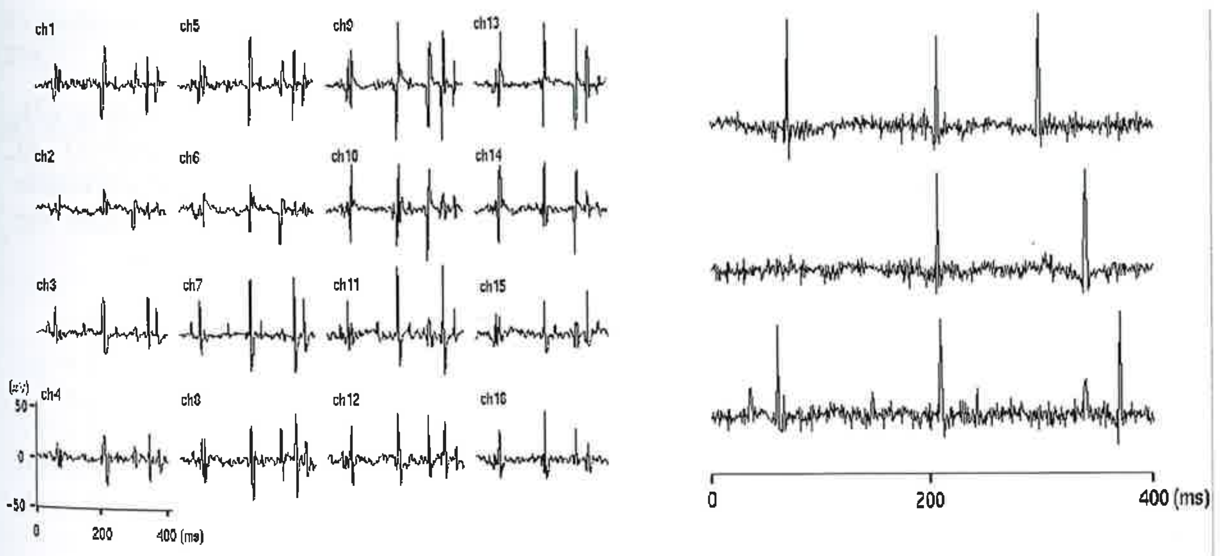

(a) Surface EMGs (b) Decomposed Signals

Figure 1: Multichannel surface EMGs (a) recorded with square array of electrodes, and separated and deconvolved outputs (b) which were selected and regarded as three MUs.

DETECTION OF MOTOR UNIT ACTION POTENTIALS IN MULTI-CHANNEL SURFACE EMG DATA USING 2-D SPATIAL AND TEMPORAL INFORMATION

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INTRODUCTION

Knowledge about individual motor unit characteristics (MU) gives the most important information about the muscle's function and structure. In addition, the diagnosis of neuromuscular disorders is largely based on abnormal MU characteristics. Information about active MUs can be obtained by recording their action potentials using EMG. This is usually done by needle EMG, but surface EMG is non-invasive and therefore preferable. However, during a voluntary muscle contraction the potential distribution on the skin-surface is a superposition of all active MU action potentials (MUAPs), a so-called interference pattern.
This work presents a new method to detect MUAPs in 130-channel surface EMG recordings and to estimate the corresponding muscle fiber orientation and conduction velocity (CV). Since MUAPs propagate along the muscle fibers, a grid of 13 by 10 electrodes can be used to sample potentials from muscle activity at the skin's surface. The aim is to use such sequential amplitude distributions to detect moving components. Detection is carried out using 3-D cost analysis. However, in order to achieve estimation precision in features of the EMG, additional processing has to be done. An approach to estimate CV and fiber orientation is done using sequential 2-D Gaussian function fitting and linear regression.

METHODS

The MUAP detection was based on 3-D cost analysis (1) and dynamic programming (2). Basically, the detection process found all possible MUAP paths (trajectories) in spatiotemporal data, using weighted path conditions. The conditions were based on amplitude and deviation criteria, and the process associated each path with a specific cost where the optimal path had the minimum cost. The problem can be described mathematically as

$$\min\ C_{sum} = -\sum_{i=1}^{O}\sum_{j=1}^{J} w_j C_j(p_i)\,,$$

where $p_i = \{r,c : r \in [1,M] \wedge c \in [1,N]\}$. M, N and O are the dimensions of the spatiotemporal data and r and c refers to rows and columns of the spatial dimensions. $C_j \in \mathbf{R^3}$ are the path condition functions, w_j are weight constants and J is the number of conditions.

To reduce the complexity of the calculations, Bellmann's principle and dynamic programming were introduced. The detection step filtered out the moving MUAPs that existed during a given interval of time. However, the positions of the resulting path data were integer numbers in the frame space and provided poor spatial resolution. To increase the estimation precision of the EMG features, the positions of the detected MUAPs had to be improved. The shape of the detected MUAP peaks resembled the 2-D Gaussian in a local

surrounding. Using this observation a sequential 2-D Gaussian function-fitting procedure was tested. The MUAP path data, i.e. row, column (proximal-distal and medial-lateral positions) and time location, given by the cost analysis, were utilized to make a fit in a local surrounding area of the amplitude distributions. The fit was carried out in a least square sense to the Gaussian distribution.
Finally, CV and fiber orientation was estimated using linear regression on the improved position data.

RESULTS AND DISCUSSION

The method was tested using real multi-channel surface EMG data, from the *m. biceps brachii*, sampled for different fiber orientations relative to the electrode grid and at low contraction. Figure 1a, shows an example of the trajectories of two detected MUAPs and their position-improved traces in spatiotemporal space. Figure 1b shows the histogram distributions of the fiber orientation angle and CV estimates.
Since the data can be seen as a sequence of images, the spatial location and propagating behavior of the MUAPs is visually evaluable. On this basis, the detection works satisfactorily. It should be stressed that, at this point, the approach of the fiber orientation and CV estimation is preliminary.

CONCLUSION

This work presents a method to detect MUAPs in multi-channel EMG data. The detection is carried out using sequential potential amplitude distributions from the skin and returns spatiotemporal trajectories of the MUAPs. EMG features may be extracted if position-improvement of the trajectories is done.

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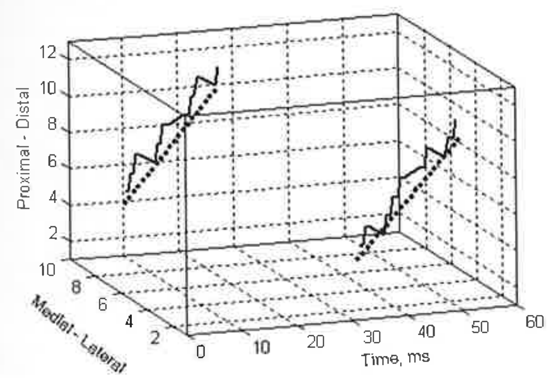
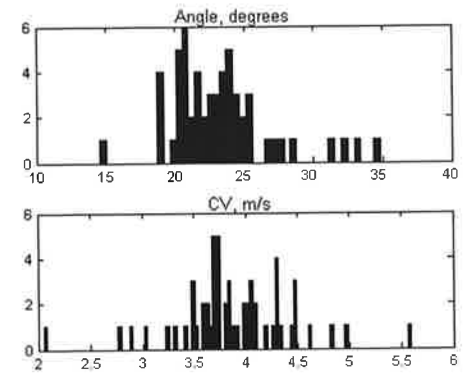


(a)



(b)

FIGURE 1: Examples of spatiotemporal trajectories of detected MUAPs (here from two different MUs), **a**, and the histogram distributions of fiber orientation and CV estimates, **b**.

POWER SPECTRUM ESTIMATION OF EMG SIGNALS VIA CHIRP-Z TRANSFORM

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INTRODUCTION

Power spectrum estimation has been widely used as the standard technique for EMG analyses. In the past few years, a number of researches have been developed, using spectral analyses, to predict contractile fatigue, to estimate the fibre type composition of muscles, to identify pathologies etc. However, accurate estimation of the power spectrum function of stochastic signals, such as EMG, is not a trivial task. Traditional methods, like DFT or FFT, have some drawbacks: 1) they generally provide imprecise estimations, when processing stochastic signals; 2) the DFT calculations are very slow, overriding the requirements when real time signal processing is necessary; 3) the FFT needs an integer power of two number of samples; 4) both FFT and DFT have limited resolution. Faced with those difficulties, the authors decided to investigate the use of the *Chirp-Z Transform*, developed by Laurence Rabiner in the 1970´s. In this paper, the concepts related to the Chirp-Z are reviewed and its performance is evaluated against traditional techniques such as DFT, FFT and Autoregessive (AR) models. The results show that, the *Chirp-Z* matched the traditional methods for EMG power spectrum estimation, but has also a number of advantages: 1) it does not require an integer power of two number of samples. For instance, if an specific case study requires the data acquisition and processing of 10000 samples, that number would need to be increased to 16384 (2^{14}) for FFT calculations. However, the Chirp-Z will need only those 10000 samples; 2) it is almost as fast as the FFT; 3) Although the authors used the Chirp-Z to obtain only the integer components of the EMG spectrum, it is possible to improve that resolution to obtain non-integer components. This can enhance the analyses of signals with low signal to noise ratio, that is often the case with EMG signals. Therefore, the authors would like to suggest the Chirp-Z Transform as the standard method for EMG power spectrum estimation.

METHODS

In order to carry out the operations required to use the Chirp Z Transform Algorithm (CZT), a series of steps must be performed to estimate the power spectrum density of EMG signals. Rabiner in (1) summarizes them as follows:

1. Choose L, the smallest integer great than or equal to (N+M-1) and compatible with any available FFT algorithm. L denotes the size transforms that must be calculated in order to perform the required high-speed convolution. N is the number of points in the input sequence and M is the number of points at which the Z transform is evaluated.
2. Form the L-point sequence y(n) as $y(n)=\begin{Bmatrix} A^{-n}W^{n^2/2}x(n), & n=0,1,\ldots,N-1 \\ 0, & n=N,N+1,\ldots,L-1 \end{Bmatrix}$
3. Compute the L-point DFT of y(n) using the available FFT routine; Call this result Yr.
4. Define the L-point sequence v(n) by $v(n)=\begin{Bmatrix} W^{-n^2/2}, 0\le n\le M-1 \\ W^{-(L-n)^2/2}, L-N+1\le n<L \\ \textit{arbitrary}, \text{other n, if any} \end{Bmatrix}$

5. Compute the L-point DFT of v(n). Call it Vr.
6. Multiply Vr by Yr, point by point, to give Gr = VrYr.
7. Compute L-point IDFT of Gr; call this g_k.
8. Multiply g_k by $W^{k^2/2}$ to give X_k defined as $X_k = g_k W^{k^2/2}$, $k=0,1,\ldots,M-1$. The values of g_k for $k \ge M$ are not meaningful and are discarded.

RESULTS

Figure 1 show an EMG signal, along with three different power spectrum estimations and error signal. The signal was collected at 1000Hz during an isotonic elbow flexion contraction. The total processing time to calculated the spectrum using those three techniques has been recorded and the error signal (between the DFT and Chirp Z spectrums) was calculated.

DISCUSSION

The results show that the Chirp Z transform (0,26 s) is faster than DFT (15,47 s) and the AR model (1,87 s) when estimating the spectrum of EMG signals. It also leads to the same values calculated by a traditional method such as the DFT (error signal almost zero - Figure 1(b)). In this specific example we could not use the FFT to determine the spectrum, as the number of samples were not an integer power of two.

CONCLUSION

The various tests carried out during this work show that the Chirp Z transform can be used as a very good option (sometimes the most suitable one) for EMG power spectrum estimation, as it gives us a handy solution for some drawbacks of traditional methods such as FFT and DFT.

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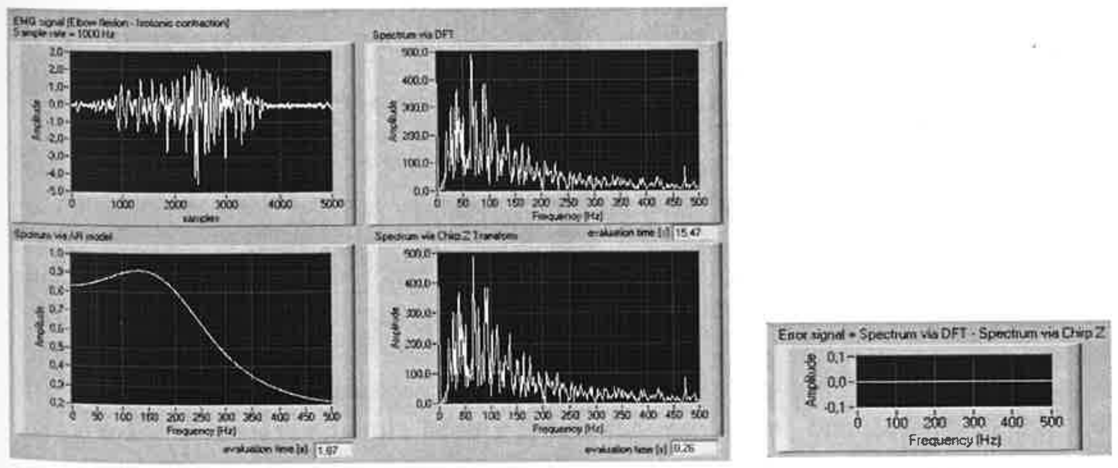

(a) (b)

Figure 1. (a) EMG signal, Spectrum estimation via DFT, AR model and Chirp Z transform. (b) Error signal between the estimated spectrum via DFT and Chirp Z transform.

CROSSTALK ASSESSMENT IN SURFACE RECORDED MYOELECTRIC SIGNALS

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INTRODUCTION

Surface electromyography is widely used to investigate muscle activation and timing during motion, muscle control strategies, and muscle fatigue. However, in many practical situations it is difficult to discriminate the signal produced by the muscle of interest from crosstalk due to neighboring muscles. The problem of crosstalk assessment is complicated by the fact that, during dynamic contractions, the degree of crosstalk affecting myoelectric signals may be different in different time instants, due to geometrical changes of the source and of the surrounding tissues. In the past, several strategies have been suggested to lessen the effects of crosstalk during signal recording, such as using double differential probes or even more complicated spatial filters, but these techniques must be applied when the signals are recorded. Many researchers have large archives of signals detected with conventional probes and would like to process them off-line, for lessening the effects of crosstalk. It is then important to develop techniques able to compare two signals, collected from neighbouring muscles, for detecting those intervals in which a signal is corrupted by crosstalk coming from the other muscle. Moreover, characterizing crosstalk should facilitate its reduction. This presentation describes a technique that allows detecting crosstalk between signals collected by means of conventional probes in the time-frequency domain.

METHODS

This technique is based on the mathematical properties of the auto- and cross-time-frequency transforms belonging to the Cohen class [1]. Two series of samples, which represent two surface recorded myoelectric signals (SMES), constitute the input of the algorithm. The time-frequency auto transform allows for estimating the instantaneous power spectrum of the input signals. The cross-time-frequency transform yields the instantaneous cross-spectrum between them. These time dependent spectra are then used to estimate the instantaneous log-squared coherency, which shows, on the time-frequency plane, the regions in which the two input signals are linearly dependent.
This information is of paramount importance, since crosstalk is due to low pass filtering (tissue filtering) of a signal generated by a neighbouring muscle. It follows that, if the istantaneous log-squared coherency is close to zero in a given region of the time-frequency plane, one of the two signals is affected by crosstalk from the other. The extension of the region allows for estimating the time interval and the frequency bands in which crosstalk is present.
This technique has first been characterized through computer simulations. These were carried out simulating different levels of crosstalk, ranging from totally correlated to completely uncorrelated signals. To this end, we generated two realizations of uncorrelated stochastic processes $s_1(t)$ and $s_2(t)$, then we obtained a filtered signal $s_f(t)$ by bandpass filtering $s_1(t)$ between 50 Hz and 200 Hz, and finally we generated a third signal $s_3(t)=\alpha s_f(t)+(1-\alpha)s_2(t)$. The two signals $s_1(t)$ and $s_3(t)$ were used as the input signals of our algorithm. The α parameter ranged from 1 (totally correlated signals) to 0 (totally uncorrelated signals).
We also tested this technique on real data, by using SMESs recoded from the rectus femoris and the biceps femoris of a subject during biking. These myoelectric signals were acquired by means of single differential active probes with Ag-AgCl pre-jelled electrodes, recorded with a sampling frequency equal to 2 kHz, and digitized by means of a 12 bit A/D converter.

RESULTS AND DISCUSSION

By means of the synthesized signals we demonstrated that the time-frequency representation of the instantaneous log-squared coherency shows the time instants and the frequency bands in which the input signals are linearly correlated. Figure 1 reports the instantaneous log-squared coherency estimated for values of α equal to 1 (left image) and 0 (right image).
It is evident that when the two signals are completely correlated (α=1), in the frequency band ranging from 50Hz to 200Hz the instantaneous log-squared coherency is close to zero. When the two signals are uncorrelated (α=0) it is not possible to find a region of the time-frequency plane in which the instantaneous log-squared coherency is zero. Similar results have been obtained by processing real myoelectric signals. Moreover, considering the values of the log-squared coherency in the time-frequency plane we have also been able to estimate the value of α, which measures the amount of crosstalk between signals.

CONCLUSION

In conclusion, the technique herein presented is a valuable tool for assessing the presence of crosstalk between surface myoelectric signals recorded from neighboring muscles. By observing the regions of the time-frequency plane in which the instantaneous log-squared coherency function is close to zero we can demonstrate the presence of crosstalk between the input signals. Since this technique is based on time-frequency transforms, it is can detect crosstalk also between nonstationary signals or even when the amount of correlation between signals is variable. We believe that this study represents a first step towards the development of a procedure for the automatic crosstalk quantification and reduction, also between already recorded signals.

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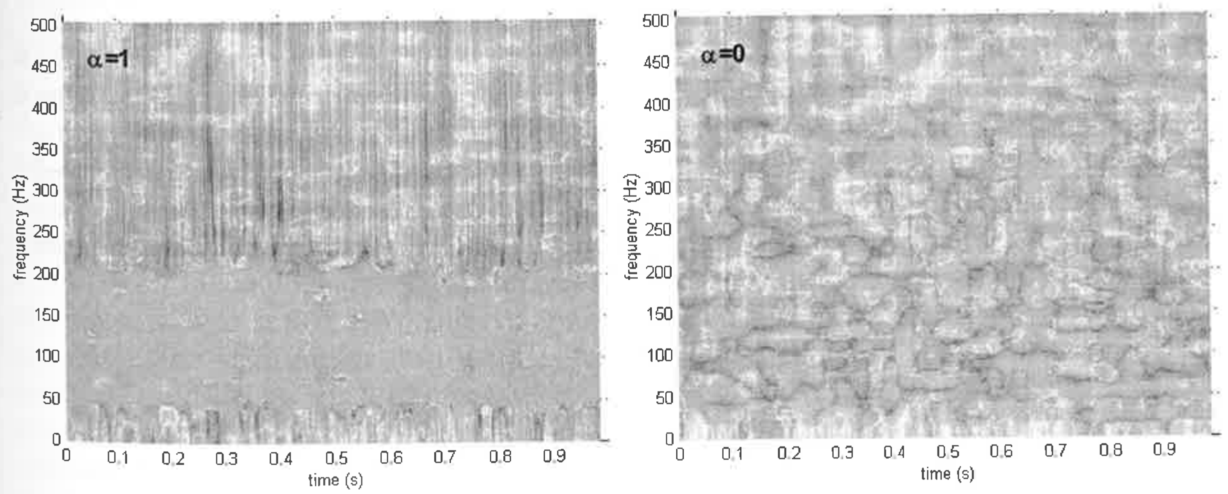

Figure 1: bidimensional representation of the instantaneous log-squared coherency function estimated in the case of correlated signals (left image) and uncorrelated signal (right image).

PRECISION DECOMPOSITION II FOR EMG SIGNALS – AN NIH BIOENGINEERING RESEARCH PROJECT

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We propose to develop a new automatic system for decomposing the electromyographic (EMG) signal into the constituent action potentials corresponding to the firing of individual motor units. It will be designed with the following characteristics:

1) Decomposition time for typical contractions will be decreased from dozens of hours to a few minutes.

2) The automatic decomposition accuracy will be increased from 70% to 95% - with provisions for assisted editing to reach 100% accuracy.

3) It will be able to decompose signals from dynamic as well as static contractions, which is a current limitation.

4) It will weigh less than 10 kg and have a notebook computer configuration.

5) Most importantly, the decomposition algorithms will be completely rewritten using a newly developed knowledge-based Artificial Intelligence language blackboard platform developed by us.

The development of the new system will be carried out in concert with two research projects and two clinical projects designed to provide challenging data to the new Decomposition system while revealing new clinical and physiological information. The hardware and software of a functioning prototype of the new portable decomposition system has already been built and is currently being alpha tested. Initial tests indicate an accuracy performance superior to the old version with a speed increase of several hundred-fold.

DETECTING CHANGES IN MUSCLE ACTIVATION FOLLOWING A FATIGUING SUBMAXIMAL CONTRACTION

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INTRODUCTION
Myoelectric signal (MES) amplitude increases seen in a reference voluntary activation (RVE), or test contraction, have been used in ergonomics research in order to detect neuromuscular fatigue resulting from prolonged or repeated low-level contractions. This is problematic in that amplitude increases, in the absence of a downward shift in the frequency spectrum, may not indicate muscle fatigue, but rather alterations in motor unit recruitment. The purposes of this study were (1) to determine if increases in the amplitude of myoelectric signal recordings at several different sites over and within the upper trapezius muscle could be detected using surface and/or fine wire electromyography after subjects performed a prolonged arm-holding task to fatigue, and (2) if these increases corresponded with upward (due to de-novo recruitment) or downward (due to fatigue) shifts in the frequency spectrum.

METHODS
Six normal healthy female volunteers participated in this study. Surface (S1-S4) and fine-wire (FW1-FW4) myoelectric signal recordings were made at four sites on the upper trapezius muscle while a series of three repetitions of each of two reference contractions (90° and 30° unweighted shoulder flexion in the scapular plane) were performed by each subject both before and after a static arm-holding task (30° flexion in the scapular plane) held to subjective fatigue. For each data file, the corrected root mean square [RMS (mV)] amplitude was computed over a moving window of 250 ms. The representative RMS values were determined as the mean RMS value computed across the windows of each RVE contraction. A Fast-Fourier Transform was used to compute the power spectrum (FFT^2) of each data file, and the median power frequency (MDF) was used to represent the central frequency. The data were normalized by taking the ratio of the post-test measures (RMS and MDF) to the pre-test measures. One-way ANOVAs on these ratios were used to determine significant differences in RMS amplitude and MDF values after the arm-holding task.

RESULTS
Root Mean Square Amplitude (RMS)
Surface channels: At the arm angle of 30 degrees, the RMS amplitude increased only at site S2 ($p<0.013$) , where the post-test values were higher than the pre-test values. With the arm at 90°, the RMS amplitude increased both at site S1 ($p<0.005$) and site S2 ($p<0.005$).
Wire channels: The wire sites demonstrated increases in amplitude at sites FW1 ($p<0.02$), FW4 ($P<0.035$), and marginally at site FW2 ($p<0.09$) when the arm was held at 30°. When the arm was held at 90°, FW3 showed a significant change in RMS amplitude, however the pre-test values were higher than the post-test values. Both sites FW1 ($p<0.066$) and FW2 ($p<0.059$) showed marginal changes, where the post-test RMS values where higher than the pre-test values.
Median Power Frequency (MDF)
Surface channels: The MDF analysis showed that site S1 demonstrated a decrease in MDF in the 90° reference contraction. Site S2 demonstrated a decrease in MDF (in both the 30° ($p<0.0005$) and 90° ($p<0.0009$) positions) after the arm-holding task.

Wire channels: Site FW2 demonstrated a decrease in MDF between pre and post-tests, at both the 30° ($p<0.001$) and 90° ($p<0.0005$) positions. Site FW3 (with the arm at 30°) demonstrated significant ($p<0.023$) changes in the MDF, however the direction of the change indicated an increase in MDF. At FW4 there was no significant change in MDF corresponding to the increase in RMS amplitude found with either the 30° or the 90° test.

DISCUSSION

For the surface electrode sites used in this study, comparing pre- and post-test contractions revealed that only site S2 showed consistent evidence of fatigue, where an increase in RMS was accompanied by a decrease in MDF. These findings are surprising given that this site was 2 cm superio-anterior to the standard electrode position used by most researchers[2,4].At the standard site (S1) an increase in RMS amplitude with the arm at 90° was not accompanied by a decrease in spectral parameters. This finding at S1 is consistent with Oberg et al[3] who found no corresponding decreases in the power spectrum of the signal despite finding increases in amplitude during similar loads. This RMS increase may not be indicative of peripheral muscle fatigue.

CONCLUSIONS

Site 2 demonstrated consistent fatigue effects in surface recordings (an increase in amplitude with a co-incident decrease in MDF), and the fine-wire recordings made at the same site were also tending to demonstrate similar findings. It appears that wire electrodes are of little benefit for the detection of fatigue post-activity. The findings of this study, coupled with the companion reliability study also reported in these proceedings, suggest that site 2, located 2 cm supero-anteriorly to the conventional electrode site used on the upper trapezius[2], is more reliable and sensitive (than the other channels used in this study) to neuromuscular fatigue recorded using post-activity test contractions.

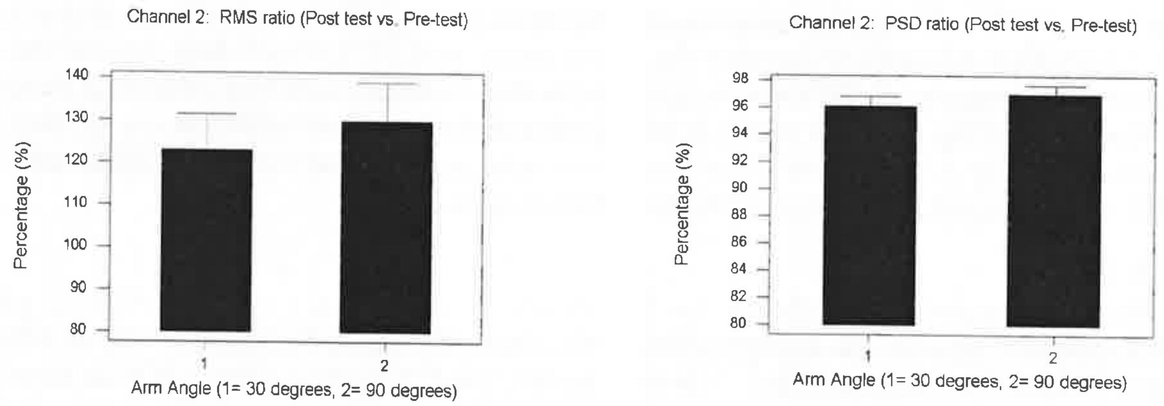

Figure 1: The percentage of post-test to pre-test RMS and MPF values recorded during the test contractions performed at surface channel site 2 (S2).

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METHODS TO IMPROVE MEAN FREQUENCY MEASUREMENTS OF SURFACE EMG SIGNALS DURING DYNAMIC CONTRACTIONS

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INTRODUCTION

Analysis of frequency spectra from EMG signals has been widely used to characterize muscle fatigue and has attracted great interest in the fields of rehabilitation medicine, ergonomy and sports physiology. The mean frequency (MNF) is commonly used as a spectral change indicator. During a static contraction the surface EMG signal is often assumed to be stationary and the Fourier transform is applied. During dynamic contractions that assumption can be questioned, and therefore in recent studies time-frequency representations (TFRs) has been used. One critical problem during dynamic contractions is the dependence of the MNF estimate on the signal-to-noise ratio (SNR). In dynamic contractions the amplitude of the surface EMG signal changes, but the noise remains almost constant. Therefore, when the signal amplitude is low the SNR will also be low and the estimates will be more affected by the noise.

METHODS

For EMG signals recorded from dynamic contractions the time-dependent or instantaneous MNF (IMNF) is an important parameter. The IMNF is often estimated from a TFR according to equation 1, where $[f_L, f_H]$ is the bandwidth on which the TFR has been calculated. The TFR can be seen as an estimate of a power spectrum for every time instant, t.

$$IMNF = \int_{f_L}^{f_H} f \cdot P(t,f)df \Big/ \int_{f_L}^{f_H} P(t,f)df \quad (1)$$

Since it is known that the MNF estimate is mostly affected by the high-frequency noise and that the surface EMG signal is bandwidth-limited, it has been proposed that the IMNF should be calculated from f_L to the maximum frequency at time t ($f_{max}(t)$) where the EMG signal is stronger than the noise. In (1) it was suggested that a method invented by D'Alessio, called threshold crossing method (TCM), should be modified and used to find f_{max}. The TCM finds f_{max} by comparing the frequency bins, starting from the high frequency end of the power spectrum, with a threshold calculated from the statistics of the spectrum by specifying a specificity of the method, p_s. If, in a sequence of r frequency bins, there are at least m bins that surpass the threshold, then f_{max} is chosen as the highest frequency bin in that sequence. In a power spectrum the frequency bins are chosen in successive order but we suggest that in a TFR the frequency bins should be chosen with an offset from each other that corresponds to the dependency in that representation.

Another way to calculate f_{max} could be with the hybrid method (2), where f_{max} is the lowest frequency that satisfies

$$\int_{f_L}^{f_{max}(t)} P(t,f)df = \int_{f_L}^{f_H} P(t,f)df - k\left(f_H - f_{max}(t)\right) \quad (2)$$

where k is a constant larger than or equal to the power noise density, N_0.

In this study we introduce a new method and compare it with the TCM and the hybrid method. This new method finds f_{max} by, with start from the high frequency end of the power spectrum, adding r frequency bins together, chosen with an offset from each other that corresponds to the dependency in that representation and compare the summation with a threshold. The threshold is calculated from the statistics of the spectrum by specifying a specificity of the method, p_s. When the summation is greater then the threshold then that value is taken as f_{max}.

To compare the three different methods 100 signals with 16 000 samples each were made by filtering white noise with a fourth-order butterworth filter with cutoff frequencies 50 and 100 Hz, with a assumed sampling rate of 2 kHz. In real EMG signals the noise can often be estimated from the first part of the recording when there is no myoelectric activity. To simulate inactivity the first 2000 samples of the signals were zeroed before white Gaussian noise was added to the signals. A scalogram (squared magnitude of a continous wavelet transform) calculated with a Morlet wavelet from 15 to 700 Hz was used to estimate the IMNF.

RESULTS AND DISCUSSION

In Figure 1 the normalized root mean square (RMS) errors of the estimated IMNF, with reference to the IMNF calculated from the noise-free signal, are shown for the different noise-reduction methods. From the figure it is clear that all methods improved the IMNF estimate for low SNR. With the parameters chosen the TCM and the new method outperformed the hybrid method. The difference in results between the new method and the TCM were small but the new method gave slightly better results, especially for very low SNR.

CONCLUSION

This study showed that MNF estimates of EMG signals recorded from dynamic contractions could be greatly improved by applying any of the presented methods. The methods could be used to estimate IMNF from other TFRs than the scalogram. These methods could also be used to estimate higher order spectral moments, which are known to be even more affected by noise than is the case for the MNF.

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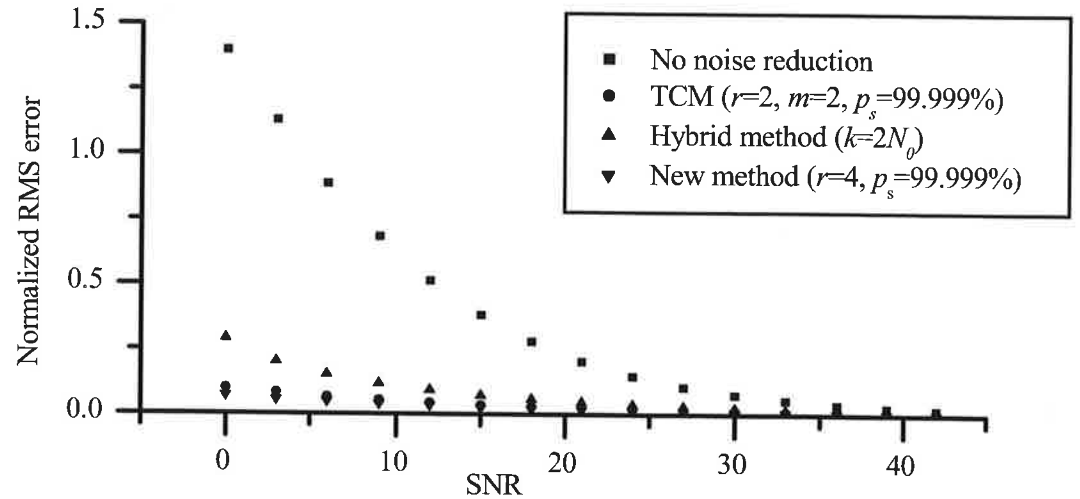

Figure 1: Normalized root mean square (RMS) errors of IMNF estimate for different noise reduction methods calculated for different SNR.

A SIMULATION STUDY OF SURFACE EMG CROSS-TALK AND CROSS-CORRELATION

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INTRODUCTION

Cross-talk, the detection of volume conducted signals from muscles other than the muscle of interest, can be a significant limiting factor in surface electromyography (EMG). In particular, it is an important concern when studying co-contraction of adjacent pairs of muscles. Unfortunately there is no clear method of distinguishing between signals generated by the muscle directly below the recording electrode and those which originate in more distant muscles. One approach which has been proposed is to examine the cross-correlation function of two EMG signals to assess their commonality (1, 2). Although concerns have been raised about this method on the basis that the volume conduction process dramatically alters the characteristics of each motor unit action potential, the cross-correlation function has continued to be widely used as a test for surface EMG cross-talk. One of the difficulties in assessing the usefulness of such a measure is that it is difficult to obtain an independent measure of EMG cross-talk experimentally. In this paper, a model of the surface EMG signal is used to explore EMG cross-talk at different electrode locations around the limb and to compare the amplitude of the volume conducted signals with cross-correlation based cross-talk indices.

METHODS

Using a finite element model (3), surface EMG signals were generated for a muscle composed of 250 motor units. The motor units were randomly distributed throughout a territory of radius 30 mm, centred on the vertical axis. Motor units fired with mean firing rates that ranged from 15-25 Hz. Each bipolar electrode pair was orientated parallel to the direction of the muscle fibres, and was separated by an inter-electrode distance of 20 mm. The root mean square (RMS) amplitude and median frequency of the surface signal detected at bipolar electrodes at 7.5 degree intervals around the surface of the volume conductor were examined. At each electrode pair, the maximum of the cross-correlation function of the EMG signal detected at that electrode and that detected above the centre of the muscle was examined. This was repeated as subcutaneous fat thickness was varied between 0 mm and 18 mm. The RMS value and cross-correlation peak were then examined when a second muscle, adjacent to the first was simultaneously activated.

RESULTS

The rate at which the EMG RMS amplitude decayed around the surface of the model decreased as subcutaneous fat thickness increased, Fig 1(a). The rate of decay of the median frequency of the EMG power spectrum also decreased with increasing subcutaneous fat thickness, Fig 1 (b). Both the EMG RMS value and the median frequency remained relatively constant at electrode locations directly above the active muscle. Peak cross-correlation values also increased with subcutaneous fat thickness, however, in each case, the maximum of the cross-correlation function decayed much more rapidly than either the median frequency or amplitude of the surface EMG signal, Fig. 1 (c). The RMS amplitude of the surface EMG signal at an electrode 30 degrees from the edge of the active muscle was 4 %, 14 % and 20 % of the EMG amplitude directly above the active muscle, in the 3 mm, 9 mm and 18 mm fat models respectively. Corresponding cross-talk estimates based on the square of the maximum of the normalised cross-correlation function were 0.8 %, 2 % and 8 %. When a second muscle was simultaneously activated, cross-correlation values changed very little, tending to increase slightly.

DISCUSSION AND CONCLUSION

The simulation results indicate that relatively large amounts of EMG cross-talk can be detected above regions of inactive muscle tissue adjacent to the active muscle. However, cross-talk indices based on the cross-correlation function did not reflect the level of cross-talk at a given electrode site. Furthermore, it was not possible to distinguish between volume conducted signals and signals generated by the muscle directly below the electrodes using cross-correlation measures. As individual action potentials propagate through different volumes of tissue, the amplitude, frequency content and shape of the waveforms change dramatically. Surface potentials from motor units lying in the common pick-up volume of two electrodes do not have the same phase, shape or magnitude. Therefore, an absence of a peak in the cross-correlation function of two EMG signals does not necessarily indicate that there is no cross-contamination between the two. Similarly, a large peak in the cross-correlation function may be due to factors such as a common neural drive to both muscles, rather than a common source. Furthermore, not only does the shape of the volume conducted signal vary as it passes through different tissues, but different motor unit dominate recordings at different electrodes, even if all belong to the same muscle. While the exact nature of the relationship between the rate of decay of the surface EMG amplitude and the median frequency of the EMG power spectrum is not immediately clear, significant differences in the median frequency of the EMG signal above the active and inactive tissues were observed, suggesting that the spectral content of the surface EMG signal may provide a better means of distinguishing between EMG cross-talk and co-activation of adjacent pairs of muscles.

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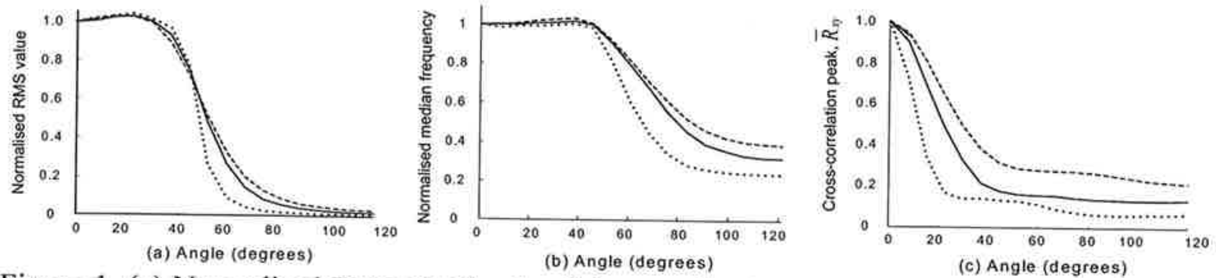

Figure 1. (a) Normalised EMG RMS value (b) normalized power spectrum median frequency and (c) peak cross-correlation value, at different locations around the surface of the model due to a muscle, radius 30 mm, centred on the vertical axis (0 degrees). Values are presented for 3 mm (dotted line), 9 mm (solid line) and 18 mm (dashed line) thick fat layers.

NON-NORMALIZED EMG DRIVEN MODEL TO QUANTIFY TRUNK MUSCLE MOMENT CONTRIBUTIONS IN BACK PAIN PATIENTS

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INTRODUCTION

EMG amplitudes obtained during maximum voluntary contractions (MVC) are commonly used for normalization of EMG data. However, low back pain (LBP) patients are often unable or unwilling to produce maximum muscle activation (1). This severely hampers studies of trunk muscle recruitment in LBP patients and probably constitutes one of the causes of divergent results in studies on trunk muscle recruitment in LBP patients. The aim of the present study therefore, was to develop and test a model to estimate trunk muscle forces driven by non-normalized EMG linear envelopes.

METHODS

Sixteen patients with chronic idiopathic LBP and 16 matched healthy control subjects volunteered for this study. Patients had no neurologic deficits, structural deformities, genetic spinal disorders or previous spinal surgery. Patients suffered from LBP for periods ranging from 6 months to 35 years. On a 10 cm visual analog scale, patients expressed their LBP on the day of testing as 2.1 (SD=1.2). Healthy control subjects had never experienced back pain lasting longer than three consecutive days and matched the experimental LBP group by gender, age, weight and height.

Subjects were placed in a semi-seated position in an apparatus that restricted hip motion and performed isometric ramp contractions in left and right lateral bending, flexion, and extension up to a level experienced by the subject as effortful but tolerable. A steel cable attached to a chest harness at approximately T9 and to a load cell provided the resistance. All ramp contractions were performed twice.

EMG signals were collected using surface electrodes placed on the following trunk muscles bilaterally: rectus abdominis, external oblique, internal oblique, thoracic erector spinae spinae, and lumbar erector spinae. The EMG signals were band passed between 20 and 500 Hz, amplified, and sampled at 1.6 kHz. EMG signals were filtered with an adaptive filter algorithm to attenuate ECG contamination and subsequently full-wave rectified and 2.5 Hz low pass filtered (2).

An EMG driven trunk muscle model was used to estimate forces produced by the muscles of interest. The geometry and outline of the model have been described earlier (3). 98 vectors crossing the L4/L5 joint were used to represent internal and external obliques and rectus abdominus and erector spinae and multifidus muscles. Muscle force was estimated as the product of maximum muscle stress, muscle activation, and muscle cross-sectional area. Conventionally, EMG signals are normalized to MVC values to estimate activation and maximum muscle stress is iteratively adjusted to obtain maximum agreement (least squares) between the time series of muscle moments and net external moments. In the present model no normalization was used and gains were estimated for each muscle group that represented the maximum muscle tension divided by the maximum EMG amplitude. Constrained optimization was employed with the cost function being the sum of the squared difference

between net moment components measured and muscle moment components predicted by the model. The constraints were based on the assumption that maximum tension is between 20 and 100 Ncm^{-2} and that the maximum EMG amplitude would be larger than the maximum found in the ramp trials and lower than 3 times this maximum. A third constraint limited the difference between the maximum and minimum of the maximum muscle tension estimates for the different muscles to a factor of 3.
Gains were estimated on the combined data of the first ramp trials in four directions (gain A). Gain A values, coefficients of determination (R^2), and average absolute errors were compared between patients and controls. To test the reliability, gain values were estimated on the set of repeat trials (gain B). To determine whether the level of effort would affect estimates, data of the first set of contractions were selected up to where 50% of the maximum moment was produced and gains were estimated by fitting the model to this limited data set (gain C). The three gain values were used to estimate the total muscle force at the instant of the maximum net moment during the first set of ramps.

RESULTS
The medians of the peak moments produced were 51, 50, 45, and 37 Nm in extension, flexion and left and right lateral bending, respectively. Differences between patients and controls were not significant (Mann Whitney: $p = 0.47$, $p=0.40$, $p=0.18$, $p=0.09$, respectively). The model in general fitted the time series of net moments well. The median coefficient of determination was 0.80 corresponding to a median of averaged absolute errors of 9.3 Nm. The coefficients of determination, average absolute errors and gain values were not significantly different between back pain patients and healthy controls (Mann-Whitney U: $p=0.59$, $p=0.52$, $p=0.47$, respectively).
Reliability of force estimates was good with an ICC between estimates based on gains A and B of 0.82. The estimated total muscle force appeared lower (median 449 N or 17%) when based on gain C as compared to gain 1, indicating a dependency on the level of effort in the gain estimation procedure (Wilcoxon: $p < 0.001$).

DISCUSSION
The model yielded good predictions of muscle moments in both the control and patient group. Estimation procedures would thus not systematically affect group comparisons. Average performance and reliability of comparable models that used MVC normalized EMG amplitudes and healthy subjects was similar to our results.
The gain estimates and resulting force predictions were affected by the level of effort during the calibration contraction. This may be indicative of non-linear EMG force relationships and suggests that at least standardisation of this level during calibration trials is required.

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SURFACE EEG-DERIVED MOVEMENT-RELATED CORTICAL POTENTIAL IS GREATER FOR HUMAN ECCENTRIC THAN CONCENTRIC MUSCLE CONTRACTIONS

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INTRODUCTION
Despite considerable evidence suggesting that different nervous system control strategies exist for human concentric (shortening) and eccentric (lengthening) muscle actions [1], little data are available to indicate that the central nervous system (CNS) signals differ for controlling these two types of muscle activities. Knowing more about CNS signals that mediate concentric and eccentric muscle actions would assist our understanding of neural mechanisms underlying human voluntary movements and development of more effective rehabilitative interventions and athletic training programs. The purpose of this study was to evaluate electroencephalogram (EEG)-derived movement-related cortical potential (MRCP) and to determine whether the level of MRCP-measured cortical activation differs between the two types of muscle activities.

METHODS
Eight volunteers (6 men and 2 women, 27.8 ± 7.2 y/o) performed 50 voluntary eccentric and 50 concentric elbow flexor contractions against a submaximal load. Scalp EEG signals from four locations overlying sensorimotor-related cortical regions in the frontal and parietal lobes were measured (Cz, C3, C4 and Fz) along with the applied force, electromyograms (EMG) of the biceps brachii (BB), brachioradialis (BR), triceps brachii (TB), and deltoid (DT) muscles, and elbow joint angles during the movements. MRCP was derived from the EEG signals associated with the eccentric and concentric muscle contractions. MRCP was divided into two major components, negative potential (NP) and positive potential (PP) [2].

RESULTS
The load applied during eccentric and concentric tasks was the same and the force exerted by the subjects for eccentric and concentric tasks measured 54.50±6.03 N and 69.33± 8.30 N, respectively. The difference in force values originated from non-uniform friction between the pulling cable and the pulley for the tasks. Eccentric contractions were less steady with greater force fluctuation (1.14±0.61 N) than these of the concentric trials (0.76±0.25 N). Although the level of elbow flexor muscle (BB) activation (EMG) was lower during eccentric (20.0±2.46%) than concentric (40.55±11.45%) actions, the amplitude of the two major MRCP components - one related to movement planning and execution (NP) and the other associated with feedback signals from the central and peripheral systems (PP) - was significantly greater for eccentric than concentric actions. The peak NP values for the Cz location were 3.78±0.74 μV and 2.80±1.19 μV for the eccentric and concentric contractions, respectively. The peak PP values were 6.06±0.40 μV and 4.74±0.66 μV for the two corresponding movements. The MRCP onset time for the eccentric task occurred earlier than that for the concentric task. The values of NP onset time (relative to the EMG onset) for the Cz location were -859±48 ms (eccentric) and -768±62 ms (concentric). All the above mentioned values were significantly different between the two tasks ($P<0.05$). An example of the data is given in Figure 1.

DISCUSSION
The main findings of this study were that for eccentric muscle actions, the MRCP amplitude was greater and onset time was earlier than the corresponding values for concentric

measurements. The EMG level of the elbow flexor muscles, however, was significantly lower for eccentric than concentric muscle activities. Possible explanations for greater magnitude and longer preparation time of cortical activities for eccentric movements: i) *Degree of difficulty.* Eccentric contractions are more difficult to perform than concentric ones, and thus, may require more cortical control [3]. ii) *Preventing muscle damage.* Eccentric actions are characterized by greater tissue damage as compared with concentric ones and thus, the CNS may need to plan ahead for damage reduction. iii) *Different control strategy.* Motor unit recruitment order during eccentric contractions differs from the "size principle," [4] indicating that the CNS control strategy for eccentric movements may be different from that for concentric ones.

CONCLUSION

It has long been speculated that the nervous system poses unique strategies for controlling eccentric muscle actions. This study shows, for the first time, that the brain plans eccentric movements and processes eccentric-related sensory information differently than it does for concentric muscle contractions. Because eccentric movements are more complex, make muscles more prone to damage, and perhaps require a unique motor unit recruitment strategy to carry out the actions, the greater NP may reflect additional cortical planning activities or effort to deal with these "special problems." The higher magnitude of MRCP PP for the eccentric contractions may indicate that a larger amount of sensory information is being processed in the brain, and additional reflex-induced cortical activity resulted from stretching the muscles.

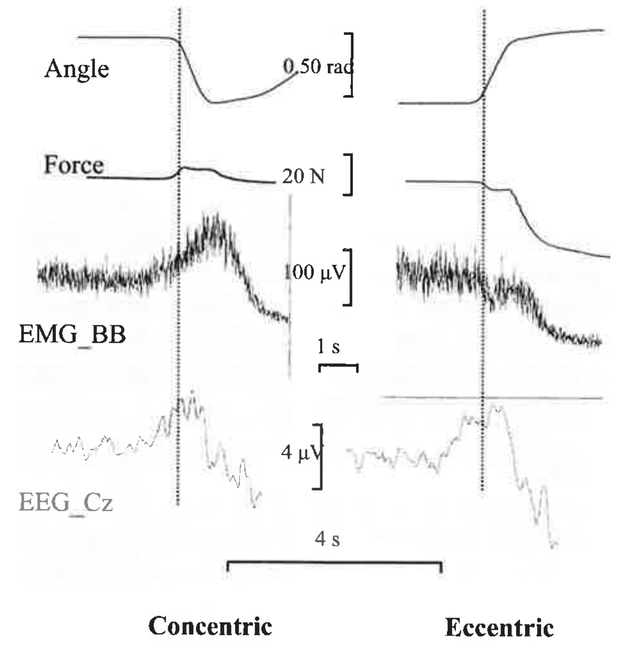

Fig. 1. Joint angle, force, EMG (BB), and MRCP (Cz electrode) for concentric (left) and eccentric (right) contractions of a subject. Dotted lines indicate the timing of trigger. The data show an increase in MRCP but a decrease in EMG. The discrepancy in force was due to different friction between the two tasks.

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VOLUNTARY AND ELECTRICAL STIMULATION-EVOKED SEGMENTAL AND SUPRASPINAL CROSSED EFFECTS IN HUMANS

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INTRODUCTION

The functional significance of unilateral motor and sensory activity is illustrated by adaptations (i.e., cross education) that are highly specific to the opposite homologous structures and occur after chronic mental rehearsal, skill practice, high-force voluntary and electrical stimulation-evoked contractions, and peripheral nerve damage (1). The substrate of these adaptations, cortical map plasticity and neuronal plasticity, plays an important role in mediating recovery from brain injury and orthopedic deficits (2).

In an attempt to elucidate the mechanisms of these chronic adaptations, we have previously reported in acute experiments that 5-s-long voluntary isometric left wrist flexion contractions (75%MVC) reduced H reflex amplitude to 50% of control in the right flexor carpi radialis (FCR) and increased 2.5-fold the motor evoked potentials (MEP) produced by transcranial magnetic stimulation (TMS) over the left motor cortex (3). Because unilateral stimulation of cutaneous afferents in spinalized animals can facilitate the homologous contralateral motoneuron pool, we tested the hypothesis that crossed segmental effects may vary according to the nature of the unilateral stimulus, i.e., whether the contraction is produced by voluntary action or by electrical stimulation (4). The aim of this study was to compare voluntary and electrical stimulation-evoked segmental and supraspinal crossed effects in humans.

METHODS

In 7, right-handed subjects (6 M, 1 F, age 24) we examined the effects of 50% and 75% left wrist flexion MVCs, electrical stimulation-evoked muscle contraction (50%STIM), and a combination of voluntary and stimulation-evoked contractions (75%COMB) on MEPs evoked by TMS (150% motor threshold, Magstim 200) over the left motor cortex and H reflex evoked by stimulating (1 ms, 50% of Hmax) the right median nerve (Digitimer DS7H). MEPs and H reflex were recorded in the relaxed right FCR (background EMG < 20 µV). For 50%STIM and 75%COMB, the left FCR and flexor carpi ulnaris were simultaneously stimulated with surface electrodes for 5 s with a 1-s on-ramp and 0-s off-ramp at 2 kHz modulated at 50 Hz, up to maximal stimulator output of 80 mA (Metron Multistim, MA-100). In the middle of each 5-s condition period, one H reflex or MEP was evoked and followed by 11 additional stimuli at 5-s intervals for 1 min and repeated 5 times. The last MEP and H reflex in each 1-min series was used for normalization.

RESULTS

One-way ANOVA revealed ($F = 67.7$, $p = 0.0049$) that 50%STIM significantly increased the size of the contralateral H reflex by 30% (±44) ($p < 0.05$, Tukey's post-hoc), whereas 50% wrist flexion MVC reduced H reflex size by 24% (±18) ($p < 0.05$). In contrast, 75%COMB reduced contralateral H reflex size by 38% (±0.22, $p < 0.05$). This reduction was statistically similar to the 34% (±0.20) decrease caused by 75% MVCs without electrical muscle stimulation ($p < 0.05$). MEPs increased significantly under all four conditions but the increases were greater ($p < 0.05$) during the 75% than during the 50% conditions (one-way ANOVA, $F = 11.1$, $p = 0.0084$).

DISCUSSION

The present data confirm previous findings that unilateral motor activity activates primary motor cortex ipsilateral to the contraction, an observation that modifies the classical view that the contralateral hemisphere exclusively controls voluntary movement (5). The similar increase in MEP during stimulated and voluntary contractions indicate the importance of ipsilateral ascending input to the motor cortex. The finding of a contralateral H reflex facilitation and depression at the same absolute force during stimulated and voluntary unilateral contraction, respectively, points to unique side-to-side segmental control of the homologous muscle pair. It is possible that the H reflex depression during voluntary contraction and the H reflex facilitation during stimulated contractions are the result of, respectively, reinforcement and removal of presynaptic inhibition. However, animal data indicate that cutaneous afferents may exert direct crossed influences or indirect crossed effects through propriospinal paths on he contralateral homologous motoneuron pool (4). The H reflex depression produced in the COMB condition suggests that the inhibitory crossed effects associated with voluntary effort canceled the facilitatory effects produced by the electrical stimulation. This cancellation indicates that the inhibitory effect associated with voluntary contraction is stronger than the facilitatory effect caused by electrical stimulation and that these two competing effects interact at the same crossed interneuronal system. The data from these acute experiments suggest that cross education brought about by chronic unilateral practice may be mediated predominantly by supraspinal or spinal mechanisms whether the training stimulus is produced by voluntary contraction or electrical stimulation.

CONCLUSION

Crossed segmental effects may vary according to the nature of the unilateral stimulus and the crossed-inhibition and crossed-facilitation of the H reflex during voluntary and stimulated contraction, respectively, may involve presynaptic inhibition and crossed propriospinal and cutaneous reflex paths.

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AGE RELATED CHANGES IN THE REACTION TIMES UNDER TWO CONDITIONS OF NO WARNING SIGNAL AND A WARNING SIGNAL

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INTRODUCTION

It has been reported that the simple reaction time was prolonged in the old age group compared to the young age group[1)]. In order to investigate the effect of preparatory signal on the prolonged reaction time in the old age group, the reaction times with and without a warning signal were studied in normal subjects[2)].

METHODS

The reaction times with and without a warning signal were studied in 84 healthy subjects from 24 to 78 years old, averaging 48 years. As the methods, the subjects were requested to look at the lamp(visual signal) and keep the finger of dominant hand on the key. They were instructed to push the key quickly as far as possible after the light. The reaction time was measured from the time interval between the onset of stimulus and the initiation of response. The test of reaction time was tried under two conditions: (1) a simple test with no warning signal, (2) a test with a warning signal preceding the reaction time stimulus by 1.0 second. The warning signal was an 80-dB buzz with duration of 0.5 sec[2)]. The 10 trials of reaction time were given for each person under the same condition, and the mean value and standard deviation (SD) of reaction times were calculated.

RESULTS

The quantitative dates showed that the mean values of reaction times under the condition with no warning signal were increased with the age distribution from 24 to 78 years old. In the young age group of 20-29 years old(Fig1), the mean value of reaction times was 252 msec, and it was significantly increased to 360 msec in the group of 60-69 years old and 379 msec in 70-79 years old.($P<0.01$). On the other hand, under the condition with a warning signal, the mean value of reaction times was 220 msec in the group of 20-29 years old, and it was 258 msec in the group of 60-69 years old and 260 msec in the group of 70-79 years old. These values of reaction times under the condition with a warning signal were not different between the groups of young age and old age.

DISCUSSION

The simple reaction times in the old age group have been reported to be prolonged compared to the young age group[1,3]. The similar results were also obtained in our examination. While in normal subjects, a warning signal preceding the reaction time induces the alpha wave desynchronization on the EEG before the reaction time stimulus[2]. Then it was suggested that a warning signal increased the arousal (alerting) level, and made the sensory motor response more efficient[2,3]. From these results, the reduction of reaction time to a warning signal (preceding or anticipating signal) has been used as a measure of phasic arousal or alerting. In our studies, the increased reaction times in the old age group significantly decreased after a warning signal. Therefore our results suggest that the decreased function of arousal (attention) system played the important role for the prolongation of reaction times in old age group of 60-79 years .

CONCLUSION

The reaction times with and without a warning signal were studied in 84 healthy subjects from 24 to 78 years old. The mean values of reaction times under the condition with no warning signal were increased with the age distribution. On the other hand, under the condition with a warning signal, the mean values of reaction times were not different between the groups of young age and old age. Our results suggest that the decreased function of arousal (attention) system played the important role for the prolongation of reaction times in old age group of 60-79 years old compared with the young age group of 20-29 years old.

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Age (Y.O.)	20-29	30-39	40-49	50-59	60-69	70-79
Without Warning					P<0.01	P<0.01
Mean±SD msec	252±19	253±25	253±26	274±26	360±76	379±80
With Warning						
Mean±SD msec	220±9	223±9	224±10	228±10	258±59	260±46

Fig 1:The age related changes in the simple reaction times without warning and with a warning

SURFACE EMG BASED FOLLOW-UP OF THE REINNERVATION PROCESS IN DENERVATED MUSCLES

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INTRODUCTION

Muscle denervation is typically investigated by using intramuscular needle EMG [1][2][3][4]. This technique is useful in diagnosis and as prognostic tool, but, since it is invasive, it is unsuitable for monitoring the reinnervation process. The aim of this work was to assess the feasibility of a surface EMG based follow-up of the reinnervation process in denervated muscles.

METHODS

Surface EMG signals were detected using linear arrays of 8 and 16 electrodes with interelectrode distance 2.5mm, 5mm, and 10mm, according to the length of the investigated muscle. Linear electrode arrays allow to identify the position of the innervation zones, estimate motor unit (MU) conduction velocity, investigate the recruitment strategies, identify polymorphic MUs, if they occur, and distinguish them from randomly superimposed MU action potentials (MUAPs).

The extensor carpi ulnaris and radialis, the extensor digitorum communis and the abductor pollicis longus muscles were investigated in patients with a denervation of the radial nerve. The abductor pollicis brevis was selected for the patients with a lesion of the medianus nerve. Finally, in subjects with a lesion of the sciatic nerve the biceps femuris and tibialis anterior muscles were investigated. Until now, 8 patients have been investigated in 14 experimental sessions. Surface EMG signals were acquired from the same subjects in experimental sessions separated by approximately 20 days. Each session provided for a series of maximal voluntary contractions for all muscles affected by the nerve lesion.

RESULTS

In 4 subjects it has been possible to follow the reinnervation process in a temporal range varying between 4 to 8 months. In one subject, with a sciatic nerve lesion, no reinnervation has been observed. Three subjects are currently under observation.

On the basis of the preliminary data analysis, the following observations were made:

- The surface technique does not reveal denervation potentials which can be observed from intramuscular recordings. Denervation potentials, also called fibrillation potentials, are brief spontaneous bi-or triphasic potentials that are seen in neurogenic and primary muscle disease.
- The appearance of MUAPs and the progressive increase of the amplitude of the surface EMG signal allow monitoring the reinnervation process. The increase of the surface EMG signal amplitude is concurrent with an improvement of the motor ability of the subject, as clinically observed.
- In some cases it has been possible to observe surface EMG signals at the very beginning of the reinnervation process, when only few MUs (3/4 MUs) were active in the detection volume.
- The evolution of the recruitment strategy can be appreciated starting with isolated MUAPs and evolving to a burst activity up to a complete interferential pattern, as shown in Figure 1.

CONCLUSION
Preliminary results indicate the suitability of the surface EMG technique for the follow-up of reinnervation processes of superficial denervated muscles. Further investigation is required to extend the data analysis to a higher number of subjects.

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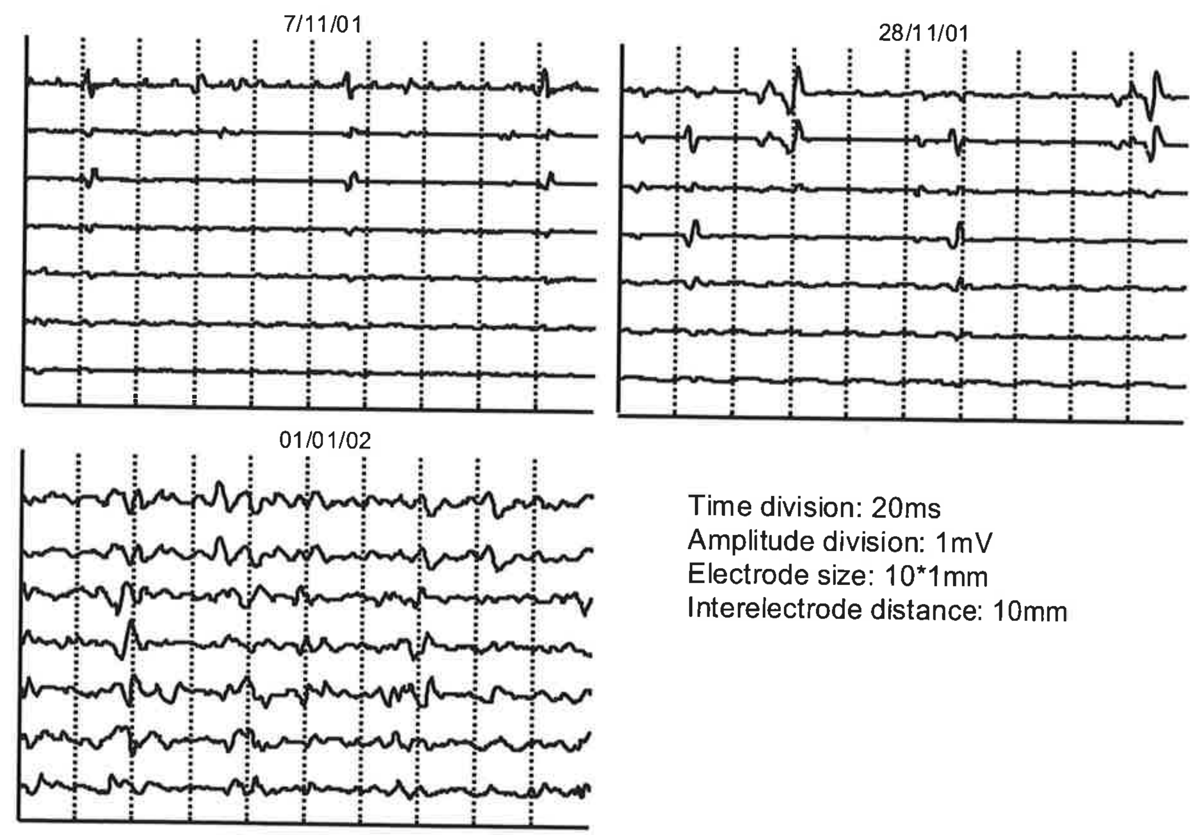

Figure 1. Surface EMG signals detected from the extensor carpi ulnaris of a subject with lesion of the radial nerve in a recovery stage, during a wrist extension exercise at the maximum voluntary force. It is possible to identify the reinnervation of new MUs..

PRINCIPLES OF STATISTICAL GAIT ANALYSIS

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INTRODUCTION
The difficulty of obtaining reliable, repeatable, and user independent data in gait analysis is one of the reasons that limit the diffusion of this technique. Gait analysis requires the collection of numerous biomechanical and myoelectric signals. Specifically, among biomechanical signals the most important and widely used are those obtained from foot switches, which describe gait timing and the modality of the foot-floor contact. Next, are often considered joint angles, acquired by means of electrogoniometers, and the acceleration of body segments, obtained thanks to mono or bi-axial accelerometers. In this presentation we focus on signals obtained by foot switches and on surface myoelectric signals.
This contribution shows that simple statistical procedures and a proper test protocol allow for obtaining highly reliable and user independent results, thus allowing for a wide clinical diffusion of this technique.

EXPERIMENTAL PROTOCOL AND SIGNAL ANALYSIS
To obtain user independent results, the test protocol must be as simple as possible. Moreover, it must be taken into account that often a proper space, for allowing patients to walk straight for at least 15 – 20 meters, is not available. To overcome this problem, we simply ask the patient to walk back and forth several times over a distance of at least eight meters, in order to collect 100 – 150 gait cycles. Signals are sampled continuously, without any need for the operator to start or stop the acquisition when the patient accelerates, decelerates, or changes direction.

<u>Foot switch signals</u>: three foot switches are positioned under each foot of the subject, respectively in correspondence to the first and fifth metatarsal heads and the heel. On-off signals from the switches are encoded to obtain four- and eight-level basography. In the following only four-level coding is considered. The only choice left to the user is the selection of the reference cycle, which is defined as the sequence of phases that characterizes the specific gait cycle the user is interested in. It must be noted that pathological subjects often show more than just a single type of gait cycle, and then the analysis described below may be repeated several times, for each specific phase sequence. Once the gait cycle has been chosen for each foot, a first software procedure selects all the cycles that are characterized by the same sequence of phases, without taking into account the statistics of the duration of each phase. In this way, most of gait cycles related to patient turns, acceleration and deceleration are rejected. To make sure of restricting the analysis to significant cycles only, a second procedure consisting on multivariate rejection of outliers (Hotelling's T-test) allows rejecting those cycles that show the correct sequence of phases but whose timing differs too much from the sample mean vector ($p \leq 0.2$). Statistics are then carried out on the reduced ensemble only.

Surface myoelectric signals: surface myoelectric signals are used in an on-off fashion, to detect muscle activation. This is obtained through a double threshold statistical detector previously developed [1]. The statistical analysis of on-off muscle activation is then restricted to the gait cycles belonging to the reduced ensemble only. In this phase, no arbitrary choice is left to the user.

RESULTS AND DISCUSSION

Data presentation: the statistics of the four phases of the reference cycles are presented in terms of mean value, standard deviation, and standard error for each phase considered as an univariate variable, as well as in terms of multiple confidence intervals of the mean vector and covariance matrix following a multivariate approach. Multivariate results allow for a sound comparison between sets of data obtained in different tests on the same subjects, to statistically demonstrate changes in his/her performances.
Muscle activation pattern is derived statistically in a similar way. To facilitate the understanding of the entire data set and of possible correlations among muscle activation, a single plot presents patterns and foot-floor contact data jointly.
Reliability of the results: following the procedures presented above, in normal subjects we obtain standard errors on the duration of gait phases as low as 2%-3%. When different operators repeat tests in different days, the repeatability of results is generally within the standard error cited above. Similarly, in normal subjects we obtain standard errors as low as a few percents when estimating the on-off and off-on transitions of muscles, with high repeatability in different days.

CONCLUSION

In conclusion, the proposed technique allows for obtaining highly repeatable and user independent data in gait analysis and we believe its diffusion could play an important role in facilitating its clinical diffusion.

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THE ROLE OF PUSH-OFF IN THE STANCE LIMB TO PREVENT FALLING AFTER TRIPPING

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INTRODUCTION

Falls and fall-related injuries are common, costly and serious medical problems, especially for the growing population of elderly people. Falling is often the result of a trip. After impact of the swing limb with an obstacle, the body gets a forward rotation. To prevent a fall, it is necessary to rapidly arrest this forward rotation to prevent a fall. Positioning of the recovery foot properly in front of the body is one means to reduce the forward rotation (1-3). Besides the swing limb, the stance limb may play an important role in the recovery reaction after tripping. Push-off with the stance limb can have two functions: first, initial counteraction of the angular momentum of the body, and secondly, gaining time and clearance for adequate positioning of the recovery limb for further arresting of the forward rotation. The present study investigated whether and how push-off of the contralateral stance limb is used to reduce the forward rotation of the body after tripping.

METHODS

Twelve subjects walked over a platform of 10 meters at a comfortable walking speed, on average at 1.64 m/s. Kinematic data and ground reaction forces were measured at 100 Hz, and EMG of lower limb muscles at 1000 Hz. Based on the online kinematic data, one of 21 obstacles, hidden in the platform, could be made to appear unexpectedly from the platform and catch the swing leg at exactly mid-swing. In about 10 of 60 trials, subjects were actually tripped over an obstacle on either left or right side. Safety was ensured by a harness attached to a ceiling-mounted rail. Data of 5 trials of normal walking were averaged over subjects and compared with averages of 5 left limb tripping trials. Joint moments in the stance limb during both walking and tripping conditions were calculated by use of an inverse dynamics model.

RESULTS & DISCUSSION

Recovery reactions appeared to be reproducible within subjects; immediately after collision the obstructed swing leg was elevated over the obstacle. Swing phase of the swing limb was prolonged, while the stance limb provided prolonged push-off. After push-off a short aerial phase was seen instead of the normal double support phase. Stride length was increased from 1.66 m during normal walking to 1.82 m after tripping.
Due to impact with the obstacle, the body rotated forwards, with a major contribution of the head-arms-trunk (HAT) segment. The averaged peak HAT angle was 42.4° with the vertical and the peak forward angular velocity was about 270°/s. This increased angular velocity was almost completely arrested within the push-off duration. The contribution of the stance limb to this counteraction is reflected by an increase in (peak) ground reaction forces in vertical as well as forward direction by respectively 33 and 61 %.
The required push-off force magnitude and direction were generated by hamstring and gastrocnemic muscles in the stance limb, which showed increased activity very rapidly. These muscle activities generated an increased ankle plantar flexion and an extension moment around the hip and a flexion moment around the knee, where in normal walking hip flexion and knee extension are generated during push-off (see figure).

CONCLUSION

Based on these results, it may be concluded that the push-off in the stance limb contributes to prevention of falling after tripping. By a rapid increase in activation of the hamstring and triceps surea muscles, adequate joint forces and moments are generated to counteract the forward rotation of the body. Additionally, push-off resulted in extra time and space (increase of swing phase and an aerial phase and increased recovery step length), which facilitate adequate positioning of the recovery limb for further arresting of the forward rotation.

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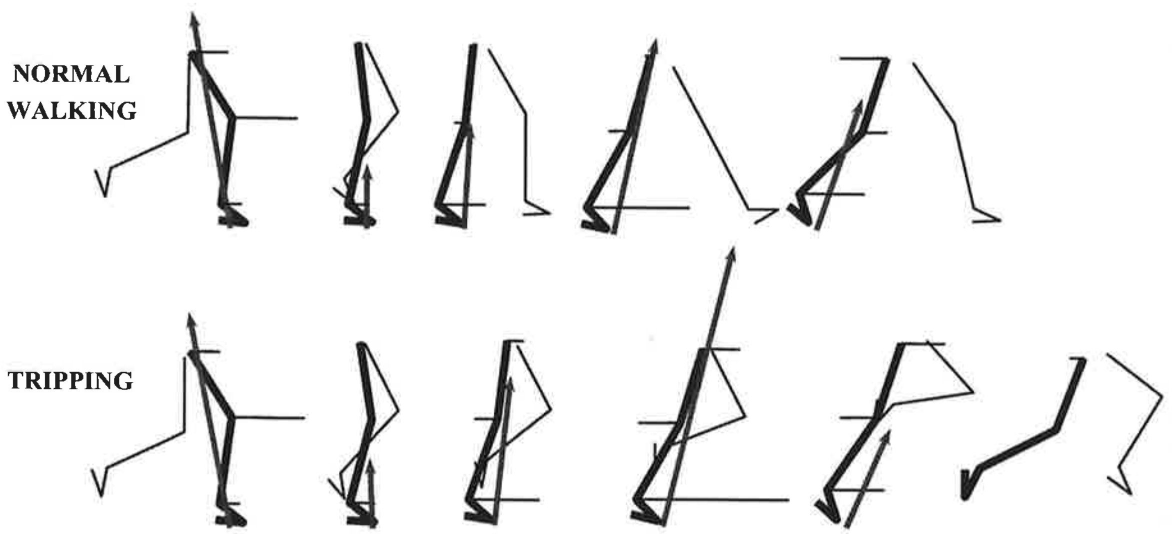

FIGURE: Stick figures of several comparable phases during normal walking or tripping. Arrow indicates the Ground Reaction Force. The GRF is increased and directed more forward after tripping. Horizontal bars in stance limb joints represent the joint moments. A bar to the right indicates an extension moment, a bar to the left a flexion moment. Note the opposite in hip and knee joint moments just during push-off of normal walking and tripping.

THREE-DIMENSIONAL ULTRASONIC GAIT ANALYSIS IN SCHIZOPHRENIC PATIENTS

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INTRODUCTION

Human locomotion is controlled by spinal and central motor programs. Schizophrenic disorders as well as neuroleptic treatment can affect common motor tasks. In general, severity of neuroleptic-induced extrapyramidal symptoms is assessed by using clinical ratings.
Aim of the study was to assess the influence of neuroleptic treatment on human gait. Therefore we used a system for precise three-dimensional ultrasonic movement analysis for the assessment of gait under neuroleptic treatment.

METHODS

The basic principle of ultrasonic three-dimensional gait analysis is the use of tiny ultrasonic markers. These markers are attached to anatomically defined positions of the body. Ultrasound impulses are detected by three stationary microphones mounted on a panel on each side of a walkway. Based on transmission times of the ultrasonic impulses between the markers and the microphones, the three-dimensional spatial position of each marker is calculated every 20 msec. Thus, several spatial and temporal gait-variables can be assessed. The basic variables calculated are stride length, cadence and velocity. Velocity of over ground locomotion is determined as the product of stride length and cadence. The cadence is the number of steps per time unit. Velocity can be increased either by longer steps or by more steps per time unit.
In our setting, the patient is asked to walk at a self-selected speed on a 8m-walkway and then to walk on a treadmill with three predefined velocities. The slowest velocity is 0.38 m/sec. Gait at this velocity requires good coordination and is little automatized. The third velocity with 0.86 m/sec represents for most probands a quite normal relaxed and casual gait. The second velocity is 0.58 m/sec.
Ultrasonic three-dimensional gait analysis was performed in 2 groups of schizophrenic patients according to DSM-IV. Group 1 consisted of 15 patients whose neuroleptic treatment was switched from conventional neuroleptics to olanzapine. 10 patients got haloperidol at a mean dose of 7.1 mg/day, 4 patients fluphenazine at a mean dose 7.3 mg/d and 1 patient 10 mg flupentixol per day. They were switched to olanzapine with a mean dose of 16.7 mg/day. We assessed gait parameters under conventional treatment and under treatment with olanzapine after the change.
Group 2 consisted of 10 neuroleptic naive schizophrenic patients according to DSM-IV who started treatment with olanzapine. Gait parameters were assessed before treatment and under mean dose of 16.7 mg/d.
Concomittant medication was limited to medication without effects on gait or motor performance. There were no significant differences in age and sex distribution, height and weight between the patients of the two groups. We used the ESRS for the clinical rating of extrapyramidal side effects and the PANSS for rating the positive and negativ symptomes of schizophrenia. The nonparametric Wilcoxon test for dependent samples was used for statistical analysis.

RESULTS

After treatment change from conventional neuroleptic to olanzapine in group 1 velocity of over ground locomotion ($p<0.05$) and stride length ($p<0.007$) increases markedly. Stride length under olanzapine treatment after the switch of the medication is similar to the stride length in the group of neuroleptic naive patients before and during treatment with olanzapine. In contrast to stride length there are no significant changes in cadence in group 1. At the predefined slowest velocity (0.38 m/sec) in both groups stride length increases and cadence decreases after the treatment change. The alterations are more marked in the group with change from conventional neuroleptics to olanzapine than in the untreated group. Assessments at the intermediate velocity (0.58 m/sec) show a similar picture. At the velocity of a normal automatized gait (0.86 m/sec), the results are similar to the results we got at the assessment of normal gait on the walkway at self-selected speed. Differences between patients without neuroleptic medication and those with olanzapine are not significant.

DISCUSSION

Dynamic gait parameters are significantly influenced by neuroleptic treatment. At normal gait velocity – the highly automatized gait – treatment change from conventional neuroleptics to olanzapine leads above all to an increase of stride-length, whereas cadence remains stable. Patients walk faster, mostly because they make longer steps and not because they make more and faster steps. The increase of stride length and decrease of cadence at 0.38 m/sec in both groups after the treatment change might partly be due to a learning effect.

CONCLUSION

Clinical findings of a reduction of the ESRS-scores under treatment change can be objectively confirmed by three-dimensional ultrasonic gait analysis. The degree of impairment can be objectively measured by testing spatiotemporal and kinematic gait parameters. Neuroleptics can be compared regarding differences of impairment. Impairment of gait is more pronounced under treatment with conventional neuroleptics then under treatment with olanzapine.

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INTERPRETATION OF THE MUSCULAR COORDINATION PATTERN SUPPORTED BY FUZZY LOGIC

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INTRODUCTION

The information about the co-ordination pattern of different muscles or muscle groups is of high relevance not only in the treatment of movement disorders but also in other fields e.g. rehabilitation medicine or sport sciences. Abnormal muscular co-ordination patterns often hinder the performance of an effective movement and results frequently in serious functional impairments.

Although the abnormal muscular co-ordination pattern is often the reason, in clinical praxis the characterisation of the movement deficiencies is still based on the analysis of the mechanics of the movement. 3D movement analysis has been turned out to be a proper tool to achieve objective data about kinematic, kinetic, forces or power of the movement but the methodology does not yield any information about the muscular co-ordination pattern. Surface-EMG signals detected simultaneous at different muscles are suitable to assess the muscular co-ordination pattern and would provide the needed information, but the interpretation of the surface-EMG signals is time consuming and often difficult even for experienced users. This is why surface-EMG is so far not been introduced to clinical routine (Winter 1984, Winter 1987).

This paper presents a new approach based on fuzzy-logic which supports the interpretation of the surface-EMG signals and facilitates the assessment of the muscular co-ordination pattern. This methodology contributes to utilise surface-EMG and the information about the muscular co-ordination pattern in clinical routine. The procedure is exemplified by the muscular co-ordination pattern of the ankle joint movement during gait in healthy volunteers and patients suffering from spasticity in gait.

METHODS

To support the interpretation of the surface-EMG signals in order to get the information about the effectiveness of the muscular co-ordination pattern, an expert-system has been created which is based on the fuzzy-inference-method. This method makes the management of uncertain, complex, logical connections with linguistic instruments possible. In a first approach the methodology has been exemplified by the muscular co-ordination pattern during gait. Thus, the surface-EMG signals of the most relevant muscles during ankle movement (tibialis anterior, soleus and gastrocnemius) have been validated by the expert system. The knowledge-base of the expert system takes into account that in those cases in which agonist and antagonist are synchronously active the movement of the joint is not effective. Those cases in which the co-activation is needed to stabilise the joint are excepted. The evaluation of the surface-EMG signals by the fuzzy-inference regards the level of activation of each muscle, the contribution of the different muscle to the resulting movement and the moment of activation within the gait cycle. This information results in 11 rules e.g. *"If the tibialis anterior and the gastrocnemius are simultaneous active, than the movement of the ankle joint is not effective"*. Each rule is weighted according to its relevance for the ankle movement. The resulting output measure of the fuzzy-system has been called *"Effectiveness"* of ankle joint movement. It indicates to which extent and at what periods of time the muscular co-ordination pattern yields an effective movement of the ankle. Figure 1 schematises the development of the different rules and their outcome measure.

11 Rules for the Fuzzy-Inference:

TibAnt active and Gastroc active	→	Effectiveness poor
TibAnt active and Soleus active	→	Effectiveness poor
TibAnt active and (Soleus + Gastro) active	→	Effectiveness very poor
⋮		⋮
TibAnt not active and Gastro active	→	Effectiveness good
TibAnt not active and Soleus active	→	Effectiveness good
TibAnt not active and (Soleus + Gastroc) active	→	Effectiveness excellent

Figure 1:
11 rules are used for the fuzzy-inference which estimates the effectiveness of the out coming movement based on the muscular co-ordination pattern. The figure gives a schematic summery in which way the different rules are build.

Conventional bipolar surface-EMG has been recorded simultaneously from the tibialis anterior, the soleus and the gastrocnemius medialis muscle according to the SENIAM recommendations (SENIAM 1999). EMG data have been rectified and smoothed by a 10 Hz high pass filter. For each muscle different EMG-sweeps have been synchronised to the gait cycle, which has been detected by food-switches. The EMG-sweeps have been averaged and normalised to the maximum EMG amplitude appearing in each EMG envelope. The normalised EMG envelopes of the different muscles have been used as the input parameters for the expert system.

RESULTS AND DISCUSSION
The expert-system has been verified at the ankle movement of healthy volunteers and patients suffering from spasticity in gait. 20 Patients aged between 5 and 12 years and 10 adult volunteers have been evaluated with the fuzzy-system.

Figure 2 shows the results of the fuzzy-logic based interpretation of the muscular co-ordination pattern of a healthy volunteer and a patient suffering from spasticity in gait. In the case of the healthy volunteer the fuzzy system shows that the effectiveness of the ankle joint motion is high during the complete gait cycle. This is due to an normal muscular co-ordination pattern where agonistic and antagonistic muscles work in an effective way. In contrast to the healthy case in patients with spasticity the effectiveness of the motion is not as good as in healthy ones. Especially in the beginning of the stance phase and at the end of the swing phase the muscular co-ordination pattern induces a motion which effectiveness is poor. In patients with less pronounced spasticity the effectiveness of the motion can increase to normal values again. The degree of loss of effectiveness and the time of recovery typical for each patient corresponds to the degree of spasticity.

The example of the ankle joint motion in spastic and healthy gait shows that the interpretation of the muscular co-ordination pattern detected with surface-EMG can be supported by the fuzzy-inference methodology. This is an essential prerequisite for the use of the information about the muscular co-ordination in clinical routine.

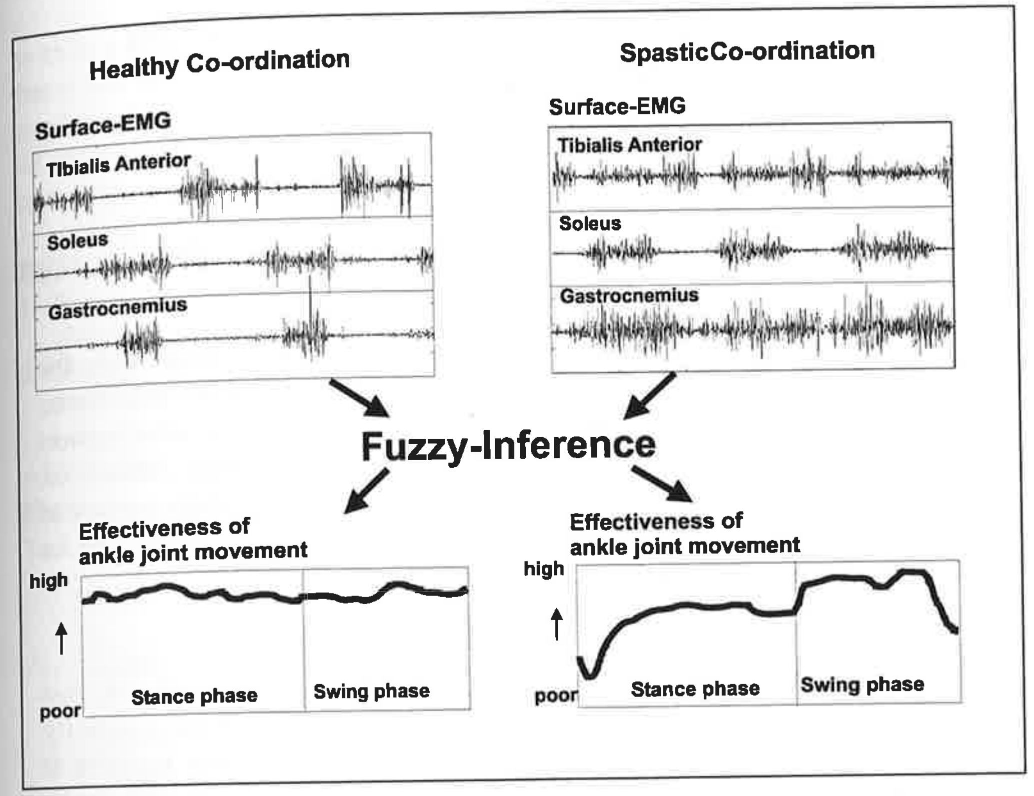

Figure 2:
Fuzzy-based interpretation of the surface-EMG signals during ankle movement. Compared are a healthy volunteer and a patient suffering from spasticity in gait.

The example of the relatively simple ankle movement demonstrates the principles' feasibility. However the structure of the methodology allows to extend it to more complex movements with a higher number of joints and muscles involved. This opens new information about the movement not only for the treatment of patients with movement disorders.

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REPRODUCIBILITY OF SURFACE EMG MEASUREMENTS IN CP CHILDREN

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INTRODUCTION

Clinical movement analysis as well as dynamic electromyography are accepted tools for evaluation of gait pathologies in CP children (1). In these patients, multi-level soft tissue surgery may be performed in order to improve gait function. The success of the intervention can again be evaluated with the above mentioned objective measurements. However, comparisons between pre and post intervention measurements rely on a certain consistency of EMG measurements. Therefore, it is mandatory to know how repeatable EMG data can be obtained in these patients. This was the objective of the present study.

METHODS

In eight patients that were submitted to our department for soft tissue surgery (Tab. 1), two measurements were recorded on separate days (PRE1 and PRE2) in order to test for reliability and reproducibility. Electromyographic measurements were performed during free walking at self-selected speed with bipolar surface electrodes from the following six thigh and shank muscles: rectus femoris, vastus lateralis, semitendinosus, tibialis anterior, peroneus longus, gastrocnemius medialis. Data was sampled and digitized at 1000 Hz with 12 bit resolution. Data processing involved an automatic burst detection algorithm (2) for determination of amplitude and timing (onset and offset) of EMG activities for at least 20 gait cycles.

RESULTS

In total, 90 PRE1 timing parameters were compared to 90 PRE2 timing parameters of eight subjects. In two subjects with hamstring tendon lengthening, the two pre-intervention recordings of the rectus femoris showed continuous activity. Therefore, no on- and offsets times and amplitudes outside the burst were found. The average difference between the PRE1 and PRE2 timing parameters was 3.5±5.7 % of the gait cycle. Fig. 1 shows an example of two repeated recordings with the detected timing parameters for the semitendinosus and vastus lateralis. No significant timing differences can be seen between the PRE1 and PRE2 measurements. Thirteen of the paired timing parameters were significantly different ($p<0.05$). In one subject (#14) all six timing parameters were significantly different. However, the profiles were almost identical except they were shifted in time (mean 7.4 %). The lengths of the bursts were not significantly different. It seems that there was an irregularity in marking the heel strike events between the two trials. From the other 7 significantly different timing parameters, 4 were within 6 % of the gait cycle. The average difference in amplitude between PRE1 and PRE2 was 2.7±2.5 µV. This is 20±15 % with respect to the amplitudes in PRE1.

DISCUSSION & CONCLUSION

By studying the two pre-intervention recordings, it can be concluded that timing parameters in the repeated measurements before surgical intervention are sufficiently reproducible. 86 % of the measurements showed no significant differences between the two measurements. From the 14 % that do show significant differences, most (43 %) are due to systematic differences in one subject and 38 % are within 6 % of the gait cycle. The average difference in timing of 3.5 % appears adequate for gait analysis. Even though the absolute amplitude differences are within a few µV, the relative differences are larger since the amplitudes in and outside the burst are relatively small. Differences in timing parameters between measurements can be explained by differences in muscle coordination and different walking speeds. Further sources of differences might be irregularities in marking heel strike events, or estimation errors in burst detection.

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Tab. 1: Subject data and type of surgical procedures that were performed in 8 subjects.

	Age [years]	Height [cm]	Body mass [kg]	Operated limb(s)	Surgery performed*
Joh04	17,9	172	53	Right	Ham
Joh05	18,6	189	69	Right	Ham, ATL, TAT
Joh06	14,9	169	54	Both	Ham, ATL, Add
Joh07	12,9			Both	Ham, Add
Joh08	13,8	158	44	Both	ATL, & others
Joh11	9,3	139	43	Both	Ham, ATL, Add
Joh13	18,0	165	55	Right	ATL, TAT,
Joh14	15,3	163	67	Both	ATL
Mean	**15,1**	**165,0**	**55,0**		
SD	**3,1**	**15,1**	**10,1**		

*Ham = Hamstring lengthening, ATL = Achilles tendon lengthening, TAT = tibialis anterior transfer, Add = Adductor lengthening.

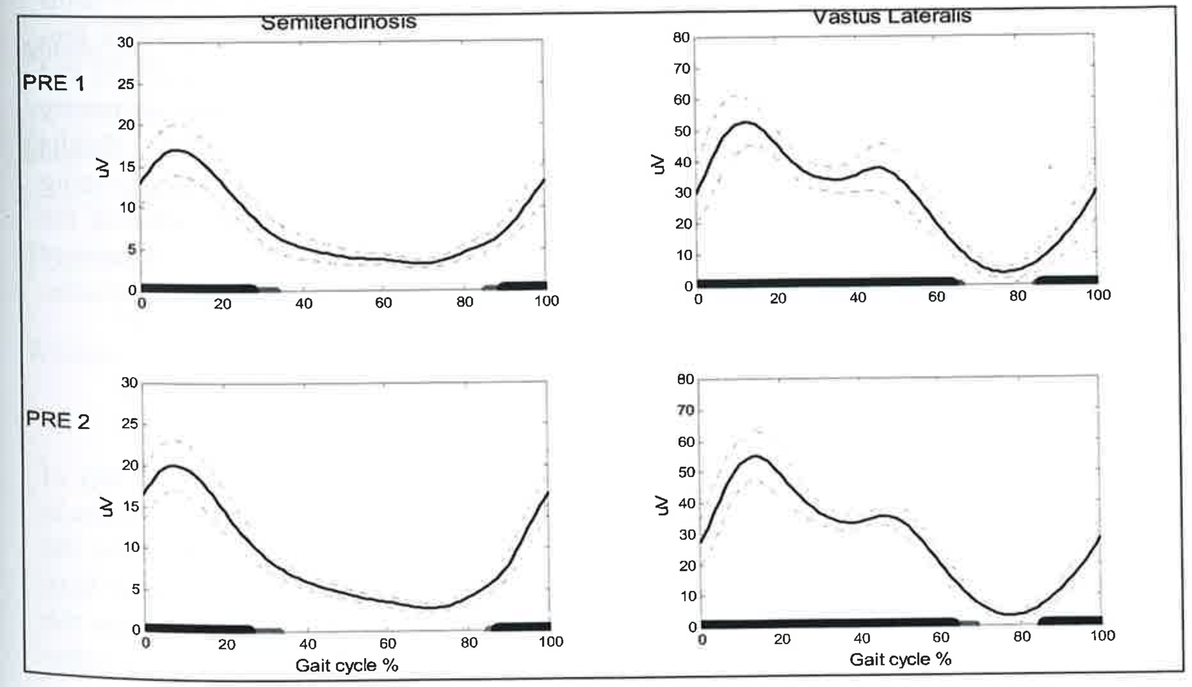

Fig. 1: Above, 1st recording; below, 2nd recording; left: semitendinosus, right: vastus lateralis.

PATTERNS OF ELECTROMYOGRAPHIC ACTIVITY DURING FOUR METHODS OF A SIT-TO-STAND TASK

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INTRODUCTION

Many studies have reported on the EMG activity of muscles during gait, but although rising from a sitting position is a compulsory pre-requisite to gait, very little research has been conducted on the muscle activity patterns of the sit-to-stand (STS) movement. Previous EMG studies found no difference between conditions of rising with or without using armrests (Kelley, Dainis, & Wood, 1976; Munton, Ellis, & Wright, 1984; Wheeler, Woodward, Ucovich, Perry, & Walker, 1985). Reduced intervertebral disk pressure during the STS maneuver was reported for rising using armrests or pushing off the thighs (Andersson, Ortengren, & Nachemeson, 1982), but muscle activity during these methods of rising has not been previously published. The purpose of the present study was to compare the EMG onset times and durations of four methods of the STS task. Although several previous studies used constrained methods of chair height (Alexander, Koester, & Grunawalt, 1996; Burdett, Habasevich, Pisciotta, & Simon, 1985; Munro, Steele, Bashford, Ryan, & Britten, 1998) and foot position (Fleckenstein, Kirby, & MacLeod, 1988; Khemlani, Carr, & Crosbie, 1999; Shepherd & Koh, 1996) to standardize experimental protocol, this investigation focused on the normal conditions every individual must use for rising from a standard seated chair position.

METHODS

One-hundred subjects (50 males, 50 females; mean age 22.4 years) performed four methods of rising from a standard height chair (43 cm). The four STS methods were rising with: arms FREE; arms CROSSed; hands on the KNEEs; and hands on the ARMrests. IOMED pre-amplified surface electrodes were attached bilaterally over the right erector spinae (RES) and left erector spinae (LES) muscles. Electrodes were also attached to the right lower extremity over the: gluteus maximus (GM); medial hamstrings (MH); rectus femoris (RF); tibialis anterior (TA); and triceps surae (TS) muscles. Subjects were instructed to rise naturally using each of the STS methods. A light directly in front of the subject signaled the initiation of the movement and simultaneously initiated recording of the EMG signals. Separate one-way analysis of variance with post-hoc analysis was used to compare conditions for each muscle. A $p < .05$ level of significance was used for all comparisons.

RESULTS

There were no significant differences between onset times over the four conditions for any of the muscles. Invariantly the TA was the first muscle recruited and the TS was the last muscle activated in the sequence Significant differences between muscle durations occurred for the RES, LES, MH and RF. The FREE condition EMG durations were significantly shorter than the other three STS conditions for the RES and RF muscles and significantly shorter than the CROSS condition for the LES. The CROSS condition was significantly longer than the other three conditions for the MH durations.

DISCUSSION

The results suggested that muscles contracting across joints are temporally linked and variations of muscle durations may be used in the lower limb and low back to coordinate the sit-to-stand action under different functional demands. The sequence of onsets were consistent across the four conditions although durations varied. The natural parameters used in this study for seat height and foot position could provide more generalizable information for the practitioner in analyzing STS movements. Also, when rising from a chair without armrests, these results indicate moving with the arms free may be more efficient than using the hands on the knees as is commonly recommended by clinicians.

CONCLUSIONS

It was concluded that each method of rising had unique EMG pattern characteristics during the STS although onset times are invariant. These data may be useful for comparing to pathologic conditions when considering rehabilitation strategies during relearning of the STS task.

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THE EFFECT OF EXTERNAL SUPPORTS ON THE AMPLITUDE OF THE SURFACE EMG SIGNAL RECORDED FROM LUMBAR SPINE MUSCLES DURING MAINTENANCE OF A FORWARD-FLEXED POSTURE

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INTRODUCTION

Forward-flexed postures have been shown to increase indicators of spinal compressive loading, such as lumbar EMG.[1,2] Maintenance of forward-flexed postures and spinal compressive loading are known to be risk factors for low back pain and dysfunction (LBPD),[3,4] Published data is scarce on effective strategies that allow people to maintain forward-flexed postures without increasing surface EMG activity of the lumbar paraspinal musculature and other indicators of spinal compressive loading. External supports such as lumbosacral corsets have been widely prescribed and used for the prevention and management of LBPD. Among the mechanisms by which they are thought to be effective is a decrease of spinal compressive load (via decrease of intra-discal pressure, and lumbar paraspinal musculature activity and contraction force).[5] However, studies that have investigated this mechanism included very small sample sizes, and are equivocal or unsupportive regarding the usefulness of lumbosacral corsets,[6-8] as are studies that have assessed injury prevention.[9,10]

The purpose of this study was to compare the effects of using an elasticized lumbosacral corset (ELSC), an electromechanical device called the Spine Sling Support (SSS), and no support (NS) on the magnitude of the surface EMG activity of the lumbar paraspinal muscles during maintenance of a forward-flexed posture. The initial value of the root-mean-square (RMS_{init}) was the dependent variable. Our hypotheses for the comparisons among the conditions of support were: 1) there would be no difference in RMS_{init} between an ELSC and NS; 2) the SSS would manifest lower values for RMS_{init} than an ELSC and NS.

METHODS

Thirty-three uninjured volunteers (24 men, 9 women), 19-40 years of age (mean 28.4 years), were recruited from the Politecnico di Torino community in Italy. Subjects maintained a forward-flexed posture in each of three conditions of support: 1) ELSC; 2) SSS; 3) NS. The position of testing, based in part on research by Schultz and colleagues,[1] was the same for all conditions of support, and was verified in each case by the experimenter. The SSS is an electromechanical back and body support.[11] In part, it consists of a motor, spool, support line, pulleys, vest, belaying track, and remote control switch. The ELSC was of similar design to corsets commonly used in the workforce, with nylon fabric, suspenders and Velcro enclosures.

Surface EMG signals were collected bilaterally from longissimus thoracis at the L1-2 spinal level, iliocostalis lumborum at L2-3, and multifidus at L5-S1, using active single differential electrodes having two parallel detection bars. Each bar was 10 mm long and 1 mm thick. Inter-electrode distance was 10 mm. During the recordings, each bar was aligned perpendicular to the direction of the muscle fibers so that the perpendicular distance between the bars was aligned parallel to the direction of the muscle fibers. The ground electrode was placed over the wrist. Signal acquisition system specifications were: bandwidth 10-500 Hz (-3 dB; 12 dB/octave roll-off), CMRR greater than 90 dB, input equivalent noise 1 μV_{rms}, linearity error < 2%. Sampling rate was 1,024Hz. Samples were digitized and stored using a computer equipped with a 12-bit A/D card (Microstar 2400). Signal processing occurred off-line.

RMS values were computed for each of 30 s of the contraction for each muscle. An ordinary least-square regression was performed on the 6th through the 30th second (epoch) of the RMS time-course, to obtain the RMS_{init} (intercept with the y-axis) along with its 95% confidence interval. The first 5 s of EMG data were disregarded to avoid possible artifacts at the beginning of the contraction due to postural adjustments. Any epoch's RMS value that was outside of the 95 % confidence interval was considered an outlier and removed. In such infrequent occurrences (less than 5%) the regression line was re-estimated after removal of the outliers, and the RMS_{init} value was computed again.

Repeated measures ANOVAs were performed to determine if the RMS_{init} was significantly different among the conditions of support, for each muscle site (i.e., six ANOVAs, total). Difference contrast tests (*a priori* planned comparisons using multiple dependent t-tests) were performed following the ANOVAs to show which pairs of support conditions were significantly different. The alpha level for the contrast tests was 0.05/3. Based on the research hypotheses, the contrast test comparing ELSC versus NS was two-tailed, and the contrasts comparing the SSS to the ELSC and to NS were one-tailed.

RESULTS

Mean RMS_{init} values for the six muscle sites ranged from 17.7 µV to 21.7 µV for the ELSC, 16.3 µV to 20.6 µV for NS, and 6.0 µV to 10.5 µV for the SSS. The SSS was significantly lower ($p<0.001$) than the ELSC and NS at all muscle sites, by about 50-75%. The ELSC and NS were not significantly different at any muscle site except one; NS was lower ($p=0.011$), but by only 1.5 µV (about 8%).

DISCUSSION

In a preliminary study, our data analyses from within-subject comparisons among the conditions of support demonstrated that there was no systematic bias affecting RMS values due to pressure exerted on the electrodes by the ELSC.[12] This is consistent with other reports.[6,7]

The mean RMS_{init} levels recorded were generally low, and those associated with SSS use were extremely low. The 95% confidence interval for the signal was approximately ±3µV. Thus, in most cases using the SSS, the signal appears to have consisted largely of noise.

CONCLUSION

The SSS was useful for reducing the magnitude of lumbar paraspinal muscle activity during maintenance of a forward-flexed posture, while the ELSC was not.

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MUSCLE ACTIVATION CHARACTERISTICS IN CHRONIC PAIN

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INTRODUCTION

In our western world, there are a large number of people with chronic pain, preventing them to participate actively in work and social life. Several models have been developed to create a better understanding of the process of the development of chronic pain as well as to answer the question why some subjects develop chronic pain while others in an apparently very similar situation do not. Most of these models recognise the multidimensional complexity of this process by involving both personal and environmental factors. A common characteristic in most models concerns changes in the motor control of the muscles in the painful (and contra-lateral) area, reflected in abnormal muscle activation pattern. To summarise some of the models with respect to this aspect:

- The pain adaptation model (Lund 1991) suggests changes in muscle activation patterns, especially a lowered activation level of muscles in the painful region and more antagonistic activity.
- The pain-spasm-pain model (Johansson 1991) suggests a circle process of pain development and higher muscle activation levels, which again contributes to more pain.
- The occupational health model of Birch links a dysbalance in workload and environmental aspects to short-term and long-term changes in muscle activation pattern
- The Cinderella theory (Hagg 1991) links a lack of sufficient relative muscle rest (EMG gaps) to damage of muscle fibres and the occurrence of chronic pain

Brutally summarising the models, one would expect that after an initial occurrence of pain, protection of the painful area is done in two ways:

- Avoiding patterns: a decrease of muscle activity and more activity of synergistic and antagonistic muscles
- Stiffening patterns: hyperactivity of the muscles and a decreased ability to relax.

Both types of patterns are found in experimental studies. Some models strongly suggest that the "choice" between these two patterns is related to personal factors (e.g. Vlaeyen 1995, Hasenbring 1996). However, it is still far from clear how and when abnormal motor control of muscles develops, who develops it, why and whether there are differences between different chronic pain disorders like e.g. whiplash syndrome and computer related muscular pain. In our centre a research program has been started towards characterisation of muscle activation patterns in subjects with chronic pain, to answer these questions and to provide starting points for a more efficient treatment.

METHODS

SEMG measurements were made from the Trapezius muscles during epochs of 15 seconds in each minute of the task (sample rate 1024 Hz; filters set at 10 and 250 Hz). Analysis was focused on amplitude behaviour (RMS calculation during 100 mS windows).
Tasks involved concerned a standardised typing task, a stress task requiring no motor activity, a gross motor task involving paced putting of marks on paper and a reference task (Mathiassen). Three groups of subjects were involved: subjects with chronic pain related to a whiplash (1), related to computer repetitive strain injuries (RSI)(2) and healthy controls (3).

RESULTS

A summary of results in chronic pain with respect to muscle activation in several different conditions is given. The terms higher and lower refer to the values obtained with a group of healthy controls.

Muscle activation during motor tasks
During the typing task higher EMG levels were found in the whiplash and in RSI groups. However during the dot task, lower EMG levels were found in whiplash patients. Remarkable is that these lower levels are found already at the very early stage of the whiplash syndrome (Nederhand). Another remarkable finding is that in one handed tasks the differences are most pronounced in the Trapezius muscle at the non-active side.
No evidence was found that the lower muscle activation was "compensated" for by more active other muscles (like Deltoid and Infraspinatus)

Muscle activation during stress task
During the stress task, muscle activation is in general higher in the whiplash and the RSI subjects. This is especially evident in the non-dominant side.

Post task relaxation
Whiplash subjects at the chronic stage show a decreased ability to relax (Nederhand 2000). However, this phenomenon is not present at early stages of whiplash, up to 6 months after the accident (Nederhand, this proceeding)

Reference task
The reference task was developed by Mathiassen (1996) to decrease inter subject variability, i.e. to enable a better comparison between subjects. However we found lower EMG values in subjects with whiplash. No clear evidence was found that this lowered activity is compensated for by higher activity in other muscles. In contrast, we found higher then normal values in subjects with RSI.

CONCLUSIONS

The results of our studies indicate a large inter-subject variability in all groups, but some patterns are becoming apparent. Subjects with chronic pain generally show more non-functional (psychogenic) muscle activity, i.e. activity not necessary for the task they are performing. This is reflected especially in activity in the Trapezius muscle at the side that requires no muscular activity and a decreased ability to relax the muscles after a task.
A second characteristic, which is less consistent, is the activity level during the task. Both higher and lower levels are found in comparison with healthy controls. This difference is probably related to differences in the task but also to differences in the chronic pain syndrome (whiplash vs computer related pain).
In a related project, a feedback therapy has been realised, in which the EMG is processed and an alarm is given when there is insufficient muscle rest (Hutten et al. this proceedings). A first study shows that muscle activation patterns can be changed using this approach. Also the experienced pain decreases. This finding supports the idea that there is some relation between experienced pain and muscle activation patterns, but more important, it also supports the concept that learned abnormal control of muscles can be changed again into more healthy muscle activation patterns.

THE INFLUENCE OF MYOFEEDBACK ON MUSCLE ACTIVATION PATTERNS AND PAIN IN WORKRELATED MUSCULOSKELETAL DISORDERS

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INTRODUCTION

Models and experimental studies show that in subjects with chronic pain, the muscle activation patterns differ from those in normal subjects. This is reflected especially in a decreased ability to relax the muscles. Subjects are not aware of this muscle activation as it often concerns rather low levels of activation. Nevertheless, according to the Cinderella theory (Hägg, 1991) these low levels of activation may contribute seriously to the development and maintenance of chronic pain, when occurring during relatively long time duration. An accurate feedback on the absence of sufficient muscle rest may contribute to a greater awareness of this undesirable muscle activation and therefore may contribute to diminishing of pain. However, up today, most myofeedback equipment has been based upon the starting point that a warning to the subject should be given when the muscle activation level exceeds a certain preset level. The Cinderella theory as well as the results of the previously described research suggests a very different approach, i.e. a subject should be warned when there is less then a preset time of rest in a muscle. In the present study a myofeedback system has been developed, based on these starting points and a first study has been carried out.

Myofeedback system

The myofeedback system consist of a two channel portable system that is combined with a harnas incorporating dry SEMG electrodes to enable a stable recording of the upper Trapezius muscles (figure 1). The SEMG signal is amplified (15x), digitised (22 bits ADC) and smooth rectified with removal of the low frequency components. Embedded software provides muscle rest detection and parameterisation. Sensory feedback is provided when a pre-set level of rest is not reached. The duration of the feedback is progressively increased when there is no adequate response on the feedback.

METHODS

Twenty one subjects with computer related disorders in their neck shoulder region received continuous myofeedback for four weeks during their normal work. Before (T0), after (T1) and 4 weeks after (T2) the myofeedback training sEMG measurements of the left and right trapezius muscles were performed and pain/discomfort was measured. sEMG analysis started with placement of the electrodes (Hermens et al, 1999) and a reference test (Mathiassen,1995). Subsequently, measurements were performed during a relaxed sitting position, a standard typing task, three rest measurements, a stress related mouse task and again three rest measurements. SEMG parameters used for analysis were:

1. The muscle activation level quantified by the RMS value.
2. Relative muscular rest time (RRT). This represents the amount of time, expressed as a percentage of the total time, during which there is total muscle relaxation. Total muscle relaxation is achieved when the RMS value is $\leq 10\ \mu V$ for at least 0.25 sec.

RESULTS

The RMS values show an obvious increase during the typing and stress task with respect to the rest measurements. There is however a large variability between individuals. 65% (left) and 71% (right) of the subject show lower RMS values during typing directly after

myofeedback and still at four weeks follow up. For the stress task 35 and 41% of the subject show an improvement directly after myofeedback and at 4 weeks follow-up.
For RRT an opposite pattern is shown as found for the RMS. The RRT is high during rest and decreases during the typing and stress task. Again there is a large variability between individuals. Directly after treatment, 24% and 35% of the subject are able to relax more during the typing task compared to 41% and 47% during the stress task. At four weeks follow up these percentages are increased; 53% and 41% for the typing task and 53% for the stress task.
Before myofeedback mean pain/discomfort scores are the highest for the neck and shoulder region but are all below four on a scale from zero till ten. Directly after myofeedback, decreases in pain/discomfort scores are found for all four (neck, shoulder, arms and upper back) regions. Remarkable is that at four weeks follow up the scores show further decreases and are significantly different from the scores before myofeedback.

DISCUSSION

Results of this study show in a considerable amount of subjects a decrease in RMS, an increase in RRT and a decrease in experienced pain/discomfort directly after a myofeedback training and even further improvements are shown at four weeks follow up. The concept of myofeedback was to increase the awareness of insufficient muscle relaxation and teach the subject how to relax. However, the increasing improvements at four weeks follow up, the rather large changes directly after myofeedback as well as the personal verbal responses indicate that not only muscle awareness but also the subject's cognition concerning pain is changing.

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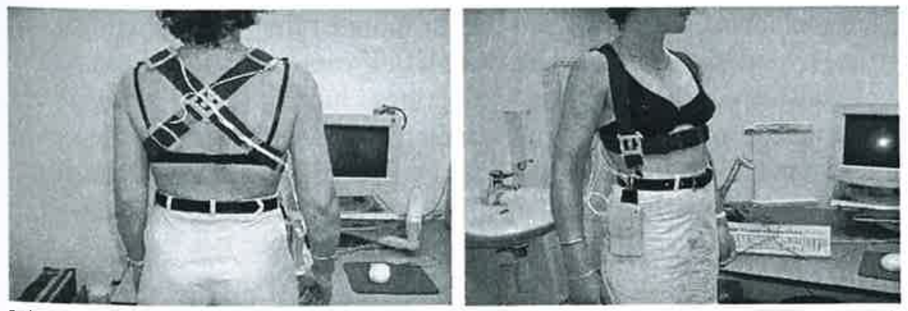

Figure 1: The myofeedback equipment consisting of the system (right hip) and the garment including the electrodes (top shoulder)

NECK AND SHOULDER PROBLEMS, TRAPEZIUS MUSCLE ACTIVITY, AND HEARING DISORDERS IN ROCK/JAZZ MUSICIANS

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INTRODUCTION

Musicians are exposed to high sound levels and many suffer from hearing disorders such as hearing loss, tinnitus ("hearing" noise and/or tones), and hyperacusis (hypersensitivity to sound). Playing an instrument often includes fine-tuned movements, awkward postures, and, especially during performances, stressful conditions, all of them factors known to contribute to neck and shoulder problems. It has also been hypothesized that hearing disorders are linked to increased muscle tension in the neck and shoulder region and/or sensitivity to stress. This study was aimed at investigating such correlations.

METHODS

In this study, hearing status and living conditions were investigated in 139 rock-/jazz-musicians by tests, physical examinations, and an extensive questionnaire (1). Forty of these subjects (22 men and 18 women) were randomly selected into two groups, one consisting of subjects having one or more of the hearing disorders hearing loss, tinnitus, distortion, hyperacusis, and diplacusis (experienced pitch depends on sound level), while the other group was free of hearing disorders. The two groups took part in an experiment where trapezius muscle activity, as measured by bipolar surface EMG, and heart rate were recorded during three different stress provocations (the Stroop test, the Norinder mental arithmetic test, and counting backwards from 1022 in steps of 13) and during attempted rest (2). Each of the three stress provocations and the resting condition went on for five minutes with short breaks in between. During these breaks the next condition was described and the subject was instructed on what to do.

The surface electrodes were placed on the upper trapezius (pars descendens) with an inter-electrode distance of 20 mm and the center of the two electrodes 2 cm lateral of the midpoint between C7 and the acromion. The muscle activity was normalized to a reference contraction where the subject held his/her arms 90° abducted in the frontal plane according to Mathiassen et al. (3), and thus is expressed in percent of reference voluntary electrical activation (%RVE).

RESULTS

Three out of four (74%) of the musicians suffered from one or more of the hearing disorders hearing loss, tinnitus, distortion, hyperacusis and diplacusis (1). Self-reported musculoskeletal problems in the neck and shoulder were more common among those having (43%), compared to those not having (31%) hearing disorders (n=139). Regarding the specific disorder hyperacusis, i.e. hypersensitivity to sound, 55% of those having hyperacusis reported problems, whilst 31% of those not having hyperacusis had problems in the neck and shoulder. This difference in neck and shoulder problems was significant (Chi-square test; $p<0.05$).

Table 1 shows group median values of heart rate and myoelectric activity during stress provocations and attempted rest. Significantly higher heart rates were registered during all three stress provocations compared to the resting condition (two-sided, paired t-test; $p<0.001$), which proves that the chosen provocations were successful in creating stressful conditions. Also the muscle activity was significantly higher during the Stroop and Norinder tests, compared to rest, in all 40 subjects and in the group without hearing disorders, but not in the group with hearing disorders (one-sided, paired t-tests; $p<0.05$).

The group without hearing disorders had somewhat higher mean muscle activity during stress provocations compared to those having hearing disorders, however, the difference was not significant. Gender differences were not evaluated due to too small groups.

DISCUSSION

The higher prevalence of neck and shoulder problems among musicians with hearing disorders, especially the significant difference among those suffering from hyperacusis, indicates a possible link between hearing disorders and musculoskeletal problems in the neck and shoulder region. Considering the nature of hearing disorders like tinnitus and hyperacusis, it is easy to assume that they can contribute to increased muscle activity in the neck and shoulder in a similar way as mental stress. Contrary to what was expected, the group of musicians without hearing disorders had higher mean muscle activity during stress provocations. Further analysis will focus on other muscle activity characteristics than mean level and test these against one hearing disorder at a time.

CONCLUSION

The hypothesis that hearing problems are linked to increased muscle tension could not be verified in the performed test were the muscle activity of the trapezius muscle was studied during stress provocations. Self-reported neck and shoulder problems were more common among those having hearing problems, and was significantly higher in the specific hearing disorder hyperacusis, which indicate that the hypothesized connection may exist.

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	Stroop test	Norinder	Counting backwards	Attempted rest
Heart rate (all) [bpm]	90	79	81	72
Hearing disorders (n=20)	91	79	82	72
No hearing disorders (n=20)	86	79	80	73
Muscle activity (all) [%RVE]	3.6	3.2	3.1	2.8
Hearing disorders (n=20)	3.2	3.1	2.8	2.9
No hearing disorders (n=20)	4.1	3.5	3.3	2.2

Table 1: Group median values of heart rate (in beats per minute; bpm) and trapezius muscle activity (in %RVE) during three stress provocations and attempted rest. The muscle activity presented is the group median values where each individual contributes the mean activity of the left and right trapezius.

VARIABILITY OF "ERGONOMIC" EMG PARAMETERS (RMS VALUES AND RELATIVE REST TIME), RELATED TO NORMALISATION AND ELECTRODE DISPLACEMENT

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INTRODUCTION

Surface EMG measurements are often used in occupational health studies to investigate e.g. the effect of ergonomic measures or muscle activation characteristics in work related disorders. The two most often used parameters are the RMS value to characterise muscle loading and the relative rest time (RRT) to characterise relative muscle rest. The RMS value is calculated during relative small consecutive time intervals to enable determination of a course of the muscle loading. RRT is calculated as the relative amount of time in which the smoothed rectified EMG signal is below a threshold. To enable a better comparison between subjects and comparison of repeated measures within a single individual often a normalisation procedure for the RMS value is used. A considerable consensus has been developed to use a submaximal normalisation test for the Trapezius muscle (Mathiassen 1996). Often this test is not only used to normalise the RMS values, but also to determine the threshold value used to calculate RRT. As a good comparison of groups of subjects or comparison of repeated measures in individual subjects is crucial in many studies, the objective of this study was to investigate the sensitivity of the RMS and RRT values for an electrode displacement and to investigate the effects of normalisation on the RMS and RRT inter- and intra-subject variability.

METHODS

SEMG was recorded of the Trapezius muscle of 12 healthy subjects. An electrode array, consisting of 8 electrodes was mounted on the Trapezius to enable simulation of electrode displacement. Inter-electrode distance was 2 cm in each direction. Using bipolar recordings, an electrode displacement of 2 cm in 4 directions could be simulated. SEMG measurements were performed during four reference tests according Mathiassen (1995) followed by a standardized typing task for 10 minutes. sEMG was bandpass filtered (30-250 Hz), sampled at 1024 Hz during 15 seconds). RMS values were calculated as well as relative values. RRT values were calculated using an absolute threshold (6 uV) and a relative threshold (5% RMS_{ref}). Variability is expressed as the ratio between standard deviation and mean value in percentage.

RESULTS

Comparing the individual results, no systematic changes were found as an effect of an electrode displacement both for RMS and RRT values, in the sense of e.g. a lateral displacement results in systematic lower values. However the results presented in table 1 show that electrode displacements result in quite large variations in RMS values within individual subjects; mean intra-subject variability's are 43% and 42% for the reference test and typing task respectively.
These variability's are much larger compared to the variability's found between subjects when considering one standardised electrode position (Hermens et al, 1999) i.e. 13% and 27% for the reference test and typing task respectively.

Normalisation of the RMS values, using the reference contraction values, decreases the sensitivity of an electrode displacement in individual subjects till 33%, but has apparently no clear effect on the inter-subject variability.
In contrast to these findings with respect to the RMS values, the sensitivity to electrode displacement is very low for the RRT values; 0.7% and 0.8% for the RRT values using a fixed and a relative threshold, respectively. The inter-subject variability on the other hand is much larger. The use of the reference contraction value for the calculation of the RRT value, as is often done in literature, did not decrease the sensitivity of RRT for an electrode displacement neither the inter-subject variability.

DISCUSSION

The results suggest that normalisation of the RMS values is useful to decrease intra-subject variability of the RMS values. This means that for investigation individual subjects repeatedly in time normalisation of RMS values enables better comparison. Normalisation of RMS values does however not decrease the inter-subject variability.
So, in contrast to the almost generally accepted hypothesis, normalisation of RMS values does not result in better group comparison. Concerning RRT values, normalisation has no effect at all, meaning that fixed thresholds can be used in its calculation. This will facilitate the use of this parameter. The small sensitivity of RRT for electrode displacement and the large intersubject variability makes this parameter less useful for direct group comparison but very well suited in studies with a repeated measure design.

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			Absolute RMS	Normalised RMS	RRT Threshold 6 uV	RRT Threshold 5% RMS_{ref}
refer. test	Intra-subject-variability (95% CI)	Different electrode placements	43% (32 - 53%)			
refer. test	Inter-subject-variability	One electrode position	13%			
typing task	Intra-subject-variability (95% CI)	Different electrode placements	42% (34 – 50%)	33% (22 - 43%)	0.7% (0.3-1.2%)	0.8% (0.4-1.2%)
typing task	Inter-subject-variability	One electrode position	27%	27%	111%	118%

Table 1 : Intra and intersubject variability and associated 95% confidence intervals (95% CI) for the reference test and typing task using absolute and normalised RMS and RRT values.

THE EFFECTS OF POSITIONING PRECISION AND MENTAL PRESSURE ON MUSCLE ACTIVATION DURING TRACKING AND AIMING WITH A COMPUTER MOUSE

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INTRODUCTION

Work related upper extremity musculoskeletal disorders have become a major problem over the last decades, with high and apparently increasing incidence and prevalence rates (1). Although the pathophysiology of these disorders is still uncertain, several factors have been associated with these disorders.(1,2). The mechanism behind these associations remain to a large extent unclear. The present study aimed at identifying the effects of two of these factors, precision and mental pressure, on the level of activity of the forearm and neck-shoulder muscles.

METHODS

Ten healthy subjects performed two computer mouse tasks: a tracking task and an aiming task. Each task was performed at two levels of precision and two levels of mental pressure. In the tracking task subjects made the cursor follow a dot moving on the computer screen in a circle at a fixed speed. The level of precision was increased by decreasing the size of the dot, and mental pressure was increased by giving feedback to the subject about the number of mistakes they made. In the aiming task subjects were asked to click on a dot which appeared in random locations on the computer screen. The level of precision was increased by decreasing the size of the dot, and mental pressure was increased by challenging the subject to act as quickly as possible and recording the end time of the task. To prevent order effects precision and mental pressure were varied in a balanced design.

Subjects were seated on an adjustable wheeled chair, with height adjustable arm rests. Also the table was adjustable in height. A standard mouse and standard monitor were used. Prior to the experiments the work place was adjusted to the anthropometry of the individual subject according to common ergonomic guidelines.

EMG signals were recorded from three different muscles at the subject's dominant side: M. trapezius descendens), abbreviated as 'trapezius' in the results, M. extensor digitorum, and M. flexor digitorum superficialis, abbreviated as 'extensor' and 'flexor' respectively. Bipolar Ag/AgCl (Medicotest, Rugmarken, Denmark) surface electrodes were used with a recording distance of 15 mm. Signals were amplified 20 times (Porti-17™, TMS, Enschede, The Netherlands, input impedance > $10^{12}\Omega$, CMRR > 90 dB), band-pass filtered (10-300 Hz) and A-D converted (22bits) at 1000 Hz. EMG data were digitally rectified, filtered (4th order Butterworth lowpass 5 Hz) and normalized to maximal voluntary contractions (MVC). Data reduction was obtained by extracting the static level (10th percentile, P10), median level (P50), and peak level (P90) from the Amplitude Probability Distribution (3).

The effects of mental pressure and precision levels on muscle activation were statistically tested with analysis of variance (ANOVA) for repeated measures.

RESULTS

Similar results were obtained for the P10, P50 and P90 levels. Therefore, only the P50 results are presented in this paper. Precision had a main effect ($F=9.342$, $p=.005$) on muscle activation with higher activation under high precision demand (figure 1). Mental pressure had a main effect ($F=7.607$, $p=.01$) on muscle activation with higher activation during high mental pressure (figure 1). Two way interactions between task and mental pressure ($F=4.200$, $p=.046$) and between mental pressure and precision ($F=4.833$, $p=.03$) were found as well as a three way interaction between task, mentalpressure and precision ($F=9.935$, $p=.004$). The activation of the muscles was higher during the aiming tasks than during the tracking tasks ($F=8.446$, $p=.007$), this effect was most pronounced in the trapezius muscle.

DISCUSSION

The results showed that both increased precision and mental pressure led to an increase in muscle activity in the arm and shoulder muscles. This effect may explain the epidemiological association of precision demands and mental pressure with the prevalence of upper extremity disorders.

The level of muscle activity in the forearm extensor muscles during these intensive mouse tasks was surprisingly high. In addition, trapezius activity appeared high especially in the tracking task. This would suggest that tasks resembling the experimental tasks in terms of mental pressure, precision demand and physical characteristics would impose a considerable health risk.

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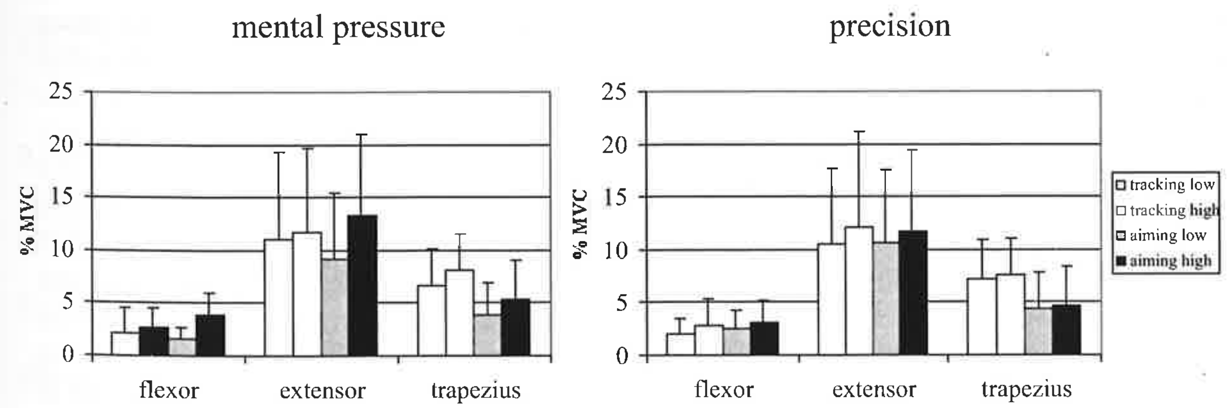

Figure 1. Effects of mental pressure and precision on muscle activity (P50) in the M. trapezius descendens, M. extensor digitorum, and M. flexor digitorum superficialis during tracking and aiming.

AN EVALUATION OF NURSING ASSISTANTS´JOB IN A NURSING HOME

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INTRODUCTION

Nursing personnel shows a relatively high prevalence of low-back pain (Stubbs et. al. 1983), and the workers compensation claims for back injury were ranked fifth among all occupations (Klein et. al., 1984). Within nursing personnel, nursing assistants (NAs) in nursing homes are at greatest risk. Lifting and transferring patients and carrying too much weight are believed to be the most important precipitating factors (Owen 1985, Scholey 1983, Stubbs et. al., Wright 1981). However, patient care and house hold tasks have been shown to be amongst the activities contributing the most to the poorer working postures. The objective of this study was to get a clearer understanding of task-specific work loads.

MATERIAL AND METHODS

The study took place in a 500 beds nursing home where all degrees of functional disability were represented. Fifteen subjects were randomly selected and observed through the morning and afternoon shifts. Physical work load was assessed by the Ovako Woking Posture Analyzing (OWAS) The recording of postures was made every 30 seconds using a small laptop computer, and the evaluation of the postural load was made with the four Action Categories as defined by Kahru et. al. (1977). The heart rate (HR) was used as a physiological measure of dynamic work load. It was monitored with Polar Electro Oy System and recorded every 5 seconds. The degree of physiological stress was classified in five categories according to the HR Astrand et. al. (1992): less than 90 (light work), 90 to 110 (moderate work), 110 to 130 (hard work), 130 to150 (very hard work) and 150 to 170 (extremely hard work).). Also, a job analysis was performed including the work environment and the use of lifting aids. And the statistic data were analyzed with *SPSS/PC+*.

RESULTS

7270 work postures were registered with OWAS. According to this method, less than 5% of the recorded postures might be associated with musculoskeletal damage (categories 3 or 4, table 1). Lifting accounted for 4.6% of the working time, however, only 15.6% of the lifting was done manually. In these occasions, 42.3% of the postures were associated with action categories 3. Pulling and assisting to toilet, which accounts for 5.3% of the working time, was also associated to high action categories, but to a lesser extent than in manually lifting (table 1). There were 77988 HR observations. The mean resting HR at work was 88 beats, and it was about 100 beats when physical work was performed, for an average increase in oxygen consumption of 5 cc/Kg/min per minute, or an increase in caloric expenditure of 82. Kcal hour for a mid weight person of 60 k.. During the work performance, the highest mean HR was 102 beats/min, and it was found associated with the following activities: patient care (washing, dressing etc), on bed postural changes of patients, changing diapers, and assisting to toilet. In 17% of the work time the HR was between 110 and 130 beats (hard work), and in about 1% was over 130 (very hard work). These results are in keeping with those found with the classification with OWAS classification method.

CONCLUSION

In this study we found that only in very few occasions NAs perform tasks with high physical demand or they adopt dangerous working postures. From these results, there is no reason to

think that NAs are at high risk of musculoskeletal, or low back, disorders. The question is if the methods have enough sensitivity for the objective. The agreement in the classification of tasks by both methods gives consistency to the findings, but both might have a poor and similar sensitivity. It might be argued that none of the methods are appropriate to study weight manipulation. But lifting was rare and OWAS takes into account weight, though in a broad way. Also, when performing brisk and short tasks, as most of the NAs tasks are, HR does not increase in proportion to the energy consumption because most of energy is produced anaerobically. It is interesting that HR is on the average high at rest while working, higher than at home. Although oxygen debt must be paid while resting, most of it should be done in the first 5 minutes, therefore this is not an explanation. In summary, it appears that NAs tasks are not as dangerous as it might be deduced by the high degree of complaints and sick leaves we observe in our practice. A possible explanation, aside of misclassification of the burden of tasks, is that as NAs job is low in content and high in emotional demand, then stress might be manifested in a higher susceptibility to back pain. Stress could also explain the moderate high HR at rest while at work. In a subsample of 7 NAs, there is a 10 beats difference between rest at home and rest at work, while there is no difference between rest at home and sleep. The study is being extended to evaluate this hypothesis.

Basic Activities	Observations Number	Observations %	Action Categories (%) 1	2	3	4
1. Assisting patient on bed (washing, changing positions, changing diaper, etc.)	1028	14,1	38,4	56,4	4,9	0,3
2. Manual lifting	52	0,7	28,8	28,8	42,3	
3. Mecanical lifting (hoist)	281	3,8	68,3	29,5	2,1	
4. Pushing and pulling	291	4	79,0	4,1	16,8	
5. Patient care (washing, dressing) in bathroom	502	6,9	57,6	33,6	7,9	0,8
6. Assisting with eating, drinking or taking medication	2846	39,1	73,3	24,7	1,8	0,2
7. Housekeeping	620	8,5	65,3	31,4	3,2	
8. Put the work's materials away	965	13,3	77,8	18,2	3,8	0,1
9. Assisting toilet	94	1,3	57,4	27,6	11,7	3,2
10. Other tasks, mostly administration	591	8,1	98	1,9	0,1	
Total	7270	100	68,7	27,1	3,9	0,2

Table 1: Distribution of basic activities and percentage of the OWAS action category.

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QUANTIFICATION OF MUSCLE POSTURAL ACTIVITY

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INTRODUCTION

Posturography provides physicians with valuable information on patients suffering from motor control disorders. Until now, the postural control system has been investigated mainly by analyzing the postural sway and the trajectories of the center of pressure (COP) and of the center of mass (COM) during standing posture tasks. Quantitative assessment of balance is typically conducted on the basis of parameters derived from the analysis of the COP trajectory [1]. However, the role played by muscle stiffness and muscle activation during standing is still controversial [2,3]. Thus, the analysis and the quantification of muscular activity can be of help in investigating motor control strategies. Surface myoelectric signal (SMES) is commonly utilized in posturoraphy to evaluate responses to externally applied disturbances [4], but SMESs detected during quiet standing have generally not been taken into account, mainly because of their very scarce amplitude that makes it difficult separating them from background electrical and biological noise. At this time, there are detection systems that allow for strongly amplifying myoelectric signals still yielding high signal to noise ratios and then studying muscle activation during quiet standing has become feasible. This contribution suggests a possible strategy to quantify muscle activation in this specific task.

METHODS

Experimental validation was carried out on a sample population consisting of 10 healthy subjects who signed an informed consent form. The acquisition system consisted of a three-component dynamometric platform that allowed for acquiring the vertical force and the moments with respect to the two orthogonal axes. This platform was connected to a system which allowed the simultaneous acquisition of platform and myoelectric signals (StepPC© by DEM, Torino, Italy). Signals were acquired with a sampling rate equal to 2 kHz and discretized by means of a 12-bit A/D converter. Subjects were asked to stand in the center of the force platform for 60 seconds, with open eyes and arms lying along their trunk. To evaluate the muscle activity responsible for the maintenance of balance, SMESa were recorded bilaterally from four muscles: tibialis anterior (TA), gastrocnemius (GSC), rectus femoris (RF) and biceps femoris (BF).

RESULTS

The modalities of muscle activation during quite standing have been classified as follows: a) no activation; b) weak and continuous activation; c) strong intermittent activation; d) activation of a few motor units only. Obviously, in case a) no muscle activity quantification is possible; in cases b) and c) quantification was attempted using the signal root mean square value and its number of turning points. By comparing the characteristics of these two parameters it is evident that the root mean square value is easier to compute, while the number of turning points is theoretically more robust during dynamic contractions. Our experimental results show, that in the specific task considered, the two parameters are strongly correlated (correlation coefficient ≥ 0.85). Finally, in case d) quantification was obtained by counting the active motor units and estimating their firing rates.

DISCUSSION

A model for the control of posture is the equilibrium point control, which states that: 1) on the ascending portion of the force/length relationship, a muscle behaves like a spring: the force it exerts once activated increases with its length; 2) the recruitment and the modulation of the firing rate of active motor units change the stiffness of each muscle; 3) stretch reflexes further enhance stiffness. Following this model, motion is described as a shift of the equilibrium point realized by agonist and antagonist springs. This is obtained by changing commands to alpha motoneurons, changing commands to gamma motoneurons, and changing the amplification of the stretch reflex. While fast movements are modeled as a sudden recovering of equilibrium, slow movements are interpreted as a gradual change in the equilibrium point towards which the system gradually moves. During these slow movements, such as during quiet standing, reflexes are not required to generate opposing torque since intrinsic stiffness is sufficient for maintaining stability [5]. It follows that the system does not need to significantly increase muscular activity. Moreover, it is demonstrated [6] that myoelectric activity of calf muscles is modulated by the displacement of the human body during superslow sinusoidal tilts of the force plate. The classification of muscular activity carried out in this work confirms the previous findings. In fact: weak and continuous activation (b) can be related to the slow movements of the center of mass of the body stretching calf muscles; strong intermittent activation (c) represents the fast movements due to quick recovering from unstable situations when the stiffness alone cannot allow stabilization; activation of a few motor units only (d) could interpreted as a way for obtaining stiffness modulation.

CONCLUSION

This contribution presents a methodology for the quantification of muscle activity during quite standing. Investigated muscles showed four different typologies of activation modality. We demonstrated that, depending on the activation modality, a quantification is possible by evaluating the SMES root mean square value or the number of active motor units and their firing rate. From a physiological point of view, our results confirm previous studies that related the different activation modalities to specific postural tasks, such as muscle stiffness modulation, fast corrections, and slow movements.

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EFFECTS OF SUDDEN VISUAL STIMULI ON POSTURAL REACTIONS TO PLATFORM PERTURBATIONS

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INTRODUCTION

Vision is an important factor in maintaining upright balance of the body. The aim of this study was to investigate the effects of short bursts of visual flow (flow-like movements of the visual environment) on the recovery of balance following fast, sudden movements of the supporting surface during quiet standing. The support-surface movements were generated using a movable platform. Previous studies on postural control using visual stimuli have mainly investigated effects of long-duration visual flow and/or used slow perturbations (1, 2, 3). It was hypothesized that reactions following perturbations of balance during quiet standing can be enhanced by sudden visual stimuli experienced during the perturbation.

METHODS

Ten healthy subjects (8 male, 2 female; age 29.3 ± 7.1 yrs; height 1.79 ± 0.06m; weight 74.4 ± 9.6kg) were exposed to 0.05 m perturbations (acceleration and deceleration: $5 m/s^2$) in either forward (fwd) or backward (bwd) direction, while simultaneously two computer screens (21") on both sides of their heads displayed stationary, forward or backward moving bar patterns for 1.0 s. Each of the 6 conditions was repeated 8 times in a randomized order during the course of the experiment. Force plate data and platform position, synchronized with kinematic data (Optotrak 3010, 18 markers) were collected for each trial. The antero-posterior (AP) center-of-mass (COM) movement was estimated from the kinematic data, and the velocity with which the COM returned toward its upright position after the perturbation was extracted (see Figure 1). AP Center-of-pressure (COP) excursions were calculated from the force plate data. The impulse generated by the COP on the COM during balance recovery was obtained by integrating the difference between the AP COP and COM excursions during the recovery phase (i.e. the large COP excursion pushing the COM towards upright position after the platform movement, see Figure 1). For each subject and for each platform perturbation direction, both the impulse and the recovery velocity values were normalized by subtracting the median of the stationary bar-pattern trials.

RESULTS

The direction of visual flow had a significant effect on the COM recovery velocity (parametric ANOVA: fwd perturbations $p=0.005$; bwd perturbations $p<0.001$), as well as on the recovery impulse (Kruskal-Wallis ANOVA: fwd perturbations $p=0.013$, bwd perturbations $p=0.020$). A post-hoc Scheffe test showed that visual flow bursts in the same direction as the support-surface movements strengthened the subjects' response by increasing the velocity with which the body returned toward its upright position (fwd perturbation $p=0.022$, bwd perturbation $p<.0001$). No such effect was found when the burst direction was opposite to the perturbation direction ($p>=0.35$). Similarly, an increase and no effect for the two respective flow directions were found for the recovery impulse as well.

DISCUSSION

Previous studies that have combined extended sequences of visual flow with postural perturbations have demonstrated that a reversal of visual flow direction resulted in a reversal

of the effect of the visual flow on the postural response (1, 2, 3). In the current study it was demonstrated that short bursts of visual flow also have an effect on postural responses. However, these effects only occurred when the visual flow direction was the same as the direction of the platform perturbation. No change in the response was found when the visual flow was in the opposite direction. It is important to note that when the visual flow direction equals the direction of the platform movement, the visual flow corresponds to the flow generated by the induced fall. For example, if the platform moves in the forward direction, the subject will fall backward and a forward visual flow is generated. When the visual flow direction is opposite to the platform movement direction, the visual flow direction is in conflict with the expected flow based on proprioceptive and vestibular input. It is hypothesized that for short bursts of visual flow the central nervous system integrates the visual flow and strengthens the postural response when the visual flow is as expected, but ignores the visual flow when it is in conflict. This hypothesis is reasonable, because the central nervous system is experienced in dealing with conflicting visual information associated with, for instance, large moving objects. When the visual flow is persistent, the initial conflict is overcome and the visually dominated responses emerge generating the previously described bi-directional responses.

CONCLUSION

This study showed that short bursts of visual flow have an effect on postural responses to platform perturbations. It was found that visual flow bursts in the same direction as the platform movement strengthened the postural responses by increasing both the AP-COM recovery velocity and the AP-COP impulse on the COM. It is hypothesized that when the direction of a short burst of visual flow is in agreement with the proprioceptive and vestibular information related to the perturbation, it strengthens the subject's response, whereas the burst is ignored when the direction of flow is in conflict.

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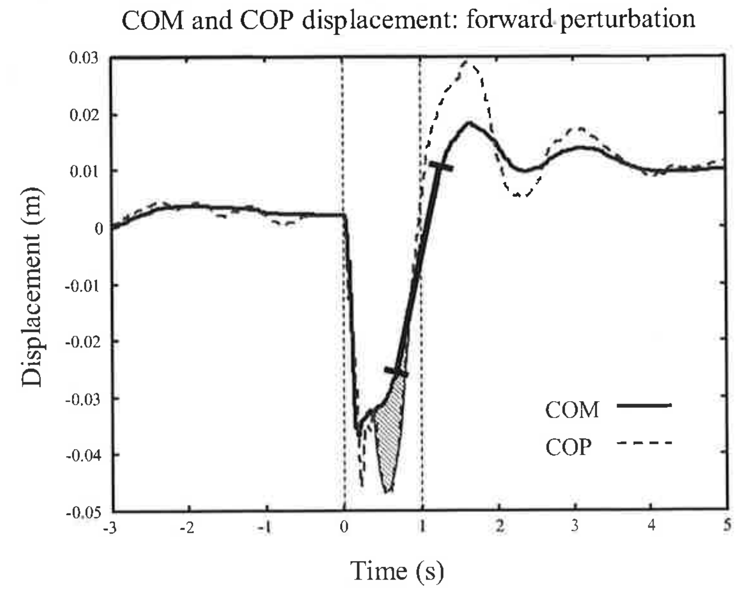

Figure 1. COM (solid) and COP (dashed) displacements following a forward platform perturbation (onset at t = 0) measured relative to the moving platform. A visual stimulus was applied for 1s. The shaded area was used as an estimation of the impulse applied by the COP on the COM. The slope of the marked line was used as the COM recovery velocity.

STANDING BALANCE CONTROL UNDER A CONCURRENT COGNITIVE TASK IN CHILDREN AND ADULTS

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INTRODUCTION

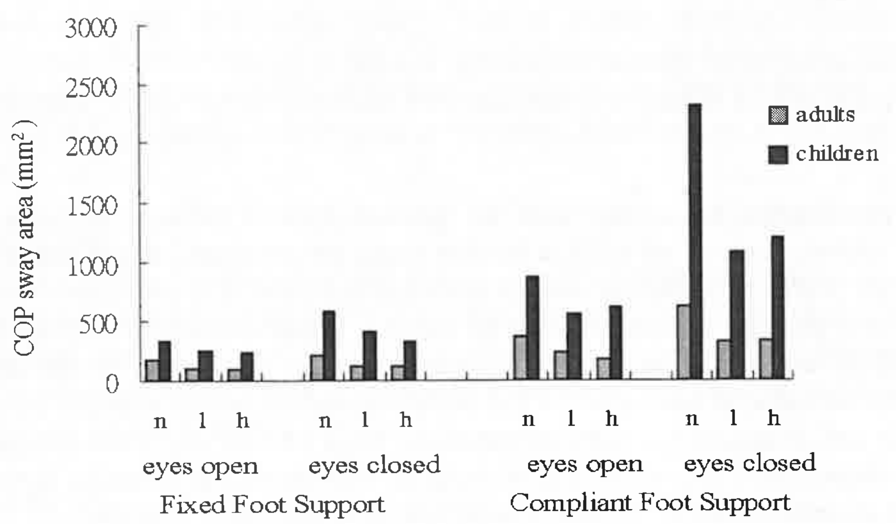
Many studies in aging have demonstrated the mechanisms for regulating balance in standing and walking interact with high level cognitive systems and may share similar attention resources (Teasdale 1993, Shumway-Cook et al. 1997). Research on children also shows that attention is related to learning and fine motor control. Whether attention affects the children's postural and balance control is less investigated. The purpose of this study was to investigate the effect of a concurrent attentional task on the standing balance performance in children and young adults.

METHODS

Ten children (age, 9.4 ± 3.5 range 5-13 years; height, 141.2 ± 20 range 116-158.5 cm; mass, 34.8 ± 11.8 range 20.5-48.5 kg) and ten young adults (age, 20.8 ± 0.7 range 20-22 years; height, 165.7 ± 5.1 range 157-171 cm; mass, 56.7 ± 7.3 range 50.5-74 kg) participated in this study. The effect of a concurrent attentional task (tone detection) on standing balance was investigated under four sensory conditions: (1) eyes open, fixed foot support, (2) eyes close, fixed foot support, (3) eyes open, compliant foot support, and (4) eyes close, compliant foot support. The fixed foot support was the metal force plate surface and the compliant foot support was a medium-density foam (40.5-cm x 40.5-cm x 7.5-cm section) placed on top of the force plate. The tone-detection task involved playing an auditory signal (a 250 Hz pure tone) to the right or left of the subjects every two seconds. The subjects were required to press a left or a right button as soon as they heard the tone. The task varied at two levels of difficulty: easy (stimulus and response were spatially compatible, e.g., left tone left button) and hard (where stimulus and response were spatially incompatible, e.g., left tone right button). A force platform (Kistler 9284) was used to collect the ground reaction force at a sampling rate of 500 Hz for 20 s. The sway area of center of pressure was calculated and used to represent the subject's standing balance.

RESULTS

Figure presents the subjects' standing balance under three attentional conditions (no concurrent task, low-demand task and high-demand task) and four sensory conditions. Results showed that children generally presented greater COP sway area than adults did ($F_{1, 18}=30.01$, $p < .0001$). The effect of attentional task was also significant. Subjects showed smaller COP sway area with concurrent attentional task than without concurrent attentional task ($F_{2, 36}=19.05$, $p < .0001$). The effect of the attentional task was not different in children than in adults (($F_{2, 36}=1.65$, n.s.).



DISCUSSION

The results showed that the concurrent attentional task did not negatively affect the balance performance. Rather it improved the standing stability of both children and young adults. The effect was not different in low demand and high demand attention tasks. Although these results are not the same as those reported in the literature with elderly subjects, they are not different from the results of Vuillerme et al (2000). Vuillerme et al. showed that postural stability was better while concurrently performing a reaction time task than while maintaining balance alone condition. The discrepancy between our results and those studies based on the elderly subjects may be explained as follows. Aging caused deterioration of the peripheral system. As a result, a task (such as standing) which normally is taken to be easy and automatic by children and young adults will require attention in the elderly. This renders the elderly especially vulnerable under a divided-attention situation. By contrast, children and young adults apparently cope with the situation with a different strategy. In our study, it appeared that the effect of the concurrent attentional task was an increase of the subjects' alertness and their body stiffness, thereby an increase in their standing stability.

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THE EFFECT OF A COGNITIVE TASK ON VOLUNTARY STEP EXECUTION IN HEALTHY ELDERLY INDIVIDUALS

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INTRODUCTION

Gait and balance impairments associated with the gradual loss of balance function with increasing age may elevate the risk of falls, a leading cause of accidental death and human suffering in the elderly population. Recent studies have demonstrated that corrective stepping is an important postural strategy to alter the base of support, preserve stability and prevent a fall (1). Once a fall is initiated a rapid step execution appears to be one of the critical parameters for successful balance recovery (2). Laboratory based studies of stepping behavior are commonly single task in nature, i.e. subjects can direct most of their cognitive attention to the motor task. The objective of this study was to examine step execution behavior during an attention-demanding cognitive task, a situation commonly encountered in real-life.

METHODS

Forward and backward voluntary steps (3 trials/direction) were performed by 10 healthy elderly individuals (68-90 years old) under two different task conditions; 1) rapid voluntary stepping following a tap cue on the heel and 2) same as 1) performed while conducting a modified-Stroop task. During the Stroop task subjects were instructed to read out loud the color of the ink of words that represented a different color than the ink (e.g. for "blue" subjects should say "black"). Center of pressure and ground reaction force data were collected with a force platform and analyzed for step reaction time (time from tap cue to first lateral deviation of center of pressure), time to foot lift and time to foot placement. Preparatory duration (time from first reaction to foot lift) and swing duration (time from foot lift to foot placement) were calculated. A t-test for paired samples was used to estimate the effect of the dual task condition on step execution parameters. A significance level of 0.05 was used.

RESULTS

Results are shown in Table 1. Performing a cognitive task during the voluntary step execution task had a dramatic effect on step reaction time. During forward stepping the reaction time nearly tripled (170ms to 453ms), whereas during backward stepping it approximately doubled (162ms to 299ms). The duration of the preparatory phase and the and swing phase was similar between the two conditions. The following observations were also noted under the dual task condition. Occasionally subjects did not react at all to the tap cue or they stepped in the wrong direction. While this occurred multiple times in some subjects under dual task condition it never occurred under the single task condition. Interestingly, subjects reported that they felt the tap cue on their heel but were confused about what to do. Furthermore, following the tap cue subjects commonly mixed up the colors in the Stroop task or stopped reading the colors when they executed the step.

DISCUSSION

The results demonstrate that a concurrent attention-demanding cognitive task significantly delays voluntary balance responses in healthy elderly individuals. In a real-life situation the amount of time available to execute a step to successfully prevent a fall is limited and any

delay in reaction may therefore cause a fall and ensuing injuries. The influence of an attention demanding task on postural control has been demonstrated in previous studies. During the modified Stroop task it appears that elderly subjects stiffen up by co-contracting postural muscles (3). Futhermore, the ability to recover balance following a perturbation requires more attentional resources in healthy elderly than in young subjects (4) suggesting that with increasing age the execution of a motor task requires increased attention. The ability to handle secondary tasks appears to be even further deteriorated in older adults with a history of recent falls (5). Although likely, it is unclear at this time whether this ability can be influenced by training.

CONCLUSION

If the ability to rapidly execute a step during cognitive attentional stress is a skill that can be improved by training it would imply that concurrent cognitive tasks should be incorporated as an important component into balance rehabilitation programs. In addition, it suggests that tests of postural function should incorporate dual tasks into their protocols as well as control for cognititve attention during testing.

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	Forward Step		Backward Step	
	Single Task	Dual Task	Single Task	Dual Task
Step Reaction Time	170 ± 53	453 ± 201*	162 ± 30	299 ± 179*
Foot Off	640 ± 141	954 ± 300*	632 ± 163	785 ± 192*
Foot Contact	960 ± 215	1294 ± 343*	926 ± 187	1093 ± 274*
Preparatory Phase	470 ± 95	501 ± 128	470 ± 1426	486 ± 91
Swing Phase	321 ± 88	340 ± 72	294 ± 71	308 ± 98

Table 1: Temporal effects of a cognitive task (modified Stroop task) on voluntary forward and backward step execution parameters in 10 healthy elderly individuals. All values are in ms. Asterisks indicate significant differences between task conditions for forward and backward stepping, respectively ($p<0.05$).

MATHEMATICAL ANALYSIS OF COUPLED LATERAL BENDING IN ACTIVE AXIAL ROTATION OF THE CERVICAL SPINE

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INTRODUCTION

Previous efforts to facilitate the organisation of the output stream of kinematic data, obtained with an electromagnetic tracking system, resulted in simultaneous graphical display of the main motion and coupled motion components in the cervical spine (1). These graphs provide insight in the time properties of coupled motion and often present individual characteristics, appearing in a repetitive fashion. A number of them reveal left-right differences. A certain phase shift of the coupled motion component with respect to the main motion component might hamper the interpretation of ipsilateral or contralateral coupling. This is also the case, when rather complex curves are registered. Aim of the present study was to compare a number of strategies, used in order to obtain a more objective analysis of these graphs.

METHODS

The following approaches were compared:

1. Registration of sign and range of coupled motion at the peak values of the main motion.
2. Integration of the area under the curve of coupled motion.
3. Calculation of the averaged coupled motion/main motion ratio.
4. Comparison of the sinusoidal functions of the main motion and coupled motion components.
5. Cross-correlation between the results of the main motion and coupled motion components.

Except for the first approach, calculations were made, using the software Matlab.

Compared criteria are the presence of: objective and consistent readings, an adequate measure of coupled motion, an adequate measure of the relationship between coupled motion and main motion, a measure of the entire phase of coupling and a noise resistant procedure. Other criteria deal with the possibility of automatic data processing and the evaluation of a phase shift of the coupled motion component, as well as with the applicability of the procedure on irregular curves.

RESULTS AND DISCUSSION

1. Registration of direction and range of coupled motion at the peak values of the main motion:
   The procedure offers a standardised measure of coupled motion, but provides only information about the degree of coupling at selected points of the curve. It may appear that maximally coupled lateral bending does not occur at the peak value of axial rotation. Thus, beside an incomplete report of the coupled lateral bending component, the results may reflect an underestimation of it.
2. Integration of the area under the curve of coupled motion:
   This approach offers an expression of coupled motion, which not only depends on the magnitude and direction of the coupled motion at the peak values of the main motion component. The entire period of coupled motion is taken in consideration. Further elaboration of the relationship with the main motion component is advisable.
3. Calculation of the averaged coupled motion/main motion ratio:
   The calculation of the averaged coupled motion/main motion ratio offers useful information about the degree of coupling, but does not provide absolute values. The data

   may become very noisy when the main motion component becomes very small or presents a zero value.
4. Comparison of the sinusoidal functions of the main motion and coupled motion components:
   Least square fitting to sinusoidal functions of the main motion and the coupled motion components allow for the study of a phase shift of the coupled motion component with respect of the main motion component. However, this methodology is not appropriated when irregular graphs are presented in case of cervical spine degeneration or pathology:
5. Cross-correlation between the results of the main motion and coupled motion components.
   In this approach, the values of corresponding points on the curves of main motion and coupled motion are submitted to a cross-correlation. Analysis of a phase shift of the coupled motion component with respect of the main motion component is possible. An important number of disadvantages, encountered with the previous approaches, can be avoided. Moreover, normalised data are available.

CONCLUSION

We concluded that a cross-correlation between the results of axial rotation and the coupled lateral bending components represents an advantageous strategy for analysing graphs of coupled motion in the cervical spine (Table 1).

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	Coupled motion component at the peak values of the main motion	Integration of the area under the curve of the coupled motion	Ratio between the coupled motion components and the main motion components.	Comparison of the sinusoidal functions of the main motion and coupled motion components	Cross-correlation between the results of the main motion and coupled motion components.
Objective and consistent readings	+	+	+	+	+
Adequate measure of coupled motion	+	+	-	-	+
Adequate measure of the relationship between coupled and main motion	-	-	+	+	+
Measure of the entire phase of coupling	-	+	+	+	+
Evaluation of a phase shift of the coupled motion component	-	-	-	+	+
Noise resistant procedure	+	-	-	+	+
Automatic data processing	-	+	+	+	+
Applicability on irregular curves	+	+	+	-	+

Table 1. Comparison of different strategies to analyse graphs of coupled motion in the cervical spine

3D INTRA ARTICULAR KINEMATICS OF THE HUMAN ELBOW JOINT: IN VITRO STUDY

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INTRODUCTION

The aim of this study was to collect quantitative information concerning intra articular kinematics of the elbow joint during flexion-extension and pronation-supination movements. For that purpose, an Euler convention and a finite helical axis approach can be used. For flexion-extension, the obtained kinematics data were specifically related to the configuration of the trochlea,whereas for pronation-supination a longitudinal line running through the radial head and the styloid process of the ulna was used.

METHODS

Five upper extremities were taken from fresh human cadavers at the level of the humeral head. The humerus was vertically fastened in such a way that the elbow was fully free to move. 3D electromagnetic tracking sensors were fixed on the humerus, radius and ulna. Subsequently, each arm was moved through a selection of directions: elbow flexion/extension with the forearm supinated, with the forearm pronated or with the forearm in the neutral position, and forearm pronation/supination with the elbow in extension, in 45° flexion, in 90° flexion or fully flexed. The positions and attitudes of each sensor were collected through the full range of motion. The individual sensor data were used to determine the parameters of the finite helical axes for discrete sampling ranges of flexion/extension and pronation/supination between the different bones: i.e. orientation (n), position (s), shift (t) along and rotation (θ) about the estimated helical axis. Estimations of these parameters were based on the algorithms described by Woltring [1]. Subsequently, the positions of local anatomical landmarks and joint surface configurations were digitized with a 3D drawing stylus. These anatomical data were used for the definition of local co-ordinate axes to refer to. To analyze the 3D intra articular flexion/extension kinematics of the humero-ulnar joint, the finite helical axes were related to a co-ordinate system based on anatomical landmarks including the center lines (C-lines) through the trochlea and capitulum. These C-lines were estimated by a least square analysis. The pronation/supination finite helical axes were related to the line running through the center of the radial head and the processus styloideus ulnae.

RESULTS AND DISCUSSION

Though the kinematics data on flexion-extension demonstrated the helical axes approximately to run through the C-line of the trochlea, the small changes in the position of the helical axes accentuate the humeroulnar joint not to act completely as a congruent joint. The finite helical axes during pronation-supination orient parallel to an axis through the radial head and the processus styloideus.

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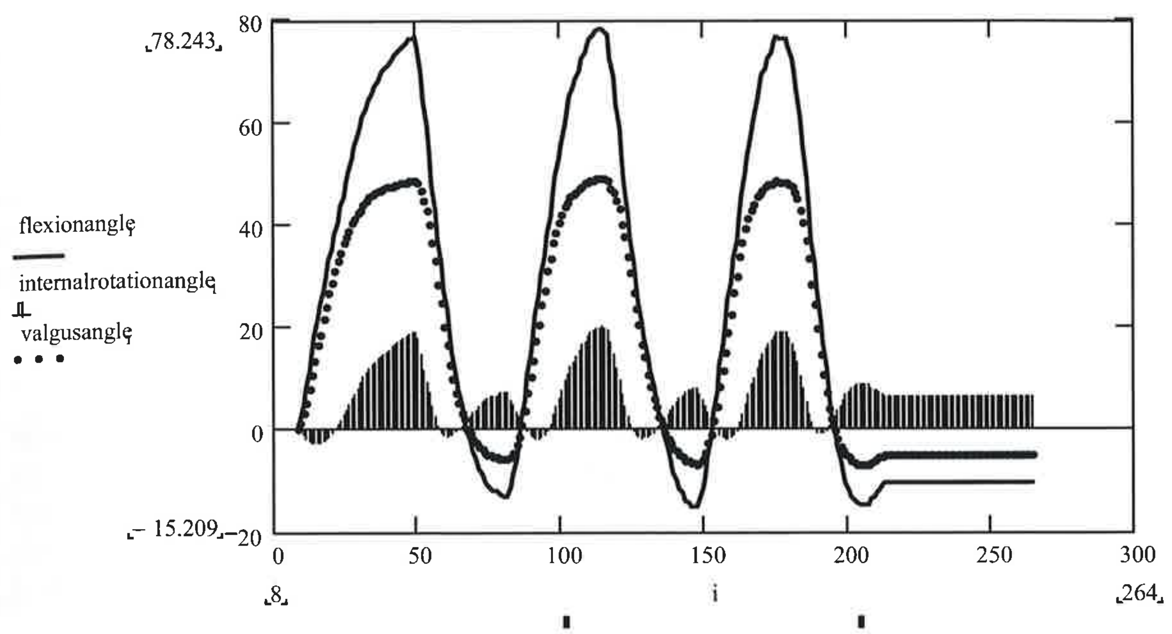

Figure 1: Euler angles during flexion in a pronated situation, referred to a trochlear embedded co-ordinate system

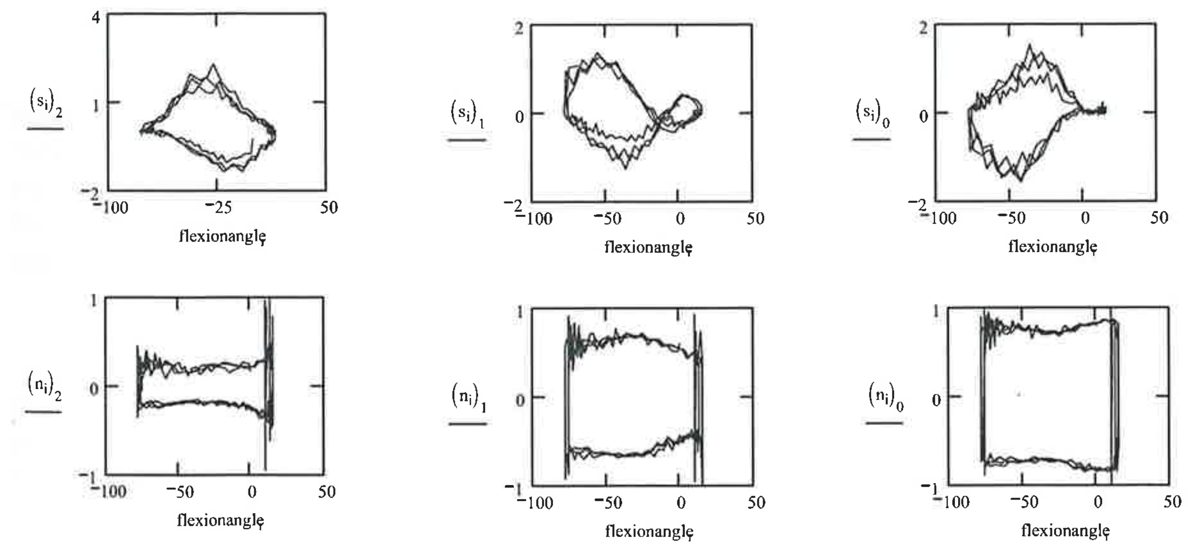

Figure 2: Decomposed values of the position (s) and the direction (n) parameter on the trochlear embedded co-ordinate system related to the flexion angle in the Euler convention

AMBULATORY MONITORING OF TRI-AXIAL NET SPINAL MOMENTS IN FREE LOAD HANDLING - FIRST RESULTS

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OBJECTIVE
To examine validity and reproducibility of a method that estimates the triaxial net spinal moment in manually handling unknown loads applying an artificial neural network, driven by surface EMG envelope and kinematic trunk data (ANN based method) [1]. The ANN is trained in supervised mode with target net moment data estimated applying a simple linked segment biomechanical model (ISB method) [3]. This model is driven solely by trunk kinematic data assessed through 3D inertial sensor modules [2].

DESIGN
For a group of 12 subjects net spinal moment curves estimated with the ANN based method were compared to the ones attained directly with the ISB method and secondarily to a simultaneously applied laboratory based reference method (Optotrack motion analysis, Kistler force plates, 3D full body biomechanical linked segment model) in an extensive set of single lift trials, pushing, pulling, and other load handling activities.

BACKGROUND
The ANN based method is the second key element of a newly proposed ambulatory method for mechanical back load exposure estimation. It estimates the "Load Handling Contribution" to the net sagittal spinal moment for a priori unknown load handling. The "Trunk Contribution" is estimated directly with the ISB method. Both methods together should facilitate a fully ambulatory method for load exposure assessment with an accuracy comparable to state of the art laboratory based methods.

METHODS
The basic and robust feed forward type ANN contains 2 layers and 2x7 trunk surface EMG signals plus full trunk kinematics as inputs. It is trained in supervised mode using directly estimated net spinal moment contributions, applying the ISB method, in a limited set of random calibration movements with controlled arm and load kinematics.

RESULTS
First analysis results for the lifting trials indicate a good estimation validity and robustness with typical average RMS differences between ISB method and reference data ranging from 10 to 40 Nm. First results of the ANN based method show that valid training within the same order of error magnitude is possible (Figure 1). Earlier findings from a similar study, assessing only sagittal net moments in lifting, seem to be confirmed, in that with decreasing relevance input signals EMG, angle, angular and linear acceleration and angular velocity all contributed significantly to the moment estimation validity.

DISCUSSION
First results indicate that the ANN based method seems able to perform well enough for application in the field, provided training session is performed with enough coverage of the input range, lasting about 10 minutes. Optimization is required for the training protocol
The ANN based and ISB method use only data from miniature inertial movement sensors and surface EMG data, both recorded with a portable device, and together seem to facilitate accurate (ambulatory) monitoring of dynamic net spinal moments and possibly spinal compressive forces in manual load handling during actual work under natural conditions.

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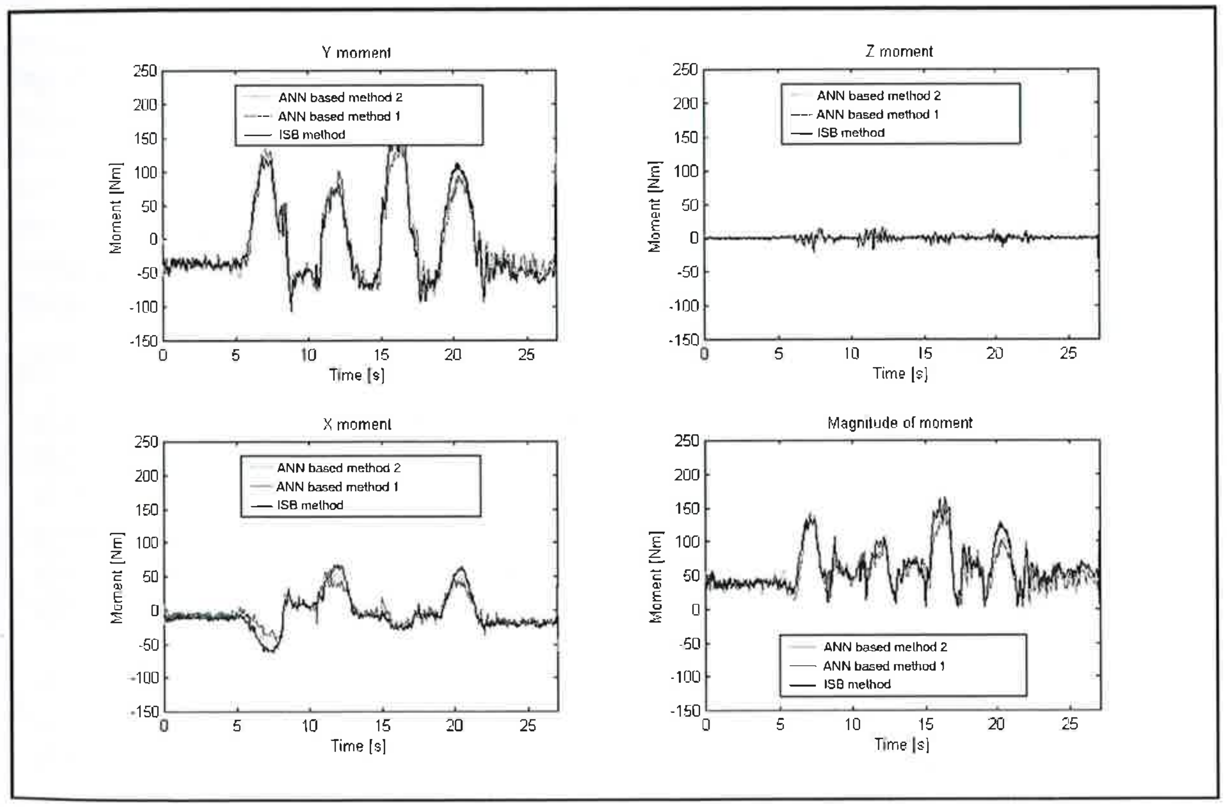

Figure 1: Example of 3D net moment estimation with the ANN-based method compared to the ISB method. The (asymmetric) task at hand was simulating moving boxes from one pile to another resulting in a movement. The ANN method is performed with prewhitening (ANN method 1) and without prewhitening (ANN method 2) Shown are the net moment components around Y, X, and Z axis of an inertial coordinate system (resp. 'flexion', 'lateroflexion' and 'rotation')and the net moment magnitude.

STRAIN RATE EFFECT ON THE TENSILE PROPERTIES OF FLEXOR TENDON

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INTRODUCTION

Tendon, ligament, bone and other biological tissues are known to be viscoelastic material. Their biomechanical properties change with different loading rates. The change in failure properties of soft tissues at different strain rates is of particular interest in analyzing mechanisms of injury. Bhatia et al.[1] estimated the extension rates at 60 to 70 mm/min close to tendon speed during gentle active mobilization while Crowninshield and Pope[2] reported that the traumatic loading of the knee joint in man might occur at strain rates ranging from 50 to 150000 %/s. However, the actual physiologic strain rates during soft tissue injury still remain unknown.

Many studies have been conducted to investigate the effect of strain rate on the structural properties of ligament-bone complex but contradicting results were obtained. Further, relatively few studies have evaluated the effect of strain rate on the mechanical properties of tendon. Flexor tendon works under dynamic loading, thus the dependence of tensile properties on strain rate should be determined.

MATERIALS AND METHODS

Seventy-five chicken Flexor Digitorum Profundus (FDP) tendons were used in this study. FDP tendons of the second digits were retrieved and wrapped by saline-soaked gauze before being sealed in airtight plastic bags. Tendons were stored at -40°C within 6 hours after death for 16 – 20 days before testing.

A 5565 Instron Universal Testing Machine was employed for the tensile tests. The main difficulty was fixating tendons onto the machine so that the tendons will not slip off nor fail prematurely when the tendons were loaded. Several methods to solve this problem were attempted and these included using roller grips, serrated jaw, sandpaper, epoxy glue potting and frozen ends. It was found that 1 kN pneumatic grips (Shimadzu Co.) lined with 2 pieces of gross fibre cardboard provided an adequate grip without slipping and damaging the tendons.

Tendons were tested at 15 different strain rates, namely 0.05, 0.1, 0.25, 0.5, 0.75, 1, 2.5, 5, 7.5, 10, 25, 50, 75, 100 and 150%/s, and samples were tested at each strain rate in quintuplicate. Mean values of Ultimate Tensile Strength, Strain at Maximum Stress and Elastic Modulus were compared at 99% confidence level using the Student's T method.

RESULTS

Results showed that strain rate has little effect on the shape of stress-strain curve. No significant changes existed in toe regions but the slope of linear region increased with increasing strain rate.

The lowest mean value of Ultimate Tensile Strength was observed at the lowest strain rate, 0.05 %/s. Generally, Ultimate Tensile Strength increased as strain rate increased. In particular, the mean values obtained at 0.25, 1, 2.5, 10, 50 and 100 %/s were significantly higher than the value obtained at 0.05 %/s. The values of Strain at Maximum Stress of all

groups did not differ significantly from that at 0.05 %/s except at 1 %/s strain rate. In addition, the increase of strain rate resulted in higher Elastic Modulii and the differences were statistically significant against 0.05 %/s except for lower strain rates of 0.1, 0.25, 0.5 and 1 %/s. However the linear regression between Elastic Modulus and strain rate ($r = 0.642$) was not significant for the range of strain rates.

DISCUSSION AND CONCLUSIONS

Strain rate was found to have little effect on the shape of stress-strain curve. The toe regions were not affected by strain rates but stiffness increased with increasing strain rate. The hypothesis advocated by Piolett et al.[3] that the strain rate effect mainly acted in the toe region with minimal influence on the linear part of the stress-strain curves was not justified in this study.

Based on the results obtained, we conclude that strain rate affects the tensile properties of tendon significantly. Within the range 0.05 to 150 %/s, the stiffness of tendons increased with the increase of strain rate, with no change in Ultimate Tensile Strength and Strain at Maximum Stress. These findings were consistent with the work of Danto and Woo[4]. However, the change in tendon tensile properties was not significant with small change of strain rate.

The reason for the strain rate sensitivity of tendon is unknown. A rate sensitive interfibrillar matrix of ground substance is suggested to help maintain the structural integrity of tendon at high strain rate. It is anticipated that lower strain rates permit shearing or gliding between collagen fibres. The shearing eliminates stress sharing among the collagen fibres and this results in uneven stress distribution. As such, only part of the collagen fibres contributes to resisting the applied load. On the other hand at higher strain rates, the rate sensitive interfibrillar matrix of ground substance helps distribute stress among the collagen fibres by limiting interfibrillar shearing. With stress sharing, more collagen fibres are involved in resisting the applied load, thus resulting in stiffer tendon.

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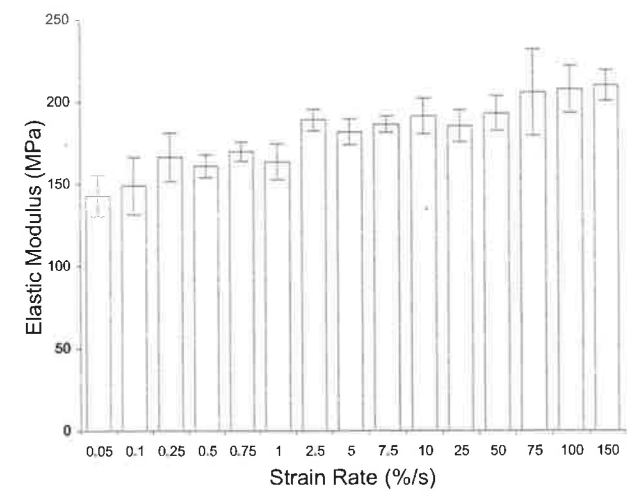

Figure 1: Correlation of tendon stiffness with strain rate.

COMPARISON OF THREE DIFFERENT CAMERA CALIBRATION TYPES

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INTRODUCTION

Video-based systems for three-dimensional reconstruction during the course of biomechanical analysis are today very popular and common. One of the pre-conditions when using such a systems is so-called camera calibration. The objective of calibration is to determine a set of camera model parameters that describe the mapping between 3D reference coordinates (calibration points) and 2D image coordinates. Traditional approach is by the means of calibration cage. Although very effective and still present in many today's operating systems, calibration cages tend to be replaced more and more through calibration plane or stick (wand). Using calibration plane or wand make such a 3D reconstruction system a lot more user-friendly [1]. On the other hand, calibration/reconstruction algorithms become more complex and the reconstruction accuracy using only plane/wand is likely to be impaired. The aim of this paper was to investigate these three different types of calibration.

METHODS

All three different types of calibration were performed for cameras put on four various positions. Looking from the birds view, angles between camera optical axes and symmetry of calibration volume were approximately –45°, -30°, +30° and +45°. The calibration volume involved was 90cm(length) × 180cm(height) × 150cm(depth). The accuracy reconstruction analysis was preformed in the following manner. The calibration wand of length 54cm was placed randomly on more than 1000 positions throughout the volume. For each type of calibration lengths of the wand were reconstructed and root mean square values (RMS) were calculated of differences between its true value (54cm) and reconstructed ones.

RESULTS

Table 1. Root mean square values [cm] between true wand length and reconstructed one.

Camera pair	Wand calibration	Plane calibration	Cage calibration
(-45°, +45°)	0.4153	0.3956	0.1552
(-30°, +30°)	0.4097	0.5188	0.2269
(-30°, +45°)	0.3931	0.6041	0.1500
(-45°, +30°)	0.4220	0.3010	0.2148

For the reconstruction analysis only the camera pairs were considered. Specifically, in Table 1 are given results of four camera pairs which optical axes angles were app. 60°, 75° and 90°. All values are in centimeters. In order to obtain more realistic view on different calibration types performances no calibration points (wand positions) were also used later for reconstruction. Each experiment trial was repeated several times and given results represent typical outcome.

DISCUSSION

Intuitively the best results were expected when using traditional calibration cage. Such an expectation is based on relatively simpler mathematical calculation during the calibration and

reconstruction itself. Namely, projection of points between 3D space and image plane is mathematically described (modeled) by some kind of function - camera model. Today's camera model are almost all based on so-called pinhole model which is usually up-graded with additional parameters for non-linear (lens) distortions. In our case were used two parameters for tangential distortion and two parameters for radial distortion, which is generally considered optimal [2]. Even though solving the non-linear equations of camera model during the calibration phase, when using the calibration cage, may seem complex enough, choosing the plane or wand calibration, as a rule, introduces even more complexity.
For example, in the case of plane calibration usually it is necessary to construct so-called virtual 3D object after which is then traditional (cage) calibration possible [3]. But calibration points obtained from such virtual 3D object are usually inferior to those acquired from the true calibration and consequently calibration results are impaired. Besides, at least approximate values of camera internal parameters are required which very likely introduces another source of error. Still, there are possibilities to avoid construction of so-called 3D virtual object and calculate camera model parameters only from plane position in space [4]. But, then again such an approach suffers from the lack of 'depth information' once we move from vicinity were calibration plane was originally.
Calibrating with wand leads to calculation of fundamental matrix. Fundamental matrix in itself has embedded information of camera model parameters [5]. Determination of fundamental matrix is challenging at least as traditional camera calibration itself. And that is only half way, since one needs yet to extract information from fundamental matrix. Here, we have little or none information whatsoever about the scene and practically entire calculation of camera model parameters is based on exploitation of fundamental matrix properties. Those properties may or may not be closely related to conic (quadric) which are geometrical entities defined from the epipolar geometry [5].

CONCLUSION

For the highest accuracy demands calibrating camera with cage has still its justification regardless of possible discomfort for the end-user. Plane calibration has slightly better results than wand calibration, but considering easiness of wand calibration that could be discarded in most cases. Different algorithms and methods for wand calibration are nowadays constantly developing and in the near future it is very likely that they will match accuracy of traditional calibration cage.

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SENSORY MOTOR CONTROL OF THE LOWER BACK

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Low back pain is a common disabling musculoskeletal disorder, whose treatment and prevention are problematic. Although most LBP syndromes lack a specific diagnoses, there is increasing evidence that muscle function plays an important role in the management of such patients. Back extensor strength and endurance are often severely impaired in such patients and pain or inhibition can further alter muscle activation. These impairments may compromise the structural integrity of the spine and leave it susceptible to further injury, prolonged recovery, or even chronicity of pain. When such impairments are revised by exercise, improvements in physical capacity, functional outcomes and pain can result. However, so far there is a lack of knowledge about the physiological and pathophysiological aspects of the spine and trunk sensory motor control system and how they may relate to the evolution to LBP. In this overview the current state of research of the sensory motor control mechanisms utilized for trunk stabilization and postural control of the lower back will be presented.

Biomechanical investigations of the lumbar spine often emphasize the vertebral column's requirement for stability to minimize conditions that could give rise to low back dysfunction. Mechanical stability of the lumbar spine requires muscles activity because the lumbar spine can buckle under physioplogic loads. Trunk muscles have to develop a sufficient force to provide adequate stability and prevent tissue failure of the spinal complex. Muscle activation must be optimally timed and of sufficient magnitude in order to protect the vulnerable osteoligamentous structures from injury during a lift.

ANATOMY OF SPINAL STABILIZATION

Contraction of the paraspinal muscles, the built-up of abdominal pressure, and the simultaneous contraction of antagonistic muscle pairs have been identified as the major mechanisms that actively contribute to spinal stabilization [for review see ref. 1]. Spinal stability is provided in a mutual interaction among all muscles of the trunk. For didactic purposes four major functional groups will be described:

1) The local paravertebral muscles directly stabilize the segments of the spine (internal stabilization). Their stabilizing role is by serving to stabilize adjoining vertebrae controlling their movement during motion of the vertebral column and providing for a more effective action of the long muscles.
2) The role of the global, polysegmental, paravertebral muscles has been suggested to provide general trunk stabilization and to balance external loads and to help minimize the resulting forces on the spine.
3) Muscles that contribute to the pressure facilitation within the abdominal cavity provide global stabilization of the spine. The contraction of the abdominal muscles , in particular the transversus abdominis together with the diaphragma and diaphragma pelvis correlate closely with increased abdominal pressure in a variety of postural tasks. To minimize displacement of the abdominal contents within the abdomen and the pelvis, it is necessary to elevate intra-abdominal pressure by simultaneously contracting the diaphragm, the pelvic floor and the abdominal muscles.
4) Muscles that facilitate the pressure within the fascia tube system of the thoracolumbar fascia may contribute to improved spinal stability in addition to the paravertebral and abdominal muscles. Thereby contraction of the abdominal muscles, in effect the obliqus internus and transversus abdominis muscles as well as the musculus latissimus dorsi, increase the tension of the thoracolumbar fascia tube, which may result in enhanced stiffness of the lumbar spine.

PHYSIOLOGY AND SENSORY MOTOR CONTROL OF THE SPINE

The stabilizing function of the trunk muscles is especially important around the neutral posture, where the spine exhibits the least stiffness. Panjabi considered the neutral zone of a segment as a sensitive region for the stabilization of joints [12, 13]. The neutral zone is defined as a small range of displacement around the mid position of the segment/ joint, where little resistance is offered by passive spinal restraints [12, 13]. Stability of the motion segment is provided by the stabilizing system, which adjusts in a way, that the neutral zone remains within certain physiological thresholds to avoid clinical instability [12, 13] that may result in low back pain.

The presence of a ligamento-muscular reflex has been proposed for the automatic control of the motion segment. Changes in the proprioceptive, kinesthetic, and nociceptive information content from mechanoreceptors in the spinal ligaments, discs and joint capsules may project to motor neurons that activate muscles which can stabilize the joint. In fact, recent animal models have found electromyographic activity in both the short and long paraspinal muscles following electrical stimulation of mechanoreceptors in the disc, facet joints and supraspinal ligaments [8,9,15]. Furthermore, the presence of a spinal ligamento-muscular reflex has been demonstrated to exist also in humans [14].

Further mechanisms have been suggested to explain how the muscular system can stabilize the vertebral segments.

MUSCLE STIFFNESS

The gamma spindle system is known to facilitate the alpha motorneurons that control the tonic slow twitch muscle fibers [10]. The degree of muscle stiffness may be regulated autonomically via the gamma system. Feedback information from the joints and ligaments, via their effects on the gamma spindle system, have been suggested to regulate the stiffness of these muscles. Probably, segmental (internal) stabilization would be most effectively achieved by the deep mono- and oligoarticular intervertebral muscles.

CO-CONTRACTION

Another mechanism that may provide stiffness to spinal segments to maintain stability in the presence of external and internal loads, is co-contraction of antagonist muscles that lie on each side of the intervertebral joints [11]. A specific cortical system (motor program) for control of joint stiffness has been proposed in man [4]. Theoretically, antagonistic trunk muscle coactivation is necessary to maintain the lumbar spine in a mechanically stable equilibrium. Thereby the muscles act as guy wires that stiffen the intervertebral joints that they span. The control of spine equilibrium and mechanical stability requires appropriate muscle recruitment and timing.

PREPROGRAMMED MUSCLE CONTRACTIONS

When reactive forces due to limb movement challenge the stability of the trunk, some muscles contract prior to the agonist limb muscle in order to minimize (compensate) perturbing effects of posture. Such precontraction during limb movement has been shown for the transversus abdominis muscle and the diaphragm [2,5,6]. Precontraction of these muscles is considered to prepare the body for the impending disturbance of the position of the center of gravity relative to the base of support and the reactive forces imposed on the adjacent spinal segments. It has been suggested by Hodges, that these preparatory postural responses in advance of limb movements may be preprogrammed by the central nervous system and initiated as part of a motor command for movement [5]. Alternatively, the activation of the TrA could simply represent a non-directional specific component of the central pattern generator (CPG) within the spinal cord.

ECCENTRIC ACTIVATION OF PARASPINAL MUSCLES

The findings of a preactivation of the abdominal muscles and the diaphragm thus increasing the pressure in the abdomen in postural tasks support a hypothesis that the activation of the paravertebral back muscles in postural control tasks is an eccentric activation mode. Thereby both preactivation of the abdominal muscles and the pressure increase within the abdomen may lead to a kyphosing effect in the lumbar spine. This kyphosing effect may release stretch impulses within paravertebral back muscles thus activating reflex muscle contractions of the paravertebral muscles of the lower back. The stretch reflex within the back muscles was first described in detail by Dimitrijevics et al. [3]. Thus paravertebral stretch reflexes may play a major role in postural adaptations of the trunk in order to compensate for internal or external perturbation.

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EVALUATION OF LOW BACK MUSCLE FATIGUE USING ELECTROMYOGRAM, MECHANOMYOGRAM AND NEAR-INFRARED SPECTROSCOPY ANALYSES

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INTRODUCTION

Previous studies have indicated that mechanomyogram (MMG) amplitude and frequency components might represent the underlying motor unit (MU) recruitment and firing rate (rate coding) (Barry and Cole, 1990; Orizio et al., 1993, 1996; Yoshitake and Moritani, 1999; Bichler, 2000). We have recently investigated the etiology of lower-back muscle fatigue by means of simultaneous analyses of electromyogram (EMG), mechanomyogram (MMG), and near-infrared spectroscopy (NIRS) in an attempt to shed some light on the electrophysiological, mechanical, and metabolic characteristics of low back muscle fatigue (Yoshitake et al., 2001).

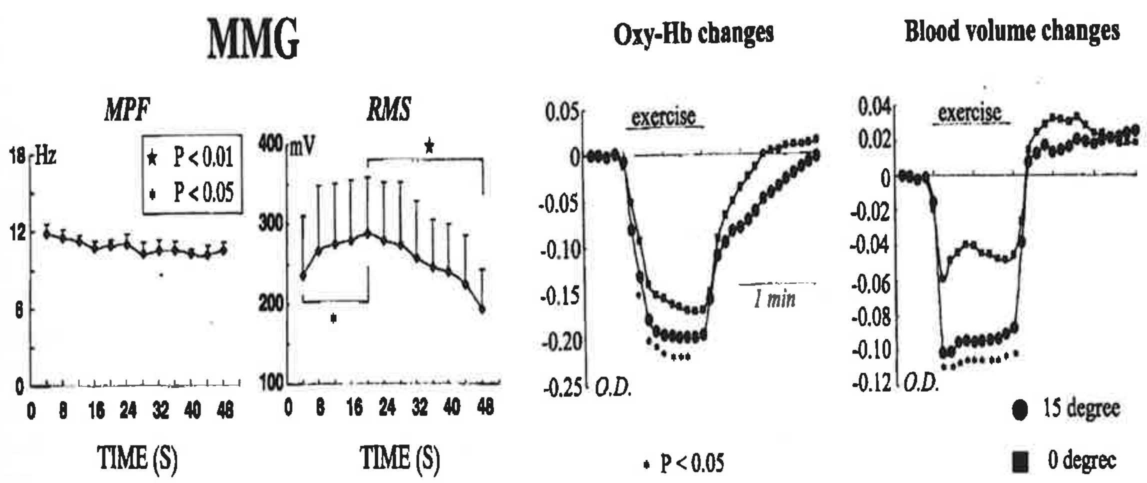
METHODS

Eight male college students volunteered for this study. The subjects were asked to lie in a prone position and were strapped to a rigid table so as to support the legs and lower trunk up to their anterior superior iliac spine. The subjects were instructed to cross their arms in front of the chest and to extend the upper body until the thoracic spine was elevated. The subjects performed isometric back extension at an angle of 15 degrees with reference to the horizontal plane for a period of 60 s. Surface EMG, MMG and NIRS signals were recorded simultaneously from the center of the erector spinae at the level of L3. A microphone sensor (10 mm diameter, mass 5g, bandwidth 3-2000 Hz) for MMG recording was fixed to the center of the belly of the erector spine at an equal distance from the two EMG electrodes, which were positioned to line up the microphone sensor on the longitudinal axis. NIRS was employed in order to determine the level of muscle blood volume (BV) and oxygenation (Oxy-Hb) during the sustained contraction.

RESULTS

The root mean square amplitude value (RMS) of the EMG signal was significantly increased at the initial phase of contraction and then fell significantly, while mean power frequency (MPF) of the EMG signal decreased significantly and progressively as a function of time. There were also significant initial increases in MMG-RMS that were followed by progressive decreases at

the end of fatiguing contractions. MMG-MPF remained unchanged. Muscle BV and Oxy-Hb decreased dramatically at the onset of the contraction and then remained almost constant throughout the rest of the contraction.



DISCUSSION

These results obtained by simultaneous recordings of EMG, MMG, and NIRS tools demonstrate that restriction of blood flow due to the high intramuscular mechanical pressure is one of the most important factors to evoke the muscle fatigue particularly in low back muscle. In addition, our simultaneous recording system described here can obtain more reliable information regarding the mechanism(s) of low back muscle fatigue.

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SUPPLEMENTING PARASPINAL WORK CAPACITY EVALUATIONS WITH INSTRUMENTED MEASURES OF MUSCLE FUNCTION

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INTRODUCTION

Back injuries in the workplace are most prevalent in jobs that require repetitive or forceful lifting. Most standardized work capacity evaluations are based on subjective measurements of lifting capacity or other manual work tasks. Although helpful from a qualitative perspective, they fail to identify the underlying mechanisms causing performance deficits. This paper describes a method of supplementing qualitative measures of physical work capacity with quantitative EMG and biomechanical measures. The technique is based on the hypothesis that when a patient with low back pain (LBP) is performing a task that involves repetitive flexion and extension of the trunk, or maintaining relatively long periods of a static position, they may modify their posture, weight distribution, or muscle activity to compensate for muscle fatigue and increased pain. These compensations may result in a modification of joint loading, or of paraspinal muscle activity, which may predispose the person to recurrent back injuries.

METHODS

Two on-going studies are described: one laboratory based (n=8 healthy control subjects with no history of LBP; mean age 35.6±9.1) and the other clinically-based (n=8 chronic LBP patients in a multidisciplinary work rehabilitation program; mean age 45.6±6.1). For both studies, surface EMG signals were detected from 8 contralateral paraspinal muscle sites (UT, T10, L2, L5) during 1) cyclical lifting of a weighted-box in the sagittal plane (12 lifts/min, 4 minutes total time) from mid-shank to waist height, and 2) standing in one position while assembling nuts and bolts at a table (4 minutes total time). The weight of the box was set to 15% of ideal body weight for control subjects, and 50% of maximal safe lifting capacity for LBP subjects. For control subjects in the laboratory setting, body kinematics and box position during repetitive lifting were measured using a stereophotogrammetric infrared system. Mechanical switches located on the box were used in the clinical setting to monitor phases of the lifting cycle.

EMG Analysis: For assessment of muscle fatigue during the box-lifting task, the EMG data were first divided into segments for the lifting phase of the task. The Cohen-Posch representation of the surface EMG signal was computed to derive the Instantaneous Median Frequency (IMDF) of the signal (Bonato et al. 2001). The time-course of the IMDF was plotted using the phase of the lift cycle that corresponded with the lowest standard deviation of the IMDF. Percentage change in IMDF during the lifting task was computed as a fatigue index. Changes in EMG activation and muscle imbalance during the static work task (i.e. assembling nuts and bolts) were computed from the Root Mean Square (RMS) value of the signal.

Biomechanical Analysis: Biomechanical changes in the lifting task were assessed among control subjects on the basis of the box acceleration in the vertical and horizontal directions and changes in time of the peak acceleration (P1Y and P2X) during the four minutes of the task. Changes in knee, hip, trunk and elbow joint angular displacement were calculated at the occurrence of the P1Y and P2X peaks in box acceleration, to monitor possible fatigue-related changes in the lifting strategy. The effect of these strategies on the loads exerted on the spine was estimated by implementing an inverse dynamics method to compute moment, compressive force, and shear force at L4 (Frigo 1990)

RESULTS

The overall change in IMDF resulting from the lifting tasks was greater in LBP patients than controls, with statistically significant differences at the L5 spinal level. Patients with higher impairment scores showed the greatest differences. Secondly, IMDF recovered periodically during the lifting task. This pattern of behavior was not observed for control subjects. For the static work task, significant muscle imbalances were observed in the LBP subjects, as well as significant increases in RMS activity, which were not present in the control subjecs.

The kinematics of lifting were modified during the course of the lifting task, such that subjects adopted different strategies as they fatigued. Some subjects increased the amount of flexion of the elbow joint, which resulted in a decreased distance between the box and the subject's body. Another strategy adopted was a progressive transition from a stoop to a squat lift. When subjects adopted these strategies, the estimated loads on the spine did not significantly change during the task, whereas when subjects struggled to maintain the same kinematics during the lifting task, the loads increased substantially in their peak value. The increase was likely related to the observed increases in box acceleration.

DISCUSSION

Recent developments in time-frequency analysis procedures to compute the IMDF of the EMG signal were utilized in this study to overcome previous limitations of the technique to isometric activities. The usefulness of this analysis procedure for assessing LBP is demonstrated by the implementation of these techniques in a standardized work-capacity evaluation. Muscle use, and subsequent patterns of fatigue, differed in control subjects and patients with LBP. The recognition of these differences may be of importance to clinicians in designing exercise therapies to improve paraspinal muscle endurance and overcome pain-related "fear" and "avoidance" behavior in patients with LBP. Among researchers, combining these tools with body kinematic measurements can help in understanding back pain injury mechanisms for specific work tasks.

CONCLUSION

Findings from these studies indicated that 1) the time course and behavior of surface EMG parameters (IMDF, RMS) during functional activities in LBP subjects was different than in control subjects, and 2) biomechanical parameters of lifting were adversely modified during the standardized lifting tasks. These modifications appear to be associated with the development of paraspinal muscle fatigue as measured by IMDF parameters.

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MYOELECTRIC MANIFESTATIONS OF MUSCLE FATIGUE IN LOW LEVEL, LONG DURATION CONTRACTIONS

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INTRODUCTION

The recruitment of new MUs during fatiguing constant force isometric contractions has been demonstrated in a number of studies. Examples of replacement of MUs during low level prolonged contractions of the trapezius muscle have been reported (1,2). The objective of this study was to verify this behavior in the biceps brachii.

METHODS

Five male healthy subjects (ages between 22 and 28 years) participated in the study. Each subject sat on a chair with his elbow joint at 120 degrees in an isometric brace and sustained a contraction at 5%, 10% and 15% MVC for 10 minutes (one test per day). The target level was maintained by means of visual force feedback. Skin temperature was kept at 31.5 ± 0.5 C°. Surface EMG signals were detected using linear arrays of 16 electrodes (silver bars, interelectrode distance 10 mm, 5 mm long, 1 mm diameter) which covered the entire muscle length (3). The EMG signals were amplified (gain of 5000 or 10000, 3dB bandwidth 10Hz-500Hz, 40 dB/decade), sampled at a rate of 2048 Hz and stored on a PC after 12 bit A/D conversion. The signals were analysed with a recently developed technique (4, 5), based on the automatic detection of single MUAPs and classification based on a neural network approach. Because of the computational time required by the classification procedure, only three seconds every 30 were decomposed. Spectral (mean and median frequency) and temporal (averaged rectified and root mean square value) variables as well as global conduction velocity were computed every second.

RESULTS

The Figure reports an example of signals detected during a 15% MVC contraction. The progressive recruitment of new MUs during the contraction was evident for all subjects even just by visual inspection of the multi-channel surface EMG signals. The automatic MUAP identification technique adopted permitted to assign individual MUAPs to a MU and consequently observe the recruitment of new MUs during the contraction time, as indicated in the Figure. New MUAPs appeared in the signal when the first recruited MUs started to show manifestations of muscle fatigue (evaluated by the decrease in characteristic frequencies and mean conduction velocity). This phenomenon was more pronounced with higher contraction levels.

In two out of five subjects EMG global variables (spectral variables and conduction velocity) showed a clear increase after the initial decrease due to fatigue. This behaviour was evident in two subjects, questionable in one and absent in the other two, even if consistent recruitment of new large MUs was evident in all subjects. The Figure reports an example of motor unit identification for a contraction at 15% MVC. MUAPs belonging to the same MU were very similar along the signal. The masking effect of the larger MUAPs did not allow to verify if the new MUs were recruited in addition or replacement of the previously active ones.

DISCUSSION AND CONCLUSIONS

Multiple channel EMG detection allowed the identification and classification of single MUAPs from the interference signal and the extraction of information about the recruitment of new MUs. Spectral and amplitude variables and mean conduction velocity reflect the physiological phenomena occurring during fatigue, including MU pool changes. The increase in mean conduction velocity, following the decrease due to fatigue, observed in some subjects and not in others needs further investigation.. It might reflect individual differences related to different fiber type distributions and/or force control strategies. Modifications of the active motor unit pool during the contraction are evident in all subjects.

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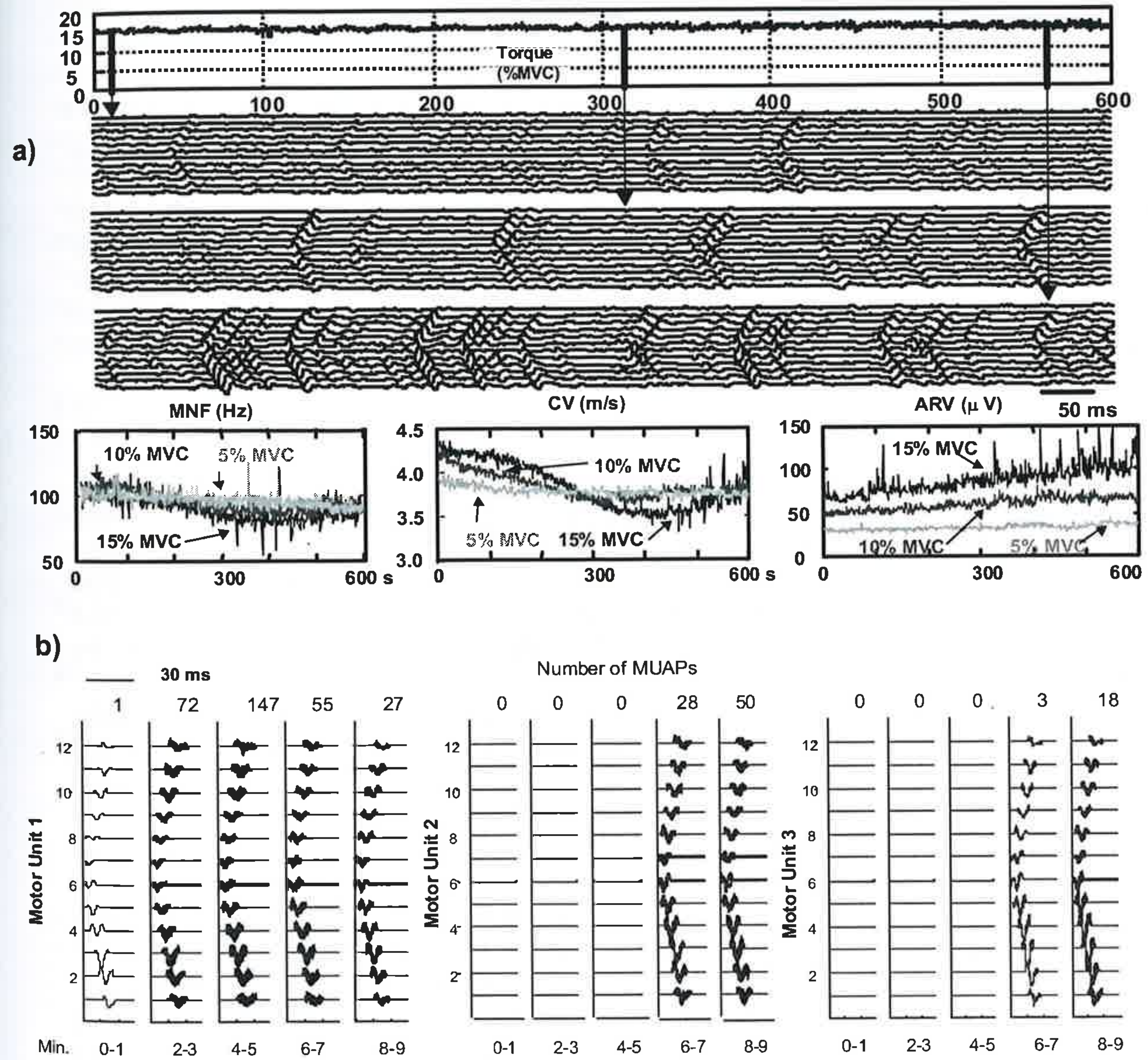

Figure. a) Force, raw signals and EMG variables of the biceps brachii during an isometric contraction sustained at 15% MVC for 10 minutes. b) Number of isolated firings of three motor units identified during three segments (each three seconds long) during two minutes (0-1, 2-3,.....,8-9). The firings identified by the algorithms on each 3 s interval do not represent the total number of firings, since superpositions are not resolved, and increase with time. For this reason the decreasing number of firings of motor unit 1 in 8-9 may not reflect the firing rate of this motor unit.

MUSCLE RECRUITMENT IN LOW BACK PAIN PATIENTS, RELEVANCE FOR FATIGUE ASSESSMENT

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INTRODUCTION

Though the evidence in the literature is not consistent, many studies have shown increased trunk extensor muscle activity in low back pain patients and possibly related, increased fatigability (1,2). Conventional conservative therapy, interprets this hyperactivity and the concomitant fatigability as a source of the perpetuation and aggrevation of back pain and is aimed at muscle relaxation (1). However, one could question whether hyperactivity of trunk muscles in low back pain patients is indeed an unwanted aspect of pain behaviour or rather a functional adaptation to changed mechanical demands. By mechanical necessity the increased activity must be accompanied by cocontraction of the antagonistic muscles. Accepting Panjabi's thesis of instability of the spine as a potential source of low back pain (3), cocontraction might provide protection against further damage and pain (4). In addition, it may be expected that the moment contribution of segmentally inserting muscles (e.g. lumbar erector spinae and internal oblique) would be increased relative to muscles inserting on thorax and pelvis only (e.g. thoracic erector spinae and rectus abdominus), as this would also enhance spinal stability (5). The aim of the present study was to test whether such adaptive changes in trunk muscle recruitment can be found in low back pain patients.

METHODS

Sixteen patients with chronic idiopathic LBP and 16 matched healthy control subjects volunteered for this study. Patients had no neurologic deficits, structural deformities, genetic spinal disorders or previous spinal surgery. Patients suffered from LBP for periods ranging from 6 months to 35 years. On a 10 cm visual analog scale, patients expressed their LBP on the day of testing as 2.1 (SD=1.2). Healthy control subjects had never experienced back pain lasting longer than three consecutive days and matched the experimental LBP group by gender, age, weight and height.

Subjects performed slow trunk motions about the neutral posture in three planes and isometric ramp contractions in two planes, while seated upright. Trials were performed unloaded and with an additional trunk load of 8kg in females and 16 kg in males. EMG signals were collected from the following trunk muscles bilaterally: rectus abdominis (RA), external oblique (EO), internal oblique (IO), thoracic erector spinae spinae (TES), lumbar erector spinae (LES).

Three ratios of absolute EMG amplitudes were calculated: ratio 1: sum of antagonists over sum of agonists, ratio 2: LES over TES, ratio 3: IO over RA. Data obtained over a range of motion from minus 5 to plus 5 degrees in each plane of movement were averaged. EMG ratios from the beginning of the ramp up to the instant where 17 Nm of moment was produced were averaged. Using a method described in a separate paper presented at this conference ratios of moment contributions of separate muscles analogous to the EMG ratios were calculated.

ANOVA was used to test for effects of health status, movement direction, plane of movement and trunk load on EMG and moment ratios.

RESULTS

The ratios of LES over TES EMG amplitude and the corresponding ratios of estimated moment contributions were significantly higher in patients as compared to controls in both motion and ramp trials (main effects over all conditions EMG ratio: $p = 0.006$ and 0.007; moment ratio: $p = 0.03$ and $p = 0.033$; Figure 1). The ratio of antagonist over agonist EMG and the corresponding moment ratio were higher in patients in motion trials only (main effects over all conditions; $p = 0.006$ and $p = 0.011$; respectively). No effect of health status on the ratio of IO over RA activity was found.

DISCUSSION

The trunk muscle recruitment pattern found in the group of low back pain patients in the present study is characterized by an increased moment contribution of the lumbar erector spinae relative to the thoracic erector spinae muscle and an increase of cocontraction. This pattern appears functional with respect to enhancing spinal stability. However, this recruitment pattern is also energetically inefficient. It will increase the overall rate of fatigue development, though not necessarilly in all trunk muscles. In addition, these changes may remain after their functional significance has disappeared, because injured structures have recovered. The fact that muscle may either be activated more (e.g. LES) or less (e.g. TES) in low back pain, suggests that in clinical assessment of muscle recruitment and fatigability information on multiple muscles needs to be obtained. The data presented suggest that adaptive changes may underly abnormal recruitment and fatigability of trunk muscles in low back pain patients. Given the potentially adaptive nature of these changes, caution should be exercised when rehabilitating low back pain patients with sole purpose of restoring normal muscle recruitment pattern.

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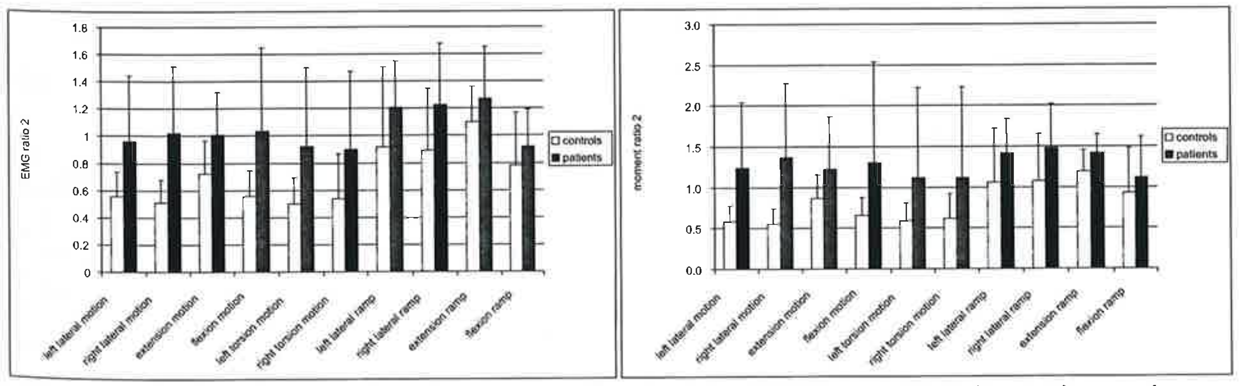

Figure 1. EMG ratios (left) and moment ratios (right) of LES over TES in motion and ramp trials. Values shown are averaged across subjects and loaded and unloaded trials. The error bars indicate one standard deviation.

THE ELECTROMYOGRAPHIC ASSESSMENT OF BACK MUSCLE WEAKNESS AND FIBER COMPOSITION

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INTRODUCTION.
Chronic low back pain (CLBP) is often associated with back muscle deconditioning (reduction in muscle mass, changes in muscle fiber characteristics) [3, 8]. These alterations in the structure of back muscles might lead to weakness and fatigability, two back muscle impairments recognized as potential causes of the recurrent nature of LBP [8]. Back muscle capacity is usually assessed during volitional effort using dynamometry and muscle composition could be measured through biopsy techniques. Unfortunately, the measurement of maximal muscle capacity is limited by motivational and psychological factors and biopsy is generally considered too invasive. However, the assessment of muscle fatigue is possible through the use of electromyography (EMG) [8]. Previous findings suggest that it may also be possible to assess muscle weakness [9] and muscle composition [4] with surface EMG. Thus, the purpose of this study was to assess the reliability and sensitivity to low back pain status of different EMG indices developed for the assessment of back muscle weakness and fiber composition.

METHODS
Subjects and tasks.
Twenty healthy (Age: 38 ± 13 yr; Height: 1.75 ± 0.04 m; Mass: 73 ± 9 kg) and 20 CLBP (Age: 41 ± 14 yr; Height: 1.77 ± 0.08 m; Mass: 80 ± 13 kg) male volunteers were evaluated on three different days (at least two days apart within 2 weeks) to assess the reliability of EMG indices and the influence of low back status. CLBP was defined as a daily or almost daily lumbar or lumbosacral pain with or without proximal radicular pain (limited distally to the knees) for at least three months. Thirteen healthy females (Age: 26 ± 4 yr; Height: 1.67 ± 0.05 m; Mass: 61 ± 7 kg) were also assessed once and compared to twelve healthy males (Age: 27 ± 5 yr; Height: 1.75 ± 0.05 m; Mass: 74 ± 11 kg) of similar age. The subjects performed, while standing in a dynamometer (Fig. 1, details in [5]), three 7 s static ramp contractions (trunk extension) ranging from 0 to 100% of the maximal voluntary contraction (MVC), using the L5/S1 extension moment as visual feedback.

Electromyography.
The EMG signals from four pairs of back muscles (Fig. 2.) were collected (bandpass filter: 20-450 Hz; preamplification gain: 1000; sampling rate: 2048 Hz) with active surface electrodes (Delsys Inc., MA). The MF (250 ms windows, 512 points; Hanning window, fast Fourier transform) and the corresponding root mean square (RMS) values of the EMG signals were computed at each 5% force level from 10 to 80% MVC.

Data analysis and statistics.
Back muscle weakness was assessed using the neuromuscular efficiency (NME) concept which was defined as the slope of the relationship (quantified by linear regression) between

the extension moment at L5/S1 (Y axis) and RMS (X axis) across the force levels (NME_{10-80}) [7]. More efficient muscle contractions should present steeper (more +) slopes (larger NME_{10-80}). Back muscle composition was assessed by two EMG parameters presumably sensitive to muscle fiber composition [2]: (1) the slope (MFT_{10-80}) and (2) the intercept ($IMFT_{10-80}$) of the relationship between the MF (Y axis) and the extension moment at L5/S1 (X axis) across the force levels. Each EMG parameter (NME_{10-80}; MFT_{10-80}; $IMFT_{10-80}$) computed for each ramp and each muscle was averaged across bilateral muscle pairs and the three ramps. The power contained in different frequency bands (20-60, 60-120, 120-180, 180-240, 240-300 Hz) of the power spectrum were also examined between healthy and CLBP male subjects. Between group comparisons were performed with unpaired t-tests ($\alpha = 0.05$). Reliability was assessed for each EMG parameter, group (healthy and CLBP male subjects) and muscle pair by the intra-class correlation coefficient (ICC) and standard error of measurement (SEM) expressed as a percentage of the grand mean (across days) [6].

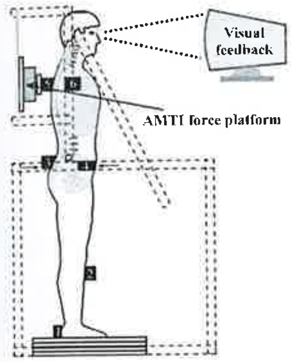

Fig. 1. Experimental setup

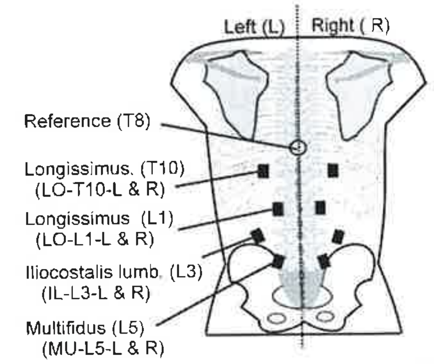

Fig. 2. EMG recording

RESULTS
Description of subject samples and reliability.
Age, height, mass, BMI and subcutaneous tissue thickness at each electrode site were not significantly different between the healthy and CLBP subjects. Males and females subjects samples were different according in height, mass and subcutaneous tissue thickness at the L3 and L5 electrode sites (females > males). At the first day of testing, the CLBP subjects showed a perception of functional capacity score of 18% (SD: 14) [1] and a back pain intensity equivalent to 1.9 cm (SD: 2.4) on a 10 cm visual analogue scale. The NME_{10-80} and $IMFT_{10-80}$ EMG parameters showed excellent reliability for both groups (ICCs from 0.74 to 0.96; SEMs from 6 to 23%) while modest results were achieved for MFT_{10-80} (ICCs from 0.36 to 0.71; SEMs from 189 to 598%).

Back muscle weakness.
The healthy subjects showed higher (t-test, P = 0.004) peak extension moment at L5/S1 (252 Nm, SD: 40) than CLBP subjects (202 Nm, SD: 62) but the NME_{10-80} was not different between groups (Fig. 3). Likewise, men showed a significantly (P = 0.000) higher peak extension moment at L5/S1 (274 Nm, SD: 38) than women (191 Nm, SD: 26) and the NME_{10-80} showed no between group difference, even when accounting for differences in subcutaneous tissue thickness (ANCOVA). However, accounting for skinfold thickness increased the estimated marginal means of the men and decreased the estimated marginal

means of the women so that a nearly significant difference was obtained for the LO-L1 muscle.

Back muscle composition.
The MF based parameters (MFT_{10-80}, $IMFT_{10-80}$) were not sensitive to the expected between group differences in muscle composition (Fig. 4). Between group discrepancies were observed in the MF data across the force levels (Fig. 4) but a large inter-individual variability was present.

Frequency banding analysis.
The power contained in the different frequency bands of the EMG power spectrum was equivalent for the same extension moment at L5/S1.

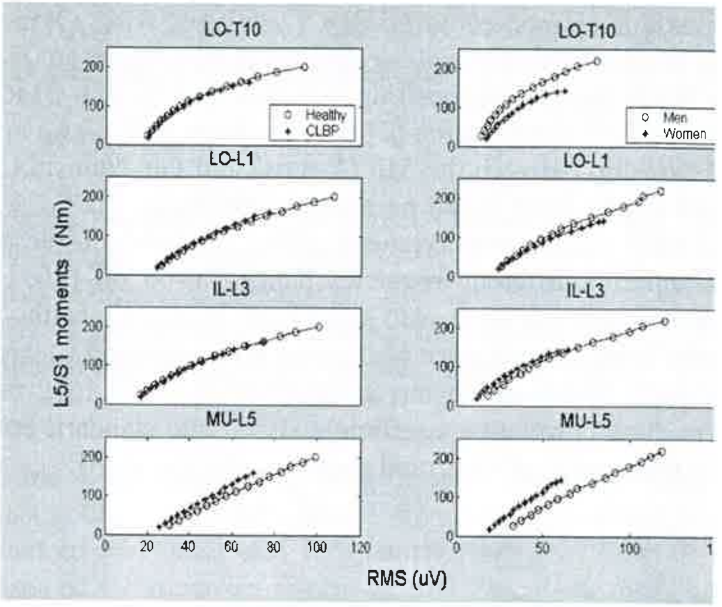

Fig. 3. Trunk extension moment at L5/S1 (Nm) as a function of RMS amplitude (μV)

DISCUSSION

Back muscle weakness.
Excellent reliability was observed for the NME_{10-80} parameter for both groups. However, even though CLBP males and healthy females were weaker than the healthy males according to the dynamometry results, the NME_{10-80} EMG parameter showed no between group difference. Pain and fear of injury may have affected the strength measures for CLBP subjects. However, this possible bias was unlikely between healthy males and females, which confirmed that the NME parameter was not sensitive to back muscle weakness. The NME concept might be a useful tool to assess weakness at simple joints but might not be applicable to the spine where multiple muscles span several joints.

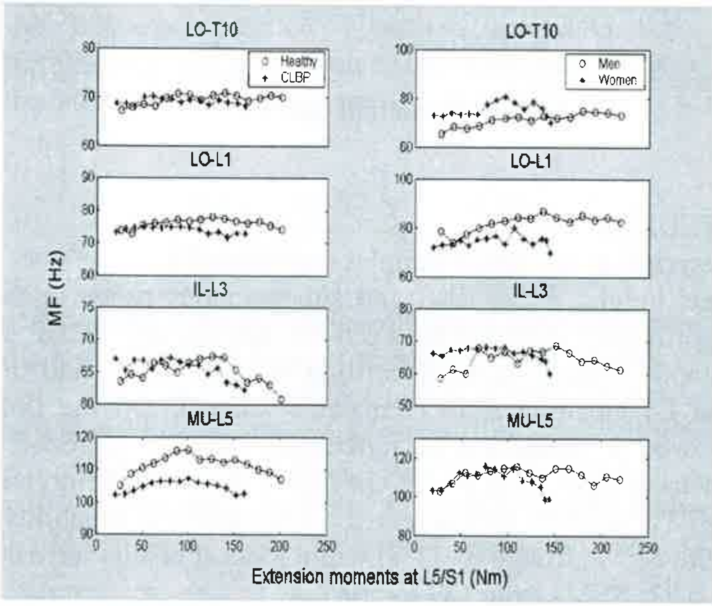

Fig. 4. MF (Hz) as a function of the trunk extension moment at L5/S1 (Nm)

Back muscle composition.
The $IMFT_{10-80}$ parameters showed excellent reliability while modest results were achieved for MFT_{10-80}. The MFT_{10-80} slopes were close to zero, thus explaining the high SEM relative values. It is possible that the CLBP subjects involved in the present study were not impaired enough to have alterations in muscle fiber size and/or proportion that could be detected by EMG spectral parameters. In the present study, the CLBP subjects were working on a regular basis at the time of testing and had minimal disability according to the gradation proposed by

Fairbank et al. [1]. However, because females should have smaller muscle fibers than men, irrespective of fiber type or specific back muscle [8,10], the lack of difference between men and women suggested that these spectral parameters were not sensitive to muscle fiber size.

Frequency banding analysis.
Given the discrepancy between the MF curves of healthy and CLBP (Fig. 4, MU-L5 muscle), it was initially hypothesized that between group differences could appear within specific frequency bands of the power spectrum. Unfortunately, this was not the case. However, time-frequency analyses such as wavelet transforms might give better results, given their better ability to analyze the EMG signal with specific time resolutions.

CONCLUSION
The assessment of muscle weakness and fiber composition through surface EMG, as previously applied for limb muscles, was questionable in the case of back muscles. More advanced EMG processing tools should be developed and tested on more severely affected CLBP subjects in terms of duration and pain before the use of spectral parameters as a noninvasive alternative to biopsy of back muscles is rejected.

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REPEATABILITY OF SURFACE EMG VARIABLES IN THE STERNOCLEIDOMASTOID AND ANTERIOR SCALENE MUSCLES

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INTRODUCTION: Assessment of the repeatability of surface electromyography (sEMG) measurements and the identification of accurate electrode positioning criteria based on the localisation of the innervation zones (1, 2) are recognised as factors necessary to achieve standardisation of sEMG based muscle assessment. These factors represent the main goal of the standardization process initiated by the European concerted action SENIAM (Surface EMG for Non-Invasive Assessment of Muscles) and continued with the European shared cost project NEW (Neuromuscular assessment in the Elderly Worker).
A recent review of the literature concerning the use of sEMG for the analysis of neck muscle activity (3) highlighted the need for further research to establish the reliability of sEMG measures for the neck musculature. Further investigations are essential to enhance the value of sEMG as a tool for assessing neck muscle activity and to standardize sEMG methodology in order to allow comparison of data collected in different laboratories and by different operators. The purpose of this study was to examine the repeatability of sEMG-derived indices of muscle fatigue for the sternocleidomastoid (SCM) and anterior scalene (AS) muscles in healthy volunteer subjects during fatiguing voluntary contractions. The repeatability of cervical flexion force values was also examined during maximal voluntary contractions (MVC).

METHODS: Nine volunteers (7 males, 2 females) aged between 23 and 39 years (Mean 28.0, SD 5.0 years) participated in this study after giving informed consent. Subjects were included if they were free of neck pain, had no past history of orthopaedic disorders affecting the neck and no history of neurological disorders. Subjects attended three testing sessions over non-consecutive days.
The repeatability of the sEMG mean frequency (MNF), average rectified value (ARV) and conduction velocity (CV) were examined for SCM and AS during sustained isometric cervical flexion contractions at 50% of the MVC. Repeatability estimates were obtained for the initial values and rates of change of the MNF, ARV and CV, by using both the intraclass correlation coefficient (ICC) and the normalised standard error of the mean (nSEM).

RESULTS: Good levels of repeatability were obtained for the initial value and slope of ARV for the SCM (ICC = 70.6% and 69.6% respectively). Data from the AS demonstrated high repeatability for the initial value of MNF (ICC=74.3%) and the slope of ARV (ICC=76.5%). Excellent values of nSEM (from 2.6 to 7.2%) were found for the initial values of CV and MNF for both SCM and AS, providing evidence of high repeatability for these variables in terms of repeated measure precision. An ICC of 92.5% and a nSEM of 8.7% were obtained for the repeatability of the maximum voluntary cervical flexion force.
The Mann-Whitney non parametric test indicates that for all the sEMG parameters, except ARV slope and ARV normalized slope, the estimates are statistically different between SCM and AS. The initial values (of MNF, CV and ARV), the rate of change (of MNF and CV) and the normalized slope (of MNF and CV) are greater in the SCM.

DISCUSSION: The results of this study indicate that for the SCM, because of the low values of nSEM, initial values of all the sEMG variables tested (MNF, ARV and CV) demonstrate

reliability or constancy that supports their use in clinical and research evaluations. The initial value of ARV for the SCM showed good repeatability as described by the ICC, indicating that this measure can be used to monitor individual subjects. The low values obtained in this study for both between and within-subject nSEM for MNF and CV initial values demonstrate that these variables could be estimated with very high repeated measures precision. However, it also shows that there was very little variation of the initial values of MNF and CV obtained across subjects and trials, so these measures might not be able to detect different muscle properties among uniform groups (such as asymptomatic subjects). The initial values of MNF and CV in the SCM muscle might therefore be best utilised in providing a reference range for a uniform group of subjects (e.g. normative data).
For the AS muscles, the initial values of MNF and CV show high levels of reliability, as indicated by the low nSEM, thus demonstrating that these variables could be useful measures in clinical and research applications. The low value obtained for the nSEM of the MNF intercept for the between-subject analysis, in combination with the high ICC demonstrates that this measure is both repeatable and reliable. These results suggest that this measure will be suitable for describing differences between asymptomatic subjects, distinguishing between healthy subjects and subjects affected by pathologies that modify MNF, or monitoring the efficacy of rehabilitation treatments. As found in the SCM, high levels of repeated measure precision were also evident for the initial value of CV for the AS. However, because of the low ICC, the initial value of CV for the AS has limitations in detecting differences among groups of similar subjects.
For both the SCM and AS muscles the slope of the ARV showed good repeatability as indicated by the ICC values obtained. This finding indicates the suitability of this measure for clinical and research applications. In contrast, MNF slope and both initial value and slope of CV displayed low ICC values and greater values of the nSEM for the within-subject analysis compared to the between-subject analysis.

CONCLUSIONS:
Initial values of MNF and CV are the most reliable parameters (i.e. the parameters that can be estimated with the highest precision) for SCM, AS.
ARV (both initial value and slope) in the SCM, and the initial value of MNF and slope of ARV in the AS have demonstrated high repeatability.
Slopes of MNF, ARV and CV show, in general, poor repeatability although are often associated with an acceptable reliability (nSEM<15%, mainly for MNF and ARV) that allows such parameters to be used in the assessment of manifestations of localized muscle fatigue.
The low value obtained for the nSEM of the initial value of MNF for the AS, in combination with the high ICC indicates that it is suitable for specific clinical and research applications and represents the sEMG variable of choice for future evaluation purposes.

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SURFACE EMG ALTERATIONS INDUCED BY UNDER WATER RECORDING. A CASE STUDY

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INTRODUCTION
Although rehabilitation treatments in exercise pools are widely used, there are no reports in the scientific literature aimed to validate and standardize the methodological aspects of the EMG recording in water. Different recording techniques are found in the literature of water surface EMG. Some authors isolate the electrodes from water by means of water resistant adhesive film, others claim that no extra protection on the electrodes is necessary in the water.

Considering this apparent lack of uniformity and standardization, this work aims to assess:
- If water introduces an alteration in surface EMG signal recordings
- How the electrodes interact with water
- How water modifies both amplitude and spectral characteristics of the EMG signal
- The best procedure to reduce wet recording artifacts.

MATERIALS AND METHODS
Surface EMG signals during isometric contractions of the abductor digiti minimi muscle were recorded in one subject in several different conditions using pre-gelled commercially available adhesive electrodes (Arbo Kiddy® by Kendall, USA). The subject was asked to abduct the fifth finger against a spring and shorten the spring to a fixed length. This approach assured repeatable isometric efforts at the same force value. Signals were recorded both in dry condition (D) and in water. Two different water solutions were tested. Samples from hydrotherapy pool (PW) and distilled water (DW) were maintained at the constant temperature of 30 °C during the experiments. Contractions were performed with and without water resistant adhesive tape (T) (Cutifilm® Beiersdorf AG, Germany).
In all the above conditions the following EMG variables were estimated to determine the water induced effect on the signal recording: Average Rectified Value (ARV), Root Mean Square value (RMS), Mean frequency (MNF, the centroid of the spectrum) and Median Frequency (MDF, that divides the spectrum into two parts of equal power).
EMG signals were recorded for ten seconds in single differential configuration using a pair of electrodes placed on the skin (electrode diameter equal to 10 mm; inter-electrode distance equal to 20 mm, patient ground on the controlateral wrist). Special care was taken in electrode location to avoid artifacts induced by the innervation zone. The best location for the adhesive electrode placement was chosen distally because of the greater signal amplitude with respect to the proximal side.

RESULTS
Modification of the signal amplitude
The most evident finding is that the amplitude of the signals recorded in PW and DW is strongly diminished (up to 5-12% of the corresponding value obtained in D). This result was confirmed also by measuring the electrode impedance in both D and PW conditions in eight subjects. The impedance between the electrodes in D was 73,4±28,6 kΩ (mean ± std) and in PW was 6,8±3,3 kΩ. These estimates were obtained with a current of 3.9 μA at 20 Hz. On

the basis of a simplified model of the skin-electrode impedance, we estimated the ratio between the output signal in D and the output signal in PW as 13,6±6,7. This value, due to the interelectrode impedance introduced by water conduction volume, explains the reduction of ARV and RMS variables in the wet condition reported above.

Modification of the signal spectral content
The movement of water introduces significant spectral components in the low frequency range [0-20 Hz] with the consequence of underestimation of MNF and MDF values. The Figure shows how such artifacts affect the signal power spectrum and how noise spectral contributions become important because of the dramatic EMG amplitude reduction (described above). MNF, MDF and RMS alterations can be totally avoided with the use of water-resistant adhesive taping.

DISCUSSION AND CONCLUSION
Differences between the D and PW modalities described above, albeit from a single case study, can be compared with the range of EMG variables available from the literature. On the basis of such values it is possible to attribute the amplitude decrease in PW and DW with respect to D to water effect, and not to the experimental noise and/or the between subject variability.

In conclusion, to properly perform EMG acquisition in water, this work supports:
- The need for water-resistant adhesive taping technique to avoid large signal artifacts both in the time and in the frequency domain, avoiding also the need for high pass filtering techniques;
- The need for the design and the industrial production of water proof EMG electrodes

ACKNOWLEDGMENTS
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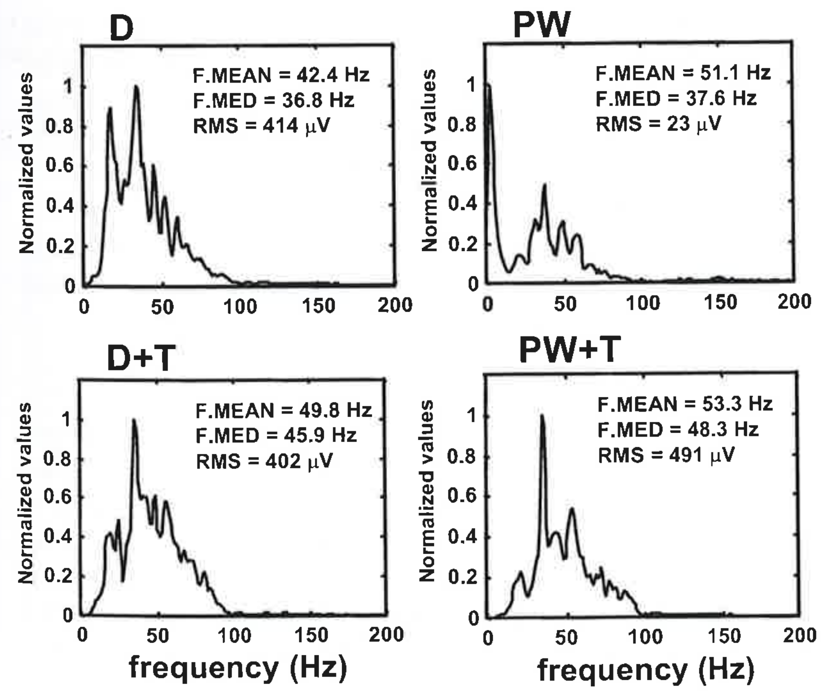

Signal spectra of the raw (10-450 Hz) signals recorded in the two different modalities (D and PW) compared with the signal spectra of raw signals recorded in the same modalities, but using the taping technique.
Please note that:
1) although the MNF estimate in PW condition seems unchanged, it is actually strongly affected by two opposite artifacts: the low spectral component in the range 0-20 Hz and the increased importance of the noise spectral contribution
2) the T modality reduces artifacts and leads to a spectrum with similar characteristics as in the D case.

QUANTIFYING TRUNK MUSCLE CO-ACTIVATION USING PATTERN RECOGNITION TECHNIQUES

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INTRODUCTION

Dynamic stability exercises for the trunk musculature focus on co-activating muscles in a synergistic manner in an attempt to improve spinal stability. Onset times and amplitude measures from the electromyogram (EMG) are often used as indicators of muscle co-activation, however, muscles may have similar onset times, or similar amplitudes but very different temporal EMG patterns. The correlation between two EMG waveforms has been used to quantify co-activation, however, this technique is limited to comparing only two muscles at one time. The purpose of this study was to evaluate the effectiveness of a pattern recognition technique applied to temporal EMG waveforms to quantify synergistic co-activation among the agonist and antagonist trunk muscles in response to a sequence of perturbations with increasing difficulty.

METHODS

Nineteen healthy subjects (7 males and 12 females) had surface electrodes (Ag/ AgCl) attached in a bipolar configuration over the lower and upper rectus abdominus, three sites corresponding to anterior, medial and posterior fibers of the external oblique, the erector spinae, and the multifidus muscles on the right side. Subjects performed, in random order, five levels of an exercise progression aimed at challenging trunk stability[1]. Levels two to five used single and double leg extension tasks of increasing difficulty while maintaining a neutral spinal posture though an abdominal hollowing maneuver[1], and were the levels examined in this study. Level 2 included a supported single-leg extension, level 3 an unsupported single-leg extension, level 4 a supported double-leg extension and level 5 an unsupported double-leg extension. An event marker synchronized the tasks with the EMG data and a video system was used to monitor lumbar and pelvic motion. The event marker and EMG signals were digitized at 1000Hz using a National Instruments digital interface card (12-bit resolution) and Labview software. The EMG signals were full wave rectified and low pass filtered at 6 Hz to yield a linear envelop profile. The profiles were amplitude normalized to the average amplitude recorded during each task and time normalized to 100% time using a linear interpolation routine. The Karhunen-Loève expansion was applied to the matrix of ensemble average EMG profiles recorded from the 7 muscle sites (101*1548) and eigenvectors representing the principal features were derived. The weighting coefficients for each pattern were the main dependent variables in separate three-factor (trial, level and muscle) repeated ANOVA models ($\alpha = 0.05$). Bonferonni post hoc analyses were performed on significant effects.

RESULTS

Ninety one percent of the variance in the 1548 patterns was explained by the three principal patterns illustrated in figure 1. The residuals for all three weighting coefficients for the three patterns were normally distributed. There were no significant trial interactions for the first principal pattern and no significant trial main effects or interactions for patterns 2 and 3. The ANOVA found statistically significant muscle by level interactions ($p<0.05$) for the three principal patterns. The between level ($p<0.083$) differences showed that pattern one (EV1) was not different among levels for all muscles, whereas pattern 2 (EV2) was different for the abdominal muscles among the four levels, but not for the two trunk extensor sites. The rectus abdominus muscles had high positive weighting coefficients for pattern 2 that included the burst of activity for the unsupported leg extension-flexion task. The between muscle differences ($p<0.024$) illustrated that more significant differences were found among muscle sites for the unsupported tasks compared to the supported tasks for pattern 2, but for pattern 3 more differences were found for the unsupported task.

DISCUSSION

The Karhunen-Loève expansion effectively reduced the EMG profiles into three principal patterns. The first pattern (EV1) was dominant for all of the muscles and for all of the levels. It essentially determines the mean amplitude. The between level differences for the second and third patterns showed that the neuromuscular control pattern was different between the two unsupported and two supported leg-extension tasks. In particular the second principal pattern (see figure 1) that depicted a burst of activity peaking at full leg extension was prominent for the two rectus abdominus sites for the unsupported tasks, but only for the external oblique during the double-leg unsupported task. The differences between the single and double leg extension tasks were most prominent for the supported task, illustrating a drop in activity at full extension. The differences illustrate a lack of synergistic co-activation among the agonist sites and between agonist and antagonists for some of the conditions examined. Of particular interest was that all three external oblique sites did not produce similar temporal patterns, in particular the anterior fibers, indicating inhomogeneous recruitment within this muscle.

CONCLUSION

The key characteristics from the EMG waveform data were captured by three principal patterns. The differences among muscles and levels illustrate that the degree of difficulty alters the co-activation patterns, and does not simply result in an increase in neural gain. It was concluded that co-activation among muscles during sequenced tasks can be quantitatively evaluated using this technique, and can be used to assess whether the neuromuscular control strategies differ from normal.

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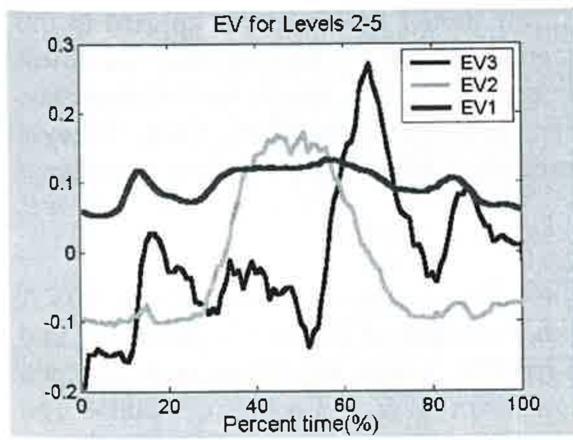

Figure 1. The three principal patterns: EV1 captured three small bursts of activity (86.4%), EV2 captures the large burst during the leg extension phase (3.4%), and EV3 captures the variance throughout the exercise (1%). 0-20% time is the leg-lift phase to 90 degrees, 20-50% is leg extension, 50-80% is leg flexion and 80-100% is leg lowering.

ELECTROMYOGRAPHIC ANALYSIS OF THREE MODES OF SHOULDER ABDUCTION THERAPY

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INTRODUCTION

Physical therapists must consider several factors when selecting the most beneficial exercise modality for patients; these factors include availability of and biomechanical properties of resistance equipment. One important biomechanical property is how closely the resistance coincides with the muscle strength curve, defined as "any plot that describes the maximum muscular capability at a joint as a function of joint angle."[1]

There is a paucity of data concerning muscle activation during exercise involving the most common modalities - pulleys, free weights, and elastic resistance. The purpose of our study is to assess muscle activity throughout a fixed range of motion of shoulder abduction using a standardized load with the above three modalities. We hypothesize that there will be no significant difference in EMG activity across each modality in a given arc of motion; therefore, the three modalities studied can be used interchangeably in a rehabilitation program

METHODS

Ten healthy young women, age 21 to 45, participated in this study after giving their informed consent as approved by the Institutional Review Board. Two Ag/AgCl surface electrodes were placed longitudinally with a 2.5 cm interelectrode distance over the bulk of the middle deltoid muscle on the subject's right side. An additional electrode was placed over the subject's lateral epicondyle as a reference. The electrodes were connected to an EMG amplifier with gain of 1000, CMRR of 90 dB, and frequency pass-band between 6-500 Hz. A plastic vest containing a shoulder electro-goniometer was worn so that the axis of the goniometer most closely approximated the glenohumeral joint's axis of motion. The goniometer was taped to the upper arm in line with the lateral epicondyle. The EMG signal was acquired at a rate of 1000 samples per second and stored for further processing and analysis. A computerized tracking device was used to standardize velocity throughout the range of motion.

Three trials of shoulder abduction in the range of 40° to 140°, using a five-pound load with each of the following modalities were performed in random order: free weights, pulleys, and Theraband®. EMG data was recorded during a 12-second period, including 2 seconds before the movement began, 8 seconds during the movement, and 2 seconds after it ended.

The range of motion was divided into angle bands of 20°, and categorized as concentric or eccentric. Mean MAV levels were extracted and then normalized by dividing each value by the EMG signal during MVC at 90°. A general linear model ANOVA was applied to the MAV for each angle band to determine the effect on the MAV of the following variables: subjects, angle band, concentric/eccentric condition, resistance type, interaction of resistance type for either condition, and interaction of resistance type and angle band. Then, Tukey's Studentized Range Test was used to analyze experimental variables contributing to significant differences.

RESULTS

Table 1 shows the mean MAV values for each angle band in each condition The ANOVA indicated that the variables range of angular position, eccentric or concentric condition, and interaction of range of angle and type of resistance ($p<.0001$) were significant with an alpha criterion of 0.05. The type of resistance was significant with a p value of .0005. The interaction of condition and resistance variable showed marginal significance with p value of .084.

Post-hoc tests revealed a significant difference in the MAV mean values under the condition of the pulley (lower mean value), while no significant difference was apparent between the free weight and Theraband®. Further post-hoc analysis indicated a significant difference of the eccentric portion of the pulley from the Theraband® and free weight. There were also significant differences among the mean MAV values for the angle bands, and between concentric and eccentric conditions. The mean MAV was consistently lower during the eccentric portion of the activity than the concentric portion with all loads. THE mean MAV above 120° with the Theraband® was significantly greater than that with the other two types of resistance.

DISCUSSION

The most important finding in this study is that elastic resistance requires the same amount of muscle activity as pulleys and free weights except beyond 120° of shoulder abduction.

An issue addressed here is the assumption that more force is required when using Theraband® than when using pulleys or free weights of comparable load. This assumption is based on the elastic properties of Theraband®; as it lengthens, tension increases, requiring greater force to oppose the increasing resistance. As found by Hughes et al, the angle between the tubing and the arm coupled with the angle of shoulder abduction offset the increasing tension of the tubing [2]; however, our study revealed an increase in muscle activity above 120° of concentric motion with Theraband® as opposed to pulleys and free weights.

In corresponding band angles of motion, the mean muscle activity was less in the eccentric portion. When comparing the three resistances, the MAV value of the pulley was significantly different from the other two resistances. The interaction between the condition and resistance variables revealed that the difference occurred during only the eccentric portion of shoulder abduction. We attribute this to friction between the rope and pulley.

In summary, our results show increased muscle activity above 120° using elastic resistance, and similar elsewhere in the range of motion; we hence recommend that end range of motion be avoided when prescribing this type of resistance to the patients for home use.

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ACKNOWLEDGEMENT

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Table 1

Angle Band	Free weight	Pulley	Theraband
40-60 Concentric	0.30±0.07	0.20±0.07	0.17±0.05
60 - 80 Concentric	0.52±0.09	0.46±0.10	0.34±0.09
80 - 100 Concentric	0.65±0.10	0.63±0.11	0.54±0.16
100 - 120 Concentric	0.78±0.14	0.76±0.12	0.77±0.16
> 120 Concentric	1.13±0.22	1.09±0.16	1.31±0.32
>120 Eccentric	0.66±0.15	0.57±0.08	0.79±0.15
100 - 120 Eccentric	0.45±0.07	0.30±0.07	0.49±0.14
80 - 100 Eccentric	0.36±0.05	0.26±0.05	0.31±0.10
60 - 80 Eccentric	0.28±0.05	0.20±0.04	0.19±0.07
40 - 60 Eccentric	0.20±0.05	0.13±0.04	0.12±0.03

AMPLITUDE AND TIMING DIFFERENCES BETWEEN CONTROL AND LOW BACK PAIN SUBJECTS DURING SELECTED EXERCISES

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INTRODUCTION

Muscle activation amplitudes have been used to evaluate the neuromuscular gain associated with function, and to determine impairment associated with injury and pain. Adequate standardization and reporting of the EMG methodology are needed to allow for comparisons among studies [1]. While normalization to maximal contractions has been criticized, in particular for patient populations, a rationale has been presented to support such techniques with respect to studies of low back pain [2] . At the present time no standard procedures have been agreed upon and studies use different exercises, different procedures and different criteria for selecting the segment of data to analyze from a normalization exercise. These differences have contributed to the controversy among studies assessing function and dysfunction of the neuromuscular system associated with low back pain using surface electromyography (EMG). The present study addresses the issue of repeatability of results and the differences between those with low back pain and those without, for commonly used normalization exercises. We compared timing and normalized amplitudes from EMG waveforms while subjects performed four exercises aimed at preferentially recruiting muscles to different amplitudes based on function.

METHODS

Thirty-nine men volunteered; 25 had no history of LBP (CON) and 14 had a history of chronic LBP (one year or longer). Surface electrodes (Ag/AgCl) were placed in a bipolar configuration over the right lower and upper rectus abdominus; the right external oblique; the right erector spinae; and the right and the left multifidus muscles. Four maximal effort exercises were randomly assigned: a resisted "curl up" aimed to recruit the rectus abdominus; a "trunk rotation" aimed to recruit external oblique; an "isometric abdominal contraction" aimed to recruit both rectus abdominus and external oblique and a "trunk extension" aimed to recruit the erector spinae and multifidus. EMG data were digitized at 1000 Hz during the 4s exercise. Root-mean-square (RMS) amplitudes were calculated for each trial using a computer algorithm that determined the 500 consecutive samples of EMG data, within the 4 s sample, with the highest RMS amplitude for each exercise, for each muscle. The algorithm determined the time in the 4 s. sample at which the maximum amplitude window occurred. RMS amplitudes for each trial were normalized to the maximum amplitude produced for that muscle site regardless of exercise. Three-factor mixed model ANOVAs tested all two and three way interactions and main effects for each muscle site separately, for time and amplitude. Post hoc analyses were performed on all significant effects ($\alpha = 0.05$).

RESULTS

There were no significant ($p>0.05$) trial main effects or interactions for any dependent measure. There was a significant group difference ($p<0.05$) for time for the two multifidus sites, with the LBP group taking on average .5 s longer to reach maximum amplitude. Examining the 25^{th} to 75^{th} percentiles illustrated that 50% of the LBP subjects were delayed by .75 to 1 s in producing their maximums for all muscle sites except the erector spinae

compared to 50% of the CON. The normalized amplitude showed a significant group by exercise interaction ($p<0.05$) for all three abdominal muscle sites, but only significant exercise effects for the three trunk extensor sites. Pair wise significant differences among muscles were not the same for the CON and the LBP group indicative of different amplitude recruitment patterns. They are presented in Table 1 below.

DISCUSSION

The results show that the timing and amplitude measures were consistent among trials. The timing differences between groups, muscles and the variability within groups clearly illustrate that choosing a predetermined window in which to calculate the maximal amplitude would not consistently detect the maximal amplitude window of data within the waveform. Both groups produced their maximal activity for the two rectus abdominus sites during the curl up exercise. While the CON group produced the highest external oblique amplitudes during the trunk rotation, the LBP subjects produced the same external oblique amplitudes for both the trunk rotation and the curl up exercises. Therefore LBP subjects were not able to consistently recruit the external oblique to high amplitudes during the trunk rotation. The trunk rotation exercise may selectively recruit high amplitudes for those with out low back pain, however, it should not be assumed that the same pattern for activation is present for those with LBP.

CONCLUSION

This study provides further evidence of neuromuscular differences between those with LBP and those without performing simple normalization exercises. Since the four exercises are used in amplitude normalization procedures, the differences found in both timing and amplitudes have implications for improving normalization procedures.

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Table 1 Pair wise significant differences for significant interactions: CU is curl up, TR is trunk rotation, IAC is isometric abdominal contraction and TE is trunk extension.

Muscle	CON	LBP	Between group
Lower rectus abdominus	CU>TR>IAC=TE	CU>TR=IAC=TE	CONTR>LBPTR
Upper rectus abdominus	CU>TR>IAC=TE	CU>TR=IAC=TE	CONTR>LBPTR
External oblique	TR>CU>IAC>TE	TR=CU>IAC>TE	CONTR>LBPTR

> Mean is significantly different from exercise to the right.
= Mean is not significantly different from exercise to right.

THE ROLE OF NECK MUSCLE DYSFUNCTION IN THE ACUTE STAGE OF THE WHIPLASH ASSOCIATED DISORDER (WAD)

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INTRODUCTION

In acute musculoskeletal pain syndromes the onset of pain initiates muscular responses to 'guard' the injured area in order to prevent more pain or injury. The role of these responses in the transition from acute to chronic pain however is still not clear. Earlier surface EMG studies show that in the chronic stage of the 'Whiplash Associated Disorder (WAD)' there is an inability to relax the trapezius muscles after a physical exercise[1]. However, there are no studies which support the presence of this or other muscular reactions in the acute stage of WAD. Therefore we performed a prospective cohort study of patients admitting in an emergency room of a general hospital with acute neck pain due to a motor vehicle accident (MVA) with the aim to study:

1. Muscle activation patterns of the upper trapezius muscles in acute WAD.
2. Differences in course of muscle activation patterns in acute WAD patients who develop chronic pain disability and patients that will recover.

METHODS

A cohort of 100 subjects were included in the study. Smooth Rectified Electromyography (SRE) of the upper trapezius muscles were obtained at 1, 4, 8, 12 and 24 weeks after a motor vehicle accident. The Neck Disability Index (NDI) was used to rate the final outcome, with a subdivision into 4 categories of disability (recovered 0-4, mild 5-14, moderate 15-24, severe 25-50). The EMG parameters obtained at 1 week post MVA were used to study the muscle activation patterns in acute neck pain. The EMG parameters obtained during 24 weeks follow up were used to study differences in course of muscle activity between the four NDI categories. EMG recordings proceeded in 4 stages:

1. *Pre-exercise muscle activity*: obtained by recording 10 epochs of upper trapezius SRE in a sitting position in a desk-chair.
2. *Isometric muscle activity*: obtained by recording 4 epochs of upper trapezius SRE in a sitting position in a desk-chair, with the arms held straight and horizontal in 90° abduction in the frontal plane of the body.
3. *Dynamic muscle activity*: obtained by recording 1 epoch of SRE in a sitting position in a desk chair at a table while performing a manual exercise. For this exercise the subject was asked to move his or her dominant arm continuously between three target areas by marking circles with a diameter of 10 milimeter using a pencil. A metronome was used to maintain a constant pace of 88 marks per minute. This exercise was performed for approximately 2 minutes. After 1 minute the SRE activity was obtained
4. *Post-exercise muscle activity:* obtained in a similar way as the pre-exercise activity.

Each epoch lasted 15 seconds with an interval of 1 minute in between. Three EMG arameters were used to study the muscle activity:

1. The isometric muscle activity, computed as the mean muscle activity during the performance of the isometric physical task
2. The dynamic muscle activity.

3. The muscle hyper-reactivity, computed as the mean pre-exercise SRE level subtracted from the mean post exercise SRE level. Hyper-reactivity was considered present if this computation resulted in a positive value (i.e. an increase in muscle activity after the exercise).

RESULTS

1. In acute WAD there is no sign of muscle hyper-reactivity after physical exercise (results not shown) but there is a reduction of muscle activation of the upper trapezius muscles during physical exercise (figure 1a and 1b).
2. During the transition from acute to chronic neck pain, the muscle activation level is different between subjects that have recovered and subjects that are suffering from moderate or severe disability. Furthermore, the reduction in muscle activation level seems to be inversely related to the extent of pain disability, in a sense that subjects suffering from highest pain disability show the most reduced recruitment of upper trapezius muscles during physical exercise(figure 1a and b).

DISCUSSION

The onset of acute pain in WAD does not initiate 'muscle guarding' by means of an increased but rather by a decreased activation level of the upper trapezius muscles. These results indicate a dynamic reorganisation of the muscular activation of neck shoulder muscles possibly aimed at minimising the use of painful muscles. Future research should focus on analysing the muscle activation of surrounding neck shoulder muscles acting as synergist in order to confirm this reorganisation.

CONCLUSION

In acute WAD as well as during the transition from acute WAD to a state of chronic neck pain disability, the muscle activation level of the upper trapezius muscles is reduced. The reduction in muscle activation level seems to be inversely related to the extent of neck pain disability.

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Figure 1a:
Dynamic activity

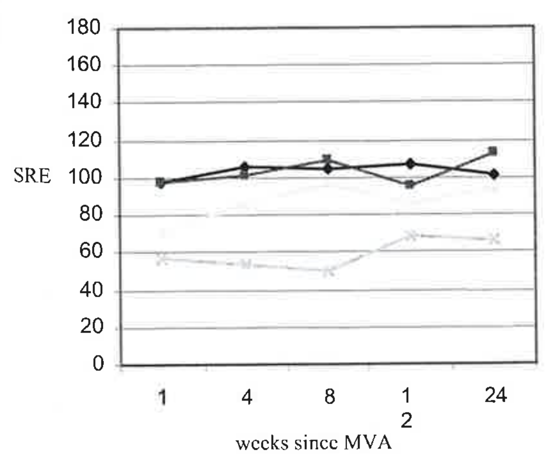
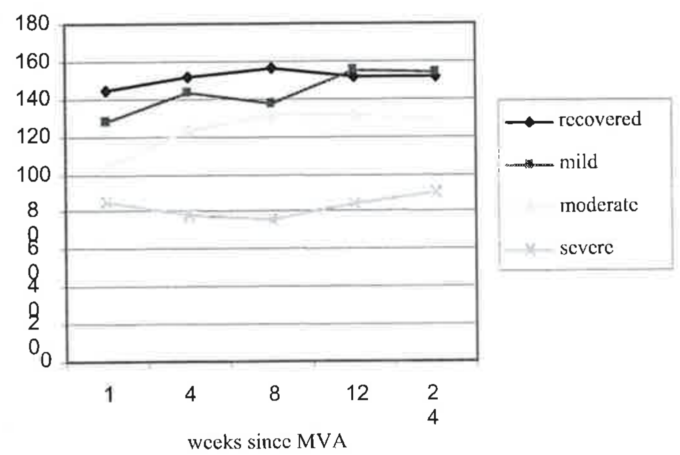


Figure 1b:
Isometric activity



SUB-CLINICAL EVOLUTION OF MOTOR CONTROL AFTER SPINAL CORD INJURY

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INTRODUCTION

Following spinal cord injury (SCI) in humans, motor dysfunction is typically assessed using clinical instruments such as the American Spinal Injury Association Impairment Scale (AIS)[ASIA, 1992; Maynard et al., 1997]. However, such clinical examination lacks sensitivity to describe the features of motor control, especially in the acute phase when emerging functions are likely to begin to appear at very low strength. Further, there exists some controversy over the presence and duration of a "spinal shock" period that is thought to begin immediately after injury, during which time motor activity is completely suppressed [Hiersemenzel et al., 2000] and which, if understood and monitored, could provide opportunities to improve long-term outcomes [Ko et al., 1999]. These issues become more important as we face the possibility of surgical treatment of the injury itself [Hulsebosch et al., 2000]. Our group has worked on surface EMG methods for assessing motor control for the past several decades, with particular emphasis on sub-clinical motor control and spasticity [Sherwood et al., 1992]. The advent of neurophysiological assessment tools such as the brain motor control assessment (BMCA) has provided a view of motor events with greater sensitivity than the clinical examination of post-SCI motor control [Sherwood et al., 1996]. Using the increased sensitivity of the BMCA, we were able to track the return of motor function from the first few motor unit firings to the development of clinically visible motor events. The current study was undertaken to examine the re-emergence of motor activity beginning as soon after injury as was medically possible using this sensitive, electromyographically-based technique.

METHODS

The procedures used include the Brain Motor Control Assessment (BMCA) protocol, which is a rigidly presented series of volitional and reflex motor tasks designed to elucidate evidence of residual brain influence on motor activity in the lower limbs while recording from 12 muscle groups [Sherwood et al. 1996]. The tasks included relaxation, reinforcement maneuvers which demonstrate mostly postural motor function, specific voluntary movements, passive movements, tendon and vibration reflexes, manually elicited clonus, responses to plantar stimulation without and with volitional attempts to suppress those responses. As a part of the BMCA study, transcranial magnetic stimulation was used to attempt to elicit motor evoked potentials below the lesion. In addition to the AIS and BMCA, the Functional Independence Measure (FIM) was administered to document subjects' level of independence. In addition to the BMCA, the AIS and the Functional Independence Measure (FIM) were administered to document subjects' motor and sensory recovery and levels of independence.Entry into the study was determined by the physician in charge and occurred within two months of injury. Subjects were assessed serially with BMCAs repeated on a weekly basis until three months post-injury. They then were seen for BMCA assessment twice a month for two months, and once a month for seven months. The FIM was repeated at the same frequency as the BMCA recordings but the AIS was repeated on discharge and at the end of the study. Analyses were made with subjective [Sherwood et al., 1996] and quantitative methods [Sherwood et al., 1997].

RESULTS

This presentation will detail a case study following recovery over a six month period. It was possible to record the presence of low amplitude motor activity in response to BMCA maneuvers and to monitor as that activity formed into patterns that were eventually clinically recognizable. The various elements of the BMCA protocol showed different rates of progression. The patient also developed spasticity after several months, requiring pharmacological management whose effectiveness was also documented with these methods.

DISCUSSION

Acute spinal cord injury is often characterized by a period of greatly reduced excitability or responsiveness to external stimuli. Possible reasons for this reduced responsiveness would include long tract demyelination, loss of charge gradient across axon membranes from the leakage of intracellular fluids into the interstitial space, damage to peripheral nerves, or other mechanisms. As these processes resolve, the re-emergence of motor output will certainly appear at low amplitudes and be quite difficult to recognize by palpation as used in current clinical examinations such as the AIS (Maynard et al., 1997). We chose the BMCA protocol to provide maximum sensitivity to sub-clinical motor event, because this method could make us monitor even single motor unit firings in multiple muscle groups simultaneously.

With the BMCA, it is possible to monitor motor activity too slight to be picked up by observation, interview, or palpation of only a few muscles as typically done in a clinical examination. With serial recording, it was possible to follow changes over time in motor activity. These findings show it is possible using the BMCA to demonstrate the re-emergence of motor activity as it develops toward a functional or non-functional form of motor control. With a larger sample size would come the possibility of recognizing some of the rules of motor return that occur in the early phase of SCI. This method offers the possibility of classifying patients who experience a spinal cord injury on the basis of types of motor patterns seen.

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EFFECTS OF EMG-TRIGGERED FEEDBACK ON ARM-HAND FUNCTION IN SUBACUTE STROKE PATIENTS

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INTRODUCTION

A controllable arm-hand function (AHF) is essential in the execution of activities of daily life. In literature several techniques have been described which might influence and enhance motor recovery. Conventional EMG feedback featuring audiovisual feedback has proven not to be effective in the long term in the treatment of arm paresis in stroke patients. The absence of any generalisation or remaining of effects may be explained by the fact that only peripheral muscles are stimulated without influencing cortical and subcortical areas representing sensorimotor function. More recently, a technique called EMG-triggered feedback has been reported to positively affect motor performance in stroke patients in the chronic phase, i.e. after 1 year post-stroke (1). Also Kraft et al. (3) investigated the effects of EMG-triggered feedback on restoration of AHF, relative to several other therapies. However, both studies have (major) methodological flaws. In view of these shortcomings it was decided to set up a randomised, placebo-controlled, single-blind, longitudinal clinical study involving stroke patients treated in an earlier, acute and sub-acute phase post-stroke. The purpose of this study was to identify possible effects of EMG-triggered feedback training of the wrist extensor muscles using the Danmeter Automove AM800 (Danmeter A/S, Odense, Denmark) on restoration of AHF in post-stroke patients, in contrast to effects of conventional electrostimulation. The present study is a multi-centre study involving 27 acute/sub-acute stroke patients, referred to either the stroke unit of the Hoensbroeck Rehabilitation Centre in Hoensbroek, the Netherlands or the MS and Rehabilitation Centre in Overpelt, Belgium.

METHODS

Approximately 300 patients diagnosed with a stroke entering one of the participating centres were considered for this study. Inclusion criteria were a) central paresis of the arm/hand after a first ever stroke, b) post stroke period > 3 weeks, c) active wrist muscle strength between grade 2 and grade 3 on the MRC scale, d) fair cognitive level (MMSE score > 23), e) no additional severe rheumatologic, neurologic or orthopaedic problems prior to the stroke, f) no pacemaker or severe cardiopulmonary complications, g) no history of epilepsy. Eventually, 27 stroke patients who met all inclusion criteria participated in this study. They were randomly divided over 2 groups. On average, subjects who had a first-ever uncomplicated ischemic stroke based on occlusion of one of the cerebral arteries, entered the study 54.8 days (sem=7.6) after stroke onset. One group received EMG-triggered feedback therapy, using the Automove AM800, additional to the 'treatment as usual' in the rehabilitation centre (=experimental group). The second group received 'conventional electrostimulation', using the Automove AM800 without EMG-triggered feedback, additional to the 'treatment as usual' in the rehabilitation centre (reference group).

Apparatus and movement task

For the EMG-triggered feedback and conventional electrostimulation the Danmeter Automove AM800 (Danmeter A/S, Odense, Denmark) was used, in combination with Biostim CF5050 pre-wired, reusable surface electrodes (Biomedical Life Systems, Vista, CA) with a contact surface of approx. 50 x 50 mm. Optimal electrode location, i.e. such a location that wrist extensor muscles contracted optimally during application of either stimulation technique, was established in a pilot study prior to this study and was checked for each patient participating in the study. EMG/stimulation electrodes were positioned over the wrist extensor muscle group of the paretic

side. The reference electrode was positioned at an electrically neutral position near the elbow joint. Each subject sat in front of a table. Both forearms were placed on the tabletop.

Training protocol

All therapeutic interventions were carried out 5 days a week during a period of 12 weeks for 30 minutes per day. Each treatment session was additionally to the usual rehabilitation treatment for stroke patients. The apparatus picked up the wrist extensor EMG signal which was reinforced and was subsequently used as a stimulation signal. The latter led to an (assisted) contraction of the wrist extensors of the paretic arm and a clear dorsiflexion of the wrist over the total range of wrist extension motion, starting from the neutral position.

Measurements

In order to evaluate AHF status of each participant both the Brunnstrom-Fugl-Meier (FM) test (2) and the Action Research Arm (ARA) test were used (2). Measurements took place upon entry of the participant in this study (=T1), immediately following the cessation of the training phase (=T2) (i.e. three months after T1) and 12 months after T1 (=T3). All ARA and FM tests were administered by one of three experienced occupational therapists of the SRL who were trained in using both tests. Statistical analyses, performed using SPSS (SPSS Inc, Chicago, IL) included Wilcoxon signed ranks tests and Mann-Whitney U-tests.

RESULTS

Twenty seven patients entered the study, i.e. 14 in the experimental group (designated to receive EMG triggered feedback training) and 13 in the reference group (designated to receive conventional electrostimulation). AHF in both groups significantly improved during therapy, i.e. between T1 and T2, both on the ARA and the FM test ($p<0.001$). This improvement persisted and increased after one year follow up. AHF improvement did not differ significantly between groups, neither regarding the ARA test results nor regarding the FM test results. This was found for the results after 3 months and at one year follow up period. In addition, no differences on the ARA and FM sub-scales were found at 3 and 12 months follow up.

DISCUSSION/CONCLUSION

In this study it was shown that AHF performance improved markedly after several weeks of therapy and continued to improve even after 1 year, but differential effects between groups did not attain significance level. An explanation could be that the stroke patients performed a very localised movement (i.e. wrist dorsiflexion) void of an added functionality, thus possibly making movement imagery more difficult for participants. On the basis of our findings and the very recently reported body of evidence regarding effects of movement imagery on augmentation of task performance it is hypothesised that movement imagery applied in a training protocol involving functional AHF tasks will lead to a significant and long lasting improvement in AHF in stroke patients. Our future research will focus on this theme.

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EFFICIENCY OF BOTULINUMTOXIN IN THERAPY OF CHILDREN WITH OBSTETRIC PLEXUS LESION

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INTRODUCTION

Patients with a lesion of the nervus plexus brachialis frequently suffer from serious impairments in upper extremity movement. An example of this, is the restricted range of motion observed during flexion-extension of the elbow joint. One possible cause for incorrect performance of this movement may be the coactivation of m. biceps brachii and its antagonist m. triceps brachii. The nerve poison Botulinumtoxin, which paralyses the muscle for a certain period of time, has been used successfully in the treatment of spastic gait patients, who presented with coactivation in certain leg muscle groups. The question therefore arose, as to whether patients with plexus lesion could also be successfully treated through the use of Botulinumtoxin. To estimate its efficiency, a measurement procedure has been developed which permits a quantitative evaluation of the impact of Botulinumtoxin on both range and quality of motion.

METHODS

At the Helmholtz-Institute, a procedure has been developed which synchronizes bipolar surface electromyography and 3D motion capture in the measurement of upper extremity movement. As part of the measurement procedure, the patients were asked to perform a flexion-extension movement of the elbow joint. EMG data of m. biceps brachii and m. triceps brachii were recorded according to the SENIAM recommendations. The EMG data obtained was subsequently rectified and enveloped. Crosstalk between the two muscles could be identified by comparing the healthy and affected sides. If no crosstalk occurred during measurement of EMG data for the healthy side, no incidence of crosstalk for the affected side was supposed. The 3D motion data were collected using the video-based Vicon 370 motion analysis system. By applying the motion data to a kinematic model of the arm [Schmidt, 1999], relevant joint angles were calculated. All patients were in the age range 2 to 6 years and suffered from a lesion of the n. plexus brachialis. 100 units of Botulinumtoxin were injected into the m. triceps at the transition from proximal to middle third. Measurements were performed prior to the injection of Botulinumtoxin and afterwards, at intervals of 6 weeks, 3 months and 6 months. Additionally the patients received physiotherapy over these time periods.

RESULTS / DISCUSSION

All patients' EMGs showed clear coactivation of m. biceps brachii and m. triceps brachii during measurement before the injection of Botulinumtoxin. 11 of the 15 patients benefited from the application of Botulinumtoxin. They showed a promising reaction with respect to EMG recordings and 3D motion data, which is examplarily illustrated below (Fig. 1) for one patient: After 6 weeks the triceps showed almost no muscular activity. The biceps had lost its previously clear activation pattern to some extend and flexion-extension movement in the elbow was not improved significantly. After 3 month however, the triceps still did only show very little activity, whereas the biceps got back to its clear activation pattern. The elbow flexion was performed with only slight improvement. After 6 month the picture was still the

same in most of the cases, although some patients already showed signs of alternating activation of biceps and triceps. Flexion-Extension movement in most cases was performed much more fluently than prior to the application of Botulinumtoxin.

CONCLUSION

The analysis of the 3D motion data showed that the application of Botulinumtoxin together with the physiotherapy significantly reduced functional impairments during elbow flexion for the majority of patients. Within a period of 3-6 months after treatment with Botulinumtoxin an increase in range of motion was recognizable. Additionally, EMG measurements indicated that the paralyzing effect of Botulinumtoxin had ceased. Consequently, a temporary paralysis of the antagonistic muscle seems to allow time for a permanent muscular improvement in the agonist. This results in the overall enhancement of muscle coordination and range of movement for most patients with a plexus lesion. Additionally, the results show that by the method used, an objective analysis of upper extremity movement becomes possible. This methodology includes both – the detection of motion performance and the analysis of the underlying muscular coordination pattern. Moreover it can be easily expanded to other pathologies of upper extremity movement.

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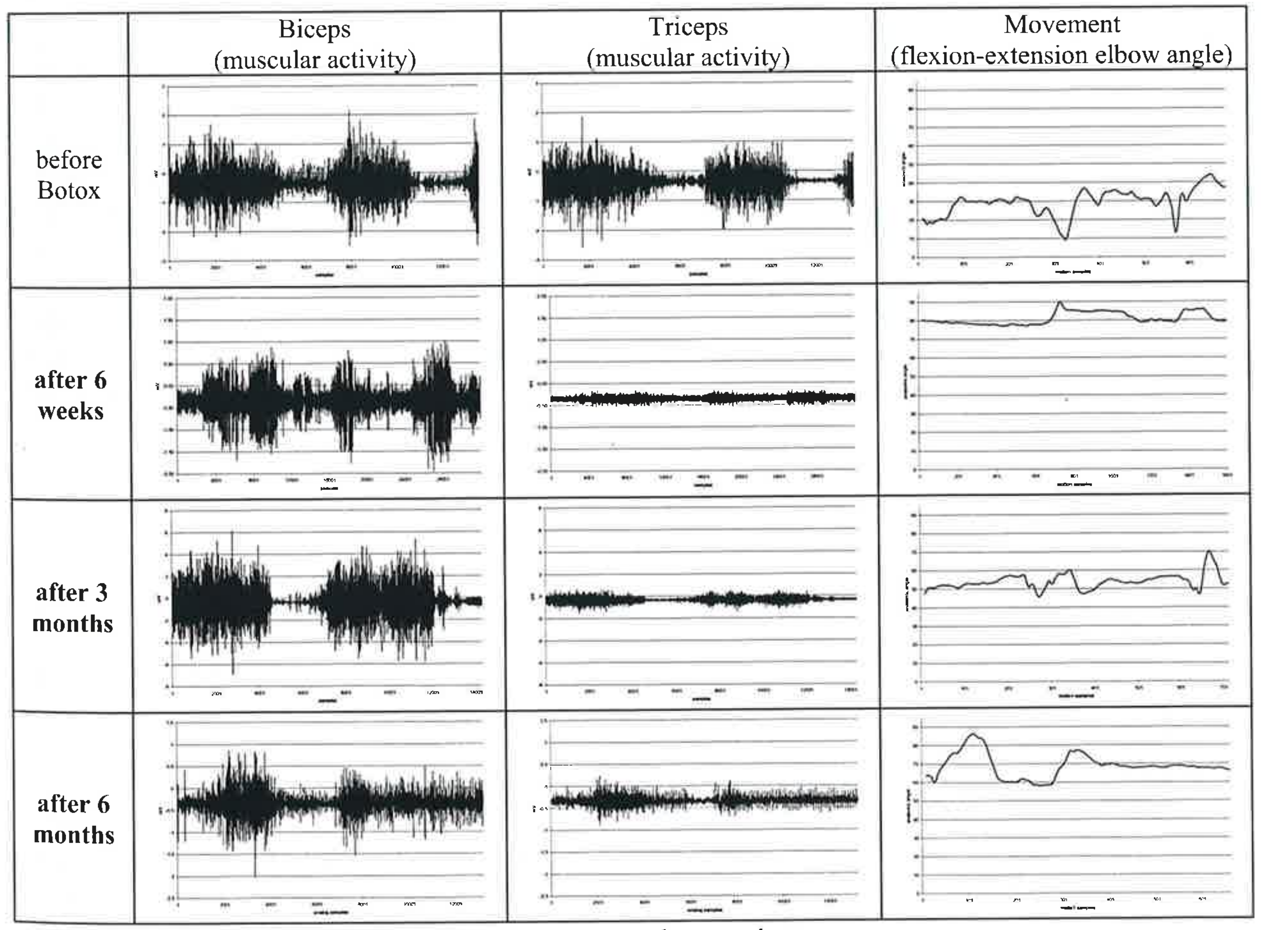

Fig. 1: Reaction on Botolinumtoxin for an examplary patient.

ELECTROMYOGRAPHIC STUDY OF PATIENTS WITH MASTIGATORY MUSCLES DISORDERS, PHYSIOTHERAPY TREATMENT (MASSAGE)

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INTRODUCTION

In 1990, the American of Academy Orofacial Pain (AAOP) it published the classification of TMD. Thus, according to AAPO, TMD can be classified etiológicamente as artrôgenica, miogênica and mixed. The etiologia of TMD is as varied as its sintomatologia, because countless factors can affect the dynamic balance of the components of the system mastigatório.

According to Okeson (1992) and Steenks & Wijer (1996), the factor more common etiológico in the DTM miogênica is the muscular hiperatividade. This hiperatividade, for its time, can contribute to the internal desarranjos of TMJ. The intensity and the presence of those signs and symptoms can vary of an individual for the another. The muscle hyperactivity can be a significant characteristic in subjects with Temporomandibular Disorders, which can be associated with pain, muscle fatigue. According to Bérzin (1999), the electric activity of the muscles mastigatory in individuals carriers of TMD, associated the pain miofacial was verified in 88% of the cases studied by the author, being associated to the hyperactivity of an isolated muscle or in different types of combinations.

Several types of treatments are proposed for the muscles dysfunctions. Ferreira(2001), Abekura et al. (1995) among other, they use the splint as therapeutic resource for the musles dysfunctions originating from of the bruxism, with the purpose of relaxing the musculature mastigatory, even so just the first author, it really uses the splint with this objectify, not concluding, however, the treatment. The second author just recommends the use of the splint with the purpose of reception of the sign EMG, not contemplating the objective proposed in its research. This last situation is a constant in most of the works (as for example Abekura et al., 1995) that use the splint as a study object during the collection of the sign EMG and don't treatment.

The aim of this study was to compare the behavior of the the electromyographic activity with major chewing muscles (anterior temporalis, and masseter muscles) by using Parafilm material (Biasotto, 2000) through analysis of electromyographic signals before and after massage.

METHODS

Sixty female young adult subjects, Graduation students of the Course of Physiotherapy of University of Mogi das Cruzes, aged from 17 to 28 years, they being, twenty with normal occlusion and no history of temporomandibular disorder, fourty , with signs and/or symptoms of TMD from parafunctional habits (excessively hard chewing or repeated forced mandibular opening and bruxism), which being twenty submitted to the physiotherapy treatment (massage) and the other twenty were't submitted this treatment. For the study of the masticatory activity the electromyographic signals were processed through rectification, linear envelope, so that the coefficient of variation obtained from the procedure was comparatively analyzed. For the study of the mastigatory active the electromyographic signals were stored and analyzed for RMS values (Root Mean Square) in the isotonic contraction phases of each muscle. In according to the ethical principles, the purposes of this study were carefully explained to the volunteers, who signed a formal consent.The recording of the electric activity

was carried out on the temporalis (anterior portion) and masseter muscles, bilaterally, through EMG exam.

RESULTS

The results of this study indicated that the massage showed decrease hyperactivity and the best relax and that the 87% of the subjects showed improvement signification of the symptoms and sings. In a general way, this treatment is indicated for the accomplishment of the other treatment and the electromyographic exams are very important to analyze of the muscles activity.

DISCUSSION

At first it is not possible to do any comparison with the literature on the effect of the massage in patients that present bruxism, because there is not in the literature a protocol with this treatment. The great majority of the researched works that they make use of a treatment, they use as treatment, for example, the splint (Ferreira, 2001; Abekura et al. ,1995) For this reason it was researched in volunteers with bruxism the effect of the physitherapy treatment, making use of the massage technique, and the possible variations of the sign eletromiográfico during the bilateral isotonic mastication. The results demonstrate that the massoterapia was effective to decrease the electric activity of the studied muscles, that were current hiperativos of the bruxism. Although it has not been found in the literature studies about the action of the massage in musles with hyperactivity, as in certain DTMs, the results agree with every classic literature on massage (Beard 1952; Bell, 1964) and, that sustains the theory that promotes an increase of the sanguine and lymphatic circulation favoring a muscles relax.

CONCLUSION

The results of the present work, inside of the used experimental conditions they allow to end that: - Had an improvement of the volunteers' symptons; -The reduction of the activity eletromyografic of the studied muscles was significant in the treated group. - Had effectiveness in the physiotherapy treatment through the tool massage on the Disorder Temporomandibular miogenic; - The 15 sessions contemplated the proposed objectives

Therefore, this way, the proposed therapeutic protocol, it can come to add the other treatments that present the same objectives, where one will complement the other with its owed importance.

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THE RELIABILITY OF MVE AND RVE VALUES USED FOR THE NORMALIZATION OF UPPER TRAPEZIUS MUSCLE ACTIVITY

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INTRODUCTION

Studies of ergonomic risk exposures, functional activities and therapeutic exercise protocols require that the myoelectric signal amplitude be normalized with respect to the signal amplitude recorded during some reference activity. The best combination of electrode site and test position for the generation maximal voluntary activations (MVEs) of the upper trapezius muscle has been investigated by several researchers[1], and is generally accepted to be a point approximately 2cm lateral to the midpoint along a line drawn between the C7 spinous process and the lateral acromion. There has been sparse information presented regarding the test-retest reliability of this site, especially for recordings of the submaximal contractions typically used in ergonomic studies[1,2]. The purpose of this study was to determine the reliability of maximal voluntary activation (MVE) and reference voluntary activation (RVE) contractions commonly used for normalization purposes in upper trapezius muscle investigations using both surface and fine wire electromyography and several recording sites.

METHODS

Four upper trapezius electrode sites were studied simultaneously using both surface and fine-wire electrodes on six female volunteers. Participants performed (in random order) three repetitions of four maximal isometric contraction efforts (level 3) and three repetitions of two commonly used RVEs: isometric unweighted arm holding at 30°(level 1) and 90° (level 2) flexion in the scapular plane, respectively. All data files were corrected for subject and instrumentation bias, and in all cases, root mean square (RMS) amplitudes were computed over moving windows of 250 ms. The representative value of the MVEs were determined as the maximum RMS values (in mV) recorded during each contraction. The values for the RVEs were computed as the mean RMS value computed over the duration of each 4-second contraction. For both the MVEs and RVEs, between-trial reliability was assessed in two ways – using an Intra-Class Correlation Coefficient (ICC) and using the Coefficient of Variation (CV). For each measure, a three-way repeated measures ANOVA was used to determine if significant differences in signal reliability existed between electrode type (surface vs. wire), recording site (1 through 4), and contraction level (1, 2, or 3).

RESULTS

The ICC values for all electrode types, electrode sites, and contraction types were not significantly different for either the MVEs or the RVEs (ranging from 0.808 to 0.999). The CV results for MVEs and RVEs, are presented below.

i. MVEs: The electrode type data (surface vs. wire) were analyzed separately (due to unequal variances). The RMS amplitudes revealed no significant differences in MVE between any of the electrode sites or movements for neither surface nor wire electrodes, although there was a trend ($p=0.058$) towards movement 1 (pure abduction) resulting in higher amplitudes. The CV values demonstrated no interactions between the contraction positions and electrode sites. There were no contraction type main effects for either electrode type. For the surface electrodes, there was a significant electrode site main effect ($p<0.0005$), where site 1 was less reliable (CV = 77.0%) than all other sites, and site 3 was better than all other sites (CV = 19.7%). The wire electrodes also had a significant electrode site main effect ($p=0.002$), where site 2 was more reliable than the other sites (CV = 64.7%), albeit quite poor.

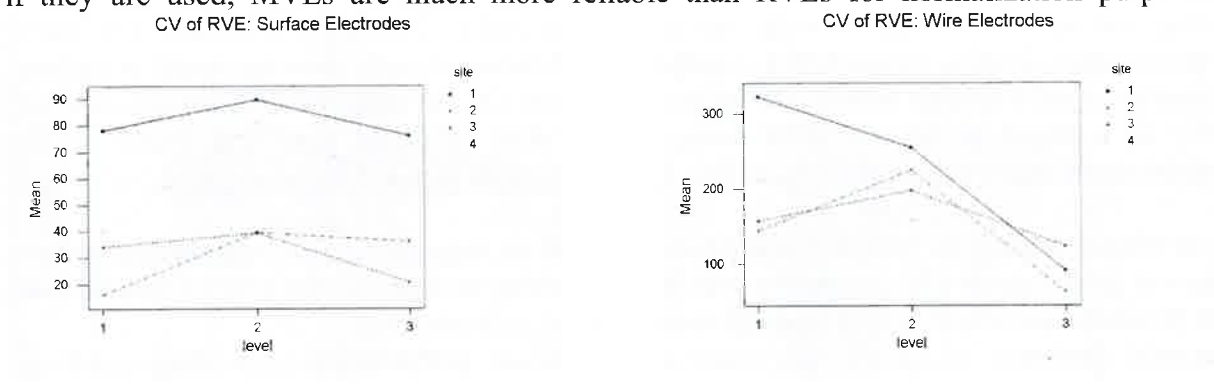
ii. RVEs: For each subject, the contraction that produced the largest MVE recorded at each electrode site was used as the level 3 RVE. ANOVA revealed a significant electrode type by contraction level interaction ($p<0.013$). Electrode type accounted for approximately 2/3 of the variance in the model. Bonferroni post-hoc analyses (using an adjusted α to account for multiple comparisons) revealed that, fixing type to surface electrodes, site 1(CV =77.0%) was significantly less reliable than sites 2 (CV = 31.2%) and 3 (CV = 19.7%), but not 4 (54.6%). Sites 2 and 3 were not significantly different. FW2 was more reliable than FW3 and FW4, but all sites had poor reliability (CV's ranging from 64.7% to 119%). For the surface electrodes, contraction level 2 (90°) was generally more reliable than contraction 1 (30°). The wire electrode data revealed that contraction 3 (MVE) was more reliable than contraction 1 (30°).

DISCUSSION

Test-retest reliability of surface electrodes has been reported at a CV of 6 to 14 % for MVEs of the upper trapezius muscle[2,3]. Mathiassen et al. claim that CV values for RVEs are comparable to this under carefully controlled circumstances in the laboratory[1]. This claim is not supported by the current study. More consistent with the present findings, Attebrant et al. showed that the test-retest repeatability of a ramp normalization was only 46% at 5% MVC[4].

CONCLUSIONS

ICC is not a sensitive measure in determining reliability of the RMS of myoelectric signals. Although Channel 1 tends to generate higher amplitude MVEs and is the most commonly used site in the literature, MVE and RVE values at this site have poor reliability as compared to other recording sites on the upper trapezius muscle. Fine wire recordings are very difficult to standardize despite careful electrode placement, arm positioning and contraction level, but, if they are used, MVEs are much more reliable than RVEs for normalization purposes.



(a) (b)

Figure 1: Coefficient of Variation by electrode site and contraction level.Surface (a) and wire (b) electrode recordings plotted against contraction level at each electrode site. Site 1 is the standard electrode position located at 2 cm lateral to the midpoint between C7 spinous process and the lataral acromion, site 2 is 2 cm supero-anterior to that site 1, site 3 is 2 cm inferior to site 1, and site 4 is 4 cm medial to site 1. The contraction levels are described in the text.

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DO TRAPEZIUS MUSCLE MOTOR UNITS PREFER MOUSE OR PEN?

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INTRODUCTION

It is poorly understood why computer users frequently develop neck pain. Therefore, it may be of interest to register the physiological burden on single motor units (MUs) in the trapezius muscle during computer work, and to assess eventual differences between using mouse or pen as pointing devices.

METHODS

13 subjects performed a randomised set of six standardised computer tasks (9 min. each) with typical interactions (single and double clicking, drag-and-drop) using a mouse vs. a pen. The tasks were set to achieve semi-natural behaviour. Muscle load was assessed using root-mean-squared surface EMG (sEMG) and intramuscular EMG (iEMG), decomposed with a fully automated programme in the trapezius muscle, pars descendens. Finally the signals were triggered on the mouse/pen clicks. Cusum (cumulative sum) method was used to check for firing rate changes.

RESULTS

In the measured part of the trapezius muscle the level of sEMG, the total number of motor unit action potentials (MUAPs) and the number of detected MUs did not significantly depend nor on the mode of interacting with the computer nor on the pointing device (tab 1). On average the activity was significantly increased compared to the resting condition. The level of activity considerably varied between subjects, with three subjects showing nearly complete relaxation of the trapezius muscle during measurements. On the other hand, nearly continuous firing of a single or several MUs during the 9 min. recording time was found in 11 measurements that were obtained from four subjects (5 with pen, 6 with mouse use).

Time triggered analysis revealed significantly higher m. trapezius sEMG during the down-phase of mouse single clicking in two and double clicking in four subjects (fig 1). Similar but less pronounced event triggered findings were attained with pen use.
Standard deviation of sEMG was markedly increased immediately after drag-and-drop activity in 6 subjects using the mouse and in 7 subjects using the pen. In one specific subject, time triggering of MU firing revealed a decrease in firing rate before the single clicks and a significant increase after the clicks in most of detected MU's. Significant relationships were found in other subjects, with both similar and different temporal behaviour. Firing rate increase could often be explained by a raise of doublet rate.

DISCUSSION

Although the experiments were done at a workplace with optimal ergonomic dimensions, trapezius muscle activity (number of detected MUAPS) on average was ten until 15 times higher than during the resting condition and in four subjects continuously active MUs were demonstrated. Thus the computer task induced both a possibly minor increase of muscle load in the majority of subjects and a potentially important strain on continuously active MU in a few subjects.
We did not observe a regular significant effect on trapezius muscle activity neither by the kind of device used nor by the type of the prearranged task. Three out of the 13 subjects were

capable to constantly relax the trapezius muscle. We conclude that such a different behaviour of subjects demands for a careful consideration of interactions between personal and work-related factors when studying workload. Indeed in some test subjects we could demonstrate a intermittent component in the activation of the trapezius muscle that was significantly correlated with the clicking movement. It was demonstrated both on the muscle (sEMG) and MU (single MU firing rate) level.

Table 1. Average activity in the right trapezius muscle of 13 subjects during resting and various computer tasks (mean ± standard deviation)

Task	Device	Surface EMG [% max. voluntary activity] 5th percentile	50th percentile	95th percentile	MUAPs number	MUs number
Resting		4 ± 5	6 ± 7	10 ± 10	782 ± 1418	2 ± 4
Single Click	Mouse	13 ± 23	25 ± 40	53 ± 71	10656 ± 11994	9 ± 7
Double Click	Mouse	14 ± 18	28 ± 34	51 ± 55	10067 ± 11007	9 ± 7
Single Click	Pen	10 ± 10	19 ± 16	34 ± 25	9701 ± 14398	8 ± 11
Double Click	Pen	9 ± 8	17 ± 13	37 ± 29	8834 ± 11749	10 ± 11
Drag & Drop	Mouse	16 ± 21	29 ± 33	57 ± 60	13119 ± 16071	12 ± 8
Drag & Drop	Pen	11 ± 12	25 ± 29	54 ± 55	12467 ± 19713	12 ± 10

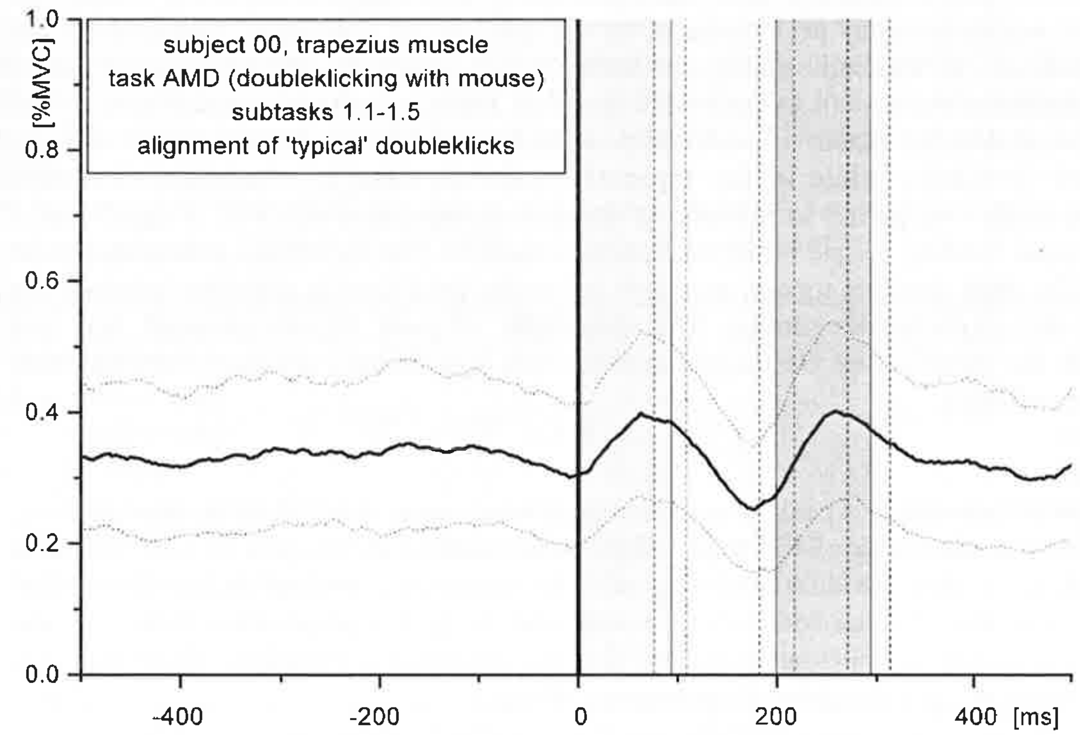

Figure 1. Example from a single subject. The time-triggered activity of the right trapezius muscle during double-clicking with mouse is clearly related to the finger movements.

EFFECTS OF UNEXPECTED LATERAL MASS PLACEMENT ON TRUNK MOVEMENT AND TRUNK LOADING IN LIFTING

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INTRODUCTION

Epidemiological studies suggest that sudden, unexpected loading leads to low back pain (1, 2). Also, asymmetrical loading is considered to be harmful to the spine (3, 4). It can be anticipated that unexpected asymmetrical loading will increase the risk of injury even more. The aim of the present study was to investigate whether unexpected asymmetric loading increases the loading of the spine compared to expected asymmetric loading in lifting.

METHODS

Ten subjects lifted in an upright posture a crate of 2 kg in which a mass of 10 kg was placed laterally to the left side. The subjects either knew the location of the mass, or it had been laterally replaced without them knowing so. Kinematics and force data were recorded throughout the experiment. Trunk muscle activity was measured and muscles were grouped according to their mechanical function. The compression force, net torque and trunk lateral flexion were calculated. The effects of expectation were tested with paired t-tests.

RESULTS

The peak total moment and the extension moment were significantly increased with approximately 10% compared to the expected condition. In contrast, the loading of the low back in terms of compression force did not appear to be increased when the mass in the crate was unexpectedly placed laterally. This was explained by the co-contraction level, which was lower at the instant of total peak moment in the unexpected condition compared to the expected condition, as was indicated by the lower ratio between the summed muscle forces and total moment at the instant of peak total moment. Further analyses revealed that in the unexpected condition the instant of peak compression force occurred after the instant of peak total moment ($p=0.004$), while in the expected conditions these two instances coincided ($p=0.2$). The trunk was pulled to the side of the perturbation, as a result of a significantly decreased lateral bending torque in the unexpected condition. No significant differences were observed in the peak absolute torsion moments and in the peak muscle activities between the unexpected and expected conditions. The magnitude of peak torsion moment was not different, but the direction of the torsion moment was significantly different between both conditions ($p = 0.025$).

DISCUSSION

In the unexpected condition a peak in co-contraction was seen as reaction to the perturbation, probably aimed at stiffening of the trunk. High co-contraction in reaction to a perturbation was also found in other studies analyzing sudden asymmetric loading (5, 6). Our initial expectation was that this co-contraction would lead to high compression forces in the unexpected condition. It appeared however, that the resulting compression force was not higher than that produced in expected asymmetric lifting.

Mechanical stability of the spine must be assured during lifting objects, since loss of stability will impose an injury risk (7). In the unexpected condition, the moment direction produced is not adequate for the imposed perturbation. In addition, the stiffness of the trunk at the instant of peak total moment appears to be lower compared to the stiffness in the expected condition, indicated by a lower value of summed muscle forces. Both the inadequate direction of the moment produced, and the lower stiffness result in a decreased stability of the spine compared to the expected condition. This stability appears to be too low to sufficiently resist the perturbation due to the laterally placed mass, as is seen by a lateral trunk movement of 2 degrees in the unexpected condition. Although the total trunk movement is small, the insufficient muscular stabilization may lead to excessive rotations at the segmental level. Such rotations in combination with a considerable compression force may cause injury to the ligaments or intervertebral disc. Therefore, although the peak loading of the trunk in terms of compression forces does not appear to be increased, the decreased stability due to the unexpected lateral loading may increase the risk of injury.

CONCLUSION

Unexpected lateral placement of the mass does not substantially increase the peak loading on the low back. However, the stability of the spine is insufficient to withstand the perturbation, which may entail an injury risk.

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	Expected	Unexpected	P-value
Peak compression force (N)	1933 (142)	1941 (142)	0.88
Peak moment (Nm)			
Lateral bending	23.7 (2.3)	17.6 (1.7)	0.005
Extension	59.3 (2.6)	66.6 (3.2)	0.01
Torsion (absolute value)	5.2 (0.5)	5.3 (1.3)	0.90
Total	62.6 (2.6)	68.8 (3.2)	0.01
Peak lumbar lateral angle (deg)	-1.5 (0.3)	0.7 (0.8)	0.01
Ratio muscle force - Mtot at instant peak Mtot (m^{-1})	34.0 (3.4)	27.2 (3.7)	0.05

Table 1: Average values (standard error of mean in parentheses) of the peak compression force, lateral bending moment, extension moment, absolute torsion moment, total moment (=Mtot), the ratio between the summed muscle forces and peak total moment at the instant of peak total moment, and lumbar lateral angle.

KINETIC AND KINEMATIC EVALUATION OF POSTURAL LIMITS TEST: THE EFFECT OF AGE

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INTRODUTION

The postural limits test has been used as an indicator of balance control during functional tasks. For example, smaller postural limits during reaching and leaning tasks in the elderly have been interpreted as poorer balance control. The position of the center of Pressure (COP) and center of mass (COM) has a profound effect on maintenance of balance, suggesting that it is a key factor for balance control in many activities of daily living. The aim of this study was to investigate the ability and the strategies in which the elderly regulate the movement of COP and COM in relation to base of support (BOS) during voluntary postural limits test, the boundaries beyond which the position of body's center of gravity will exceed the base of support provided by the feet.

METHOD

Hundred-seventeen volunteers, seventy-seven 75-84 years old (mean 78.2±2.1), and forty 20-34 years old (mean 25.2±3.2) who did not have neurological or psychiatric disorders or signs of serious cognitive dysfunction. The stability limits were measured during standing on single force plate: maximum voluntary excursions of center of pressure while leaning forward, backward, left, and right in upright standing, to the outer limits of the stability margins, Each extreme position was held for 1-2 seconds. The test was made in two trials, once standing with wide BOS (the feet placed apart), and narrow BOS (the feet placed together).

Surface electromyography was recorded from the Tibialis anterior, Soleus, Rectus-femoris and Biceps-femoris muscles. The test was videotaped from side view to evaluate strategies used by the subjects during stability limits test. Static two-point discrimination test at the plantar surface of 1st toe was made to evaluate the innervation density of the slowly adapting fiber.

DATA ANALYSIS

The COP based parameters of balance performance was expressed as the length of center of pressure (COP) path sway in antero-posterior (AP sway) and medio-lateral directions (ML sway) in cm. The electromyographic parameters of balance performance were expressed as percentage from MVIC in young and old subjects. Joint angles (in degrees) of each of the major joints were measured during the stability limits test. Statistical evaluation was carried out using two-tailed students t-test the results presented as mean±SEM taking a two-tailed probability of 5 percent as the level of significance.

RESULTS

COP based measurements

Forward-backward incline of 20-34 years old subjects was 16.8 cm during standing in wide base and 16.1 cm in narrow base, significantly higher than 75-84 years old subjects (9.5 cm and 8.3 cm respectively). Lateral postural limits in the young subjects were 21.1 cm in wide

base and 11.2 cm in narrow base, significantly higher then 75-84 years old subjects (12.5 cm, and 7.3 cm respectively).

Postural limit/Feet Length Ratio (PLFLR) is the forward-backward incline value during stability limits test expressed as a percent of foot length. The PLFLR in the young subjects was 70% in wide BOS and 67% in narrow BOS, significantly more then in elderly individuals (39.5% and 34.5% respectively).

KINEMATIC MEASUREMENTS

The decline in stability border in old age is accompanied by changes in the strategy of postural control. 20-34 years old subjects used their ankle joint, 8^0 of Dorsiflexion during forward leaning. No Dorsiflexion was performed by 75-84 years elderly individuals during forward incline. Instead, elderly individuals used hip strategy during performance of forward-backward incline, 32^0 of flexion at the hip joint compared with only 2.6^0 flexion at the Hip joint in young subjects during stability limits test.

KINETIC MEASUREMENTS

The shift from ankle to hip strategy can be viewed also by the EMG finding, while young subjects showed low levels of recruitment in hip muscles 9.5-6.4% of MVIC in RF muscle during backward incline and 29.8-42.9% of MVIC in BF muscle during forward incline, elderly individuals used significantly more of those muscles 37.4-34% for RF muscle and 58.4-65.4% for BF muscle.

Elderly individuals also used 65.8%-73.8% of Sol muscle MVIC during forward incline and 75-76% of TA muscle MVIC during backward incline significantly higher then young individuals that used only 34.4-35.2% of Sol muscle MVIC and 25.0-26.9% of TA muscle MVIC.

TWO-POINT DISCRIMINATION

There were significant differences in Two-point discrimination (TPD) at the plantar surface of the great toe, while TPD of 20-34 years old subjects was 8.1 mm, only 13.1 mm, was found in old group subjects.

DISCUSSION AND CONCLUSIONS

It can be argued whether the decrease in sensation as seen from TPD measures is causing the changes in strategies or decrease in muscle strength around the Ankle joint. Elderly individuals are less likely to sense their toes and then they will choose a more secure strategy, performing stability limits test in hip strategy. Hip strategy enables the elderly individuals to perform forward reaching tasks with minimal movement of their center of mass to a place that they are less likely to sense their foot or due to luck of ankle muscles strength or both. We concluded that functional exercises such as close chain exercises in standing that stress the stability limits of the elderly individuals could strengthen ankle muscles in their functional position, teach the subjects how to recruit their motor units in more effective way and at the same time training the proprioception and cutaneus sensations. Future studies should test this hypothesis enhancing postural stability of elderly through functional training that consist exercises that imitate "real life" activities and effect on fear of falling.

FATIGUE OF POSTURAL MUSCLES HAS MINOR EFFECTS ON POSTURAL ACTIVITIES ASSOCIATED WITH VOLUNTARY WRIST MOVEMENTS

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INTRODUCTION

Voluntary movements are always preceded and accompanied by unconscious postural adjustments. Since muscular fatigue alters muscle contractile efficacy, proprioceptive informations, and cortical control (Belhadj-Saïf et al., 1996; Bigland-Ritchie et al., 1983; Gandevia, 1998) it is likely to affect movement and posture control. Nevertheless, how fatigue may alter postural control has been poorly documented.
The purpose of the present study was to examine whether fatigue induced by long-lasting submaximal isometric contraction induces changes in the coordination between segmental posture and movement.

METHODS

Seven healthy adults performed series of fifteen fast wrist flexions and extensions while being instructed to keep a given upper limb posture as constant as possible. These series of voluntary movements were performed before and after a 20-30% MVC isometric contraction of the elbow flexors, exerted as long time as possible. This task was stopped as soon as the exerted force declined. Two postural conditions were also examined: with or without the help of an elbow support.
Surface EMG from shoulder (Deltoïdeus anterior: DA), elbow (Biceps brachii: BB, Triceps brachii: TB) and wrist (Flexor carpi ulnaris: FCU, Extensor carpi radialis: ECR) prime-mover muscles were recorded. Hand, forearm and arm tangential accelerations and wrist and elbow angular displacements were simultaneously recorded. Onset time of wrist acceleration was used as time zero (t0) for all data processing. EMG analysis was the same for all muscles. The onset of EMG burst was visually defined as the first rectified EMG value above the background activity level. Integrated EMG (iEMG) was measured over two periods of time: 1) a 100 ms time period to measure the background activity corresponding to the static conditions of the initial posture and 2) from onset of EMG burst to the end of the wrist acceleration phase. Muscular fatigue was evidenced by a shift of the elbow and shoulder muscles EMG power density spectra towards low frequencies.
A repeated measure analysis of variance ($P \leq 0.05$) was used to compare the subjects' averages before and after LLFC for each support condition (with or without support). Paired t-tests or Wilcoxon signed rank tests (when the normality test failed) were performed to compare the onsets of elbow or shoulder movements and onsets of postural EMG bursts within each movement series ($P \leq 0.05$).

RESULTS

Subjects were able to keep the 30% elbow flexion force for 2-6mn depending on subjects and experiments. For all subjects, MPF from the 3 postural muscles (i.e. DA, BB, TB) decreased between the beginning and the end of each long lasting isometric elbow flexion, RMS values were generally unchanged or increased at the end of each session for all muscles.

Kinematics of wrist movements and corresponding activations of wrist prime-movers, as well as the background of postural muscle activation before wrist movement were not modified by "fatigue". There were only slight changes in the timing of postural muscle activations (fig. 1).

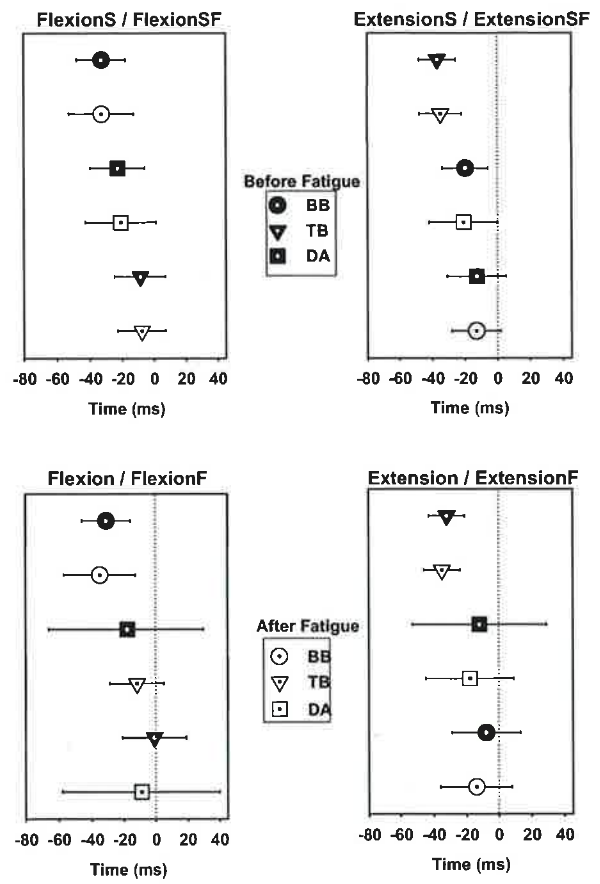

Figure 1

Onset latencies of the EMG bursts of Deltoideus anterior (DA), Biceps brachii (BB), Triceps brachii (TB) for wrist flexions and extensions with support (S) or without support before and after (F) fatigue.
Vertical dotted line indicates the onset of wrist movements (t0). Mean and standard deviation values were calculated over all trials for all subjects.

DISCUSSION AND CONCLUSION

These data indicate that postural fatigue induced by a 20-30% MVC isometric contraction has no effect on the kinematics of voluntary movement and requires no dramatic adaptation in postural control. This result suggests that most of the "fatigued" motor-units were not implied in the postural adjustments needed by the voluntary movement. It also suggests that the importance of postural muscle fatigue in occupational muscle disorders might not be over-estimated.

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THE EFFECT OF SCAPULAR TAPING ON SHOULDER IMPINGEMENT SYNDROME (SIS) PATIENTS

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INTRODUCTION

Shoulder Impingement Syndrome (SIS) is one of the most common pain disorders of the shoulder. It is defined as the pinching of soft tissue involving the supraspinatus tendon and the subacromial bursa. It can also include the tendon of the long head of the biceps muscle. Although SIS affects all age groups, it tends to afflict those over forty years old. A predisposition to the pathology can be found in individuals who engage in extensive overhead arm activity. The primary evidence for impingement is found when the arm is flexed, or abducted and internally rotated[1,2]. Some have proposed that scapulo-thoracic muscle imbalance is a factor that contributes to impingement. Little is known about the effect of scapular taping on muscle performance in SIS patients.

Objective: The aim of this pilot study was to assess the effect of scapulo-thoracic taping as a non-operative means of treating SIS.

METHOD

An orthopedic specialist referred 10 patients with identifiable shoulder pain disorders for biomechanical testing in double-blind fashion. Upon isokinetic testing, only five of the subjects were found symptomatic of SIS. The symptomatic subjects were then divided into four groups: Group A – Uninvolved shoulder of SIS patients, Group B - Involved shoulder of SIS patients before taping, Group C - Involved shoulder with true taping, Group D - Involved shoulder with false taping (control). The subjects were evaluated while seated in an isokinetic dynamometer. Then the subjects were requested to perform five isokinetic abduction-adduction repetitions in the scapular plane at 180 degrees/sec. Surface EMGs (sEMG) from the four shoulder girdle muscles: Deltoid (D), Infra-Spinatus (IF), Upper Trapezius (UT) and Lower Trapezius (LT). The following dependent variables were measured: 1) muscle strength – a comparison of the isokinetic abduction/adduction torque profiles in the scapular plane of the involved and uninvolved shoulders; 2) workload- expressed as a % of the average sEMG (AEMG) generated by the four muscles.

RESULTS

The involved shoulder showed overall weakness; however, the difference between the shoulders was greater for abduction than adduction. This muscle imbalance of weaker adductors than abductors was clearly expressed in our findings for the agonist/antagonist ratio. We also found muscle imbalance between two forced couples on the involved side of SIS patients (UT & LT at the scapulo-thoracic articulation and D & IF at the gleno-humeral joint).

The workload results showed reduced AEMG activity of the UT muscle in the involved shoulder of SIS patients, as compared with the uninvolved shoulder (Fig. 1). True taping resulted in a reversal of 60-90 degrees for each group of the activity levels of the UT muscle in the involved shoulder as opposed to those in the uninvolved shoulder. False taping showed the same activity levels of UT in both the involved and uninvolved shoulders of SIS patients.

The muscle imbalance in the gleno-humeral joint as manifested in the forced couple (D & IF) on the involved side of SIS patients were similar to the imbalance found in the S-T region. There was no change in the activity patterns of gleno-humeral muscles (D and IF) occurred before and after taping. The effect of ROM on muscle activity was minimal between groups C and D (those with true and false taping) for all muscles except UT, which showed meaningful improvement with true taping.

Conclusions: Scapulo-thoracic taping changed the activity patterns of muscles in the scapulo-thoracic area. There was enhancement of the activity of the UT muscle in the involved shoulder of SIS patients. Scapulo-thorcic taping did not change the activity patterns of muscles in the gleno-humeral region. It appears that scapulo-thoracic taping can reduce symptoms in SIS patients. Future investigations need to study the effect of long-lasting taping on pain, range of motion, strength, and EMG behavior in SIS patients.

sEMG, together with isokinetic testing, provides an objective assessment of shoulder muscle performance that can help the clinician/ researcher distinguish between various shoulder problems.

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	Involved		Uninvolved	
Abd/Add ratio (%)	91.7		98.4	
	Abd	Add	Abd	Add
Peak Torque (Nm)	50	54.5	55.2	56.1
Total Work (Nm)	1441.5	1597.6	1403.3	1639.7
Average Power (watts)	65.9	73.8	67.1	69.4

Table 1. Mean isokinetic dynamometer results of SIS patients before taping.

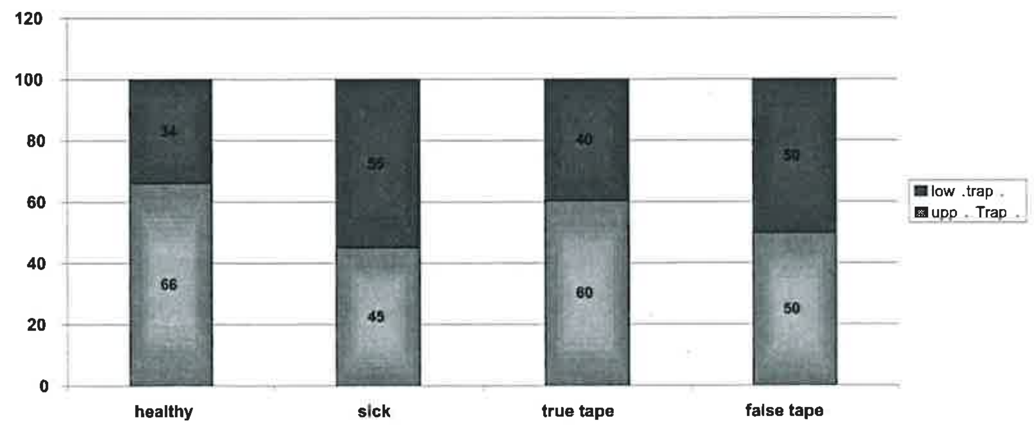

Fig. 1. AEMG ratios of UT and LT during abduction between the ROM

LINEAR AND NON-LINEAR ANALYSIS OF BICEPS BRACHII SEMG DURING RAMP ISOMETRIC CONTRACTIONS

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INTRODUCTION

The term "motor unit activation" refers to the combination of recruitment and discharge properties of motor units (MUs) within muscles (Sale 1987). The relative role of MU recruitment and increased firing rate as mechanisms for increasing force exertion has been debated since the early seventies (Clamann, 1970; Milner Brown et al, 1973). During incremental force isometric contractions, it has been shown that the median frequency (MDF) of the EMG power spectrum increased linearly with orderly recruitment (stimulation) of MUs until complete recruitment was obtained (Solomonow et al., 1990). The purpose of the present study was to investigate the dependency of sEMG time and frequency domain parameters on the speed of ramp contraction. In order to add further information on sEMG changes during ramp contractions, we also applied a non-linear tool, the recurrence quantification analysis (RQA), which already provided interesting information related to the ongoing manifestations of muscle fatigue as well as on MU synchronisation (Felici et al, 2001).

METHODS

Six subjects participated in this study. They performed isometric contractions with their right biceps brachii (BB) at an elbow angle of 90° and the hand fixed halfway between pronation and supination. After identification of the motor point, a four electrodes linear array (interelectrode distance: 10mm) was positioned over the longitudinal axis of the BB between the motor point and the distal tendon. After measuring the MVC, the sEMG was recorded in a single differential mode from BB muscle during linearly increasing isometric contractions from 0 to 100%MVC (ramp phase), and during the subsequent 10s of sustained MVC (constant phase). Three single differential sEMG signals were recorded (SD1-SD2-SD3). Two double differential sEMG signals (DD1-DD2) were then computed and used for average muscle conduction velocity (CV) estimate. For each subject 5 trials were accepted at each speed of contraction using acceptance criteria described elsewhere (Bernardi et al., 1999).

The SD2 RMS and MDF were computed over the ramp and constant force phase on subsequent epochs overlapped one another by half their length. The sEMG was segmented in epochs of different length as such that the correspondent force variation was equal to 2,5% MVC. Thus, window lengths were: 250ms for 20% MVC s^{-1}, 500ms for 10% MVC s^{-1} and 1000ms for 5% MVC s^{-1} ramps respectively. Epochs were windowed with a Hanning window and, when appropriate, zero padded up to 2048 points to obtain a frequency resolution of 1 Hz (Bernardi et al, 1999). All MDF and RMS data were normalised with respect to the maximum value obtained. A grand average was then computed over all RMS and MDF curves for each ramp speed.

FMUR was considered as the force value (%MVC) at which the highest MDF value was reached (Solomonow et al 1990). The dependency of FMUR on the speed of contraction was tested by means of a one-way analysis of variance applied on FMUR points obtained on all subjects at 5, 10 and 20%MVCs^{-1} at a level of confidence $p<0.05$ (Sachs, 1984). A linear regression was performed over the MDF values computed during the constant force phase, and the slope of the regression lines thus obtained was normalised to the intercept and used as an index of localised muscle fatigue. According to Felici et al (2001), muscle fibre action potential CV was estimated by the cross-correlation function technique between DD1 and DD2 using the same windowing as for sEMG analysis. Estimates of CV were accepted only when the sEMG cross-correlation function values were higher than 0.6. As far as non-linear analysis is concerned, the percent of determinism (%DET) was computed on all the sEMGs using the same segmentation as for linear analysis (Felici et al., 2001).

RESULTS AND DISCUSSION.

The main results are reported in Table. Analysis of variance showed that FMUR is dependent upon speed of contraction ($p<0.01$). CV modifications mirrored those reported for MDF.

Contraction Speed	**5%MVC s^{-1}**	**10 MVC s^{-1}**	**20 MVC s^{-1}**
FMUR (%MVC)	52.3 ± 7.3	58.63 ± 15.1	77.6 ± 15.1
Fatigue Index(% s^{-1})	-1.64 ± 0.1	-2.24 ± 0.5	-2.74±0.6

During the ramp phase, %DET reached a maximum preceding the FMUR and then levelled until the constant force phase started. Afterwards, %DET increased linearly during the entire constant force phase.

These findings suggest that MU recruitment strategies are significantly related to the speed of contraction. While no difference was found in MDF peak between ramp performed at 5 and 10% MVC s^{-1}, both are significantly different from contractions at 20% MVC s^{-1} ($p<0.01$). During slower ramp contractions MU recruitment is complete at a lower percentage of MVC (50-60% compared to 70-80%) and this fact is also associated to increased neuromuscular fatigue during ramp phase (slower MDF decay during steady state).

Farina et al (2002) warned about limits of sEMG spectral analysis as a technique for the investigation of muscle force control. However, our findings suggest that 1) MU recruitment strategies, as can be judged on MDF data, seem to be related to the speed of contraction; 2) Differently, %DET confirmed to be more sensitive in revealing dynamical changes within sEMG, as well as in providing evidence of increased presence of MU common drive (F. Felici et al., 2001).

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THE CONTRIBUTION OF MUSCLE PROPERTIES IN THE CONTROL OF A PERTURBED LIFTING MOVEMENT

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INTRODUCTION

Movement performance is surprisingly robust. Think for example of walking on uneven surfaces. While lifting an object unexpected events may also occur. However, these unexpected events do not always result in major movement disturbances: Subjects are able to lift a box which is 10 kg heavier than they expect (1, 2). Surprisingly, changes in trunk muscle activity are seen only after at least 100 ms in these experiments. Possibly muscle properties, i.e. the force-length-velocity relationship, provide a mechanism to offer resistance to the perturbation before muscle activity is adapted. The aim of the present study was to investigate whether muscle dynamics, which represent a zero-delay peripheral feedback system (3), reduce the effects of a perturbation in a whole body lifting movement.

METHODS

Numerical simulations were performed using a 2D-model of the musculoskeletal system, consisting of seven segments representing the feet, lower legs, upper legs, pelvis, trunk, upper arms and forearms. All segments were connected in frictionless hinge joints. The skeleton is actuated by agonistic muscle groups, which were represented with a lumped Hill-type muscle model (3). Two kinds of open-loop control signals are compared: with and without muscle dynamics. The control signals were chosen such that a measured whole body lifting movement with a box of 1.6 kg was closely mimicked. Given these control signals, a perturbation was applied by an instantaneous increase of box mass with 2 kg, 5 kg, 10 kg or 15 kg. The analysis focuses on the trunk, by far the heaviest segment. The momentum developed by the trunk before a perturbation is applied, may be sufficient to keep it extending after the perturbation. To prevent the arms to reduce the effects of the perturbation on the trunk, the orientation of both arm segments was fixed in space during the simulations. During all simulations, the joint angle between pelvis and trunk was recorded. At 100 ms and 200 ms after increasing the box mass, the deviation of the pevis-trunk angle with respect to the unperturbed movement was calculated as a measure of the effects of the perturbation.

RESULTS

The perturbation affects the lifting movement in both models (i.e. with and without muscle dynamics) and in all mass conditions. The higher the mass, the more the movement was disturbed. For both control signals this was a monotonous increase. However, the model including muscle dynamics was better able to withstand the perturbation in all mass conditions (Figure 1). In the 10 kg condition, the model lacking muscle dynamics showed a deviation in pelvis-trunk angle of 0.235 rad after 200 ms, whereas this deviation was only 0.082 rad in the model incorporating muscle dynamics. For the ten male subjects that participated in study (1), this deviation lay between 0.039 and 0.0925 rad. The simulation values from the model with muscle dynamics lay within these ranges. The same was true for the values after 100 ms. Additional analyses revealed that the results were not critically dependent on the pelvis-trunk angle where the erector spinae produces its peak forces.

DISCUSSION

The momentum of upper body alone could not resist the perturbation. Incorporating muscle dynamics in the model clearly increased the resistance to the perturbation. Unfortunately, even with muscle dynamics the effects of a perturbation increases rapidly after 100 ms. An active response, which we have previously found to occur after about 100 ms (1), is required to fully compensate for the perturbation. Thus, especially shortly after the perturbation muscle properties diminish the effects of the perturbation. In absence of the effect of muscle properties, the perturbation would cause such a large deviation from the planned movement trajectory that active correction might be infeasible or the perturbation might already have caused injury before an active response can be initiated. Therefore, muscle properties appear to constitute an important stabilizing factor in the performance of movement tasks.
The simulation data lay within the range of the experimental data. This resemblance gives us confidence that our model of the musculoskeletal system leads to valid results.

CONCLUSION

Muscle properties provide a mechanism to offer resistance to a perturbation before muscle activity is adapted. It can be concluded that muscle dynamics are an important factor in stability of movements.

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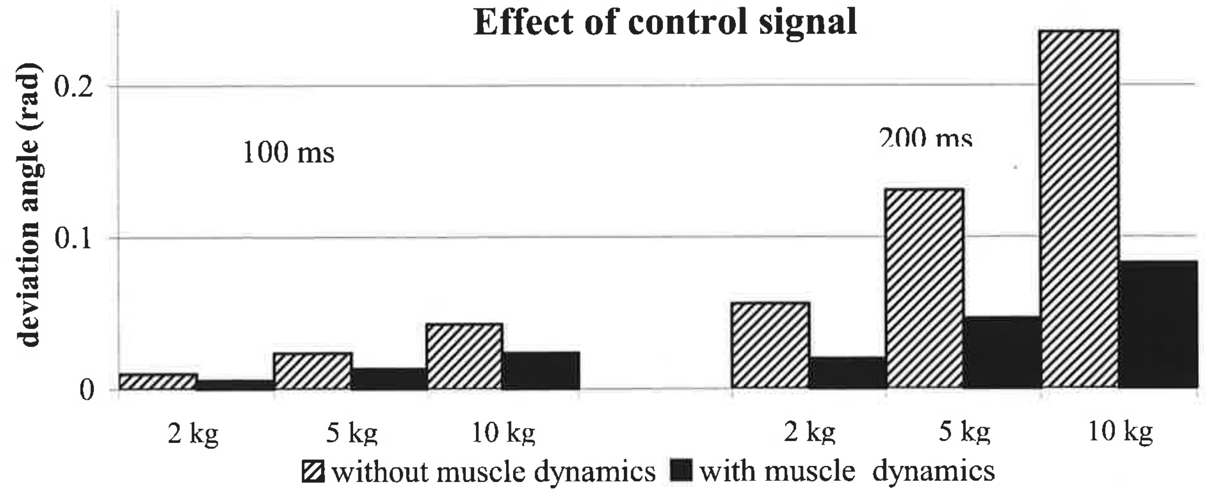

Figure 1: Deviation in pelvis-trunk angle between the unperturbed and perturbed movement for the model lacking, and the model including muscle dynamics (in rad).

NEURAL ADAPTATIONS DURING LEARNING A STRENGTH TASK

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INTRODUCTION

Learning can be defined as a set of processes associated with practice or experience leading to relatively permanent changes in skilled behavior. In a strength task, several processes are considered to contribute to a relatively permanent improvement in force production: a) hypertrophy (MacDougall, 1992), b) increased myofibrillar protein density (Phillips, 2000), c) several biochemical processes (Bell et al., 1992) and d) neural adaptations (Rutherford & Jones, 1986). The aim of this study is to quantify the neural adaptations following the learning of a strength performance. To be able to quantify such neural adaptations, the other processes involved in increasing strength need to be excluded. In the present study, prevention of hypertrophy is obtained by performing the study within a short time period (five days), and the influence of increased myofibrillar density is excluded by keeping the subject in a fasted state during the training period. Besides neural adaptations, only possible biochemical processes have not been excluded as a cause for strength improvement.

METHOD

Maximal isometric dorsiflexion of the ankle joint was chosen as the strength task. During five days, fourteen male subjects performed two training sessions each day. Each session consisted of 5 sequences with 5 trials each sequence. Each trial included 4 seconds of maximal isometric dorsiflexion of the ankle. The strength-training device consisted of a pedal, with the center of rotation passing through the center of rotation of the ankle joint. During all trials in all sessions, force was recorded with three force-cells, attached to the pedal. The dorsiflexion force was recorded with one force-cell, the other two force-cells were connected to the point of rotation of the pedal, measuring force exerted on the point of rotation itself (pulling/pushing with the leg in any direction). The subjects received continuous feedback of the generated dorsiflexion force on an oscilloscope.

Surface electromyography (SEMG) measurements were performed at the first session each day (active electrodes, sampling frequency 2048 Hz / channel, band pass filter 3-400 Hz). The m. tibialis anterior (TA) was investigated with a 130-channel grid electrode (13 by 10 electrodes, inter-electrode-distances (IED) 5 mm). Electrode pairs (IED 20 mm) were placed above the m. extensor digitorum longus, the medial part of m. gastrocnemius and the m. soleus. All recordings were made unipolar with a reference electrodes placed over the knee joint. Different electrode configurations were reconstructed digitally. In the first session, the electrode positions were marked by pen, such that the same electrode position could be used during the different sessions. To allow normalization of the SEMG signals, three recordings per session were made while the subjects were standing on their toes.

RESULTS AND DISCUSSION

From the first to the fifth day, the subjects increased dorsiflexion strength gradually, resulting in a total increase of 15% of the initial strength ($p<0.05$, paired t-test). In addition, the

subjects showed an increased pushing force ($p<0.05$, Sign test), despite the lack of mechanical profit of this on the dorsiflexion force. In contrast to the dorsiflexion force, this pushing force showed a large variation within each session of each subject. This variation was not related to the dorsiflexion force. Why the subjects increased the pushing force while learning dorsiflexion cannot be explained directly from the present data. The spatial distribution of (unipolar and bipolar) amplitude and frequency over the TA was very stable within each subject during the different trials in each session and during the different sessions (figure 1), while large inter-individual differences were found. However, the amplitude and frequency (absolute and relative to an other standard contraction) of the unipolar SEMG varied largely from session to session, without any relation to the strength training. Bipolar SEMG characteristics varied less between sessions, while no consistent relation to the strength training could be found either. The absolute and normalized synergist and antagonist muscle activity did neither have a relation with training sessions.

CONCLUSIONS

Despite that the dorsiflexion task contained a few degrees of freedom, the subjects obtained significant increase in dorsiflexion force as a result of 4 days of training. How the subjects managed to increase this force remains uncertain. More intensive data analyses might reveal some of it.

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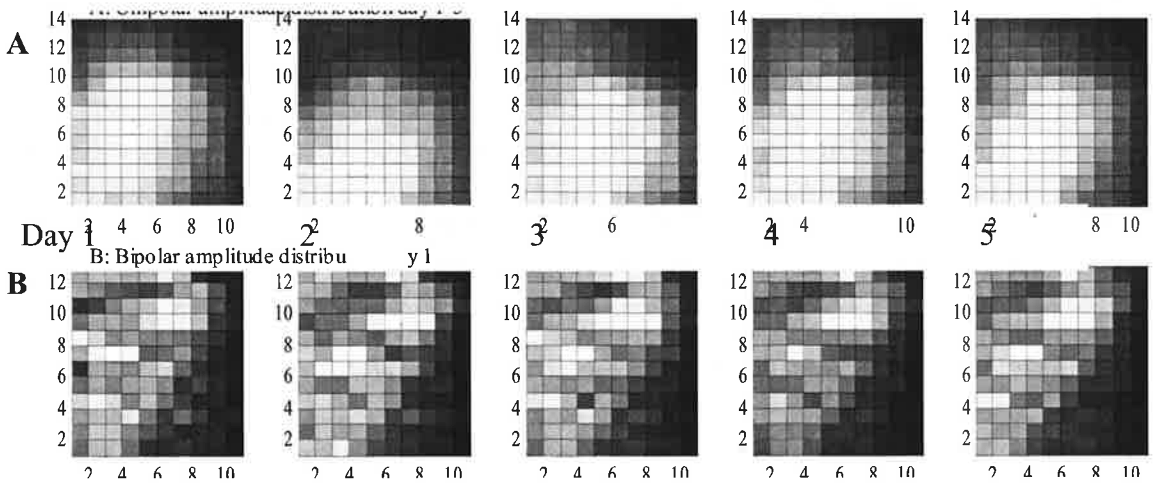

Figure 1:Typical example of spatial distribution of unipolar (A) and bipolar (B) amplitude over the TA from day 1-5. The lower left corners indicate the most medial and proximal part of the muscle. Each square is the root mean square value calculated from 0.5 seconds around the time instant with maximal dorsiflexion force of that session. Light is high and dark is low amplitude (each plot is scaled individually).

ARM MOTION AND EMG IN CHILDREN WITH DUCHENNE'S TYPE MUSCULAR DYSTROPHY; A PILOT STUDY

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INTRODUCTION

Surface EMG has been used to examine changes in the power spectrum during isometric and isotonic contractions to discriminate neurophysiological differences between healthy and MD subjects [1-3]. In lower motor disturbances such as Duchenne's muscular dystrophy (DMD) EMG has been primarily used as a method of disease detection. The aim of this study was to determine if patterns in muscle activation and the kinematics of point-to-point planar arm movements are preserved in children with DMD.

METHODOLOGY

Subjects were seated at a test table with their arm in a support device that floated the arm above the table surface, allowing for semi-frictionless movements. Subjects were then asked to perform a series of target acquisitions throughout a horizontal planar workspace from various start positions. Each movement task was done at a self-selected pace with trunk movement unrestricted. Electromyographic activity of eight upper extremity muscles were monitored with two BioResearch™ EMG amplifiers systems having a resolution of 0.1µV and a band width of 30-600 Hz with a fixed gain of 5000 at a sampling frequency of 3000 Hz. Kinematic data were collected with a MacReflex 50 Hz passive-reflective marker system with markers placed on strategic anatomical landmarks.

RESULTS

Data handling and analysis were performed using Matlab®. EMG data were filtered using a 4th order Butterworth low pass filter at a cutoff frequency of 6 Hz, linear enveloped and then normalized to the maximum value. The EMG activity level was in the 50 µV range for DMD muscles and was significantly lower with peak levels reported in available literature for healthy muscle performing similar tasks. Hand trajectory and tangential velocities (Equation 1) were obtained from the hand coordinates (x, y), which were smoothed before being differentiated. Mean V_{peak}/V_{mean} ratio that falls in the reported range of 1.60-1.90 however, in the DMD subject this ratio increased to a mean ratio of 2.35 [4, 5-7].

$$V_t = \sqrt{\dot{x}_w^2 + \dot{y}_w^2} \quad \text{Equation 1}$$

DISCUSSION

Movement along a straight trajectory is accomplished by utilizing upper body motion to propel the shoulder and produce "whipping" action of the arm. The path of the hand tends to remain along a straight point-to-point trajectory, despite the inertial component of the arm motion. This suggests that straight hand trajectories between point-to-point are an important goal in arm movements. Moreover, in the presence of weakness, compensations are made throughout the movement by muscle activation and joint rotations in other body segments.

From EMG data it is clear that flexor and extensor groups were often active regardless of direction of joint rotations or direction of the hand path. That is synergistic or agonist-antagonist activity is not as clearly discernable as in healthy subjects. This result is the not due to muscle contraction dynamics alone but also to the use of the whole body as a means for projecting the hand in a desired direction.

CONCLUSION

In this study the residual motor activity potentials of MD muscles realized with EMG as obtained under isometric and dynamic conditions were at significant levels [1]. Moreover, potentials emerged at sufficient levels for use in studies for the purpose of determining muscle coordination strategies, estimating muscle contribution, and agonist selection at onset of movement [8-10]. For the MD subjects tested the EMG potential levels generated are reduced as anticipated [2-3] but they are also large enough to suggest when and perhaps how much a muscle is contributing to an effort. This information is useful in understanding how the CNS copes with a neuropathology affecting muscle pathology as well as function.

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AN EMG PATTERN RECOGNITION SYSTEM BASED ON FUNCTIONAL LANGUAGES FOR PROSTHESES CONTROL

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INTRODUCTION

In spite of the enormous technological advances of the past decades, we still can not produce prosthetic devices capable of perfectly mimicking their natural counterparts. The major drawbacks are not necessarily related to the development of anthropomorphic structures, capable of performing complex tasks. Those kind of devices have already been developed, with a number of degrees of freedom (3). However, the control of such devices is not an easy task. There is a very pronounced lack of sensory feedback and it is normally very difficulty to control more than three or four degrees of freedom - since it requires a great amount of conscious effort. In this work, the authors describe their myoelectric strategy for prosthesis control, based on a hardware/software system, capable of detecting and acquiring EMG data. As an attempt for easier control, it is proposed that the EMG signals, from remnant muscle groups, can be modeled by an autoregressive (AR) model. In order to develop the necessary software it is proposed the use of the functional paradigm which leads to a fast developing stage and a very compact set of programs that will run as fast as (sometimes even faster than) traditional C/C++ programs.

METHODS

Figure 1 shows the strategy for EMG pattern discrimination used in this work. The signals were collected by surface passive electrodes fixed on the midline of the muscle belly and further amplified and filtered (gain = 4950; band pass filter = 15Hz – 4500Hz) for data acquisition (sample frequency = 1000Hz; resolution = 12 bits). The next task is to discriminate what type of contraction has been developed. In this study we were interested in four different arm movements: elbow extension, elbow flexion, wrist pronation and wrist supination (1). To do so, the EMG electrodes were placed above two main muscle groups: biceps and triceps. An artificial neural network (as described in (2)) has been used to classify the current contraction into one of those four groups. In order to "focus the attention" of the neural network into the correct section of the EMG signal, an initial stage of signal windowing was necessary. The windowing makes sure that the features will be extracted during a proper EMG activity. The first step for pattern discrimination is to represent the biomedical process *y(n)* by a set of parameters that will be fed to the neural network. Among the various methods to extract those parameters, it has been elected the autoregressive (AR) model. This model allow us to represent the EMG signal by a set of internal coefficients. All the necessary software has been written using the Clean language (functional). Two concurrent processes were developed for the software. The first process embeds the data acquisition and pattern discrimination. The results are then fed to a pipe. The other process receives the command (one of four movements) from the pipe and uses it to control the limb. In order to support this scheme, one of the authors created a minimalist version of Linux, which consists in a boot kernel image and a basic file system to support the programs.

RESULTS

The rates of success of the proposed scheme ranged from 95% to 100% (as shown in 1) when discriminating four different arm movements. As a Clean program can be written as an almost perfect mirror of the formula found in theoretical works, it was possible to develop the prototype very fast.

DISCUSSION

The control of a prosthetic device requires intensive computer processing. Hence, in order to improve the performance of such control system, a computer language must have the following features: an efficient implementation of vectors and matrices; facilities for importing libraries produced by other languages; facilities for producing and exporting libraries to other languages; fast compilation; a syntax and semantics that eases the implementation and analysis of seminumerical and symbolic algorithms; ports to the main operational systems in use today (Linux, Windows, MacOS, Sun OS, and Oberon); and be able to generate a fast and small code that can be executed in simplified, light and dedicated hardware for field deployment. The authors noticed that through a judicious use of matrix and vector manipulation (essential for this type of work), a functional language can perform as efficiently as traditional procedural languages like C or Oberon. Besides, the possibility of reasoning about algorithms, in order to prove that the programs are correct, gives functional languages such as Clean and Haskell a clear advantage over procedural languages. The features of Clean that were decisive to the successful implementation of the system are constant time access to vectors, the existence of unique types and a very sparingly use of the heap, which is an asset in the kind of application in hand.

CONCLUSION

In this article the authors showed that the use of functional languages has a number of advantages over traditional procedural languages, when developing applications such as those required for prosthetic control. It must also be stressed that the functional paradigm should be used not only for the kind of application described, but also for any work where fast and safe design is a paramount requirement.

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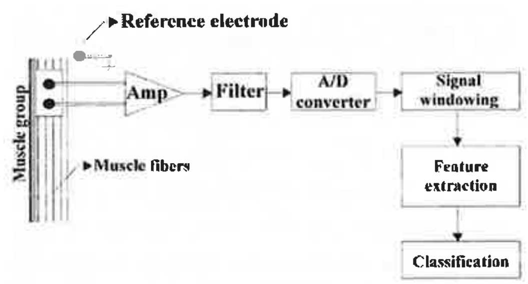

Figure 1. An approach for EMG pattern discrimination.

THE TOMPAW MODULAR PROSTHESIS - A PLATFORM FOR RESEARCH IN UPPER LIMB PROSTHETICS

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INTRODUCTION

The design of an upper limb prosthesis presents many challenges, mechanical, electrical and cosmetic. An underlying problem is that every user is different, so a high degree of customisation is involved. Variations include degree of loss, side, size, gender, and preferred method of control. Often it is difficult to select a suitable control strategy because there are several degrees of freedom to be controlled with only a small number of separable control signals available. It is therefore important that different strategies can be tried out easily. Microprocessor based controllers can fulfil this objective (1); however, up to now there has been little standardisation in this area. The ToMPAW (Totally Modular Prosthetic Arm with high Workability) system described here aims to provide a modularised set of components from which prostheses to fit a wide range of users can be assembled. The ToMPAW hand has four independently controllable digits, providing a variety of grips, and the wrist can flex/extend as well as pronate/supinate. The system is modular not only mechanically but also in terms of electronics and software. This is facilitated by the use of a network of microprocessors, each controlling one or two joints. Each processor is close to the sensors and actuators that interface with it. This has reliability advantages as well as the low stock-keeping costs and ease of maintenance associated with modular parts.

METHODS

Taking into account the requirements of modularity and the need to keep interconnections to a minimum (as these tend to be one of the most unreliable features of an electronic system), the preferred approach was to use a microprocessor network of the fieldbus type. The network has a number of inbuilt protection mechanisms against data corruption, and allows for graceful degradation where a fault in one processor need not affect the others. The technology selected for this project was Echelon's LONWORKS®, which implements a full OSI seven-layer communications protocol model (2). It is supported by Neuron® microprocessor devices from Toshiba and Cypress. Each device has a processor that runs programs compiled from Neuron C, (a variant of C) that provides multitasking and supports the I/O features of the Neuron device. Two additional processors in the device operate the bus protocol; these are inaccessible to the programmer, who does not need to be concerned with the details, but simply needs a reliable link along which variables can be transmitted from one device to another. The network also provides a convenient channel through which the system may be set up or monitored, either directly by a PC or through a radio link to avoid the inconvenience of cables (3). It is possible to use an Internet connection for remote diagnostics. The PC runs a clinical support system which provides an effective human-computer interface, displaying EMG data graphically in real time. The electronics of the ToMPAW system consists of four-layer printed circuit boards with surface-mount components. A hand board controls a complete hand with four active digits. It contains two Neuron devices, together with the

interface electronics for force and slip sensing by force sensing resistors and acoustic transducers respectively. Another (distal) board drives two motors in the wrist, and a similar board drives the motor in the elbow joint. A master or proximal board accepts up to eight fast analog inputs and can be used for overall control of a system. However, the hand board can also accept two EMG inputs, so a hand prosthesis will not normally require a master board in addition. This allocation of Neuron devices to physical functions means that the same device handles the motor and sensors associated with each joint. This minimises the network traffic.

RESULTS

The electronic subsystems of the arm were tested using a development system consisting of the existing mechanical assemblies of the Oxford Intelligent Hand (4) and the Edinburgh Arm (5). Two users operated the hand alone and two further users employed arm systems composed of hand, wrist and elbow. The first arm user employed ON-OFF operation with an EMG amplifier and the second used proportional control. Both used a pull switch to switch between the different axes. Training progressed over several visits, with different strategies tested in an effort to find the best one for the particular user. The second user could produce myoelectric signals from two muscle sites, but could not fully separate the two channels. This problem was addressed by creating a "winner-takes-all" strategy where the signal that was smaller in magnitude was ignored, and the larger magnitude signal was used for control, until the muscles relaxed and both signals fell below pre-set thresholds.

DISCUSSION

The design of the ToMPAW system was informed by the results of a user survey (6) and this was later used in the evaluation phase to establish from user questioning whether the design aims had been met, in addition to informal user feedback. An objective functional assessment will be made using a protocol that is based on both abstract object handling as well as simulated activities of daily living (7). Integration of the electronics subsystems into the final mechanics is continuing.

CONCLUSION

An upper limb prosthetic system with control effected by a network of microprocessors, each local to the joint it controls, has been successfully demonstrated. Assessment of the system is continuing along with further engineering development towards a manufactured product.

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FEASIBILITY OF A NEURO-ADAPTIVE SYSTEM FOR KNEE JOINT CONTROL USING QUADRICEPS STIMULATION

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INTRODUCTION

In recent years Artificial Neural Networks (ANNs) were used always more frequently for designing efficient control systems in several technological areas. ANNs have very interesting properties, such as: non-linearity, generalization and especially ability to learn from experience and to modify itself according to the changing environment. These characteristics become fundamental when controlled system was highly non-linear and time variant such as biological systems. Not many authors applied this technique to the problem of Functional Electrical Stimulation (FES)[1]. Methods and results, here descript, concerned the study of feasibility of an adaptive controller, based on neural architecture, using computerized simulations.

METHODS

We focused our attention on the control of the knee angle during swinging leg movements of seated subject by means of quadriceps stimulation. This experimental condition was often adopted because it allows to perform several experimental sessions in a safety and relatively simple condition[2, 3]. As target system, we adopted a dynamical and physiological based model of the lower limb, previously developed by Riener et al.[4, 5] Five muscular groups were considered involved in knee flexo-extension: biarticular and monoarticular knee flexor muscles, biarticular and monoarticular knee extensor muscles and biarticular ankle plantarflexors. Inputs to the model are the modulated pulse widths (controlled variable) and pulse frequencies (fixed) produced by stimulator. Maximum pulse width was fixed at 500 μs. Model output is the knee angular trajectory as resulting from torque produced by stimulated muscles (Rectus Femoris and Vasti) and by the passive action of the others.

The block diagram of implemented control strategies is illustrated in figure 1. It consists of an ANN, off-line trained to obtain the inverse dynamics of the implant, on the feed-forward control line, and a PID controller on the feedback one. In order to allow at the controller to follow variations in muscle's force generation, induced by fatigue, we implement a modified version of History Stack Adaptation (HSA) algorithm[6], that change ANN's weights according to the control signal ($PW=PW_{ANN}+PW_{PID}$), modified by feedback information (PW_{PID}). The neural network that simulate inverse model is a multilayered feed forward perceptron. It has seven input neurons, eighteen neurons in hidden layer and one neuron in the output layer. Number of neurons in hidden layer was chosen in order to obtain optimal generalization performance. Inputs are the pulse frequency, the actual and two time delayed samples of desired angular trajectory of the knee joint and her derivative (angular velocity). The output was the normalized between 0-1 pulse width. Activation functions are hyperbolic tangent for the hidden layer and sigmoid in the output neuron.Identification procedure for the neural inverse model was performed using Nguyen-Widrow initialization method and Levenberg Marquardt training algorithm.Examples used to train ANN were collected stimulating the implant model with several triangular and pseudo-sinusoidal pulse widths, sampled at 20 Hz, with a duration variable between 2 and 10 seconds and a maximum value variable between 200 μs and 500 μs.

RESULTS

In order to evaluate the neural controller we compare its performances with two other different control schemes: one consisting of the PID controller alone, the other composed by the static (not on-line adapted) inverse model and the PID controller. Furthermore in order to simulate presence of an external disturbance, we constrained controllers to follow a normal desired angular trajectory, the plant having an external 1 kg weight positioned near the ankle. In this way we changed the dynamics of the plant and furthermore we could test our systems, when a significant effect of muscles fatigue was present. In table 1 are shown RMSE for each control strategy, at the first cycle and at the fifth cycle of a repeated sequence of the same desired trajectory.

DISCUSSION

The results indicate that adaptation of inverse model, in the way here shown, perform a good matching with changes induced by simulated fatigue. In this way, during repeated sequences of movements, contribution of PID is reduced in favour of an increase of the feed-forward contribution. Furthermore, when a disturbance was present this effect appear more relevant, but not at all desirable. The designed system, in fact, can't distinguish between an external disturbance, such as a weight, and internal variations induced by fatigue. In this condition performance of controller make worse than the static strategy.

CONCLUSION

In this work a neuro-adaptive controller for the knee joint position during quadriceps stimulation has been developed. The interesting results obtained in computer simulations, prelude an experimental session, on healthy and pathological subjects, that could confirm the applicability of this control strategy. In a near future, furthermore, to a functional movement can be expected.

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RMSE [°]						
Condition	PID		ANN+PID		Ad-ANN+PID	
	1°cycle	5°cycle	1°cycle	5°cycle	1°cycle	5°cycle
Free swing	4.47	4.59	2.37	2.75	2.33	1.95
Weighted swing (1 kg)	15.36	--	10.34	11.49	10.25	13.52

Table 1: Root Mean Square Errors (RMSE) in the first and fifth consecutive repetition of a desired trajectory, not used during training session, for the different control strategies and in two conditions: free swing and weighted swing (1 kg).

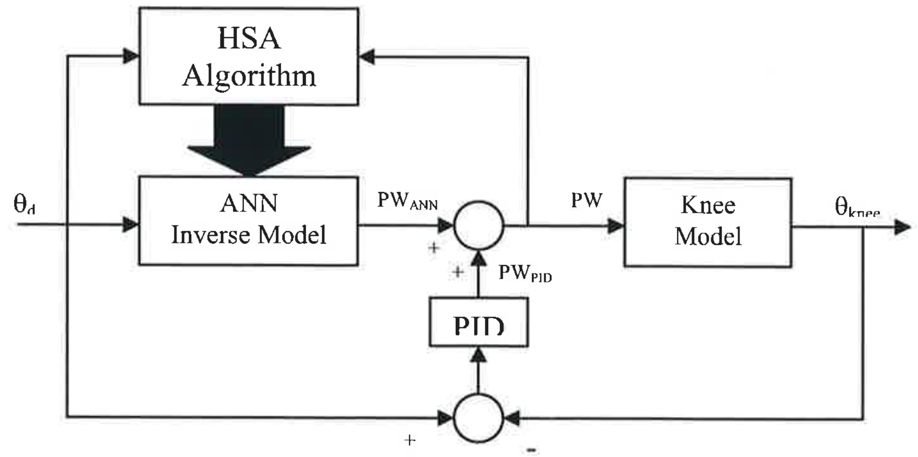

Fig. 1: Block diagram of neuro-adaptive control system: θ_d is the angular trajectory of the knee joint, θ is that simulated.

NON-INVASIVE ESTIMATION OF MOTOR UNIT CONDUCTION VELOCITY DURING ISOMETRIC AND DYNAMIC CONTRACTIONS

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INTRODUCTION

Spatial selectivity of surface EMG signals can be enhanced by the use of spatial filters. Thereby it becomes possible to track single motor units (sMUs) over time and thus to investigate peripheral and control properties of the neuromuscular system. In particular, the estimation of sMU conduction velocity (CV) may provide information on fiber type and muscle fatigue. The main aims of the present study were 1) to non-invasively detect sMU activity during isometric and dynamic contractions, 2) to compare the spatial selectivity of one- and two- dimensional (1-D and 2-D) spatial filters, 3) to investigate the effect of the spatial filter on the estimation of sMU CV on the basis of experimental data.

METHODS

Subjects: Five male healthy subjects participated in the study, after providing an informed consent. Experimental protocol: A 2-D point electrode array with inter-electrode distance (IED) 5 mm was used. The array consisted of 16 electrodes, arranged as shown in the Figure. Measurements were performed on the biceps brachii and the upper trapezius muscle. From the biceps brachii, recordings were obtained at elbow joint angles of 160°, 140°, 120°, 110°, 100°, and 90° (180° being the full extension of the forearm) with the arm 90° flexed. For the recordings from the upper trapezius, the arm positions were 22°, 45°, 68° and 90° abduction and flexion. The experimental protocol consisted of 2 second static contractions without load followed by two seconds of slow movement, changing the joint angle from one to the next position, and finally another two second static contraction in the new position.

Signal processing: Single and double differential filters in longitudinal (LSD, LDD) and transversal (TSD, TDD) direction with respect to the muscle fibers and a two dimensional Laplacian filter (NDD, see the Figure) were applied to the recordings. From the NDD filtered signals sMU activities were detected by automatic decomposition. From the detected MU potentials (MUPs), longitudinal selectivity was evaluated from the duration (based on energy) of the spatially filtered signals.

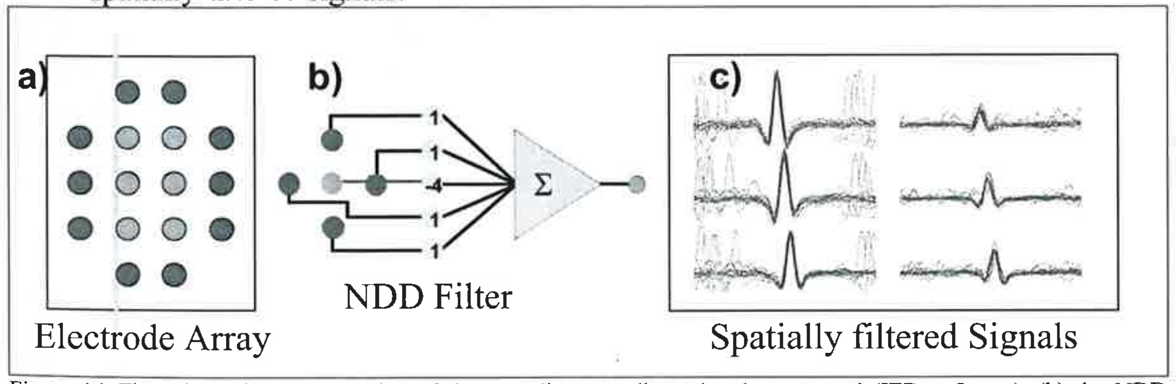

Figure (a) The schematic representation of the recording two-dimensional array used (IED = 5 mm), (b) the NDD filter applied in each of the six points indicated with light grey on the left, and (c) examples of MUPs detected and classified as belonging to the same MU (biceps brachii muscle). The CV of single MUs can be estimated from these potentials from the delay of propagation along the fiber direction (three delayed signals for each column). The spatial selectivity can be evaluated on the basis of the attenuation of the potentials in the two columns (transversal selectivity) and of the duration of the detected potentials (longitudinal selectivity). The CV has been estimated from the potentials in the column resulting in the highest amplitude (left column in this case). The MUP duration was assumed as the mean of the durations of the six detected potentials.

Transversal selectivity was defined as the relative attenuation of the MUPs detected from two points over the skin with a lateral distance equal to the IED (difference between the peak to peak amplitudes normalized with respect to the higher amplitude and expressed in percentage). Finally, CV of sMUs was estimated for all the spatially filtered signals from 1) the delay between the peak positions without interpolation and 2) after interpolation around the peak by a parabola, by the maximum likelihood estimator 3) using two delayed signals, and 4) using three delayed signals.

RESULTS

In all cases it was possible to detect single MUPs from the interference signals and classify them as belonging to a number of MUs which varied in the range 3-27 for the five subjects and the two muscles investigated. Selectivity: The 2-way repeated measures analysis of variance (ANOVA) of transversal selectivity with independent factors the muscle (biceps brachii and upper trapezius) and the spatial filter used (NDD, LDD, TDD, TSD, LSD), was significant for both factors ($F = 15.3$, $p<<0.001$, $F = 35.0$, $p<<0.001$, respectively). The systems were in average more selective in the transversal direction for the biceps brachii than for the trapezius (mean attenuation 28.8% for the biceps and 25.7% for the trapezius, $p<<0.001$). The average transversal relative attenuations for the different filters were 33.6% (TDD), 31.6% (NDD), 26.4% (LDD), 23.03% (TSD), and 21.81% (LSD). The 2-way ANOVA of longitudinal selectivity with independent factors the muscle and the spatial filter was not significant ($p=0.11$ and $p=0.41$, respectively). The average MUAP duration was 22.4 ms for the trapezius and 22.8 ms for the biceps brachii.

CV estimation: The 2-way ANOVA of CV estimates (on averaged potentials), with independent factors the method used for the estimation (peak method with and without interpolation and maximum likelihood estimator with two and three channels), and spatial filter adopted (NDD, LDD, TDD, TSD, LSD) was significant for the two effects ($F = 15.5$, $p<<0.001$, $F = 22.5$, $p<<0.001$, respectively). The post-hoc Student-Newman-Keuls (SNK) test disclosed pair-wise differences ($p<<0.001$) between the maximum likelihood method with two channels and all the other methods (average values: 4.3 m/s, 4.3 m/s, 4.6 m/s and 4.2 m/s for the four methods, respectively). The average CV estimates obtained with the five spatial filters were 4.3 m/s (NDD), 4.2 m/s (LDD), 4.6 m/s (TDD), 4.6 m/s (TSD), and 4.0 m/s (LSD). Computing the CV for each MU firing led to an average standard deviation of estimation (for all the MUs and the two muscles together) for the four methods tested (peak with interpolation, peak without interpolation, MLE with two channels, MLE with three channels) of 0.6 m/s, 0.8 m/s, 0.5 m/s, and 0.3 m/s, respectively. For the following results we used the NDD filtered signals and the maximum likelihood method with 3 channels for CV estimation. The 1-way ANOVA of CV estimates (averaged potentials), with independent factor the arm position, was not significant for both the trapezius and the biceps brachii muscle. The Student t-test for independent samples showed a statistically significant difference between the CV values of the trapezius (mean 4.0 m/s) and the biceps brachii muscle (mean 3.7 m/s, $p<<0.001$).

DISCUSSION AND CONCLUSIONS

CV of sMUs can be estimated from spatially filtered EMG signals. However, CV estimates depend on the spatial filter used and the method adopted. Thus, studies which report CV estimates with different types of recording systems can be hardly compared. One of the main results of the study is that CV estimates may have a large bias if low variance methods are used in association with spatial filters which do not remove end of fiber components (especially filters in the transversal direction). These filters should thus be avoided in CV studies. To increase spatial selectivity, limit CV bias and reduce CV estimation variance, a NDD filter with multi-channel MLE CV estimation methods can be effectively used.

ACKNOWLEDGEMENT

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EVOLUTION IN IMPEDANCE AT THE ELECTRODE-SKIN INTERFACE OF TWO TYPES OF SURFACE EMG ELECTRODES DURING LONG-TERM RECORDINGS

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INTRODUCTION

The observed shift of the power spectrum of a surface EMG signal to lower frequency bands during fatiguing contractions can be used to identify the onset and progression of muscular fatigue. The presence of noise in a SEMG signal shifts the power spectrum towards higher frequency bands (white noise), possibly masking any fatigue-related spectral changes during long-term postural activity. Impedance at the electrode-skin interface is a potentially large source of noise. The aim of this investigation was to assess the evolution in impedance at the electrode-skin interface of SEMG electrodes during long-term recordings, in order to improve the detection and processing of postural SEMG signals.

METHODS

Data were recorded from tibialis anterior on the right leg of 13 subjects using two types of Ag/AgCl surface electrodes (reusable Beckman and disposable Red Dot 2237 electrodes), each of which was arranged in a bipolar configuration. Electrodes were applied to the skin after the electrode site had been shaved, cleaned with alcohol, and left to dry. Impedance was measured by recording the voltage differences of a pair of electrodes while simultaneously applying an alternating current through the electrodes. The applied waveform was constructed of sinusoids of known frequencies from 1-16,384 Hz, evenly distributed on a logarithmic scale. Impedance data were acquired at 65,536 Hz and SEMG signals at 1000 Hz with an anti-aliasing filter of 400 Hz. Subjects were seated in a chair with their right foot connected via a pulley system to a suspended mass (10% of maximum), which could be lifted by dorsiflexion of the ankle. SEMG recordings were made during isometric contractions of 30 s performed every 15 min for 2 h. Impedance was measured immediately before and after each SEMG recording.

RESULTS

There were large individual differences in impedance levels for both electrode types. Impedance values at 1 Hz were higher for Beckman electrodes than for Red Dot electrodes ($F=13.13$, $p<0.01$; ranging from 3-1411 kΩand 1-130 kΩ for Beckman and Red Dot electrodes, respectively). Impedance decreased steadily with time only for Beckman electrodes ($F=12.13$, $p < 0.01$). Impedance also decreased with increasing frequency, for both types of electrode used ($F=36.24$, $p<0.01$, and $F=46.91$, $p<0.01$ for Beckman and Red Dot electrodes, respectively). At low frequencies, large differences in impedance values between subjects were apparent. However, at high frequencies, impedance values were similar for all subjects but varied with the electrode type.

The cut-off frequency, calculated as the frequency at which the impedance level was attenuated by 3dB (half-power point) in comparison to the initial impedance value at 1 Hz, was negatively correlated with the initial impedance ($r = -0.95$, $p < 0.01$, and $r = -0.99$, $p < 0.01$, for Beckman and Red Dot electrodes, respectively). The gain-bandwidth product, calculated as the product of the resistance (initial impedance) and the cut-off frequency, remained constant for each electrode type (95% confidence intervals 146.2-148.2 and 126.1-127.8 for Beckman and Red Dot electrodes, respectively).

There was a significant effect of impedance on the EMG spectra (Figure 1). Those subjects classified as high impedance (> 100 kΩ) had a greater percentage of the EMG spectrum contained at high frequencies compared to those subjects classified as low impedance (< 10 kΩ). When subjects with impedance values between these two extremes were added to the high impedance group (data added subject by subject in ascending order of magnitude), there ceased to be a significant difference when the high impedance group contained impedance values greater than 55 kΩ.

DISCUSSION

The impedance of skin can be electrically modelled with a simple network containing a resistor and a capacitor. As the gain-bandwidth product was constant for a given electrode, capacitance depends only on the properties of the electrode and gel. In contrast, resistance depends only on the properties of the skin and underlying tissue of individual subjects. Thus, it would be possible to calculate impedance based on a measure of the resistance, provided the capacitance of the electrode type is known.

Traditionally, it has been advocated to apply surface EMG electrodes to the skin after rigorous preparation (1). However, the findings of the present study show that only moderate preparation is needed, provided the impedance levels are less than 55 kΩ.

CONCLUSION

The EMG spectrum is unaffected by impedance provided skin preparation is sufficient to reduce the impedance below 55 kΩ.

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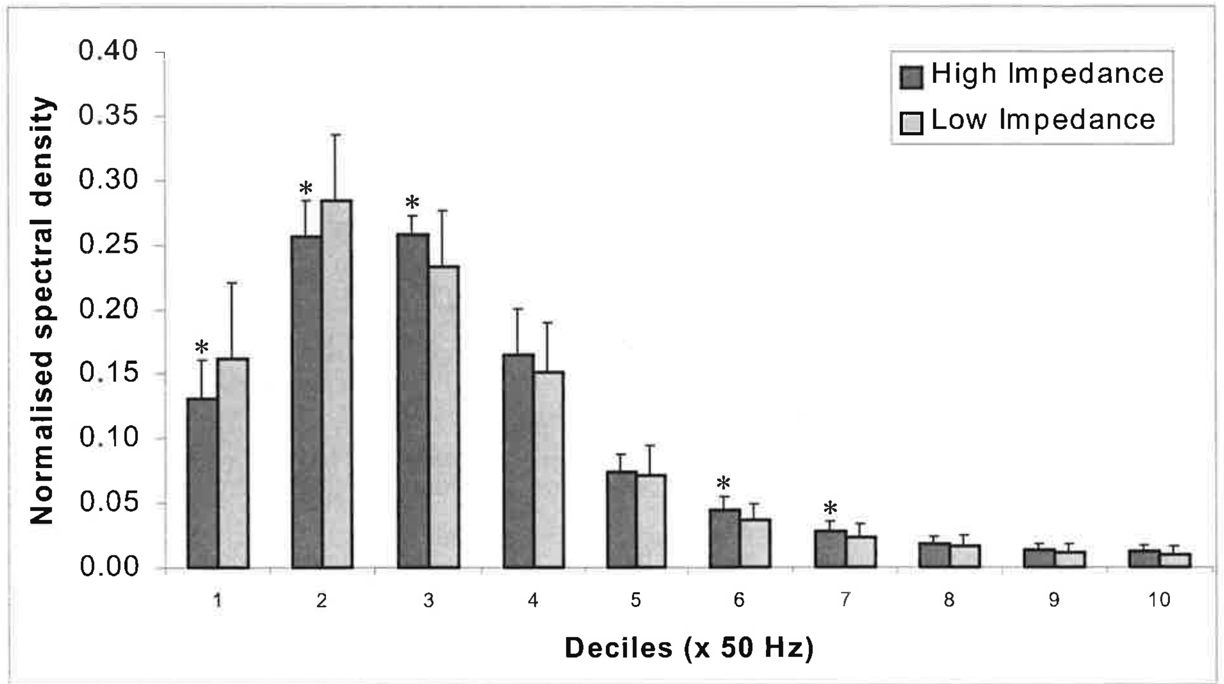

Figure 1: Spectral density by 50 Hz deciles (0-500 Hz) for high and low impedance groups. *Significantly different from low impedance (p<0.01).

FACTORS AFFECTING THE SURFACE EMG POWER SPECTRUM DURING FATIGUE

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INTRODUCTION

Computer simulations were used to explore the significance of changes in conduction velocity (CV) and firing statistics on the power spectrum of the surface electromyogram (EMG). In order to quantify changes occurring in the EMG power spectrum, specific characteristic frequencies e.g. mean (MNF) and median (MDF), were monitored. The mathematical analysis of Lindstrom (1) and others prescribes that a change in muscle fibre CV be mirrored by an equal relative shift in the surface EMG spectrum. Experimental results frequently show that changes in MNF and/or MDF cannot be fully accounted for by CV changes alone (2,3). Firing statistics, synchronisation, motor unit action potential (MUAP) shape changes and varying rates of CV changes are among the potentially significant additional factors affecting the EMG spectrum. In a previous simulation study we examined the affect of varying rates of change of CV for different motor units (MU), (4). Our results agreed with others (5), indicating that this could play a role in the divergence seen between the CV and spectral measures of the surface EMG signal during fatigue. Results in the literature from mathematical analysis and simulation studies, (6,7,8), point to the significant affect of MU synchronisation on the low frequency content of the SEMG spectrum. However, the details of MU synchronisation during fatiguing contractions are as yet unknown and much debate still surrounds this topic. In this study we have not dealt with synchronisation but sought instead to focus on and clarify the affect of changes in firing rate statistics and CV on the EMG spectrum. Analysis is carried out using both parametric and non-parametric spectral estimation techniques.

METHODS

The data used in this study was simulated from a physiological muscle model based on the biceps brachii (2,4). Firing rates were assigned randomly to MU's from a gaussian distribution with mean 30Hz, (12-60Hz). EMG signals were generated under three different circumstances; (a) decreasing mean firing rate only; (b) decreasing CV only; (c) simultaneously decreasing both CV and mean firing rates. Decreases are made in steps of 5% from 100% to 70% of the initial value. The Burg method was applied to 2 seconds of data, where 0.5 second spectra were calculated and averaged. The Fourier Transform and Welch methods estimated spectra for 0.5 second epochs, results from four successive epochs were averaged. Statistical analysis was carried out using ANOVA. See (9,10, 11) for a comparison and discussion of the different analysis methods.

RESULTS

ANOVA results indicated that changes in mean MU firing rate have no significant effect on the EMG spectral measures, MNF and MDF, for all three analysis techniques ($p = 0.3$). There was a consistently significant difference in the values of the MNF and MDF for all signals analysed with $p<10^{-6}$, (MNF > MDF). A marginal statistical difference was recorded ($p=0.0366$), for MDF and MNF values between cases (b) and (c).

DISCUSSION

The surface EMG signal represents the sum of all the MUAP trains (MUAPT's) within the pick-up region of the electrode. Each MUAPT can be modeled as a point process passing through a filter, where the filter function is derived from the MUAP shape. Linear systems analysis predicts that the power spectrum of the MUAPT is equal to the product of the power spectra of its firing statistics by that of the MUAP. The spectrum of the MU firing statistics thus forms an integral part of the spectrum of the MUAPT. The question of whether or not changes in MU firing statistics affect the spectrum of the overall surface EMG signal has been analysed in this study. Results indicate that MU firing statistics have little affect on the MNF and MDF of the EMG signal. This was found regardless of the analysis method used. These findings coincide with those of others (3,6,7).

CONCLUSION

Preliminary results illustrate that simultaneous changes in firing rates and CV, hypothesised to occur during fatiguing contractions, do not fully account for the experimentally observed changes in EMG spectra. It can be concluded that changes in MU firing statistics can be ruled out as a significant factor affecting the EMG spectrum. The issues of MU synchronization and different patterns of CV change deserve further attention.

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DO PATTERNS OF MUSCLE USE INFLUENCE MUSCLE FATIGUE?

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INTRODUCTION

It is uncertain whether different muscle loading patterns with the same overall mean force will have an influence on the resultant muscle fatigue. It has been postulated that muscle force variability may be beneficial means to reduce the relative levels of muscle fatigue. The objective of this study was to load the Extensor Carpi Ulnaris (ECU) muscle using three different loading patterns with the same overall mean force, for a fixed period of time, in order to determine whether different patterns of muscle loading would influence the relative levels of muscle fatigue. It was hypothesized that there would be no differences in fatigue between the different loading patterns.

METHODS

Using a repeated measures design, six healthy subjects participated in four contraction conditions: 1) a static continuous (SC) contraction at 5% Maximum Voluntary Contraction (MVC); 2) a static intermittent (SI) contraction alternating between five seconds at 0% and 10% MVC; and 3) a static fluctuating (SF) contraction alternating between five seconds at 2.5% and 7.5% MVC and 4) a control (CON) condition where subjects simply relaxed and remained inactive (0% MVC). Each of the contraction conditions lasted 30 minutes. The subjects' wrists were secured in a specially designed fixture containing a load cell. In order to perform the desired contraction pattern, subjects had to isometrically contract their muscle and ulnarly deviate their wrist against a load cell so the force output of the load cell matched the contraction pattern on the computer monitor. The force output from the load cell was continuously measured and stored on the hard disk of a computer at 60 Hz. During the contractions, the EMG activity of the ECU muscle was measured at 1000 Hz using two 10mm diameter electrodes (N-00-S; Medicotest; Ølstykke, Denmark) spaced 20mm apart. EMG values were normalized relative to maximum EMG values obtained at the end of the experiment and expressed as a percentage of the Maximum Voluntary Electrical activity (% MVE). With the subject's arm and wrist secured in the specially designed fixture with the load cell; 2, 20 and 100 Hz constant current electrical stimulation; 0.1 ms in pulse duration; was delivered to the ECU muscle and the resultant force response at the wrist was measured. The force response from the ECU muscle was measured before exercise, after exercise and 30, 60 and 120 minutes into recovery. At each time period, subjects also rated their level of perceived fatigue in the forearm using a visual analogue scale. The order of the four conditions was randomized and performed on separate days at least 24 hours apart. Repeated measures analysis of variance methods were used to make comparisons between the various conditions.

RESULTS

The mean forces (mean standard deviations) applied to the load cell during the various contractions were nearly identical averaging 5.1% (0.03%) MVC in the SC condition, 5.0% (2.3%) MVC in the SF condition and 5.1% (4.7%) MVC in the SI condition. There were differences in the mean MVE level measured from the ECU muscle during the contractions averaging 12.7% (1.3%) MVE in the SC condition, 7.6% (2.7%) MVC in the SF condition and 6.6% (5.3%) MVC in the SI condition ($p < 0.01$). There were also differences between conditions in the level of muscle fatigue measured with the electrical stimulation ($p = 0.03$). Muscle fatigue results were averaged over all frequencies and are presented in Figure 1. The SC contractions were the most fatiguing with fatigue being measured at both high and low frequencies, SF contractions resulted in intermediate levels of fatigue, and the SI contractions were the least fatiguing with fatigue only being measured at the low frequencies. Finally, as shown in the bottom of Figure 1, there were also differences in the subjective perceptions of fatigue measured with the VAS scale ($P = 0.02$). Theses differences mirrored the objective measurements of fatigue pre- and immediately post-exercise, but then became disassociated from the objective muscle fatigue measurements during recovery.

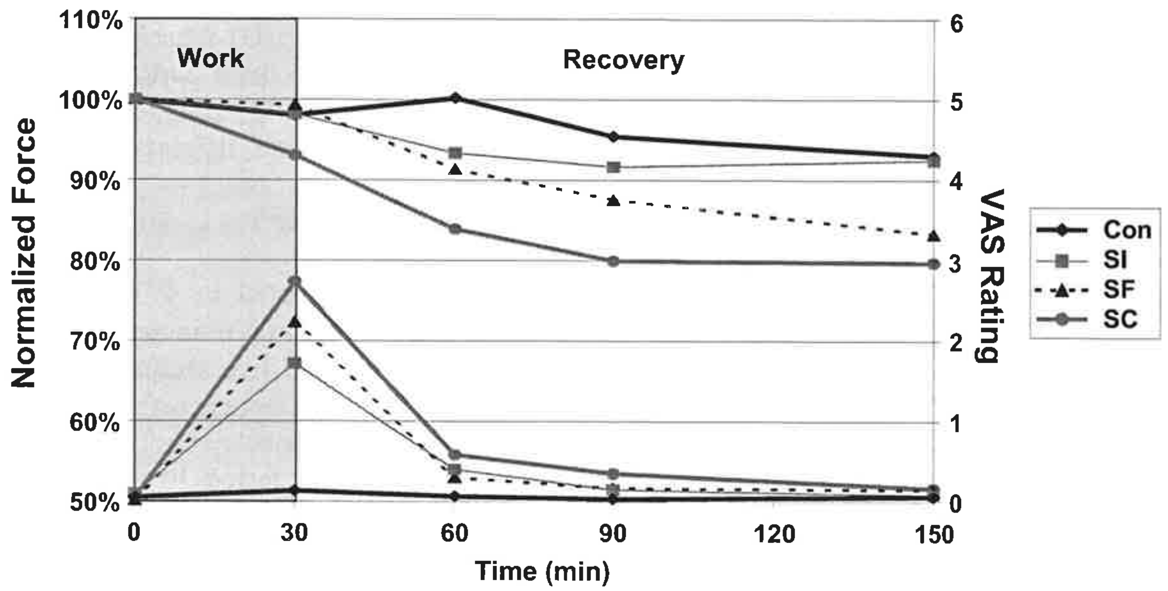

FIGURE 1 – (Top) Normalized force response of the ECU muscle averaged over all stimulation frequencies and grouped by condition. (Bottom) Subjective levels of perceived fatigue measured with the VAS scale [n = 6].

DISCUSSION

In this experiment subjects exerted the same overall mean force but in different patterns. Despite having the same overall all mean force, subjective and objective differences in muscle fatigue were measured. The static intermittent condition had the lowest level of fatigue and was the only condition where the level of fatigue was the same as for the control condition after 120 minutes of recovery. It is interesting that despite applying the same overall mean force, there were differences in the overall mean EMG levels measured from the ECU muscle. It would be interesting to repeat this experiment and investigate whether fatigue differences would result from contractions that had the same overall mean EMG level rather than the same overall force level. This would help determine whether mean EMG level or mean force level is a better predictor of fatigue.

CONCLUSION

Despite having the same overall all mean force, patterns of muscle contraction will influence overall mean EMG levels and the relative levels of muscle fatigue measured with electrical stimulation. In addition, the perception of fatigue was disassociated from the objective measurement of fatigue in the recovery period.

FEASIBILITY OF CONDUCTION VELOCITY AS A FATIGUE INDEX IN DYNAMIC CONTRACTIONS

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INTRODUCTION

The purpose of this investigation was to examine the influence of 1) action potential origination and extinction and 2) electrode position on muscle conduction velocity (CV) estimates obtained from surface myoelectric signal (MES) measurements at different joint angles. The motivation was to determine the feasibility of using CV as a fatigue index in dynamic muscle contractions, during which joint angle and muscle force may vary.

CV is often estimated by measuring the delay between MES measurements acquired at two locations along the muscle, and dividing the distance between the locations by this delay. The estimate depends upon the assumptions that both electrodes are placed on the same side of the innervation zone (IZ), along the length of the muscle fibers, and that the fibres are sufficiently long to render the effects of action potential origination and extinction (collectively known as end-effects) insignificant. Electrode position is a critical factor in meeting the assumptions required to obtain reasonable CV estimates [1,2]. Not only have striking effects on the cross-correlation been shown when electrodes align with or traverse the IZ, but since muscle fibres are finite in length, end-effects, which manifest as non-propagating components in MES, have also been shown to introduce a non-delay element into the cross-correlation which can bias a CV estimate. These effects increase as depth of the active muscle fibres increases and as the electrodes get closer to the innervation or fibre termination zones.

As joint angle decreases, muscle fibre length decreases. Also, the IZ may migrate towards the measuring electrodes. These changes are reflected in MES measurements as changes in electrode position. If substantial, they could cause increased bias and/or variability in CV estimations rendering CV difficult to measure during dynamic contractions.

METHODS

Simulated MES were generated to determine the possible influence of end-effects and IZ migration on CV estimation. Single fibre action potentials (SFAPs) were modeled according to [3] and then summated to generate motor unit action potentials. Model parameters were chosen to model a typical measurement from the biceps brachii.

To determine the influence of muscle fibre shortening on CV, estimations were made from 100 independently generated motor unit action potentials at 7 average fibre lengths (L_f). The center of the IZ of each motor unit was assumed to be located at ½ L_f and the average depth (d_{th}) of each motor unit was randomly chosen from a uniform distribution (5-30 mm). All fibre CVs were set to 4 m/sec. To account for individual fibre differences, right and left fibre termination dispersions (RT_D, LT_D), an innervation point dispersion (IP_D), and a depth dispersion (D_D) were included as fibre parameters. All dispersion values for each fibre were chosen from a uniform distribution within the limits given in Figure 1.

Each motor unit was made up of 450 SFAPs and three electrodes were located 35 mm from the center of the IZ, 5 mm apart, yielding two single differential simulated measurements. To model the effects of IZ migration, at each subsequent average fibre length, the center of the IZ was shifted 8 mm towards and eventually past the electrodes.

To determine whether or not the IZ actually migrates as joint angle changes, two channels of MES were collected from the biceps brachii at different electrode positions with the elbow held at 3 joint angles (50°, 90°, 160° with respect to 180° full elbow extension). At each angle, the IZ was located by determining the position of the electrodes which corresponded to a +/- transition in the delay of the cross-correlation of the signals. Bar electrodes (5 mm x 1mm) spaced 5 mm apart in a single differential configuration were used.

RESULTS

Figure 1depicts the results of the simulation. The white bars represent the mean of the 100 estimates with the IZ fixed. The gray bars represent the mean of the 100 estimates with the IZ shifting. The first two bars in this category have been omitted since a CV estimate could not be obtained based on the shape of the cross-correlation (IZ was too close to electrodes, or beyond end of fibre). Finally, preliminary results of the empirical experiment revealed that IZs were dispersed 20 mm ± 4mm and that a shift was evident in the 3 participants (60 mm, 40 mm and 40 mm respectively).

DISCUSSION AND CONCLUSIONS

According to Figure 1, average fibre length must decrease from 200 mm to 80 mm, or to 40% of its original length before end-effects become problematic, with no IZ migration. Double differential measurements would likely reduce the observed bias. However, IZ migration introduces bias and striking changes in the shape of the cross-correlation which may not be so easily avoided. Given that such a migration was noted in the empirical experiment as joint angle changed, accurate CV estimates are likely to be difficult, if not impossible, to obtain during dynamic contractions.

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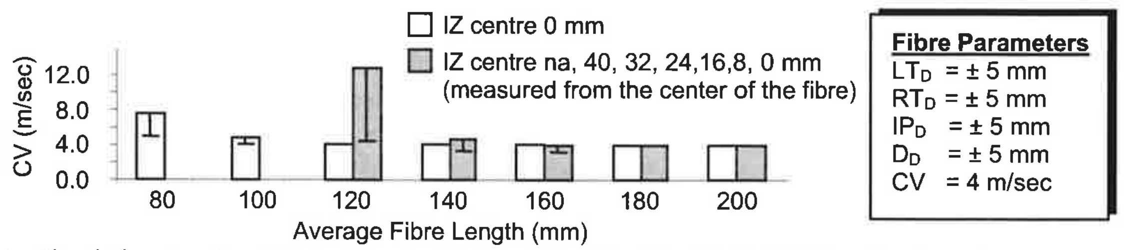

Figure 1: Simulation Results - CV vs Average Fibre length with and without IZ Migration (error bars represents 1 standard deviation)

FATIGABILITY DIFFERENCES OF THE SUPERFICIAL QUADRICEPS FEMORIS MUSCLES

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INTRODUCTION: The weakness of the quadriceps femoris muscle (QFM) has been suggested as one of the possible causal factors to knee osteoarthritis in the elderly (1). It has also been linked to the quadriceps avoidance (inhibition) mechanism during sport and everyday tasks. More recently QFM weakness has been identified as a causal factor to the frequency of anterior cruciate ligament injury in young basketball players (2). The main function of the QFM, while acting at the most unstable joint of the lower extremity, is to produce knee extension and/or control knee flexion (in its role as an antigravity muscle), and to stabilize the patella. The QFM resistance to fatigue is an important factor in exercise performance and can affect function in clinical situations. The connection to these two pathology/injury related mechanisms and the role of the QFM in everyday life activities have attracted significant attention from the scientific community. The QFM, while receiving innervation from the femoral nerve, is unique in a number of ways with respect to the morphology, function and structure of its components: (a) it exhibits varying fiber type compositions (3), (b) has dissimilar tension direction on the patella, (c) shows differential recruitment, and (d) has functional discrepancies single versus two joint span. While in sport and everyday activities the functional discrepancies are important, during an experimental bench paradigm, where the hip joint motion is restricted, only the morphological characteristics play a role in its force production and fatigue. The purpose of this study was to assess the fatigability of the superficial quadriceps muscles as a result of repeated alternating isometric, and slow and fast isokinetic bouts of exercise that resemble high intensity activity often encountered during training or competitive game settings.

METHODS: Fifteen undergraduate healthy untrained exercise science students (age 20.1 ± 1.3 years) gave their consent to take part in this experiment. All subjects were physically active without prior formal competitive training. The subjects sat in the chair of the CYBEX II testing apparatus with the torso upright against the vertical back support, strapped at the thigh, pelvis and mid chest to avoid extraneous body movements. The axis of rotation of the CYBEX dynamometer was aligned to the knee joint axis and the lower shank was securely attached to its arm. The seated subjects were asked to fold their arms across their chest and perform two familiarization contractions at each condition (isometric at 45° flexion, and isokinetic at 90 °/s and 300 °/s). Following the familiarization trials subjects were asked to produce maximum voluntary contractions (MVC) of the quadriceps in the following exercise bout (EB) sequence: One isometric for 5 s, followed by ten slow (90 °/s) and ten fast (300 °/s) isokinetic contractions. This EB sequence continued to exhaustion, and most subjects were able to realize 4 to 5 complete bouts with an extra isometric contraction at the end. The median frequency was assessed through surface electromyography (EMG) for the vastus lateralis (VL), rectus femoris (RF) and vastus medialis (VM) muscles. Preamplified bipolar surface EMG electrodes (Ag/AgCl), with a fixed inter-electrode distance of 2 cm, were placed on each muscle oriented towards the longitudinal direction of the muscle fibers. The reference electrode was placed on the right acromion process. EMG data were collected by a 16 channel Biopac unit at a rate of 1,000 Hz per muscle using the Acknowledge software (V. 3.5.3, Biopac Systems, Santa Barbara, CA). The signals were detrended, and band-pass filtered between 20 and 450 Hz. Only the isometric data were used for the analysis on the central 3072 points (~ 3 s duration) of each 5 s contraction. A power spectral analysis was done using the fast Fourier algorithm in the psd function of MATLAB (The Mathworks Inc. V. 6.0.0.88) on time epochs of 512 ms length (overlapping by 256 ms), eleven in total using Hanning window processing. The mean median frequency was calculated using the median frequencies of the 11 overlapping epochs. To examine the effects of EB and muscle on changes in the median frequency a two-factor repeated measures ANOVA (muscle x EB) was performed followed by Tukey HSD multiple comparisons test for the between factor (muscle), and by one-way ANOVAs and paired samples T-tests evoking the Bonferroni adjustment to ensure the familywise error rate for the within factor (EB).

RESULTS: The descriptive data (means ± SD) for the mean median frequency for the vastus lateralis (VL), rectus femoris (RF) and vastus medialis (VM) at each EB are presented in Table 1. The results demonstrated an overall significant main effect for muscle ($F_{2,42} = 3.42$, $p = 0.042$), with the Tukey post hoc comparison indicating that the median frequency of the VM was significantly lower than the VL, and the RF median frequency values between the other two muscles. At the initial pre exercise stage the median frequency of the VM was significantly lower than both of the other muscles ($F_{2,42} = 6.41$, $p=0.004$). There was also a significant EB main effect ($F_{4,168} = 118.5$, $p < 0.001$), as well as a significant muscle by EB interaction ($F_{8,168} = 7.5$, $p < 0.001$). These results indicate that a significantly greater decrease in the median frequency occurred between EB(1) and EB(5) for the RF and the VL than the VM. Repeated paired T-tests between consecutive EBs (on the univariate ANOVAs of the muscle factor) showed that the median frequency of the RF continue to decrease significantly up and until EB(4), while significant drops in the median frequency for VL and VM continued up and until EB(3). Even after adjustment for the familywise alpha level (0.05/12 value for comparison $p_a = 0.005$) these findings were found significant.

DISCUSSION: The findings of this study demonstrate that the QFM displays distinct differences in median frequency over the range of intense exercise setting. The most interesting differences are between muscles with respect to the rate of change in the median frequency as a function of EB and the length of time over which significant changes in median frequency persist for each individual muscle. The largest change in the median frequency as a function of single joint repeated fast and slow EB to exhaustion occurred at the RF (27%) followed by the VL (22%), with the least change found at the VM (12%).

CONCLUSION: While fatigue effect was evident in all three muscles the fatigability of the RF was the highest followed by the VL, with the VM being the most defiant to fatigue. These results indicate that fiber type homogeneity (slow twitch fibers) may exist in the VM, which displayed the least amount of change in the median frequency, and that VL may be more homogeneous with respect to fast twitch fibers as opposed to RF which may possess larger morphological (fiber type) variability.

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Table 1. Mean median frequency (means ± SD) in Hz for each of the superficial quadriceps femoris muscles at each exercise bout (EB)

‡

		EB(1)		EB(2)		EB(3)		EB(4)	EB(5)
†	**Vastus Lateralis**	110.7 ± 7.24	-*-	95.3 ± 6.69	-*-	88.9 ± 6.53		86.8 ± 5.87	87.9 ± 6.43
	Rectus Femoris	104.8 ± 6.26	-*-	84.5 ± 5.34	-*-	79.2 ± 5.31	-*-	76.4 ± 4.62	76.6 ± 4.88
	Vastus Medialis	86.8 ± 6.10	-*-	79.6 ± 5.28	-*-	76.8 ± 4.49		75.4 ± 4.78	76.7 ± 4.89

† **p** < 0.05, ‡ **p** < 0.001, -*- **p** < 0.005

FATIGUE AND RECOVERY OF ELBOW FLEXOR MUSCLES AFTER SUSTAINED STATIC SUBMAXIMAL EFFORT

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INTRODUCTION

Ergonomic studies show that surface electrode electromyography (sEMG) is a valid tool to assess the effect of exposure to physical effort and its physiological consequences at the muscle. Surface EMG signals can be used to predict relative muscle activity and even fatigue (De Luca, 1997; Luttmann, 2000). Several EMG parameters have been used to evaluate muscular fatigue. The most commonly used EMG frequency descriptor during sustained isometric contractions is the median frequency or 50th percentile value (MDF). A frequently used EMG amplitude indicator is the root mean square value (RMS). Many studies performed since the beginning of the century have shown that fatiguing contractions of a muscle coincide with an increase in the EMG amplitude and a shift in the EMG spectrum towards lower frequencies. Many authors cite that the shift in the EMG spectrum is related to a decrease in muscle fibre conduction velocity (MFCV). The increase in the EMG amplitude is mainly affected by the increased recruitment and firing rates of motor units. Until now, most studies have focused on the temporal EMG changes during fatiguing contractions. A less number of studies (De Luca, 1984; Hara Y et al., 1998; Krogh-Lund, 1993; Kuorinka I, 1988; Vaz MA et al., 1996) have looked at the recovery of the EMG parameters after a sustained submaximal effort. The aim of this study was to analyse the muscular fatigue and the muscular recovery of m. Biceps Brachii and m. Brachioradialis after a 25% MVC isometric elbow flexion task in a population of older men.

METHODS

Subjects – Forty elderly men gave their informed consent to participate in this study. They all participated in a larger study concerning the effect of different types of fitness training on their health status. Their mean ± SD age, weight and height were 61.73 ± 4.14 years, 83.59 ± 11.6 kg, 174.92 ± 5.84 cm respectively.

Procedure – After a maximal voluntary contraction (MVC) test, the subjects performed a unilateral static endurance test of the right elbow flexor muscles at 25% MVC until exhaustion. Immediately after the termination of the fatigue test, a second MVC was asked. During the recovery period, measurements of 5 second contractions (at 25% MVC) were obtained at 30 seconds, 60 seconds, 2, 5, 10, 15 and 20 minutes after the second MVC. At the end of the recovery period a third MVC was performed.

Electromyography – Bipolar surface EMG signals were obtained from the right m. Biceps Brachii and the right m. Brachioradialis. A telemetric EMG device (ME3000p, Mega Electronics Ltd., Finland) was used to register the EMG activity. The signals were filtered using a bandpass filter (15Hz – 500Hz) and sampled at 1000 Hz. Muscular fatigue and muscular recovery was quantified over time by assessing the changes in MDF (fast Fourier transformation) and RMS value (time epoch of 1s). All the EMG data were normalized against the initial (prefatigue) values of the parameters during the first 2s of the 25% MVC fatigue test.

Statistics – A one-way ANOVA for repeated measures was used to determine statistical relevance of the observed differences. The level of significance was chosen as p<0.05 for all statistical tests.

RESULTS

(1) Torque values of the three maximal voluntary contractions (before and after sustained contraction and after 20 minutes of recovery) were significantly different from each other ($p<0.01$). After 20 minutes of recovery, maximal isometric torque was still 9 % lower than the prefatigue value.

(2) During the isometric fatigue task, the increase in RMS (m. biceps brachii : +159%; m. brachioradialis : +164%) and the decrease in MDF (m. biceps brachii : -16.8%; m. brachioradialis : -14.3%) were significant for both muscles ($p<0.01$).

(3) The recovery pattern of MDF and RMS for both muscles is shown in figure one. Within 5 minutes, MDF returns to its prefatigue value. In contrast, the recovery of the RMS value was still incomplete after 20 minutes. The RMS value of the m. biceps and m. brachioradialis was still 2 times and 1.8 times greater than the prefatigue value at the end of the recovery period.

DISCUSSION

(1) Muscle fatigue can be defined as “any reduction in the force-generating capacity of the total neuro-muscular system, regardless of the force required in any given situation”. This finding is confirmed by the significant decrease in maximal isometric force after the fatigue task.

(2) The increase of the EMG amplitude and the shift of the EMG spectrum to lower frequencies are also well known phenomena of localized muscle fatigue (De Luca, 1984).

(3) The restitution of the power spectrum after complete exhaustion is in agreement with other studies (Kuorinka I., 1988; Vaz et al., 1996). An explanation of the fast recovery of MDF can be found in the recovery of the muscle fibre conduction velocity after fatigue. Several studies (Corcos et al., 2002; Hara Y, 1998; Kirsch and Rymer, 1987) have reported that the MFCV returned to normal within 10 minutes. The incomplete recovery of the RMS value is less understood. Krogh-Lund (1993) and Vaz et al. (1996) published some recovery data of the RMS value and also found an incomplete recovery of the EMG amplitude after 30 and 15 min respectively. Vaz et al. (1996) even found a further increase of the RMS value during the recovery period. In our study, we found a decrease in RMS during the first 5 minutes of the recovery period. During the remaining 15 minutes of the recovery period no further restitution of the EMG amplitude occurred. The difference between our study and Vaz M.A. (1996) may be due to the lower contraction level (25% MVC) and the use of other muscle groups with other fibre type composition in our study. The elevated RMS value probably indicates that more motor units were activated and/or higher average rates of firing are used to achieve the target force level. We may conclude that some evidence exists that the peripheral component of muscle fatigue (e.g. conduction velocity) recovered more quickly then the neural drive to the muscle (central component of muscle).

CONCLUSIONS

From this study, we can conclude that the EMG spectrum returns to normal within 5 minutes following a logarithmic function of time. In contrast, the RMS value was not fully recovered after 20 minutes of rest. There seems to be a discrepancy in the recovery of the peripheral and central component of muscle fatigue.

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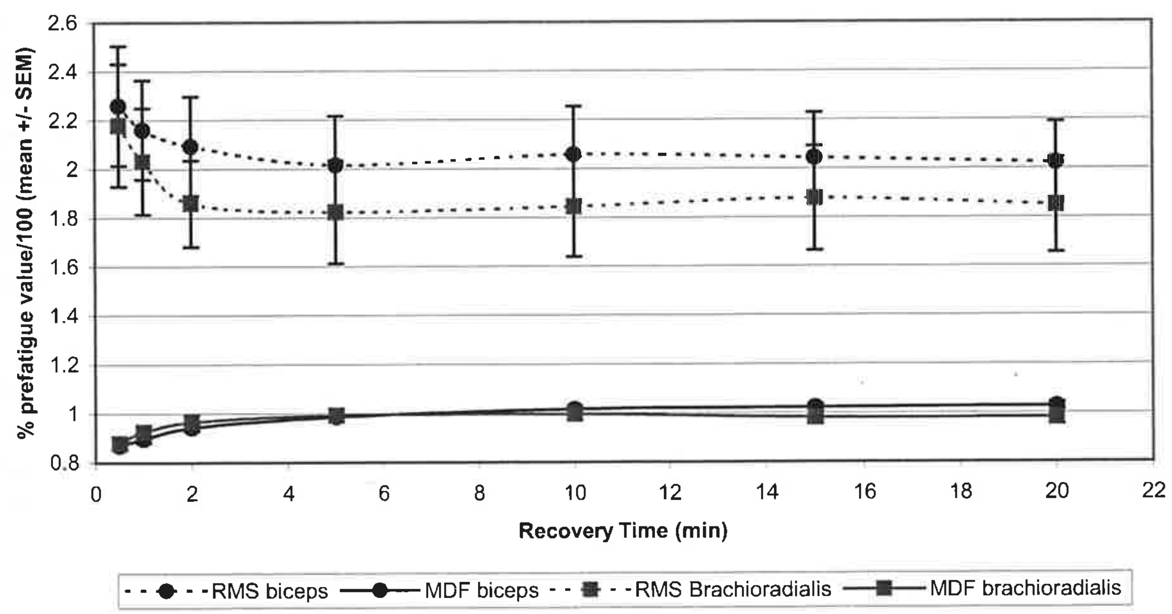

Figure 1 – Recovery pattern of RMS and MDF for right m. Biceps Brachii and right m. Brachioradialis.

MOTOR UNIT ACTIVITY IN THE TRAPEZIUS MUSCLE DURING PROLONGED COMPUTER WORK

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INTRODUCTION

Work–related muscular shoulder-neck pain, also referred to as tension–neck syndrome, is a major health risk in many different modern workplaces. The patho-physiological origin of muscle pain at low-intensive work is still unclear. During low level contraction a prevailing hypothesis postulates the continuous activity of specific motor units (MUs) that may become metabolically overloaded (Cinderella hypothesis) [1]. The present study was undertaken to assess the EMG pattern in the trapezius muscle during a 30 minutes computer task.

METHODS

Fourteen subjects, seven women and seven men with ages ranging from 23 to 38 years, participated in the study. The subjects were healthy and did not suffer neither from neck nor from shoulder pain. The subjects were asked to execute an interactive computer learning program (called ErgoLight) that was mainly mouse driven and lasted 30 minutes. ErgoLight simulates the effects of different illumination conditions at different workplaces. It teaches illumination ergonomics by letting the subjects interactively explore different combinations of light sources and positions of the working place. The task consisted of pointing the cursor to targets (e.g. light, secretary table) that were activated by clicking the mouse button, of moving and of rotating the targets until an optimal position was reached. Subjects sat in a chair with back support and the height of the table and chair were adjusted such that the upper arms were vertical and the forearms horizontal, symmetrical and parallel to each other.

Two three-channel intramuscular EMG signals were picked up from two locations in the trapezius muscle. Electrodes with four thin wires were used. An automated multi-channel long-term decomposition programme (EMG-LODEC) [3] was used for the analysis. To obtain an estimate of the over-all muscular activity, bipolar surface electrodes were attached on each side of the intramuscular insertion points. All decomposition results of the intramuscular signals were tested for accuracy using a method, as described in [2].

RESULTS

Despite of the usually low contraction level during computer mouse work (<6% MVC) 238 different MUs (range: 1-31 per person) with more than five firings were identified in the fourteen subjects. During the experimental computer task the 10^{th} and 90^{th} level of the surface EMG ranged from 0.3% to 6.4% MVC and 7.7% to 24.7% MVC, respectively. The mean number of segments during the 30 minutes experimental task was 1705 (range: 5-17294) and the mean duration of MU activity was 233 seconds (range: 1-1710). The MUs were divided into four different groups relating to their duration of activity (Table 1). The majority of the MUs were only active during a part of the experimental session (226 MUs were active during less than 50 % of the experimental task), but a systematic substitution of MUs was not observed.

In two subjects (subject 1 and 2) two and one Cinderella MUs, respectively, were found to be active during the entire experimental task, as shown in Fig. 1.

DISCUSSION

Our results provide evidence of continuously active MUs throughout 30 minutes dynamic long-term measurements while working with a computer mouse. Although the majority of the MUs only were active during a part of the experimental session, an ordered on-off behavior

(e.g. substitution) was not observed. The results from the present study support the Cinderella hypothesis so far as in some subjects long-lasting activity was verified. Thus, if continuous activity actually overloads low thresholds MUs, then there is a potential for selective fibre injuries in low threshold MUs of the trapezius muscle in subjects that are exposed to long-term computer work. Additionally, continuous MU activity seems to depend on personal characteristics.

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TABLE 1: MU activity for the 14 subjects. The four groups were defined due to the duration of MU activity.

	Duration of MU activity where the MU were x % active during the experimental task			
	x<50%	50%≤x<70 %	70 %≤x<90 %	x≥90 %
Total number of MUs	226	7	2	3
Range of duration of MU activity [s]	1-896	907-1244	1266-1297	1610-1710
Range of detected MUAPs	5-11030	5441-13351	6557-8887	11016-17294

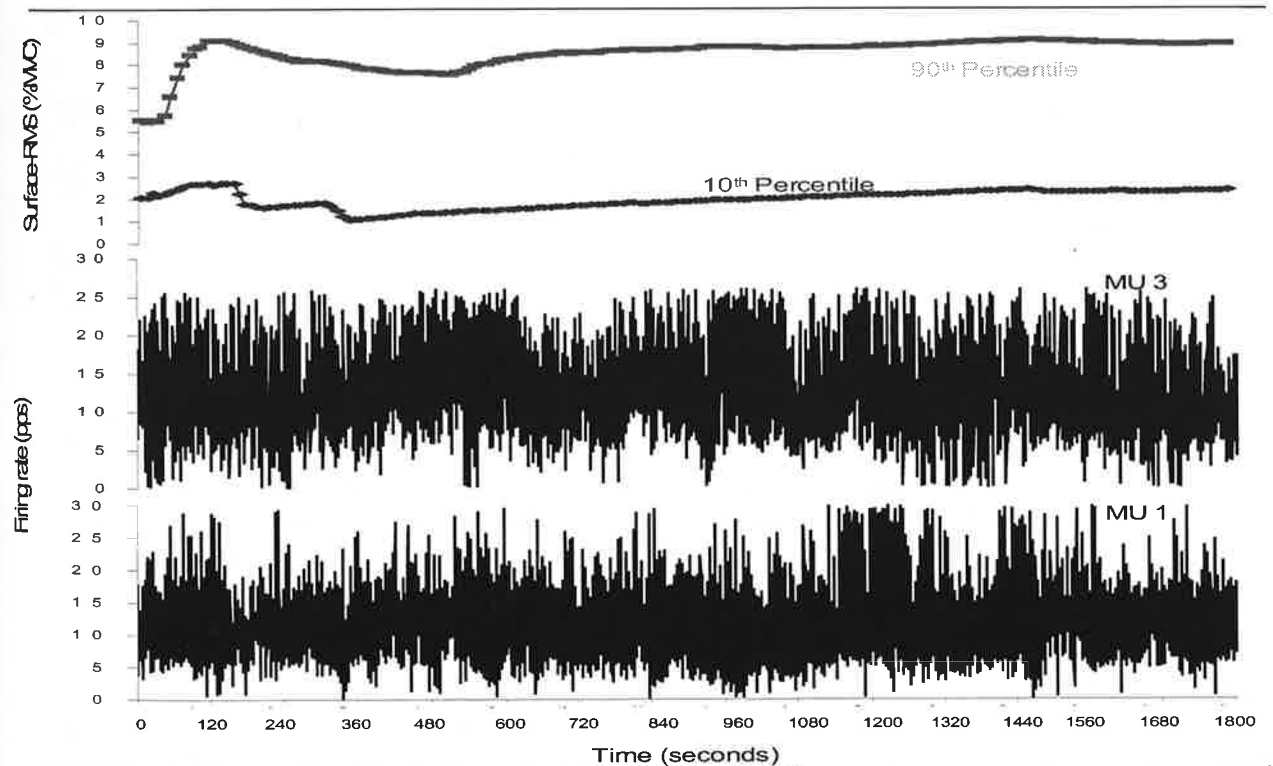

FIGURE 1: Example of instantaneous firing rate estimation and corresponding surface-RMS over time for two Cinderella MUs during the computer work task (subject 1).

MYOELECTRIC MANIFESTATIONS OF STERNOCLEIDOMASTOID AND ANTERIOR SCALENE MUSCLE FATIGUE IN CHRONIC NECK PAIN PATIENTS

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INTRODUCTION

Distinct modifications of the surface electromyography (sEMG) signal can be identified during sustained voluntary or electrically elicited muscle contractions. The analysis of myoelectric manifestations of fatigue can provide important information about physiological changes evolving in the muscle (1, 2). The evidence strongly suggests that there is a relationship between estimates of sEMG variables during sustained isometric contractions and muscle fibre constituency (3, 4).

Muscle biopsy studies of neck pain patients have established the occurrence of muscle fibre transformations in the neck flexors (5). Specifically, Uhlig and colleagues (5) demonstrated that the quantity of type-IIC transitional fibres was significantly increased in neck pain patients and there was transformation of slow-twitch oxidative type-I fibres to fast-twitch glycolytic type-IIB fibres. A reduction in the proportion of slow twitch fibres indicates a decrease of the tonic holding abilities of muscles and this finding supports the clinical mechanical measures which have identified reduced endurance in the neck flexors in patients with neck pain (6, 7).

The purpose of this study was to further the understanding of neck flexor muscle dysfunction in patients with neck pain by studying the fatigue patterns of two neck flexor muscles, the sternocleidomastoid (SCM) and anterior scalene (AS) during sustained sub-maximal cervical flexion contractions in subjects with and without chronic neck pain.

METHODS

Twenty volunteer subjects participated in this study. Patients (1 male, 9 females) were aged between 18 and 47 years (mean 28.7, SD 9.2 years) and had a history of neck pain longer than 1 year (mean 4.7, SD 3.4 years). The asymptomatic subjects (1 male, 9 females) were aged between 21 and 42 years (mean 30.1, SD 7.6 years).

Myoelectric signals were recorded from the sternal head of SCM and the AS muscles bilaterally during sub-maximal isometric cervical flexion contractions at 25% and 50% of the maximum voluntary contraction (MVC). The initial values and rate of change of the mean spectral frequency (MNF), average rectified value (ARV) and conduction velocity (CV) were computed to quantify myoelectric manifestations of muscle fatigue. To compare the rate of change of different variables and allow comparison between subjects, the time course of each sEMG variable was normalised with respect to the intercept of the regression line to produce a "fatigue plot" (1). The neuromuscular efficiency (NME), defined as the quotient of force and the integrated EMG was calculated as the ratio between MVC and the corresponding ARV, measured in N/μV. Individual NMEs obtained allowed an estimate of the amount of electrical activity that each subject produced to reach a value of force.

RESULTS

The Mann-Whitney U test revealed significantly greater estimates of the initial value and slope of the MNF for both the SCM and AS muscles of neck pain patients ($p<0.05$) (see the Figure). This difference was identified for contractions at both 25% and 50% of MVC. The

MVC's recorded in the two groups were not statistically different (52.1±15.7 N in the asymptomatic subjects versus 53.6±13.4 N recorded in the patient group, p=0.76). Despite this finding, neck pain patients demonstrated, although not significantly, less NME with respect to control group for both the SCM and AS during contractions at 25% MVC.

DISCUSSION AND CONCLUSIONS

The results of this study revealed greater fatigability of the SCM and AS muscles of neck pain patients as indicated by the significantly greater estimates of the initial value and slope of the MNF. This significant difference was identified for contractions at both 25% and 50% of MVC. These results suggest: a) a predominance of type II fibres in the neck pain patients and/or b) greater fatigability of the superficial cervical flexors in neck pain patients. These results are consistent with previous muscle biopsy studies in subjects with neck pain, which identified an increase in the number of type-IIC transitional fibres resulting from transformation of slow-twitch type-I fibres to fast-twitch type-IIB fibres, as well as the clinical observation of reduced endurance in the cervical flexors in neck pain patients (6, 7). Since small differences in subcutaneous thickness could substantially affect ARV estimates, we consider that NME should be used with caution.

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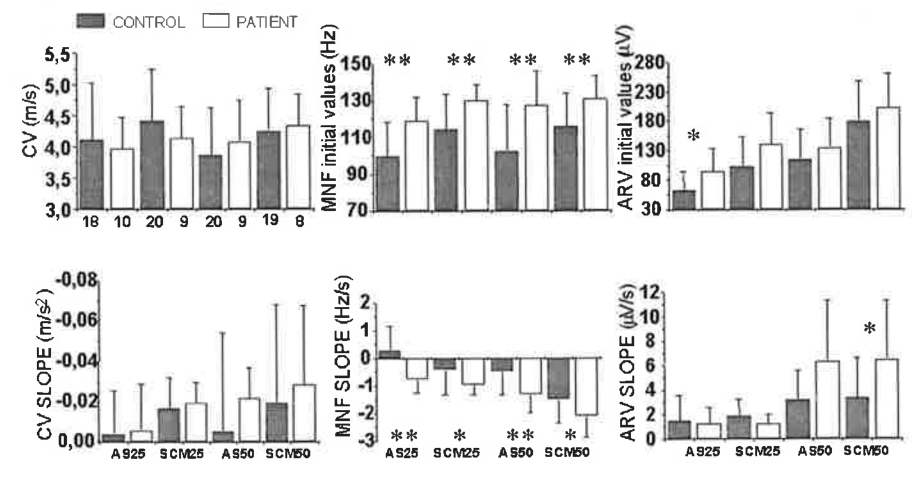

The absolute values (mean and standard deviation) of the initial values and slopes of the three sEMG variables (CV, MNF and ARV) in the two groups of subjects. The symbol * denotes a statistically significant difference (Mann Whitney U test) with $p<0.05$ between the two groups (asymptomatic vs symptomatic), while the symbol ** denotes a statistically significant difference with $p<0.01$.

NOVEL ADAPTIVE FILTER TRAINING SCHEME FOR REDUCING STIMULUS ARTIFACT IN NON-INVASIVE SOMATOSENSORY EVOKED POTENTIAL MEASUREMENTS

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INTRODUCTION

Somatosensory evoked potentials (SEPs) propagate through central or peripheral nerves in response to external stimulation. Non-invasive acquisition of SEPs poses a challenging problem as the observed SEP is obscured by stimulus artifact (SA). Time windowing, frequency filtering, and ensemble averaging cannot be used to remove the SA because the SA and SEP overlap in time and frequency and are coherent with each other. A widely used approach for SA reduction is adaptive noise cancellation (1,2). In this approach, two channels of data are acquired: the primary channel, which contains the SEP corrupted by the SA, and the reference channel, which contains only the SA. The filter attempts to map the SA present in the reference channel to that in the primary channel. This method requires segmented training (2), where only the first portion of the primary and reference channel waveforms are used to train the adaptive element. This ensures that the primary channel does not contain the SEP during training and does not eliminate both the SA and SEP at the output. Once training is complete, the entire reference channel waveform is provided to the adaptive filter to predict the entire SA in the primary channel. With the segmented approach, the adaptive filter has to learn the relationship between the reference and primary channel and must also generalize in time to portions of the waveform that are not provided during training.

METHODS

This investigation proposes a novel approach to SA cancellation, addressing the problem of temporal generalization. An adaptive filter is trained to learn the relationship between the stimulus pulse amplitude and the corresponding SA waveform generated. The input to the filter is the stimulating pulse and the output is the SA waveform observed at the recording site. Training is conducted on data collected during sub-threshold stimulation, where only the SA is present. The adaptive filter learns the impulse response of the process of SA generation, eliminating the need for temporal generalization, as the entire waveform is used. By providing several sub-threshold SA exemplars, the adaptive filter can learn the nonlinear relationship between the amplitude of the stimulus pulse and the corresponding SA waveform. After training, the adaptive filter should have the ability to predict the SA present given a supra-threshold stimulation level. A finite-impulse response neural network (FIRNN) was used as the adaptive element. This filter has demonstrated excellent prediction capabilities (4). A simulated data set was used to quantify the estimation error of the filter, which is not possible with real data, as the true SEP waveform is unknown. A quantitative model of SA generation (3) was used to create SA-like waveforms at varying stimulus current amplitudes. A set of 30 SA waveforms was generated for stimulus current amplitudes between 9.2mA and 15mA. The first 15 waveforms represented training data acquired at sub-threshold stimulation. The last 15 waveforms represented test data acquired at supra-threshold stimulation and were added to a known SEP waveform with both signals overlapping. Performance measures, including the residual error of the SEP waveform after SA cancellation (M1) and the power reduction in the area where both the SA and SEP overlapped (M2) (2), were calculated to validate the effectiveness of the FIRNN to cancel SA in the test data. Results were compared to the segmented technique (1).

RESULTS

The performance measures for both approaches are provided in Figure 1 below. For both methods, M1 improved but M2 deteriorated as the supra-threshold stimulation amplitude increased. Figure 1(a) illustrates how the FIRNN has a lower SEP residual error than the segmented approach for stimulus amplitudes less then 14.4mA. Figure 1(b) illustrates how the segmented approach has a greater SA power reduction than the FIRNN in all stimulus amplitudes.

DISCUSSION

The FIRNN has shown to have better SA cancellation abilities than the segmented approach for stimulus amplitudes 2-3mA greater than the excitation threshold. The FIRNN has more difficulties predicting the SA for stimulus amplitudes much larger than the amplitudes used for training. The FIRNN has shown to perform worse SA power reduction than the segmented approach. This can be accounted for by the ringing observed at the output of the FIRNN in the region of the SA and SEP overlap. The ringing is most likely due to improper FIRNN architecture selection and indicates that further optimization of the adaptive filter is required.

CONCLUSION

A novel approach to SA cancellation using an FIRNN to learn the relationship between the stimulus pulse amplitude and the corresponding SA waveform generated is described here. This technique addresses both the problem of segmented training as well as temporal generalization of other SA reduction techniques. Simulated data has been used to demonstrate that the FIRNN outperforms the segmented training approach to adaptive noise cancellation for stimulus levels 2-3mA greater than the excitation threshold. Optimization of the FIRNN is still requires as well as analysis of this novel technique upon measured data.

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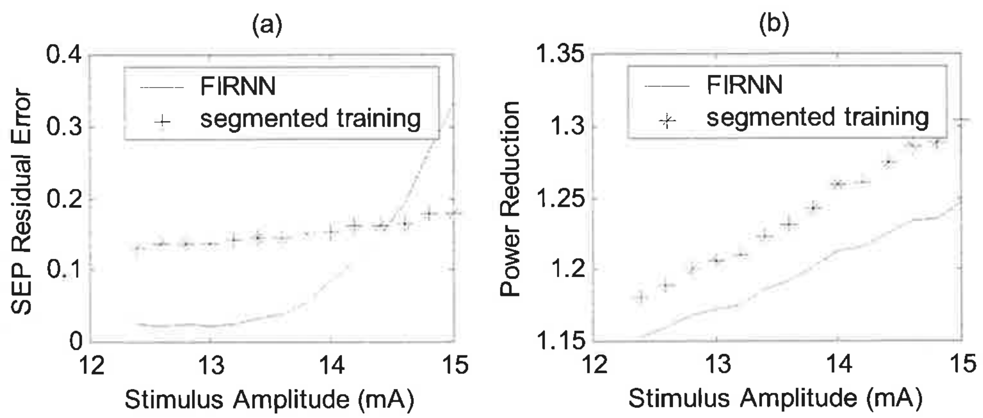

Figure 1: (a) Measure of SEP Residual Error; (b) Power Reduction Measure

EEG CHANGE DURING AND AFTER WALKING IN HEMIPARETIC STROKE PATIENTS

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INTRODUCTION

Physical exercise including gait training has a possibility to modulate the brain activity in post stroke patients. EEG recording after movement is one of ways to study acute effect of physical exercise to the brain activity. However, there are few confirmed evidences supporting such a provoking effect of exercise to EEG. Furthermore, EEG change during systemic movement is unknown because of difficulties of measuring techniques. Walking is a popular method of training for hemiparetic stroke patients and we have previously found that less artifact EEG recording was possible in normal subjects walking with slow speed. In the present study, we investigated the EEG change during and after walking in hemiparetic stroke patients to reveal the influence of walking on the brain activity after a stroke onset.

METHODS

The subjects were 9 hemiparetic stroke patients in the recovery stage. They were all admitted to Tohoku University Hospital to receive medical rehabilitation within 3 months after a stroke onset. The lesion of stroke confirmed by CT or MRI was localized in the cerebral hemisphere contralateral to the paretic limbs. The side of hemiparesis was right in 4 patients and left in 5. The mean age was 59.6 (40~78) years. The mean walking speed with maximum effort at the examination was 31 m/sec. EEG was recorded from 6 points on the scalp according to the international 10-20 system (F3, F4, C3, C4, O1and O2) with a sampling frequency of 1000Hz and digitally filtered with 4-35Hz. The EEG was monitored with telemetry system (MT11) and stored in a hard disk to analyze with PowerLab system. All subjects walked on a wooden floor with bare feet to prevent static electricity from interfering in the proper wave of EEG. EEG recording was performed on the 4 phases as follows: at rest before walking with eyes closed, during walking with preferred speed and maximum effort in 20m distance and at rest after walking with eyes closed. The recording time at rest before and after walking was at least more than 60 seconds. We analyzed the EEG from 4.0Hz to 30.3Hz with power spectral analysis and calculated the values of relative power (power of target band pass/ total power*100) on each recording point and unilateral hemisphere in three band passes: theta (3.9~7.7Hz), alpha (7.8~12.7Hz) and beta (12.8~16.6Hz). The values of relative power in the hemisphere with and without lesion during walking and after walking were compared to those before walking with Wilcoxon's signed rank test.

RESULTS

The values of relative power in theta band showed no significant changes both during and after walking. On the other hand, significant changes were found in alpha and beta band. Since the change of EEG during walking with maximum effort was as same as that with preferred speed, we regarded them as same data. 1) During walking: there was significant decrease of the values of relative power in alpha band on each hemisphere. Particularly on the front-central area of the non-affected hemisphere, the decrease of the values was remarkable. In beta band, significant increase of the values on the affected hemisphere particularly on the occipital area was observed. 2) After walking: there was significant increase of the values of relative power in beta band on the intact hemisphere and on the frontal area of the affected hemisphere. Distributions of these changes are shown in the table.

DISCUSSION

There are few studies reported on an EEG change associated with systemic movement like physical exercise. Kamp and Troost reported that a consistent and prolonged decrease of alpha frequency was observed in some stroke patients through EEG recording immediately after physical exercise on a bicycle ergometer. This is only evidence showing that physical exercise has a possibility to play a role as a provocative method to EEG in stroke patients. In the present study we showed some EEG changes relating to physical task not only after but also during walking. The decrease of alpha activity in the both hemispheres, however, was observed only during walking. Such attenuation of alpha activity may be based on an alpha blocking mechanism, since our subjects closed their eyes while the EEG was being recorded at rest. However, the beta activity increased only in the affected hemisphere during walking. It is difficult to explain this result by alpha blocking hypothesis. According to Mauro et.al., the hemiparetic stroke patients showed the increase of flow velocity during motor task in the middle cerebral artery both contralateral and ipsilateral to the paretic hand performing the task. It is natural that the functioning of the brain in walking is different from that in single hand performance but the healthy hemisphere in stroke patients can be supposed to have an important role even in walking. Therefore, the decrease of alpha power during walking suggests general augmentation of the cortical activity associated with voluntary movement like the inhibition of mu rhythm. The increase of beta activity after walking also has a possibility to show the cortical activation provoked by walking although the reason why it occurred remarkably in the non-affected hemisphere is unclear.

CONCLUSION

The EEG changes during and after walking shown in this study suggest physical exercise can activate the brain function in hemiparetic stroke patients.

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Table. Distributions on the changes of the relative power in alpha and beta band pass.

alpha / during walking

AH NH AH

W	F	F ↓	W
↓	C	C ↓	↓
	O	O	

beta / during walking

NH AH NH

W	F	F	W
↑	C	C	
	O ↑	O	

beta / after walking

W	F ↑	F ↑	W
	C	C	↑
	O	O	

AH: affected hemisphere NH: non-affected hemisphere W: whole of the hemisphere
F: F3 or F4 C: C3 or C4 O: O1 or O2
↑ (↓) : significant increase (decrease) of the relative power

ELECTROMYOGRAPHY OF A COMMON IDIOPATHIC LOW BACK DISORDER

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INTRODUCTION
The physiological or biomechanical processes associated with the development of idiopathic low back disorder are unknown, and are the objective of this paper.

METHODS
An in-vivo feline preparations were subjected to static constant load applied to the L-4/5 supraspinous ligament putting the spine into moderate flexion for 20-minutes. The load and the creep in the ligament and the lumbar spine were measured. Simultaneously, EMG from the multifidus muscles were recorded from the L-3/4, L-4/5 and L-5/6 lumbar levels (feline has seven lumbars) with wire intramuscular electrodes. Load, displacement, creep and EMG were recorded during the 20-minutes of static loading and during seven hours of rest after the static loading.

RESULTS
The results show that during the 20-minutes of static flexion, reflexive EMG in the multifidus was exponentially decreasing and then EMG spasms were superimposed at random throughout the flexion period. During the rest period, the EMG demonstrated hyperexcitability during the first hour, followed by a gradual recovery to normal level. However, the EMG increased to 2 - 3 times that of the normal EMG after the 5 - 7 hours of rest. Therefore, a four component neuromuscular disorder was identified; the loss of reflexive EMG and initiation of spasms during static flexion, initial muscle hyperexcitability immediately after flexion was terminated, and a "morning after" hyperexcitability 5 -7 hours after initiation of rest. Creep values of the supraspinous ligament show elongation of 8 - 25% over 20-minutes depending on the load magnitude. The creep never recovered fully after seven hours of rest.

DISCUSSION
It is suggested that static flexion of the lumbar spine results in creep of its viscoelastic tissues. The sustained creep causes micro-trauma to the collagen structure of the tissues, reflexive pain signals and spasms. Several hours after rest, the micro-damage initiates an inflammation which peaks some seven hours after rest began (e.g., the "morning after" hyperexcitability) and may last several days.

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ERROR DETECTION AND RESPONSE IN PATIENTS WITH PARKINSON'S DISEASE

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INTRODUCTION
Error negativity (Ne) or error-related negativity of event-related potentials may be related to action monitoring and error detection, and the Ne generator is thought to be located in the anterior cingulate gyrus (1,2). A variety of cognitive deficits can be observed in non-demented patients with Parkinson's disease (PD) and these deficits may result from the dysfunction of processes controlled by the prefrontal cortex (3). To evaluate error detection and processing in PD, the Ne components were evaluated.

METHODS
Fifteen non-demented PD (7 males and 8 females; mean =63.5 years) and 16 healthy age-matched control subjects (Controls)(6 males and 10 females; mean=62.1 years) participated in this study. The PD were in stages 1 to 3 of the Hoehm and Yahr stage. Mini-Mental State examination showed no significant differences between the Controls and PD. The patients were administered carbidopa/levodopa and/or dopamine agonist daily, but they had not taken any other drugs that might affect ERP components.
In an oddball-type paradigm, each stimulus, composed of two Japanese phonetic characters ('hiragana' or 'katakana), was presented on a computer monitor. Identical 'hiragana' characters were designated as target stimuli, and the remaining stimuli were non-target stimuli composed of frequent stimuli (different 'hiragana'), rare category deviant stimuli (different 'katakana'), and rare phonetically task-relevant but category deviant stimuli (identical 'katakana'). Each stimulus varied from trial to trial and the subjects performed go/no-go reactions (Task 1).
In a lexical recognition paradigm, stimuli consisting of S1 (word) and S2 (word or non-word) constituted of two 'kanji' (ideogram or morphogram) characters were presented on a computer monitor, and the subjects performed two-way choice reactions (Task 2).
Electroencephalogram epochs were time-locked to the response in each trial and averaged separately for correct and error trials.

RESULTS
The Ne amplitude was significantly smaller in PD than in Controls ($F(1, 29)=13.4$, $P<0.001$ in Task 1; $F(1, 29)=7.2$, $P<0.05$ in Task 2). The Ne latency did not show significant differences between the PD and Controls. In the correct trials, a small negative component with a peak latency of 40msec (CRN) was also detected in PD and Controls, and the CRN component was not enhanced in PD. Although reaction times (RTs) were significantly delayed in PD ($F(1, 29)=22.8$, $P<0.001$ in Task 1; $F(1, 29)=7.8$, $P<0.01$ in Task 2), the RTs in error trials were faster than those in correct trials in PD. In PD, error rates were higher and error correction rates detected by surface electromyograms were lower.

DISCUSSION

Patients with the frontal lesion showed enhanced Ne and CRN components (2). However, in this study, the PD showed smaller Ne and CRN amplitudes, regardless of the paradigms. These findings were comparable with a previous study using a Eriksen flanker task and a Simon-type task (4). As the P3 amplitude in averaged waveforms time-locked stimulus was not reduced in PD, the smaller Ne amplitude in PD might not result from trial-to-trial latency jitter or impaired working memory.

CONCLUSION

The smaller Ne amplitude and the lower error correction rates suggested that error detection and response monitoring were impaired in PD.

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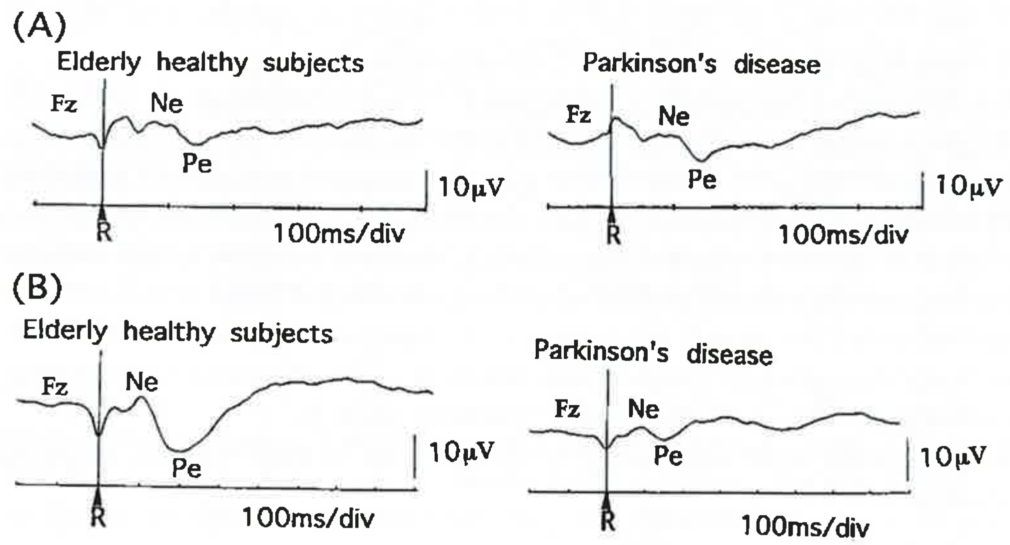

Figure 1

Grand averaged waveforms in the oddball type dual paradigm (A) and in the lexical recognition paradigm (B). The Ne amplitude was smaller in PD.

CHANGES IN FUNCTIONAL ACTIVITY WITH PREDICTION DURING CYCLING EXERCISE

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INTRODUCTION

A power-assisted-bicycle has been popular in Japan for supporting transportation on the road with ups and downs in town. By the power-assisted-bicycle, muscle force is assisted by the electric power while the speed is lower than a predetermined limit. However, a lower speed or a higher torque alone is not the total reference of physical activity during cycling. We focused on the changes in physical activities at each corner with different gradients in a circuit around campus buildings. Since the conditions of physical activity and somatic senses vary depending on the environment and as a function of time, the undertaken motor control needs prediction and it seems to be remarkable at each corner. Moreover, the autonomic nervous activity supports the exercise by controlling cardiovascular system.

METHODS

We measured the electrocardiogram and surface myoelectric (ME) signals as biosignals, and the torque, speed, and cadence as the information on the vehicle. ECG was measured on the chest by using disposable disk electrodes and ME signals were measured by an active four-bar electrode pasted on the subject's skin parallel to the muscle fibers of the vastus lateralis muscle. ECG and ME signals were sampled at 5 kHz with a 12-bit resolution. The information on the vehicle was sampled with the trigger pulses from the sensors equipped on the wheel and the crank. We used the Wavelet transform (WT) for analyzing the heart rate variability (HRV) in a trial and estimated a muscular fatigue related index, $\gamma_{ARV\text{-}MPF}$. The time-series $\gamma_{ARV\text{-}MPF}$ was obtained from the averaged rectified value and the mean power frequency of ME signals. Furthermore, we decomposed bursts of ME signals at the second corner into short-term frequency components by the Matching Pursuit method [1].

RESULTS

Fourteen subjects (11 males and 3 females with 22.9 ± 0.5 years) volunteered in the field experiments. The length of the path was approximately 980 m and there was a steep hill from the second to third corners near the middle of the path. An experiment was composed of several consecutive 2.5-minutes cycling with 2-minutes rest trials. The subjects were asked to pedal the power-assisted bicycle at 60 rpm as possible and to turn left at each corner.

The subjects were separated into two groups, a steeply decreasing in R-R interval (SDRR) group and a gradually decreasing in R-R interval (GDRR) group, referring to the changes in HRV after turning the second corner (descending-ascending). After the second corner, the variation in the amplitude of ME signals was larger at the first stroke than that at the second stroke (Fig. 1). Moreover, the time-frequency representations (TFRs) at the second stroke showed the increase of long-term lower-frequency components under 100 Hz in the last trials. According to the TFRs near the top of the hill the short-term higher-frequency components over 100 Hz appeared, then, they finally decreased in the last trials. In these conditions, the power-assist was effective for the GDRR group.

DISCUSSION AND CONCLUSION

Results seem to be related to the control strategy of individuals because the prediction is required at the turning corner. To control the vehicle noticeably after turning the second corner, the timing of muscle contraction at the first stroke was apt to vary depending on the conditions. On the other hand, the muscle contraction at the second stroke was steady strong and that finally caused muscular fatigue demonstrated in the TFRs. Power was assisted at this part to continue cycling with an appropriate speed. As a result, the torque was the same at each trial, but muscle contraction pattern changed. It means that the predetermined speed dependent control did not achieve sufficient support for muscles. It needs a new strategy of the control scheme based on biosignal information. Analyzing the results, we are planning to develop a personally fitted attractive-vehicle for human in terms of man-machine-system.

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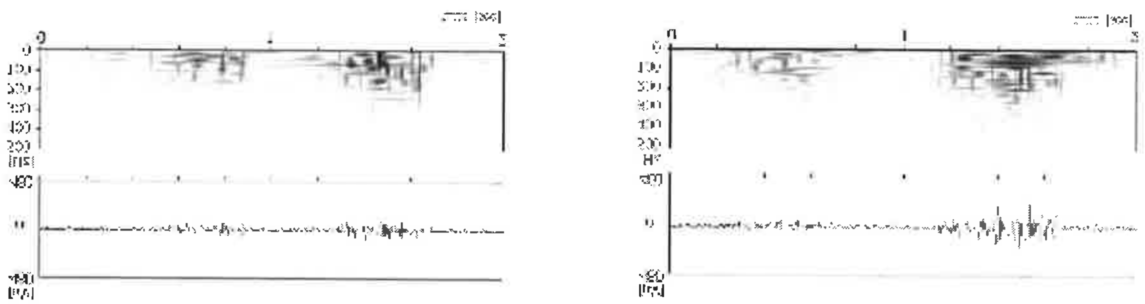

(a) (b)

Figure 1. ME signals and TFRs of ME signals at first two strokes after turning the second corner in a steeply decreasing in R-R interval group: (a) first trial and (b) seventh trial.

CHANGES IN KNEE MUSCLE FUNCTION CAUSED BY ACL CREEP

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INTRODUCTION

In addition being the primary restraint against anterior tibial displacement, the Anterior Cruciate Ligament (ACL) plays a sensory role in the control of knee stability. Like other ligaments and soft tissues, the ACL has viscoelastic properties such as creep, hysteresis and time dependent stress-strain relations. Recent reports have shown that ligament creep is associated with desensitization of the ligament mechanoreceptors, reducing reflexive muscle activity and exposing the joint to instability and injury.

Women have more joint laxity, diminished joint proprioception and higher risk of ACL injuries than men. Joint laxity is also associated with occupational and sports tasks in which the knees are in deep flexion position for periods of time (skiers, catchers, brick and carpet layers); these activities are also associated with higher rates of ACL injuries.
Therefore, the purpose of this study is to examine the relationship between ACL creep and agonist-antagonist muscle coordination in men and women.

METHODS

20 healthy young adults (10 men, 10 women) participated in this study. The skin overlying their rectus femoris and biceps femoris was prepped, and disposable 1 cm diameter Ag/Ag-Cl electrodes were placed over the muscle bellies. A reference electrode was placed over the fibular head. The electrodes were connected to an EMG amplifier with gain of 1000, CMRR of 90 dB, and frequency pass-band between 6-500 Hz.
The subjects were seated in a specially designed chair, and the knee placed at either 35E or 90Eof flexion. The subject was strapped in position. The shank was secured to a padded polypropylene cuff connected through to a load cell. A low friction linear displacement potentiometer was used to estimate ACL creep via anterior tibial displacement. Signals were sampled at 1000 Hz and stored for processing.

The Maximum Voluntary Contraction (MVC) force in isometric flexion and extension were recorded. Following, an anterior load (200 N for men, 150 N for women) was placed on the tibia via a strap and pulley system with the subject relaxed while recording EMG and anterior tibial translation. After the loading period, the subjects again performed maximum voluntary isometric extension and flexion. Then subjects walked 30 m and reported their impression of knee stability. The same protocol was repeated two weeks later at the second knee angle. The order of angles was randomized.
The EMG signals were processed via a mean absolute value (MAV) algorithm (200 msec time window). The peak force and peak MAV for each muscle in each contraction yielded agonist and antagonist activity values for each muscle and peak flexion and extension forces. These were normalized by dividing each by the corresponding value prior to the loading period. ANOVA with repeated measures was used to discern if there were effects of creep, angle and gender on the measured variables.

RESULTS

During the loading period, EMG discharge appeared in 6/20 subjects. The amplitude, frequency and timing were unpredictable, fitting the definition of spasms; these appeared in both muscles at 90E, but only in the hamstrings at 35E.

The EMG of the quadriceps (agonist) in MVC extension increased significantly following creep ($p < 0.01$), with no effect of angle and gender, indicating that the increase was similar at both angles in men and women. No changes werenoted in the hamstrings (antagonist) co-activation due to creep, gender or angle. There was a trend to increased force after creep at both 90E and 35E degrees ($p = 0.09$) with significant effect of gender - women had greater increases than men. Increases were found for hamstrings (agonist) EMG activity in flexion in all groups post-creep ($p = 0.02$), with no effect of gender or angle. There was a trend to increased flexion force ($p = 0.06$), with no effect of gender or angle. The antagonist quadriceps EMG had no change due to creep, gender or angle.

The results of the anterior displacement show significant angle effect, (35E greater than 90E), and a trend of women to have more than men ($p = 0.09$).

DISCUSSION AND CONCLUSIONS

The most important finding in this study is that creep developed in the ACL due to static anterior load on the proximal tibia develop changes in muscle function consisting of spasms and increased contractile ability of the agonist muscles resulting in increased net torque. Increased Agonist EMG activity and corresponding force is not accompanied by compensation from the corresponding antagonists. The increased force and EMG activity is also more prominent in women than in men. It was also found that the anterior displacement of the tibia after anterior loading was more prominent at 35Eof flexion than at 90E, with trends of more displacement in women than in men.

Relatively short static loading of the ACL may cause a transient neuromuscular disorder that may increase the risk of knee injury. The data available from recovery studies strongly supports the concept that athletes or workers should allow at least a 24-hours of rest between periods of activities that exposes the ligaments to creep.

In conclusion, these results suggest that ligament creep has a significant effect on neuromuscular coordination; this effect is a decrease of agonist muscle inhibition without change in antagonist muscle co-activation. We propose that prolonged ligament tension subjects joints to increased risk of instability and potential injury.

ACKNOWLEDGEMENT

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Table 1: Mean ± SD of Normalized Values Post Creep

	Men 35°	Men 90°	Women 35°	Women 90°
Quads Agonist EMG	1.10±0.29	1.05±0.28	1.27±0.48	1.16±0.32
Quads Antagonist EMG	0.14±0.06	0.14±0.08	0.16±0.09	0.17±0.08
Extension Force	1.01±0.14	0.98±0.13	1.19±0.37	1.08±0.18
Hams Agonist EMG	1.05±0.22	1.13±0.32	1.08±0.27	1.13±0.20
Hams Antagonist EMG	0.29±0.38	0.34±0.26	0.27±0.17	0.29±0.17
Flexion Force	1.01±0.05	1.20±0.36	1.05±0.15	1.00±0.15
Displacement (mm)	4.31±1.73	3.53±1.85	6.36±2.67	4.55±1.53

"FUNCTIONAL PARESIS" AFTER ANTERIOR CRUCIATE LIGAMENT (ACL) INJURY

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INTRODUCTION

After ACL- operation the restoration of normal muscle function is one of the goals in rehabilitation. Muscle function can be described by a sensorimotor model consisting of a mechanical part and a neurophysiological part. The coordination of these two parts is disturbed after any operation of the knee. In many cases this disturbance is severe enough to produce a temporary inactivity of the quadriceps femoris muscle. This can clinically be described as functional paresis. If this paresis lasts long enough all the problems of immobilisation appear. To improve the rehabilitation process one has to examine the amount of this disturbance for to set the right therapy program. Especially one has to analyse whether or not an electric stimulation of the quadriceps muscle is necessary.

During maximal voluntary contraction (MVC) additional electric stimuli can be used to recruit not voluntarily activated motor units (MU). The goal of electrical stimulation is that these electrically activated MU are lead back in the normal process of contraction and as a consequence in the process of sensorimotor function. As soon as the sensorimotor function is restored it is possible to start with a balance and endurance training process.

METHOD

To analyse the mechanical part together with the neurophysiological part of our model it is necessary to measure electromyographic activity as "input signal" and muscle force as "output signal" simultaneously.

Between 7 and 12 week after ACL operation (patellar tendon plastic) 76 male patients (29 ± 7 years old) were examined.

Patients had to sit in a custom made chair with 90 degree hip and knee flexion, adjusted to the anthropometric situation of each patient with the pelvis and trunk fixed by a belt system. After an acoustic starting signal patient had to perform an isometric extension of the knee with MVC over a period of 10 seconds, an exercise before the actual measurement took place. During this time the nervus femoralis was stimulated at the 4th and 6th second each time by a double supramaximal rectangular impulse with 500ms duration and 500ms interval. Examination started with the nonoperated site first.

EMG was recorded with surface discs electrodes (Ag/AgCl with gel, diameter of 0,5cm, interelectrode distance 2cm) after shaving, skin preparation with alcohol and application of a crème to decrease skin resistance from rectus femoris, vastus medialis and lateralis along the muscle fibres. A differential amplifier was used (CMRR >95 db).

Isometric Muscle force (IMF) was recorded by a strain gauge sensor. Both signals were A/D converted with 1kHz, EMG was bandpass filtered between 5 Hz and 350 Hz and further processed by calculating root mean square (RMS) between second 4 and 6.

RESULTS

On the injured site RMS of the r.fem., v.med. and v.lat. was reduced to 63 %, 60 % and 53 % respectively compared to the contralateral nonoperated site. IMF decreased to 51% (healthy:

605 ± 110 N; injured: 310 ± 99 N). No stimulation related increases of IMF could be found on the healthy site whereas on the injured site 77% of the patients showed an increase of IMF of 15% in the mean. Furthermore considerable oscillations could be seen in the mechanogram (strain gauge signal). Disturbances of IMF and oscillations could be found in 4 different combinations:
1. stimulus related increase alone (15% of all patients); 2. considerable oscillations alone (12%), 3. both IMF increase and oscillations (62%) and 4. IMF decrease alone without stimulus related increase during MVC (11%).

DISCUSSION
After ACL operation a "functional paresis" could be found in the majority of the patients. This kind of paresis is defined by a reduction of IMF during MVC which can be increased by supramaximal electrical stimulation of the nerve. This means that there is a reduction in voluntary activation of a certain amount of MU. The higher the amount of voluntarily not activated MU, the more IMF increases and the more IMF can be increased by electrical stimulation. On the nonoperated site we could not see any of these conditions.
Paresis also means inactivity and this is in some way immobilisation. There are a lot of problems well described after immobilisation like decrease of aerobic capability of the muscle group, decrease of capillary densitiy, increase of soft tissue, decrease of muscle fibre diameter and so on. Therefore it is evident, that this functional paresis has to be treated as soon as possible after operation if it occurs and it occurs in the majority of our patients.

Two main stages can be subdivided:

1. qualitative alteration
reduction of IMF but no stimulation related increase during MVC;
decrease of the MU firing rate and / or pathophysiological synchronisations without recruitment insufficiency (decrease of RMS, oscillations);

2. quantitative alteration
reduction of IMF with insufficiency of the voluntary MU recruitment (stimulation related effect during MVC) for all parts of the quadriceps muscle. Mostly this could be seen in combination with one of the qualitative alterations. This stimulation related increase of the strain gauge registered IMF can be as big as 40% of IMF (MVC) of the nonoperated site.

CONCLUSION
In postoperative rehabilitation it is important to know the functional status of the muscle. Especially one has to know if a functional paresis exists because paresis leads to inactivity and this can create a lot of long term problems. Different therapeutic concepts (electrotherapy and exercises) have to be applied for to activate the whole muscle. By combining EMG and Mechanogramm it is possible to detect a functional paresis when the nerve is supramaximal electrically stimulated during MVC and an increase of IMF can be found. Functional paresis stage 1 can be treated with some forms of exercise therapy. A functional paresis stage 2 can only be treated by electrotherapy and after a "learning" period exercise therapy can be applied. Training of balance or endurance which are very important in postoperative programs can only be meaningful when there is no functional paresis any more.

DEVELOPING A DRYLAND MEASURING SETUP FOR DETECTING THE MUSCLE ACTIVITY DURING GRAB START IN SWIMMING

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INTRODUCTION
For short distance races in swimming an excellent starting technique can be decisive to win the race or not to. To find the right balance between developing a maximum horizontal impulse and leaving the block as quick as possible it is necessary that the neuromuscular coordination is optimized. Depending which start technique is used (grab start, track start, ...) you can find slightly different activation patterns. The current literature provides us with less information about the electromyographical activation during the swim start. The aim was to detect the muscle activation pattern. Because of measuring the EMG signals in wet conditions makes special demands against the equipment, we had to think about possibilities to simulate the start movement.

METHODS
For save practice training gymnastics uses a pit filled with foam pieces. So our thought were that it must be possible to perform swim starts into this foam pit. To compare the starting techniques between different trials and to detect differences in the execution of jumps into the foam pit and the pool a 2-D kinematic analysis system (APAS) was applied. Also at the edge of the pit a block was mounted on a force plate to control the force development during starting (start phase) and to compare the symmetry of the two jumping conditions.
For measuring the muscle activity the subjects had to jump into a pit filled with foam pieces. For the detection of the muscle activity a 8-channel mobile EMG system (biovision) with surface electrodes and bipolar detection was used. The signals were stored in a datalogger worn on the lower back side of the subject for further processing. Force sensors on the fingers and the big toe were applied to detect leaving the block. The following timing events like starting signal, hands leaving and leaving the block were measured. The further processing were performed with self written Matlab routines. Pre activation, onset time, peak activations were calculated from 8 muscles of the lower limb.

RESULTS
The starting technique between the pool and foam pit seems to be equal after some practice time. The horizontal and vertical force development shown in Table 1 were comparable. The kinematics showed only minor differences as shown in Fig.2 for the hip angle. Also the kinematics for the knee and ankle joint were similar. So the movement into the foam pit was comparable with that into the pool.
Fig. 1 shows an overview of the collected EMG data of five selected muscles. At different swimmers different pre activation levels were found. Also minor differences of the onset time by swimmers within the same competition level were detected.

DISCUSSION
With this measurement setup it is possible to detect the activation pattern during swim starts. But some points should be considered. First the swimmer should not be afraid to jump into the foam pit. Some swimmers need more practice jumps than others but all of them should do some practice jumps to get familiar with the measurement setup. To control the execution of

the start movement two dimensional video analysis and force plate measurement should be performed for jumping into the foam pit and the into the pool. During our measurement series one swimmer was excluded from the analysis because he were not able to perform a correct jump into the foam pit.

CONCLUSION

This knowledge helps us to get a better understanding of the neuromuscular activation process during the start movement. To determining deficits or to improve the start techniques this kind of analysis is helpful and useful but practice training should be provided before applying this method.

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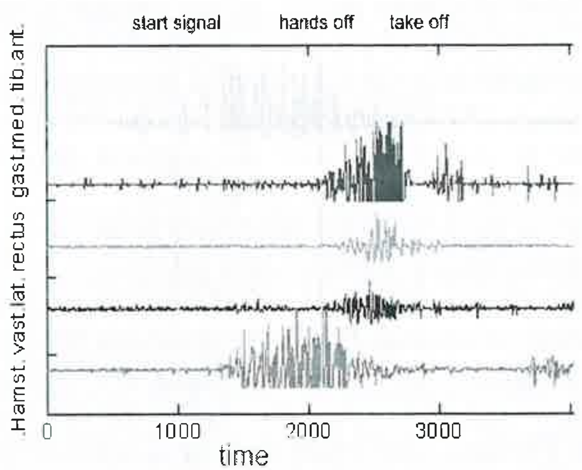

Fig 1. EMG pattern of 5 muscles

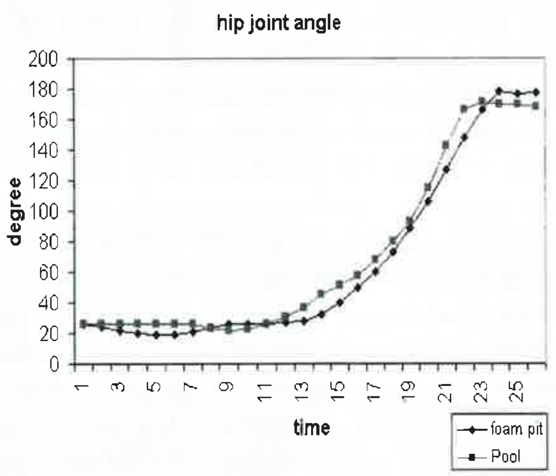

Fig 2. hip kinematic from the pool and the foam pit

	pool	*foam pit*
Force vertical mean[N]	709 ± 17	747 ± 12
Force horizontal mean [N]	498 ± 15	460 ± 13
Time of contact [ms]	633 ± 16	631 ± 21
Impulse vertical [Ns]	450 ± 26	474 ± 16
Impulse vertical [Ns]	317 ± 13	292 ± 14

Table 1: Comparison of pool and foam pit start

AN ANALYSIS OF VERTICAL JUMP FOR WOMEN VOLLEYBALL PLAYERS

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INTRODUCTION

Bad landing technique is related to a number of sport related injuries. During a normal set a volleyball player performs around 170 to 190 jumps (Kollath,1996), a basketball player performs around 80 to 100 jumps per game (Hagedorn et al., 1996), not to mention other sports such as artistic gymnastics and handball. Injuries related to landing can happen due to several factors, among them: direct contact with opponent players, muscular fatigue, loss of corporal balance control or faulty landing. Good landing techniques can reduce the risk of injury significantly (Stacoff et al., 1988; Valiant & Cavanagh, 1985). Gerberich et al. (1987) conducted an experiment with volleyball players and concluded that 63% of the injuries were related to landing deficiencies. This can be supported by Stacoff et al. (1988) that found that the forces after jumps for blocking (volleyball) are between 1000N to 2000N on the feet and between 1000N to 6500N on the heel, showing the enormous stress upon the lower limbs. In this work, the authors describe a quantitative analysis of counter-movement jumps at maximal effort with and without arms impulse. It was decided to focus the study on the evaluation of the landing phase in order to find out the correlation between vertical landing forces and the various parameters for kinematic and dynamic movements.

METHODS

A group of 15 female volleyball players, aged 16 to 20 (mean=17.5; standard deviation=1.26) was used as the experimental population. Each athlete was asked to perform three jumps without arms impulse and three jumps with arms impulse. The jumps were executed on barefoot to avoid any influence of different types of footwear in the results. The exercises were monitored via a piezoresistive platform associated to a data acquisition system and dedicated software. The reaction force on the platform was calculated by each modality of jump and correlated with anthropometric data, flight time, impulse, maximal power, damping time, landing time and impulse time. The Pearson's correlation analysis was used to determine the relation between the independent and the dependent variables. The t Student test for dependent variables was used to identify the difference between both types of jumps ($p \leq 0,05$).

RESULTS

Table 1 shows the descriptive statistic mean, minimum, maximum, standard deviation and coefficient of variability for some dynamic and kinematic variables measured during the jumps. The initial results show that 86.7% of the studied athletes have peak impact forces during landing higher than 5G (5 times their body weight), indicating inadequate landing techniques. There were also athletes reaching 11.3G and 12.6G for peak landing forces. It was then decided to study the relationship between the maximum landing force and the damping time (time lapse between the first contact with the platform - landing - and the moment when the peak force was developed - full contact). The results show a high statistical negative correlation (without arms impulse: -0,8113; with arms impulse: -0,5902) between the maximum impact force and the damping time. The results also show that jumps executed with arms impulse had a much better performance (height) then those executed without arms impulse.

DISCUSSION
As good damping is generally associated to good landing techniques, it is possible to infer that most of the players in our experimental group present a lack of good techniques, and therefore are prone to possible injuries in the future. The analyses showed a significant negative correlation between the maximum force and the duration of the damping phase for both types of jump. These results suggest that the landing force can be reduced if more time is spent in the deadening of fall (damping).

CONCLUSION
We can conclude that the damping phase plays an important role in the jump's deadening and therefore should be addressed as an important feature to prevent a number of injuries related to vertical jumping. As it was expected, the use of arms impulse also plays an important role on the performance of the jump, specially with respect to the maximum height, but its execution and synchronization must be carefully addressed in order to get the most of the impulse generate by the swinging of the arms.

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Variables	**Mean**	**Minimum**	**Maximum**	**Std.Dev.**	**CV**
Jump_N (cm)	36,54	30,46	49,17	5,190	14,20
Damping_time_N (s)	0,064	0,037	0,101	0,017	27,47
Landing Force_N (G)	7,22	4,14	12,63	2,431	33,65
Jump_Y (cm)	42,83	35,93	55,73	6,117	14,28
Damping_time_Y (s)	0,058	0,037	0,088	0,013	22,57
Landing Force_Y (G)	7,32	4,42	11,34	1,836	25,09

TABLE1 - Mean, minimum, maximum, standard deviation and coefficient of variability for the variables. (_N refer to jumps performed without arms impulse and _Y refer to jumps performed with arms impulse).

SURFACE MYOMECHANICAL ACTIVITY RECORDED ON A LASER DOPPLER VIBROMETER

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INTRODUCTION
Muscle contraction evokes surface vibrations with spectral components ranging from 6 to 50 Hz. Motor unit firing, dimensional changes and lateral bulk movements of activated fibers, as well as tissue resonant properties have been proposed as source for these vibrations, see [1] for review. Over weak contracting muscles further cardiac evoked patterns, so called "micovibrations" or "resting muscle sounds" contribute to surface activity [2]. Commonly accelerometers or contact microphones are used to record these vibrations. These pickups are rigid bodies and if coupled to the elastic skin surface this could result into spectral distortions of the source signal (excitation of a tissue-transducer resonance). Such distortions are avoided with optical sensing. In this study a high resolution Laser Doppler vibrometer (Type PDV 100, Polytec, measurement range 0.1 m/s, resolution < 0.1 µm/s) was tested to record surface vibrations in a weak and a strong contraction paradigm. For comparison the vibrations were also recorded on an accelerometer.

METHODS
The study (N = 6) was performed in a room with low seismic noise. The accelerometer (EGAX-5, Entran, piezoresitive type, measurement range 50 m/s^2) was calibrated against the Laser Doppler vibrometer by means of a piezoelectric disk translator. For skin fixation (double adhesive tape) the accelerometer was glued into a flat plexiglas disk with 10mm diameter. Myomechanical activity was recorded over the motor point of m. flexor digitorum superficialis. Isometric power grips (target contractions) with a duration of 20 s were performed. Both vibration signals were filtered (1- 100 Hz) before sampling (500 Hz). Spectral estimates were performed over eight non-overlapping 2 sec intervals (1024 point FFT). From this spectra root mean square (RMS) amplitude and mean frequency (MF) values were calculated in the 5 – 50 Hz frequency band.

In the weak contraction paradigm (0, 2, 4, 6, 8 and 10% of MVC) the Laser spot was focused on the accelerometer. To extract cardiac evoked vibrations also an ECG was recorded. The extraction was performed by R-wave triggered averaging [2], resulting into a value (SYNC) between 0 (no cardiac evoked vibrations) and 1. For comparison with the accelerometer signal, the Doppler output (= velocity) was differentiated once. The correspondence of both acceleration signals was expressed by the maximum of the absolute valued cross-correlation function (MCC), with values between 0 (no correlation) and 1.

In the strong contraction paradigm (0, 10, 20, 40 and 60% of MVC) two recordings were taken at each target level. In the first recording the laser spot was focused on the skin attached accelerometer and in the second recording the spot was directly focused on the skin surface. To determine the tissue-accelerometer resonance, slight pencil tapings at a distance of about 2 cm, proximal to the recording site, were applied during the contractions. The resonance frequency F_{RES} was estimated from the zero crossings in the response oscillation. The relative amplitude distortion (RAD) in the resonance region was estimated in a bandwidth of 5 Hz around F_{RES}.

RESULTS

Weak contractions: only at the fully relaxed state (0% MVC) cardiac evoked vibrations dominated the velocity signal (SYNC = 0.79). The RMS amplitude of this resting surface activity was 58 µm/s (MF = 11.2 Hz) and 4.72 mm/s^2 (MF = 16.3 Hz) in the acceleration domain. MCC was 0.76, due to electrical noise in the accelerometer. With contraction MCC approached to 1 (0.92 at 4% MVC) and SYNC to 0 (0.09 at 6% MVC), the RMS amplitude increased almost linearly. At 2% MVC, MF increased to 16.9 Hz (23.1 Hz in acceleration domain) as with contraction higher spectral components came up.

Strong contractions: With skin attached accelerometer the F_{RES} values were 22.3 Hz (0 % MVC), 29.8 Hz, 31.7 Hz, 37.2 Hz and 45.7 Hz (60 % MVC). Without accelerometer F_{RES} was not estimated since many tapings elicited no resonant responses on the skin surface (often only a pulse wave was recorded, especially at medium contraction levels). RAD values were +12.3% (0% MCV), +10.3%, +8.7%, +9.3% and +17.1% (60% MVC). Over the full frequency band (5- 50 Hz) no significant differences in RMS amplitude and MF values were found.

DISCUSSION

The results indicate two benefits of optical vibrometry in surface myomechanical recording. One is the high resolution achieved with velocity sensing. Consider therefore a 5 Hz surface deflection (sinus) with an amplitude of 1 µm. In a Laser Doppler vibrometer this deflection produces a signal of about 31 µm/s which is well above the systemic noise level. On an accelerometer the same deflection produces a signal amplitude of about 1 mm/s^2. For a small sensor in the 5 g range (≈50 m/s^2) this signal level approaches the systemic noise level. Such a noise component was observed during recording the relaxed musculature, with velocity sensing however, the microvibrations of the relaxed musculature were captured precisely. Therefore, when resting muscle activity or small muscles are of research interest, optical vibrometry is advantageous.

A second benefit is, that there exits no mechanical interaction during recording. The results clearly have shown, that contraction induced surface activity produces secondary resonance oscillations due to mass loading. The origin of this skin-transducer resonance lies in the viscoelastic properties of the skin and subcutaneous layers, whose oscillation dynamics is still fairly understood. By insertion of a line scanner (mirror-galvanometer) into the optical path, also spatial information can be retrieved from the skin surface. Such an advanced optical sensing method may help to better understand the skin vibration dynamics.

CONCLUSION

The high resolution and wide dynamic range makes a Laser Doppler vibrometer predestinated to resolve small and low frequency vibrations over relaxed and weakly contracted muscles. A skin-transducer resonance, emerging from tissue loading, is avoided.

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MOTOR UNIT PROPERTIES REVEALED BY MECHANOMYOGRAM (MMG) AMPLITUDE AND FREQUENCY ANALYSES

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INTRODUCTION

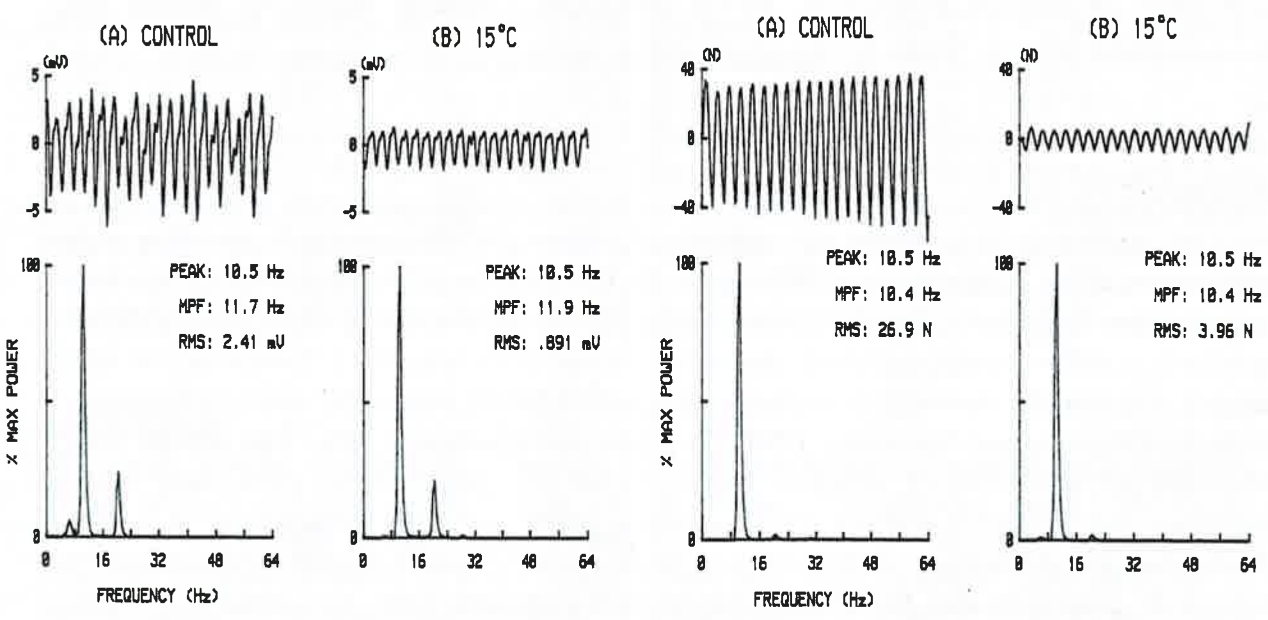
Previous studies have indicated that mechanomyogram (MMG) amplitude and frequency components might represent the underlying motor unit (MU) recruitment and firing rate (rate coding)(Barry and Cole, 1990; Orizio et al., 1993, 1996; Bichler, 2000). Interestingly, MMG amplitude actually decreases at higher force levels at which MUs might be firing at tetanic rates, causing a fusion-like contraction leading to diminished MMG amplitude, while its frequency increases (Moritani and Yoshitake, 1998; Yoshitake and Moritani, 1999). The present study examined the possibility of revealing human motor unit properties by performing MMG amplitude and frequency domain analyses during graded voluntary contractions as well as during electrical stimulations of the triceps surae muscles. Effects of muscle temperature upon contractile properties and the corresponding MMG changes were also examined during experimentally-induced hypothermia.

METHODS

Two identical microphone sensors (10 mm diameter, mass 5g, bandwidth 3-2000 Hz) for MMG recording were fixed to the center of the belly of medial gastrocnemius (MG) and soleus (SOL). Single twitch and repetitive stimulations (10 Hz) were performed during room temperature and hypothermia conditions (15, 20, and 25°C). During voluntary contractions, MU and MMG activities were recorded at 20, 40, 60, and 80% MVC. Effects of mixed micro-stimulations were also studied by stimulating two MUs at 5-10, 10-20, 8-12, and 12-24 Hz, respectively; while simultaneously recorded evoked mass action potentials (M-wave) remained constant.. In addition, isolated MU fatigue trials were performed at 12 Hz for a period of 2-min in order to determine the relationship between muscle contractile slowing and the corresponding MMG amplitude and frequency components.

RESULTS and DISCUSSION

The rms-MMG of MG increased as a function of force ($P < 0.01$). On the contrary, these values for SOL increased up to 60% MVC ($p < 0.01$), but then decreased at 80% MVC due to possible MU fusion resulting in smaller muscle dimensional changes. Similarly, a significant reduction in the muscle contractile properties (Peak force, maximal rate of force development and relaxation, contraction and half-relaxation times, etc.) caused by the experimental hypothermia also resulted in significant reduction in MMG amplitude with subsequent fusion at low stimulation frequency. Different stimulation frequency trials indicated that there were highly significant and progressive reductions in the force fluctuations from 5 Hz to 50 Hz that were almost mirrored by the similar and significant reductions in the MMG amplitudes. Mixed stimulations to different MUs clearly demonstrated that both MMG and force recordings showed two distinguished peak frequencies that were delivered to the underlying MUs.



Lastly, our fatigue study with prolonged stimulation at 12 Hz demonstrated that MMG amplitude decreased progressively as contractile slowing occurred as a function of time. These data strongly suggest that MMG analyses could be employed to study MU fusion properties together with recruitment and rate coding characteristics during various experimental conditions.

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APPLICATION OF MECHANOMYOGRAPHY TO THE ESTIMATION OF MOTOR UNIT ACTIVATION STRATEGY IN VOLUNTARY MUSCLE ACTIONS

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INTRODUCTION

During voluntary contractions, it has been found that changes in time and frequency domain parameters of the mechanomyogram (MMG) as a function of force reflect similar properties of the motor unit (MU) recruitment and firing rate (Akataki et al. 2001; Orizio et al. 1990; Orizio et al. 1992). In addition, we reported that the time-frequency analysis of non-stationary MMG during ramp contractions allows the attainment of a high force resolution in MMG/force relationship and provides more detailed insights in to the MU activation strategy in young healthy male adults (Akataki et al. 20001). The present investigation was designed, as an application of the MMG method, to examine the effect of increasing age on the MU recruitment and activation in the biceps brachii muscle, using the MMG/force relationship during isometric ramp contractions.

METHODS

Ten elderly men ranging in age from 66 to 79 years (69.8±4.7 years, mean±SD) and 15 young men (mean age, 22.7±1.8 years, range 21 to 26) were recruited for this study. All the subjects were free of any known neuromuscular or musculoskeletal diseases. The subject was seated in a chair with his right upper limb positioned on a horizontal platform. Each subject was asked to exert an isometric elbow flexion force that followed a ramp target force presented on a computer screen. The target pattern consisted of a stable contraction at 10% MVC for 3 s, followed by a linear progression of submaximal contractions from 10% to 80% MVC at a constant rate of 10% MVC/s.

The isometric elbow flexion force was measured by a strain gauge force transducer, which was coupled to a metal frame attached to the subject's wrist. The force signal was amplified through a strain amplifier. The MMG was detected by a uniaxial accelerometer with a frequency response of DC to 150 Hz, which had the shape of a rectangular prism with a base of 16 mm x 13 mm and height of 5 mm, and mass of 2.1 g. The MMG signal was amplified and filtered by an AC amplifier with a bandwidth of 1 to150 Hz.

The amplified MMG and force signals were digitized at a sampling rate of 5,000 samples/s. The short-time Fourier transform was used to calculate the root mean squared amplitude (RMS) and mean power frequency (MPF) of the non-stationary MMG during ramp contractions. From a time series of the MMG recording, short segments with a period of 0.6 s were cut every 0.1 s using the Gaussian window function with a standard deviation of 0.3 s. The power spectral density function for each segment was then estimated by the fast Fourier transform algorithm. The RMS and MPF in each data segment were computed from the power spectral density function. They were aligned with force as the RMS/force and MPF/force relationship. In addition, the rate curve was computed from the differentiation of the relationship.

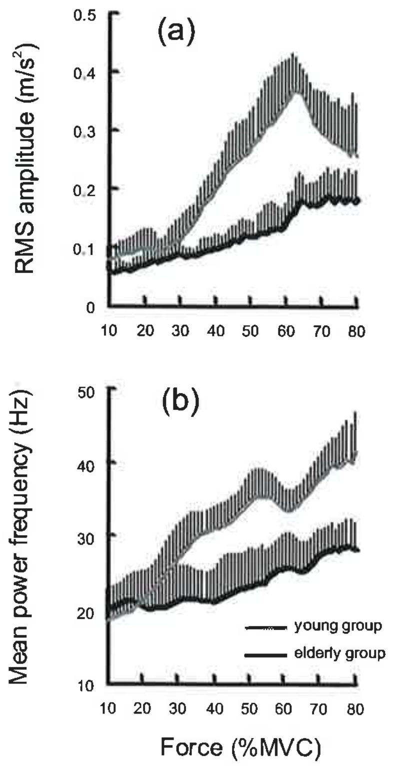
RESULTS & DISCUSION

The MVC of biceps brachii muscles in the elderly (41.3±8.4 Nm) was approximately 70% of that in the young group (59.1±6.8 Nm). The difference between the two groups was statistically significant ($P<0.01$).

Figure (a) shows the RMS/force relationship as a function of relative force (%MVC). The RMS/%MVC relationship included two inflection points in both young and elderly groups, which were identified by its rate curve. In the young group the RMS/%MVC relationship was characterized by an initial slow increase followed by a rapid increase from 25.3±2.8% MVC and a progressive decrease beyond 62.1±3.1% MVC. In the elderly, the RMS increased progressively with increasing force up to 57.6±3.4% MVC, but its increment was smaller. Then, a rapid but brief increase was followed by a stable trend beyond 63.6±3.7% MVC. As shown in figure (b), the MPF/%MVC relationship in the young group was characterized by four contiguous regions based on three inflection points as follows: (1) a relatively rapid increase up to 36.7±3.0% MVC, (2) a slow increment between 36.7% and 55.7±3.8% MVC, (3) a temporary reduction from 55.7% to 63.0±2.7% MVC, and (4) a further rapid increase above 63.0% MVC. By contrast, in the elderly only two inflection points were observed in the MPF/%MVC relationship, i. e., a slow and progressive increase in the MPF up to 59.4±2.3% MVC, followed by a slight and temporary reduction from 59.4% to 64.3±2.0% MVC, and then a progressive increase again.

The RMS/absolute-force relationship in the elderly was lower than that in the young group throughout submaximal levels of force exerted (not illustrated). Significant differences between the two groups were observed at force levels above approximately 20 Nm ($P<0.05$). The MPF/absolute-force relationship in the elderly was somewhat greater than that in the young group at absolute force levels below 12 Nm, but the difference was not significant. This trend was reversed at force levels higher than 12 Nm. The difference between the two groups was emphasized with increasing force and became significant above 16 Nm ($P<0.05$).

These findings suggest an alteration in MU activation strategy, with a predominant role for MUs with slow-twitch fibers and an effective fused tetanus induced at lower firing rate of the MUs, resulting from age-related neuromuscular changes.



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MMG BUT NOT EMG INCREASES WHEN MAXIMUM FINGER TAPPING IS PERFORMED WITH INCREASED MOVEMENT AMPLITUDE AND EXTERNAL NOISE

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INTRODUCTION

In the study of muscle activity mechanomyography has proven to add information complementary to the electromyographic information, e.g. during fatiguing contractions that may result in soreness (1,2). In the present study we challenged MMG recordings in highly dynamic movements that also lead to the generation of external noise.

METHODS

Twelve subjects performed maximum frequency of finger movement during table tapping (TT) and computer mouse clicking (MC), which elicited a "tapping" and a "clicking" noise, respectively. Movement frequency and amplitude were analysed from video recordings. Additionally, two of the subjects repeated the series of TT and MC as above in combination with three other series: 1) finger movement was performed "in the air" (no noise) with approximately the same amplitude as for TT, 2) the subjects placed their hand on the table as for TT while a colleague produced the "tapping-noise", and 3) the subjects placed their hand on the mouse as for MC while a colleague produced the "clicking-noise" close to the subject's hand. MMG was recorded above the extensor digitorum muscle using a piezoelectric accelerometer; and EMG was recorded with surface electrodes placed on either side of the MMG transducer in the longitudinal direction of the muscle. Both signals were analysed for RMS values.

RESULTS AND DISCUSSION

Maximum frequency of finger movement for the 12 subjects was similar during TT and MC and in the range of 5-6 Hz. However, the movement amplitude was in the order of 10-20 mm for TT and only about 1 mm for MC, respectively, resulting in significant differences in finger movement velocity and acceleration between TT and MC (tabel 1). EMG mean (SD) was similar during TT and MC: 7.8 (2.9) %EMGmax and 7.0 (3.2) %EMGmax, respectively. In contrast, MMG was 0.187 (0.132) ms^{-2} during TT but only about half that value during MC.

Table 1. Mean values (SD) on movement variables, EMG and MMG

	Table tapping (n=12)	Mouse clicking (n=12)
Frequency (Hz)	5.5 (0.56)	5.5 (0.84)
Amplitude (mm)	15.0 (7.7)	0.89 (0.15)
EMGrms (%max)	7.8 (2.9)	7.0 (3.2)
MMG (ms^{-2})	0.187 (0.132)	0.093 (0.066)

For the two subjects repeating TT and MC the mean MMG values (two trials for each subject) were 0.146 ms^{-2} and 0.073 ms^{-2}, respectively, and considered representative for the 12 subjects. Interestingly, the TT movement with "no noise" resulted in an MMG value of 0.078 ms^{-2}, while the "tapping-noise" produced by a colleague and without muscle activity in the recorded muscle resulted in a value of 0.022 ms^{-2}, and the "clicking-noise" correspondingly in a value of 0.009 ms^{-2}. Assuming that MMG records the combination of external noise, movement and muscle contraction, the data may be systematized as in table 2. The last two columns with italic letters are estimates of the isolated contribution to the MMG from contraction and movement, respectively. For these estimates it is assumed that the movement during MC relatively to TT is so small that it can be disregarded. Thus, the contraction fraction contributing to MMG is easily estimated to be 0.064ms^{-2} by subtracting the value for noise from the MC value. For TT the movement fraction can be estimated as: [(0.146-0.022) – (0.073-0.009)] = 0.060 ms^{-2}. From this it follows that the contraction fraction for TT is 0.146 –(0.060+0.022) = 0.064 ms^{-2}. Caution must of course be taken regarding these estimates since they are based on only 4 experiments on 2 subject, but interestingly, the contraction fraction was the same during TT and MC. This contraction fraction may correspond to the EMG activity, which was also similar during these tasks (table 1).

Table 2. Mean values of MMG (ms^{-2}) from 4 experiments on 2 subjects.

	+contraction +movement +noise	+contraction -movement +noise	+contraction +movement -noise	-contraction -movement +noise	+contraction -movement -noise	-contraction +movement -noise
TT	0.146		0.078	0.022	*0.146-0.082 = 0.064*	*0.124-0.064 = 0.060*
MC		0.073		0.009	*0.073-0.009 = 0.064*	

CONCLUSION

In conclusion, the noise produced by TT and MC did contribute to the MMG to different degrees. However, the major fraction of the MMG signal related to the muscle contraction while another significant contribution related to the velocity or acceleration of the movement that may be caused by factors other than the contraction of the investigated muscle.

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MECHANOMYOGRAPHIC AND SURFACE ELECTROMYOGRAPHIC BEHAVIOUR DURING MUSCLE FATIGUE DEVELOPMENT AT LOW CONTRACTION LEVEL

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INTRODUCTION

At high force contraction levels, the development of localised muscle fatigue is generally associated with increased amplitude and decreased frequency contents of the mechanomyographic (MMG) and surface electromyographic (EMG) signals [2, 4, 6]. At sub-maximal continuous and/or intermittent force contraction level, the changes in the time and frequency domain mentioned above are not always seen [3] underlining that some of the fundamental properties of fatigue development at low contraction level are not fully understood. The objectives of this work were to investigate the behaviour of the mechanomyographic (MMG) and surface electromyographic (EMG) signals during muscle fatigue development at sub-maximal continuous and/or intermittent force contraction level with and without feedback.

METHODS

Experimental protocol:
Part 1: the subjects (N=6) were asked to perform elbow static intermittent or continuous elbow flexion at 10 and 30% of their maximal voluntary contraction (MVC) with visual feedback or proprioceptive feedback.
Part 2: the subjects (N=11) were asked to perform 45° and 90° bilateral continuous arm flexion and abduction at approximately 20%MVC without feedback.
Data recordings and analysis: A piezoelectric accelerometer was used to record the MMG signal from the biceps brachii (part 1) and the upper trapezius muscles (part 2). Bipolar EMG surface electrodes were placed on each side of the accelerometer (37 mm apart). Both signals were sampled at 1 kHz. Prior to analysis, the MMG and EMG signal were band-pass filtered at 2-200 Hz and 10-400 Hz respectively. The rating of perceived exertion was also recorded for both parts. The root mean square (RMS) values and the mean power frequency (MPF) values of the MMG and EMG signals were computed and normalised with respect to the maximum value and/or the initial value.
Statistics: Two-way repeated measures ANOVA and three-way ANOVA with Student-Newman-Keuls (SNK) method for multiple comparisons were applied. $P<0.05$ level was considered significant.

RESULTS

At low contraction level, the general pattern (part 1 & 2) for MMG and EMG RMS value changes was an increase while the EMG MPF values decreased with contraction time in the biceps brachii and upper trapezius muscles ($P<0.05$). However, the observed patterns were different in magnitude and behaviour.

In part 1, the increase in RMS values was generally more marked for the MMG compared with the EMG while the decrease in MPF values was more consistent for the EMG compared with the MMG signal. During the intermittent contractions the main effect was on MPF for both EMG and MMG. Larger rating of perceived exertion ($P<0.05$), greater slopes of EMG and MMG RMS and MPF values vs. time were observed with proprioceptive feedback compared with visual feedback.
In part 2, The MMG RMS and MPF value changes (normalised or not) followed the subjects' rating of perceived exertion reported in the different arm positions. For the EMG, this was only true for non-normalised RMS values. Normalised EMG RMS values and normalised or not MPF values evolved with different pattern to the one observed for the MMG signal. Furthermore, MMG amplitude and spectral changes were not correlated to their EMG counterparts.

DISCUSSION & CONCLUSION
Buchanan and Lloyd [1] reported differences in EMG activity during force control or position control non-fatiguing contractions. Our results are in line with this and suggest that the mechanisms behind localised fatigue development may be dependent of the feedback modality. Moreover, different centrally mediated motor control strategies may be used with visual and proprioceptive feedback.
EMG and MMG signals bring complementary information about the muscle electrical and mechanical activity in terms of MU recruitment and firing pattern [5, 6]. Furthermore, MMG seems to provide an objective way to assess muscle fatigue development for the upper trapezius muscle and may be a valuable technique in Ergonomics.
Finally, The fact that the MMG and EMG signals responded differently in terms of magnitude and behaviour of the changes in the time and frequency domain and the lack of correlation between MMG and EMG signal during fatigue development suggest that the MMG and EMG signals may reflect different phenomena and give complementary information on localised muscle fatigue at low level contraction.

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THE INFLUENCE OF AGE ON A LEVEL OF MUSCLE ELECTRICAL AND MECHANICAL COACTIVATION DURING MVC AT DIFFERENT ELBOW ANGLE

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INTRODUCTION
During the single-joint maximal isometric contraction, electrical activity of agonist muscle is accompanied by an antagonist muscle activity (called coactivation or cocontraction), which can be affected by a joint angle (muscle length). In addition to the force developed during contraction, the mechanomyogram (sound myogram; MMG) can also be detected from a muscle. It is well known that in skeletal muscle there is a reduction in the number of motor units and muscle fibers in aged subjects (Doherty et al., 1993), which is suggested to co-exist with a motor units (MU) activation pattern similar to that in young (Esposito et al., 1996). There is also a lost of FT motor units resulting in a decrease of muscle force in old people. However the changes depend on tested muscle, i.e. in the biceps brachii (BB) Doherty et al. (1993) did not found a change in the relative proportion of the MU with type I and II fibers. If the same muscle is considered in agonist and antagonist function, the changes which take place in the muscle with aging process should similarly affect electrical (EMG) and mechanical (MMG) activity in both functions. Thus, the agonist to antagonist ratio (ANT/AGON) should not change with age.
The aim of the study was to test the influence of age on a level of muscle electrical (EMG) and mechanical (MMG) coactivation at a different elbow joint angle during maximal voluntary contraction (MVC).

METHODS
Young males (YM; age 23 yr; n=21) and females (YF; age 20 yr; n=22), and old males (OM; age 67 yr; n=18) and females (OF; age 65; n=17) were tested six times. Because there was no sex differences, data were analyzed in two age groups: young (YM+YF; age 22 yr; n=43) and old (OM+OF; age 66 yr; n=35). The first and second sessions were done to establish optimal angle (Lo) for elbow flexors and extensors. The EMG and MMG signals from biceps brachii (BB), brachioradialis (BR), and triceps brachii (TB) muscles were measured at Lo, as well as at the angles that were smaller ($Ls = Lo - 30^o$) and bigger ($Ll = Lo + 30^o$) during elbow flexion (session three and five), and extension (session four and six). All testing sessions consisted of five trials of 2 or 3-sec MVC at each angle with simultaneous recording of EMG and MMG. The BIODYNA dynamometer was used to measure torque versus time for right elbow flexor muscles. The custom-made EMG/MMG recording device was used to record EMG and MMG signals. To assess a level of coactivation, the root mean square (RMS) of EMG and MMG signals of TB, BB, and BR muscles were calculated. Then, $EMG_{(ANT/AGON)}$ and $MMG_{(ANT/AGON)}$ ratios were calculated as a RMS EMG or MMG signal of a muscle working as an antagonist divided by the muscle working as an agonist, respectively.
MANOVA for repeated measurements and Tukey post-hoc comparisons were used for statistical analysis.

RESULTS
Table 1 shows values of the $EMG_{(ANT/AGON)}$ and $MMG_{(ANT/AGON)}$. The $EMG_{(ANT/AGON)}$ for BB and BR muscle was bigger in old subjects, compared to young, and increased at Lo and Ll compared to Ls in both groups. In the TB muscle, the $EMG_{(ANT/AGON)}$ decreased at Ll,

compared to Ls in young, while it increased in old. There was a lack of significant changes of $MMG_{(ANT/AGON)}$ for BR and TB. In BB, there was a decrease of $MMG_{(ANT/AGON)}$ with age. The $MMG_{(ANT/AGON)}$ increased at Lo and Ll compared to Ls in BB and BR in young but not in old subjects. In the TB muscle the changes with the joint angle were not significant in both groups.

DISCUSSION

In contrast to what was expected, the present study showed an increase of electrical activity of a muscle working as an antagonist relative to the maximal electrical activity of that muscle working an agonist at the three joint angles in old subjects, compared to the young, with the biggest differences in BB and BR muscle, and smallest in TB. Since weakness of joint, bones, tendons and ligaments is a characteristic of elderly people, an increase of $EMG_{(ANT/AGON)}$ found in old subject, compared to young, indicate an increasing role of the antagonist with age in protection against excessive stress of e.g. ligaments.
The changes of $MMG_{(ANT/AGON)}$ with age were muscle dependent; for BB and BR the changes with joint angle were different in young and old, but for TB were the same.

CONCLUSION

The present study showed that an increase of electrical activity of a muscle working as an antagonist with age is independent of the joint angle and muscle tested, while changes of the mechanical activity is angle and muscle type dependent.

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	Age		TB			BB			BR		
			Ls	Lo	Ll	Ls	Lo	Ll	Ls	Lo	Ll
EMG ANT/AGON	Y	X	0.463	0.343	0.306	0.126	0.154	0.186	0.107	0.127	0.157
		SD	0.490	0.308	0.245	0.093	0.150	0.149	0.068	0.072	0.086
			#			* # +	* +	* +	* # +	* # +	* +
	O	X	0.308	0.356	0.445	0.267	0.311	0.366	0.193	0.203	0.227
		SD	0.207	0.344	0.372	0.178	0.298	0.316	0.111	0.131	0.131
			#			^		^	+	+	+
MMG ANT/AGON	Y	X	0.817	0.843	0.731	1.160	1.395	1.636	1,047	0.906	1.349
		SD	0.378	0.541	0.492	0.692	0.966	1.027	0,600	0.597	0.824
					*	# +	* + ^	* +	# +	#	+
	O	X	0.871	0.971	1.102	0.906	0.843	0.851	1.299	1.349	1.247
		SD	0.743	0.573	0.677	0.787	0.499	0.542	1.491	0.824	0.827
							^	^	+	+	

Table 1: The values of the $EMG_{(ANT/AGON)}$ and $MMG_{(ANT/AGON)}$ in young (Y) and old (O) subjects at optimal angle (Lo), and at the angles that were smaller (Ls) and bigger (Ll) than Lo, in TB, BB, and BR muscles. * = $P \leq 0.05$ compared to old subjects; # = $P \leq 0.05$ compared to Ll; + = $P \leq 0.05$ compared to TB; ^ = $P \leq 0.05$ compared to BR.

BICEPS BRACHII MOTOR UNITS ACTIVATION PATTERN AT FATIGUE: AN MMG INVESTIGATION

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INTRODUCTION

In the last two decades it has been explored the possibility to follow the mechanical activity of the motor units (MUs) by means of transducers (microphones, accelerometers, piezoelectric contact sensors) detecting the muscle surface displacement due to the pressure waves generated by the dimensional changes of the active fibres within the muscle. The resulting biological signal has been named surface mechanomyogram (MMG). Changes of the MMG time and frequency domains parameters, as a function of the percentage of the isometric maximal voluntary contraction (% MVC) and hence as a function of the motor units activation pattern, has been described in several papers (Akataki et al., 2001; Madeleine et al. 2001; Orizio, 1993; Stokes & Blythe, 2001). Fatigue, influencing the mechanical performance of the motor units fibres; may induce some changes in the motor units activation strategy including changes in the motor units recruitment and firing patterns (for a review see Orizio 2000). On these basis this work aims to verify if the changes in the motor units (MU) activation pattern due to fatigue may be described by the surface mechanomyogram analysis.

METHODS

Eleven male subjects, 25 - 40 years old, participate in the study. The investigated muscle was the biceps brachii during isometric contraction. The angle of the elbow joint was kept at 115°. The force was measured by a load cell strapped perpendicularly to the subject wrist. The MMG was detected by an accelerometer (Entran EGA 25 D, bandwidth 0-800 Hz, dimensions: 1 x .5 x .5 cm, mass .5 g, sensitivity 3 mV/G) fixed to the skin, over the muscle belly, by a double adhesive tape. The actual level of effort, % of the maximal voluntary contraction (MVC), was provided to the subject by means of a monitor together with the force target. The requested effort increased linearly from 0 to 90% MVC (10% MVC/.75 s). The fatiguing exercise was an intermittent contraction at 50% MVC (6 s on – 3 s off) until the subjects was not able to reach the target. At that moment a new ramp, in which the 90% MVC corresponded to the 45% MVC of the unfatigued muscle, was administered. On 16 0.5 s MMG time windows were calculated the RMS and mean frequency (MF) of the signal spectra (estimated by FFT). To avoid the analysis of transient the considered windows covered the 15 to 85 %MVC range effort step 5%.

RESULTS

RMS vs %MVC: from 15% to 55% MVC the RMS increased more than 3 times, then it decreased and halved at 85% MVC. At fatigue the RMS presented the maximum at 15% than decreased reaching a stable value in the 45-85% MVC range. *MF vs %MVC:* in the fresh muscle the MF increased in the 15-40% MVC range and beyond 60% MVC. A stable value was presented in the 40-60% MVC range. In fatigued muscle the MF values were lower that in the fresh muscle and presented an increase only in the 40-60% MVC range followed by a plateau up to 85% MVC.

DISCUSSION

The increment of the MUs recruitment and of the MUs firing rate increase and decrease the MMG RMS, respectively (Orizio et al., 1993). The firing influence may be due to the fusion process of the mechanical events. It also seems that the MMG MF reflects the global MUs firing rate (Akataki et al., 2001). On these bases our results suggest that in fatigued muscle the recruitment of the larger and more fatigable motor units, greatly contributing to the MMG generation, given their more superficial distribution in biceps, is lost. In fact a) the MMG-RMS increase up to 55% MVC is not present and b) the MF does not show the large increase beyond 60% MVC (it seems that only fast MUs attain high firing frequencies). Given the interferential nature of MMG the elongation of the mechanical response of the single motor units contributions at fatigue may also contribute to explain the lower MF values at each contraction intensity as well as an easier non linear summation of each MU contribution (Orizio et al. 1996) during recruitment determining a MMG reduction already at low level of effort.

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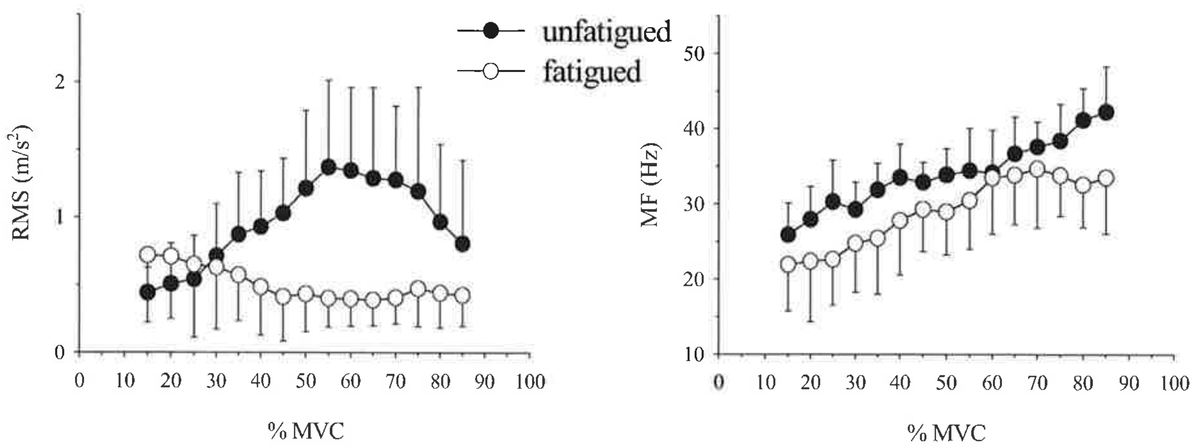

Figure 1. RMS and MF as a function of the relative effort increase. Note the lack of RMS increase in fatigued muscle and the plateau of the MF in the 60 – 85% MVC range.

ELECTROMYOGRAPHY AND MECHANOMYOGRAPHY AS INDICATORS OF LONG LASTING ELEMENT OF MUSCLE FATIGUE

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INTRODUCTION

Recently, it has been suggested that the mechanomyogram (MMG) can be used as a tool to detect long lasting element of fatigue after intermittent contractions at 50% MVC (1). The aim of this series of experiments was to investigate amplitude and spectral changes in the MMG and electromyogram (EMG) after fatiguing contractions at low force levels.

METHODS

Test contractions of 5% and 80% MVC were performed before and 10 and 30 min after intermittent fatiguing contractions at 30% (IFC30) and 10% MVC (IFC10) as well as before and after a sustained fatiguing contraction at 10% MVC (SFC10). A piezoelectric accelerometer was used to record the MMG signal from the biceps brachii and the extensor carpi radialis muscles. Bipolar EMG surface electrodes were placed on each side of the accelerometer (37mm apart). Root mean square (RMS) values and mean power frequency (MPF) values of the MMG and EMG signals were calculated for a 15 s period during the test contractions, when the force was stable and at the target level.

RESULTS

The EMG analysis of the 80% test contractions showed an increased RMS and a decreased MPF after IFC30 but no changes after IFC10. In contrast, the 5% test contractions showed an increased RMS and a decreased MPF after FC10 but no changes after FC30. No changes were found for the MMG. After the SFC10 a prolonged increase was found in the RMS values of both the EMG and MMG in the 5% test contraction but no changes were found in the MPF. Further, in the 80% tests no changes were found in any of the signals.

DISCUSSION & CONCLUSION

The difference in response between the 80% and 5% test contractions indicates that the test contraction must be performed at a force level primarily involving the fatigued muscle fibres in order to detect the long lasting element of localized muscle fatigue after low force contractions. Otherwise the changes may be masked by the contribution of the non fatigued part of the muscle. Although the more intense fatiguing contractions e.g. higher force level or sustained contraction, shows concomitant changes in the MMG and EMG signals, the intermittent fatigue protocols in this study show that the mechanical properties, as revealed by the MMG signal, are recovered after 10 min, although the electrical properties are still deteriorated. This indicates that the fatigue related changes in the mechanical and electrical properties of muscles may be dissociated and that the combination of the MMG and EMG signal most likely reveal different aspects of the localized long lasting muscle fatigue.

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ELECTROMYOGRAPHY OF THE TEMPORALIS AND MASSETER MUSCLES IN CHILDREN WITH RIGHT UNILATERAL CROSSBITE

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INTRODUCTION

Malocclusion is considered the third bigger deontological problem on public health (SILVA FILHO et al., 1989). The unbalances on muscular contraction patterns are a significant component of almost all malocclusions (MOYERS, 1991). The malocclusion approached on this study was the right unilateral posterior crossbite (RUPC), observed when the vestibular cusps of a single or more upper teeth occlude lingually with lingual cusps of lower teeth (MOYERS, 1991).

This problem is frequent on deciduous dentition and causes a unilateral chewing in young children, changing the midline to an abnormal position, and causing facial asymmetry (COHEN, 1979; PROFFIT, 1991; PLANAS, 1988). The RUPC also can produce muscular disbalance and temporomandibular joint by unilateral work (PLANAS, 1988; FELÍCIO, 1999).

This study aimed evaluates possible electromyographic (EMG) sign alterations of superficial masseter and anterior portion of temporal muscles, bilaterally, to verify muscular hyperactivity on both sides.

METHODS

Twenty girls (mean age= 7.4, standard deviation= 0.50) composed the sample. All of them had RUPC in three teeth, at least. The EMG signal were acquired with bipolar surface electrodes. The raw EMG signals were digitalized by a 12 bit A/D converter board, Butterworth filter, low-pass of 509Hz, high-pass10, 6Hz, 100 times of gain, and 1000Hz sample frequency (Lynx Electronics). Superficial EMG recordings were obtained on four masticatory conditions: habitual, right, left and bilateral chewing and maximal voluntary isometric contraction was used to normalize the electrical activity amplitude. During the recordings, two minutes of rest were realized to avoid muscular fatigue (DeLUCA, 1997). All subjects chewed pieces of rubber tubes to improve occlusal vertical dimension.

The EMG processing was realized studying ten masticatory cycles chosed by each subject on a specific situation and directing on two studies: 1) EMG amplitude normalization (Bionic routine-Matlab software); 2) RMS (square root average) to get the percentage of each muscle activation evaluated by Variance Analysis with Random Surveying using Tukey test when $p<5\%$.

RESULTS

The results signed that the right masseter muscle amplitude was higher than left masseter muscle ($p<5$) on right and habitual chewing, suggesting a prevalence of right unilateral chewing on the sample. During left chewing, the right masseter muscle didn't show significant variation of amplitude and this value was almost the same of right and habitual chewing, suggesting a muscular hyperactivity pattern on crossbite side. At right and habitual chewing, the anterior portion of left temporal muscle was hyperactive and left masseter muscle was hypoactive.

DISCUSSION

Muscular hyperactivity of right masseter muscle in RUPC subjects was related by many authors, like GOMES et al. (1998), TROELSTRUP & MÖLLER (1970) and ALARCÓN, MARTIN & PALMA (2000), however, this authors found a masseter bilateral symmetry, suggesting a masticatory adaptation in subjects with crossbite.

A hyperactivity related on anterior portion of temporal muscle when this muscle is compared with the masseter muscle at crossbite opposing side was confirmed by PLANAS (1988), SIMÕES (1985), KEELING et al. (1991) and GOMES et al. (1998)

CONCLUSION

These results may conclude that, on these experimental conditions, EMG amplitude of right masseter muscle and anterior portion of left temporal muscle were higher than others muscles studied on RUPC girls.

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A NEW MEASURE OF FOOTPRINTS IN CHILDREN WITH NORMAL LONGITUDINAL ARCH AND PHYSIOLOGIC FLATFOOT ANALYZED BY COMPUTERIZED PHOTOGRAMMETRIC PROTOCOL

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INTRODUCTION

Numerous studies[1,2,3,4] were done to identify and quantify physiologic flatfoot (PF). Various techniques are reported to assess the medial arch height, including radiographic measurements and footprint analysis, which are the most commonly used methods[4]. To evaluate the height of the medial arch using footprint, we could change the real arch due to problems with the ink, discomfort for the individuals in they have your feet demarcated with ink, possibility to stain in the paper leaves causing alteration in the format of the impression and how to determine the reliability of footprint measurements[5,6,7].

The purpose of this study was to evaluate and measure PF and normal foot with footprint analysis and a new method that we have called by Computerized Photogrammetric protocol.

METHODS

The subjects of this study were 25 children diagnosed as having PF and normal longitudinal arch. The age varied from 7 to 12 years old and thirteen were boys and twelve were girls.

First, we used the footprint analysis to see which foot was normal and PF. We put the tint in the plantar aspect of the foot and a static footprint was recorded for each foot during half body weight-bearing position. For each foot, the widest part of the arch and the heel were measured, and the former value was divided by the latter to calculate the arch index for each foot, as described by Staheli et al.[3].

In the new method (figure 1) the child was asked to stand in the glass of podoscopy and the arch was filmed and recorded using the software Alcimage 2.1. Again , we did calculate the arch index for each foot, as described by Staheli et al.[3] According to Staheli et al.[3], the normal values of arch index have broad values from 0.3 to 1.0. Values above 1.0 are indicative of flatfoot through adulthood.

The Wilcoxon test was used to verify correlation between both methods of footprints.

RESULTS

We achieved 33 normal arch and 17 flatfeets in both methods. In the footprint analysis they happened 28 cessations and 125 footprints stained that we could not measure exactly the arch index. In this method, the total time to measure all the arch index was of 12 days. In the method of the Computadorized Photogrammetry it did not happen cessations and not stain in the footprints. The total time to measure all the arch index was of 4 hours.. We did not find statistically significant differences between the two methods.

DISCUSSION

When we used ink for verification of the longitudinal arch happened many cessations and stain in a lot of footprints harming and putting back the calculation of the arch index. In the method of Computerized Photogrammetry, with ink absence, this was not verified and the time of calculation of the arch index was fast.

CONCLUSION

In conclusion, we advise this new method because it presents a lot of advantages. These advantages are practicality, comfortable for subjects, quickly in analyzes of data, electronic analyzes and registers with possibility to send these data by e-mail.

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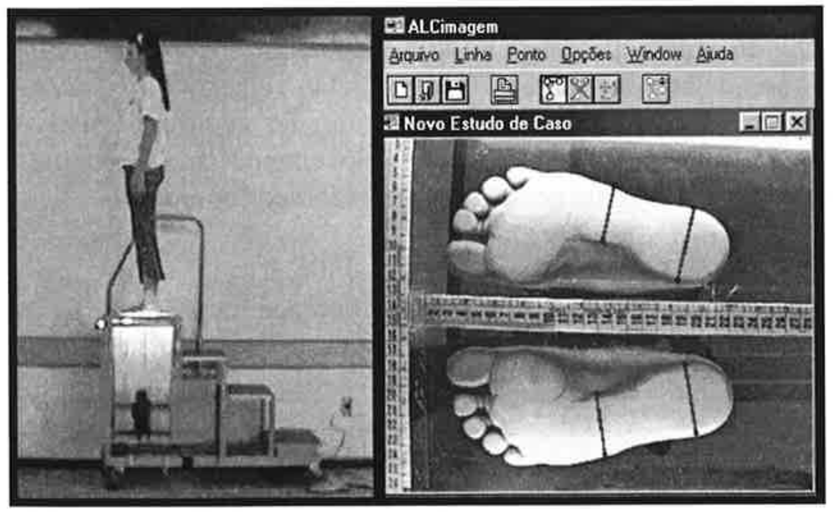

Fig. 1 – Obtaining and analysis of the longitudinal arch for Computerized Photogrammetry.

EVALUATION OF PEAK TORQUE IN CONCENTRIC ISOKINETIC KNEE EXTENSION EXERCISE

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INTRODUCTION

Patellofemoral pain syndrome (PPS) is one of the most common problems of the knee encountered by rehabilitation clinicians (6). It is characterized by a malalignment of the extensor mechanism as the result of an atrophy and general decrease of the quadriceps femoris force (7), especially the atrophy and the dysplasia of the medial compartment, characterized by the insufficiency of the Vastus Medialis Obliquos (VMO) muscle. The conservative treatment of PPS is the most usual and it is supported on the selective strengthening of the quadriceps femoris muscle (3), especially the VMO muscle (4). Isokinetic dynamometers have been used to evaluate muscle performance by the measurements of parameters such as peak torque. Therefore, this work aims to investigate the isokinetic concentric peak torque (PT) of the quadriceps femoral muscle for knee extension exercises in health and asymptomatic subjects with PPS.

METHODS

The PT (Nm) has been measured via an isokinectic dynamometer (Biodex Mult-Joint System2) for sixteen sedentary women divided into two groups for analysis - eight clinically normal subjects (22.50 ± 2.50 years) and eight PPS asymptomatic subjects (21.00 ± 1.20 years). All subjects read and signed an institutionally approved Consent Form, before proceeding onto the study (Resolution 196/96 of the National Health Committee).
Prior to the experiments, the subjects were submitted to a 5 minutes worm-up section for the lower limbs using an ergometric bicycle with no load at speeds of around 25 km/h. Seven movements of knee extension (0 to 90°), at angular velocity 90°/s, were executed on the Biodex machine by each subject in sitting position. To avoid any discrepancy due to non familiarization with the machine (irregular movements and velocities), the first two repetitions were neglected and the next five were recorded for analysis.

RESULTS

The t-test revealed that there was no significant difference (p=0.34) for the mean values of PT (Nm) obtained for the normal group and the PPS group (TABLE).

DISCUSSION

Our results showed that there was no differences in the PT between the 2 groups. These data are in consonance with (1) and (2), that used conventional strain gauges devices to carry out the experiments. On the other hand, our results do not agree with the studies of (7) that found a significant increase on the PT for normal subjects when compared to subjects with chondromalacia patellae. Nevertheless, there are methodological differences between the studies that may explain those differences. The authors evaluated the PT at 60° of knee flexion and the group evaluated was greater and symptomatic (41 subjects with

chondromalacia patellae grades 1 to 4 and 31 normal subjects). It is also possible that asymptomatic patients do not have weakness in their quadriceps muscle.
Therefore the nature and extent of quadriceps atrophy in PPS has both theoretical and clinical implications. Current technology can not show the individual contributions of the different parts of the quadriceps muscle (5). Consequently, the problem of imbalance between the VMO and VL muscles should be studied using others instruments, such as the electromyography and/or morphological variables (the proportions and fiber areas).

CONCLUSION

The results showed that, for the experimental conditions used in this work, the quadriceps femoral muscle was able to develop the same concentric isokinetic PT in both groups, suggesting that the force reduction may not be the primary etiologic factor of the PPS. However, the performance of other parameters should be also investigated, such as work (Nm) and power (Watts) associated to the electrical activity of the VMO, VLL and VLO muscles.

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ACKNOWLEDGEMENT
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	X	SD
Normal group	114.76	38.45
PPS group	106.91	23.94

TABLE : Mean and standard deviation of the concentric peak torque during knee extension movement at an angular velocity of 90°/s in the Biodex dynamometer for normal group (n=8) and PPS group (n=8).

THE EFFECT OF POSTMORTEM FREEZING STORAGE ON THE TENSILE PROPERTIES OF TENDON

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INTRODUCTION

Postmortem storage is inevitable in biomechanical evaluation of living tissues simply because it is not feasible to perform tests on live animal or immediately after sacrifice. For human specimens, it is unlikely to obtain fresh specimen rather than fresh frozen cadaveric specimens. Besides, the complex methodologies for biomechanical testing of connective tissues normally require a considerable length of time to obtain results from a substantial number of specimens.

Further, proper storage of cadaveric tendons and ligaments that are used as allografts is necessary to preserve the biomechanical properties and biological viability for successful transplantation.

Freezing is the most popular storage method for soft tissues preservation as it is relatively cheap in operation cost and easy to use. Researchers have investigated the response of soft tissues to freezing storage but results have been inconclusive. The aim of this study is therefore to elucidate the influence of postmortem freezing storage on the tensile properties of tendons.

MATERIALS AND METHODS

Two hundred and seventy-five chicken Flexor Digitorum Profundus (FDP) tendons of the middle digits were used in this study. Five pieces of tendon specimens were tested within 4 hours after death and the results obtained were recorded as fresh control. Other specimens were divided into 54 groups and stored at 0°C, -20°C and -40°C (8 groups each) within 6 hours after death. Experimental specimens were kept intact in the digits, which were wrapped in saline-soaked gauze and sealed in airtight plastic bags before being stored.

A 5565 Instron Universal Testing Machine was employed for the tensile tests and the crosshead speed was set at 60mm/min. Tendons were tested every 5 days for 3 months. Samples were tested in quintuplicate. Mean values of Ultimate Tensile Strength, Strain at Maximum Stress, Elastic Modulus and Toughness were compared to fresh control at 99% confidence level using the Student's T method.

RESULTS

Digits which were stored at 0°C decayed badly and produced unpleasant smell after 5 days. All specimens stored at 0°C were disposed and excluded for further evaluation. No statistically significant changes was found in the cross-sectional area following freezing storage and no variations in the gross shape of the stress-strain curve was recorded for all durations. The tensile properties of digital flexor tendon were not affected during freezing storage at -20°C or -40°C for up to 90 days. Generally, there was no significant deviations from fresh control in the values of Ultimate Tensile Strength, Strain at Maximum Stress, Elastic Modulus and Toughness following freezing storage. Statistics comparison between groups of same duration showed no discernable differences in tensile properties after storage at 0°C, -20°C and -40°C.

TABLE I: Comparison of tensile properties.

	No. of Specimens	Cross-sectional Area (mm^2)	Ultimate Tensile Strength (MPa)	Strain at Max. Stress (mm/mm)	Elastic Modulus (MPa)	Toughness (kJ/m^3)
Fresh	5	3.292 ± 0.094	42.720 ± 8.437	0.317 ± 0.047	203.949 ± 24.888	8.685 ± 2.629
0°C*	5	3.024 ± 0.286	47.283 ± 7.874	0.380 ± 0.023	215.862 ± 37.766	11.001 ± 2.767
-20°C*	90	3.020 ± 0.309	46.801 ± 4.768	0.383 ± 0.051	203.623 ± 19.838	10.259 ± 2.310
-40°C*	90	3.092 ± 0.267	46.753 ± 4.990	0.383 ± 0.052	203.468 ± 20.735	10.337 ± 2.387

* Average of all durations

DISCUSSION AND CONCLUSIONS

The tensile properties of digital flexor tendon were not affected during freezing storage at -20°C or -40°C for duration up to 90 days. Storage at 0°C should be avoided for storage more than 5 days since soft tissues deteriorated and produced unpleasant smell thereafter.

Based on the results obtained in this study, we conclude that freezing storage up to 90 days does not alter the tensile properties of digital flexor tendon. These findings were consistent with similar works on ligaments by Viidik and Lewin[1] and Woo et al.[2]. However, our findings were contradictory with other reported data[3-5]. Presumably, the differences could be attributed to the employment of different freeze-thaw processes. The comparison between frozen specimens and fresh control in this study has proven that tensile properties of tendon remained unchanged in unprogrammed freezing and rapid thawing processes.

Since results showed no significant changes in tensile properties through out the storage duration, it suggests that tendons can be preserved by freezing for many months, even years, without compensating its tensile properties. However, the duration limit of freezing storage should be determined empirically.

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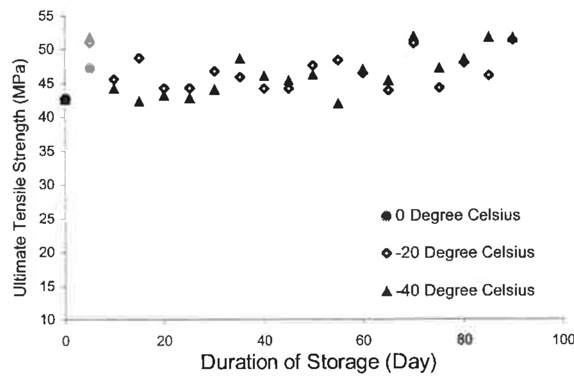

FIGURE 1: Correlation of tendon strength with the duration of freezing storage.

A METHOD OF CALCULATING ICR WITH A REGRESSION STRAIGHT LINE

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INTRODUCTION

One of the methods of analyzing human movements is the instantaneous center of rotation (ICR). The development of a 3-dimensional (D) motion analyzer in recent years allowed collection of a great deal of data, resulting in the improvement of data collection. In spite of the improvement, there have been no adequate achievements of the ICR analysis. The following reasons for the present condition are considered: 1) Any gauge mark is hardly applied to the position suitable for ICR calculation because of the limits of vital site of the mark application; 2) an error occurs at the time of data incorporation with an optical motion analyzer; and 3) since the methods of calculating ICR (Reuleax method, Spiegleman method[1)]), which have been suggested to date, indicate calculation on the 2-D plane, errors of data easily occur in determining the 3-D movements of joints. Based on the fact that errors of recording are hardly reflected by ICR calculation, we devised a method of calculating ICR by approximation of movements of gauge marks with a straight line. The results of ICR calculation with a regression straight line were compared with the results by conventional methods.

METHODS

We made a turntable of 400 mm in diameter, which was subjected to motor-driven rotation at 0.5 Hz. The axis of the turntable was set, so that it would go direct to the Y axis in the experimental space. Markers of 20 mm in diameter were applied to the following three points on the turntable: A, the center of rotation, B, the lateral margin of the turntable, and C: a point which was 100 mm lateral to the center and 10 degrees from the straight line A-B. The rotating markers were recorded for 10 seconds by means of the 3-D motion analyzer (Vicon 370). ICR was calculated by Reuleax method, Spiegleman method, and a method with a regression straight line on the basis of the thus obtained data. In the method using a regression straight line, a regression straight line of moving locus of Marker B were calculated within a fixed time period, and ICR was calculated from the intersecting point of the line going direct to the straight line. The positions of ICR calculated by these three methods were compared with those of marker A determined in terms of the accuracy of the results of calculation.

RESULTS

The values of calculation by each method and the absolute errors of the determined center of rotation(y axis) were as follows.Reuleax:35.7mm, Spiegleman:impossible to calculat, the straight line approximation method :30.0 mm . These results suggested that the calculation method with a regression straight line was excellent in the accuracy.

DISCUSSION

Some means of data processing, which include graduation, have been employed for the purpose of reducing errors occurring at the time of determination of articular movements. However, there have been no adequate achievements of the reduction. The main reason for the situation is considered to be errors occurring in calculating the 3-D movements of joints as 2-D data. Conventional methods of calculating ICR, i.e., Reuleax method and Spiegleman method, require two markers for a rigid body, whereas the distance between these markers applied to a rigid body, which should be essentially fixed, is varied by rotatory movements of the living body. The present method using an approximate straight line produces no error in this regard, because only one marker applied to the moving rigid body. All of the above-described research observations indicate that the method of calculating ICR, which uses an approximate straight line, is considered useful for the analysis of articular movements.

CONCLUSION

Based on the fact that errors of recording are hardly reflected by ICR calculation, we devised a method of calculating ICR by approximation of movements of marker with a straight line in order to minimize data errors of recording. The results of ICR calculation with an approximate straight line were compared with those by conventional methods. As a result, it was suggested that the method using an approximate straight line is more useful than the conventional ones requiring a plurality of marker in point of lowering of the frequency of errors.

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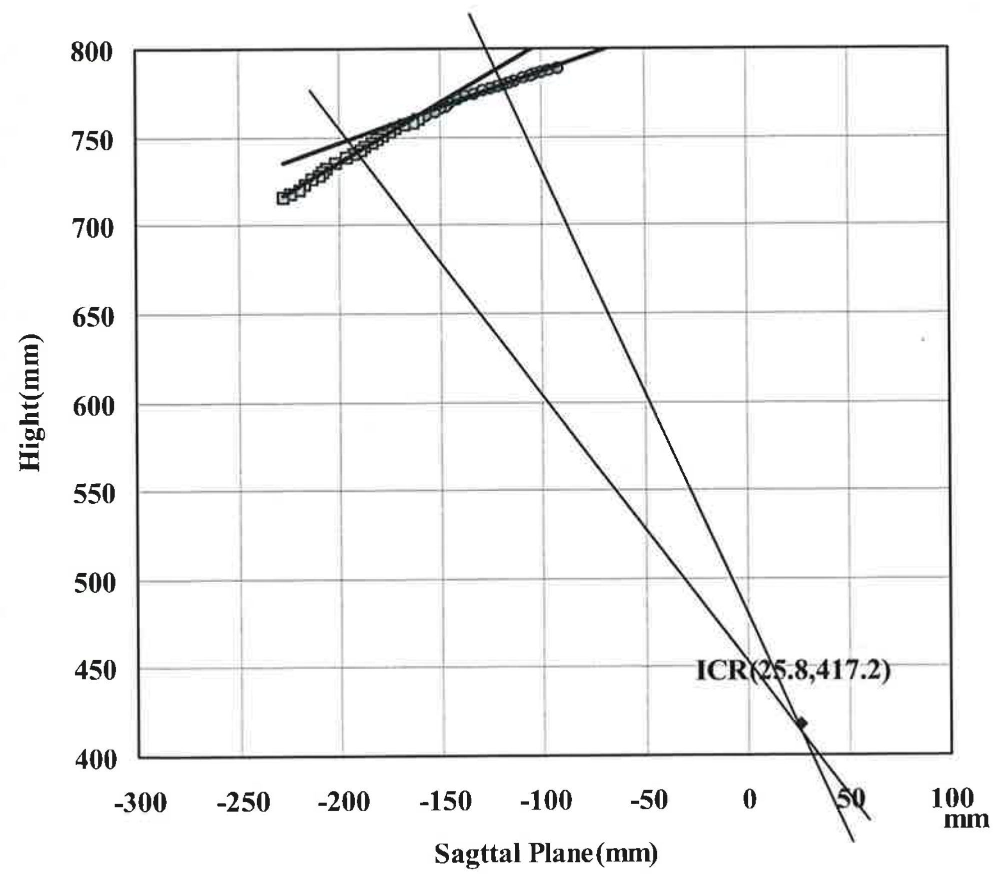

A method of calculating ICR with a regression straight line

ELECTROMYOGRAPHYC VARIATION IN SIT-TO-STAND TASKS OF HEALTHY ELDERLY ADULTS: CHANGES IN SPEED OF ASCEND

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INTRODUCTION

When studying the human movement, the relationship between the position of Center of Mass (COM) and the Base of Support (BOS) is often analyzed. Sit-to-stand (STS) task is the motion in which the BOS change from wide to narrow and moves forward. When the buttocks lose contact with the chair (lift-off), the horizontal position of the COM moves toward the area over the BOS. In the previous studies, there have been found three strategies to move the COM forward; Momentum strategy, Stabilization strategy, and Combined. Hughes et al reported that the elderly people most frequently chose Momentum strategy[1)]. In the momentum strategy, the elderly use both momentum generated by rapid forward motion of the trunk and the knee musculature to extend the knee. However it has not been cleared how this momentum was braked. Braking action is necessary to extend the body safely and to maintain the body balance. The purpose of this study is to investigate the factor of braking action by means of changing the trunk flexion speed of ascend in the STS.

METHODS

Six elderly males with no history of neuromuscular or muscuroskeltal disorders participated in the study. Six young volunteers also participated as contorol. Subjects were seated on an armless, backless chair, the height of which was adjusted to each subject's knee height. The arms were folded across the chest. The initial ankle angle was set at 20 degrees of dorsi-flexion and the feet were placed parallel on the force platforms. The chair was placed on the two platforms and each foot was placed on another platform. Subjects were instructed to stand up from the chair at different speed, preffered (Pref.) and as fast as possible (AFAP). 3D motion analysis system (Vicon512, Oxfordmetrics) was used to determine the instantaneous locations of joint centers during the STS movement(Sampling rate 60Hz). The sagittal COM positions were calculated using four-segments rigid model with ten points on the body determined as landmarks. Simultaneously, three components of the ground reaction force of the feet and the chair, as well as the center of pressure (COP) of these force were measured during STS task by means of four force platforms (AMTI). The COP displacement was defined as the distance between COM and COP. The more the displacement is, the more the momentum to rotate the subject's body increases. When buttocks lose contact with the chair (lift-off), the reaction force measured by the platforms under the chair will bocome zero. EMG signals were also collected by sanpling rate of 1000Hz, bilaterally from four muscles; rectus femoris (RF), biceps femoris longus (BFL), tibialis anteriol(TA) and soleus(So). The signals were filtered with band pass of 10-350Hz and rectified. The EMG data were normalized by 100% of the maximum activation in each muscle during the STS task performed at preffered speed.

RESULTS

We analyzed all of the parameters from the start of COP displacement to the timing when the three joints (hip, knee, and ankle) stopped the motion. The motion of STS performed by the subjects was divided into two phases. First phase was the period from the start of the motion to lift-off, and second phase was the period from lift-off to the end of the motion. Comparing the speed of AFAP to the speed of Pref., in the elders, the rate of second phase during the STS task was longer, and the peak velocity of trunk flexion increased but that of trunk extension did not change. In the youngers, the rate of each phase did not change, and the both of the velocity of trunk flexion and extension increased. At the speed of AFAP, the maximum activities of all muscles during STS task increased as same rate in the youngers. However in the elders, the maximum activity of BFL muscle was greater than three of others, and the maximum activation of BFL was seen immediately after lift-off. The horizontal displacement of COP measured with force platforms during the first phase was shown in Fig.1. At the speed of Pref., displacement was about 1cm in the elders, but was about 5cm in the youngers. There was significant difference between these two groups. At the speed of AFAP, displacement was about 8cm in the elders, but was about 6cm in the youngers. There was not significant difference between these two.

DISCUSSION

The elderly people could maintain the balance after lift-off in both conditions of speed. At the speed of Pref., they were able to perform the STS slowly, without braking after lift-off, by making initial displacement of COP small and prolongation of the first phase. This strategy can be defined as the stabilization strategy. At the speed of AFAP, they were able to lose contact with the chair by making initial displacement of the COP large, and increasing of the forward momentum resulted from fast trunk flexion. The momentum to rotate the body forward might be braked by eccentric contraction of BFL after the lift-off. This strategy can be defined as the momentum strategy. These results indicated that the elders could change the strategy corresponding to the speed of STS task. And changing the strategy was showed as amount of initial displacement of the COP backward. Because the activity of the BFL increased corresponding to the displacement of the COP in the elders, when elderly people performed STS at faster speed, the increase torque of hip extension at the time of lift off may be suppressed, by eccentric contraction of hamstrings muscles.

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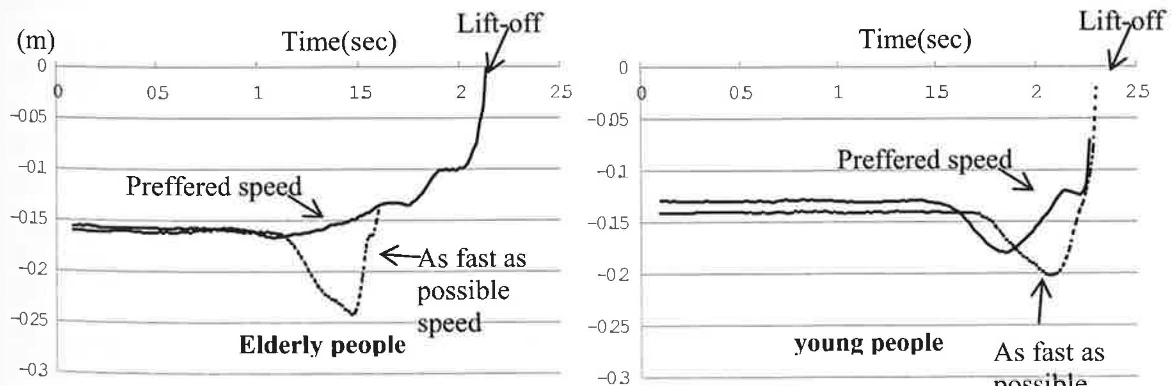

Fig 1. Comparison of Initial horizontal displacement of COP for backward

REFINING PULSE DETECTION LOGIC IN THE PULSE DENSITY DEMODULATION PROCESSING OF EMG

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INTRODUCTION

The present authors had proposed a pulse density demodulation processing of EMG for the purpose of reconstructing the neuromotor command. The adequacy of this processing had been ascertained through a computer simulation of the processes of neuromotor command generation and EMG observation. Then, the method of predicting the optimal threshold of pulse detection had been examined for real EMG experimentally (references 1, 2 and 3). In order to apply the processing in common to the EMG by various measurement conditions, pulse detection logic was refined theoretically by using synthesized EMG again in this study.
The idea which the pulse density demodulation processing based on is that the time varying pattern of the neuromotor command which controls change in degree of contraction of a muscle is transmitted to the muscle in the form of pulse density modulation of the neural pulse trains in the motoneurons connected to the muscle as a whole, so, the neuromotor command can be reconstructed by the pulse density demodulation processing. In this processing, it is necessary to detect as many pulse trains from motor units in the muscle as possible.

METHODS

SYNTHESIS OF EMG

Sequence of synthesis of EMG was as follows:
1) Specification of the on-off pattern of neuromotor command for a whole muscle,
2) generation of neural pulses in motor center,
3) regeneration of neural pulses at synapses,
4) registration of the waveform of action potential of neural pulse,
5) attenuation according to the distance from the electrode,
6) synthesis of the EMG waveform by bipolar electrodes.

Conditions of generation and observation of EMG were as follows:
1) Sensitivity, refractory period, and delay time in each neural pulse generator were set by referring to the physiological data.
2) A tri-phasic standard waveform was used for the pulse shape of action potential.
3) The observation by four channels of bipolar surface electrodes, with eight motor units in each side of the bipolar electrode, taking example of the soleus muscle in running leg action.
4) The attenuation was three steps, namely, no attenuation, 1/2 and 1/4.

PULSE DETECTION LOGIC

Peaks in the EMG waveform were detected by using logics for avoiding the influence of bi- and tri-phasic parts of the action potentials, and for separating the pulse trains of different motor units.
1) The peaks of bi- and tri-phasic parts of the waveform of action potential were eliminated referring to the ratio of their width to that of the main part of the standard waveform.
2) Pulse trains of different motor units were separated among each other, utilizing the amplitude difference due to distance difference from the electrode. When a peak of the EMG waveform was detected, the standard pulse shape with the corresponding peak amplitude was

subtracted from the EMG waveform. The amplitude of the standard pulse shape was modified step by step until the residual components became least.
3) The tri-phasic standard pulse shape was modified according to the amplitude of the detected pulse train.

STATISTICAL COMPENSATION FOR CANCELLED PULSES

By counting the number of detected pulses in the running time window, the analog time varying pattern of the pulse density was demodulated. When pulses were generated from more than one motor unit at the same time, they were cancelled each other depending on their amplitude ratio. However, they were possible to compensate statistically, by estimating the number of motor units involved and the number of generated pulses in each motor unit.

RESULTS

As the results of detailed computer simulation by combining these logics, about 95% of the given pulses were detected from the interference waveform of the synthesized EMG waveform. The pulse density was about 450 pulse/sec in average, but about 1050 pulse/sec at the highest part in the 0.5 sec duration. Furthermore, by adding the logic of statistical compensation for cancelled pulses, about 10% of the pulses were compensated at the highest density part. Thus, it became possible to reconstruct the given motor command from the analog time pattern of the demodulated pulse density accurately.

DISCUSSION

A difficulty about use of information obtained by EMG is that change in amount of the EMG does not proportionally relate to that in motor command which controls degree of contraction of a muscle. There are various factors that influence this non-proportional relationship in the process of motor control generation (reference 4). In order to analyze these factors for reducing an artificial source of discrepancy in the process of observation, such a method as developed in this study for detecting pulses from the EMG waveform that correspond one to one to the pulses of action potentials of motoneurons must be useful.

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A MODEL OF MOTOR CONTROL BASED ON PROPRIOCEPTIVE INPUT PROCESSING

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INTRODUCTION

Several models attempted to simulate motor control in man. They generally focus on the control of normal movement, sometimes with no clear reference to physiological data, and generally, fail to reproduce abnormal movement in human diseases. We here propose a model based on the opposite approach, on the basis that a model derived from the neurophysiology of common movement disorders is able to reproduce both normal and pathological motor control in man.

METHODS

Our model is primarily based on proprioceptive input processing. It is designed as a serial hierarchical structure organised in 3 levels, loaded in an appropriate software (Solid Dynamics System®, Roanne, France), to simulate rapid one-joint movement. The lowest level is a representation of segmental reflexes at the spinal cord, involving a non-linear model of the stretch reflex, previously developed as a simulation of spastic hypertonia (1). It is consistent with neurophysiological data reporting common interneuronal processing of proprioceptive inputs and supra-spinal control (2). The upper level is a "premotor" level, based on the hypothesis of central pattern generators in the spinal cord (3). This level relays the supra-spinal drive, designed as a tuning function of control parameters (stretch reflex threshold and gain) involved at the lower level. The highest level involves motor planning and learning, and has not yet been featured.

RESULTS

This simple computer model is able to simulate the well-known triphasic muscle activity reported in normal ballistic movement, as a combination of descending motor command (first agonist bursts) and stretch reflex contribution (antagonist and second agonist bursts). We demonstrate the ability and limitations of this model for simulating common movements disorders, such as resting and action tremors (figure 1), clonus, spastic movement disorder, hypokinesia and cerebellar dysmetry.

DISCUSSION – CONCLUSION

This approach points out how physiological data extracted from movement disorders' pathophysiology could help to generate a model of healthy and pathological motor control in man. The simplicity of this model also emphasises how a low-complexity representation of the neural system is able to generate complex movements. In this way, it opens on both neurophysiological (comprehensive models) and cybernetic (technology-based models) applications.

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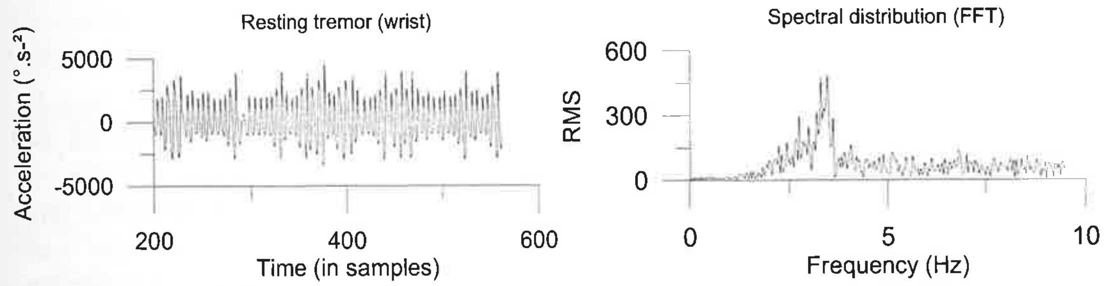

Figure 1 : simulation of resting tremor. Left : acceleration ; right : spectral distribution.

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VARIATIONS IN SURFACE EMG PICK-UP RANGE WITH SUBCUTANEOUS FAT THICKNESS

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INTRODUCTION

An important question in interpreting the surface electromyogram is what the pick-up volume of the surface electrode is. This issue is of critical importance in determining the feasibility of advanced techniques to control multifunctional myoelectric prostheses and is also relevant to a much wider range of general surface EMG applications. The relative contributions of motor units located throughout the muscle tissue to the surface EMG interference pattern remains unclear. While anatomical properties and electrode configuration are both known to affect levels of EMG cross-talk at the skin surface, it is not known how the number and location of detected motor units change as subcutaneous fat thickness varies among subjects and over different muscles. To further investigate this issue, the pick-up volume of a range of surface electrodes for different values of subcutaneous fat thickness was examined using a model of the surface EMG signal.

METHODS

Using a finite-element model of an idealised cylindrical limb composed of skin, fat, muscle and bone tissue (1), surface EMG signals from a total of 2000 motor units were simulated for different thicknesses of subcutaneous fat tissue. Motor units were randomly distributed throughout the muscle cross-section, muscle fibre start-up and end-effects were included and transmembrane currents were modelled using multi-pole sources. Each motor unit was assigned a mean firing rate and inter-pulse intervals were generated according to a Gaussian distribution about the mean inter-pulse interval. Surface EMG signals were simulated as the radius of the muscle directly below the surface electrodes was progressively increased, until the RMS amplitude of the surface EMG signal from that muscle was equal to 95 % of that detected when all units throughout the muscle were active. This was defined as the pick-up radius, *r*, Fig. 1(a). The pick-up radius was examined for bipolar electrodes with 5 mm, 10 mm and 20 mm inter-electrode distances and for a double differential electrode configuration with an inter-electrode distance of 10 mm, as the thickness of the subcutaneous fat tissue was varied between 0 mm and 18 mm.

RESULTS

The volume of muscle which contributed most to the simulated surface EMG signal increased dramatically with subcutaneous fat thickness. In the model, when the fat layer was increased from 3 mm to 18 mm, the pick-up radius of a bipolar electrode (20 mm inter-electrode distance) increased from 5.5 mm to 13.5 mm. Reducing the IED was observed to yield more focal recordings. A reduction in the IED from 20 mm to 5 mm decreased the mean pick-up radius of the muscle tissue by 19.1 %, 18.4 %, 12.9 % and 9.1 % for the 0 mm, 3 mm, 9 mm and 18 mm fat layers respectively. In all models, the double differential electrode arrangement was more selective than bipolar electrodes with an IED of either 20 or 10 mm, and was more selective than the 5 mm IED bipolar electrode with 9 mm and 18 mm thick fat layers, Fig. 1(b).

DISCUSSION

The rate at which the amplitude of the EMG signal decreases with increasing distance between electrode and muscle fibre depends on the properties of the volume conductor, the electrode configuration and the EMG parameter examined. The finite nature of the volume conductor and the different tissue conductivities are therefore critical aspects of the model for this application. The volume of muscle which dominated the surface EMG signal increased with the thickness of the subcutaneous fat layer, consistent with reported experimental findings (2). As the thickness of the fat layer increases, relative differences in the distance between the electrode and different muscle fibres are reduced. Motor unit action potentials at the skin surface appear more similar, thereby decreasing the selectivity of the EMG signal. It follows that EMG data recorded from subjects with different fat thicknesses may not represent samples from similar pools of motor units. The selectivity of the simulated surface EMG signal increased when the inter-electrode distance was reduced and with the double differential electrode configuration. Further spatial selectivity can be achieved with the use of double-differential recording techniques, and more complex spatial filters.

CONCLUSION

In conclusion, the area of muscle which dominates the surface EMG signal is dependent on the thickness of the subcutaneous fat tissue, the electrode configuration and the parameter chosen to characterise the EMG activity. Using an idealised cylindrical model of the upper limb, it is possible to examine each of these factors independently and to estimate the minimum volume of muscle tissue which is necessary for independent myoelectric control signals to be obtained.

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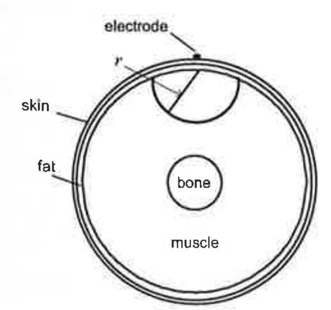
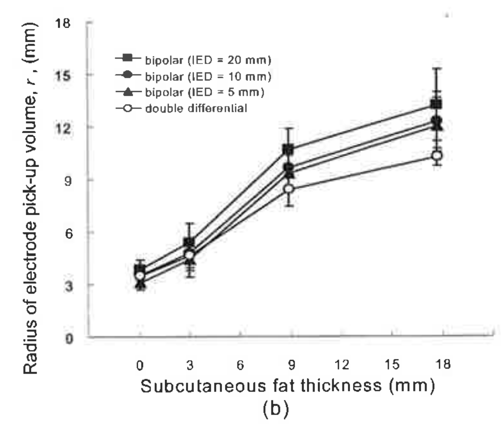


(a)



(b)

Figure 1. (a) Cross-section of muscle model (b) Pick-up radius, *r*, of simulated EMG signals for different electrode configurations and subcutaneous fat thicknesses. The mean and standard deviation of 8 sets of simulated data are presented.

EXPERIMENTAL AND THEORETICAL STUDIES FOR DISTRIBUTION OF MUSCLE FIBER CONDUCTION VELOCITY

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INTRODUCTION

Muscle fiber conduction velocity (MFCV) is usually estimated from a time shift between two EMG signals. This method is only applicable to EMG signals that propagate linearly along muscle fiber. When the recording positions are near the motor end-plate or the tendon, linear propagation is not observable[1)]. To elucidate this problem, the experimental and the theoretical studies for distribution of MFCV were performed.

METHODS

Sixteen channels of surface EMG were detected by surface array electrodes that inter-electrode were taken to be 5.0 mm. A numerical model of surface EMG generation was formulated which included MFCV, the position of motor end-plate zones, the depth of muscle fibers and length of muscle fiber as variable parameters. Action potential was expressed by current tripole[2)]. This current tripole occurred in the motor end-plate zone and it became extinct in the tendon. Many action potentials were generated in motor end-plate zones, and they propagated to ends of muscle fiber (i.e., tendon) with a constant velocity of 4.0 m/s.

MFCV was calculated by using the cross-correlation method. When correlation coefficient ($R_{xy}(T)$) reached to maximum at the shift time T_s, an amplitude ratio (AMR) between two signals was also estimated in order to evaluate the change of the amplitude.

RESULTS

Fig. 1(A) shows MFCV, $R_{xy}(T_s)$ and AMR for m. biceps brachii of a healthy adult male under 20% of the maximum voluntary contraction. MFCV increased rapidly, and both $R_{xy}(T_s)$ and AMR decreased rapidly, around the motor end-plate zone (L_e=5 mm) and the tendon (L_e>40 mm). MFCV, $R_{xy}(T_s)$ and AMR showed constant values in the region other than the motor end-plate and the tendon.

MFCV, $R_{xy}(T_s)$ and AMR for simulated surface EMGs which are calculated by the numerical model are shown in Fig. 1(B). Tow motor end-plate zones were assumed. Near the motor end-plate zone and the tendon, MFCVs showed comparatively high value more than 8.0 m/s and showed high variability, when the $R_{xy}(T_s)$ and the AMR showed comparatively low values. In the locations other than the motor end-plate zone and the tendon, $R_{xy}(T_s)$ and AMR showed high values more than 0.90 and MFCVs showed constant values of 4.0 m/s. These results by the numerical model were in good agreement with the result by the experiment.

DISCUSSION

It has been argued that the propagation velocity of action potential (i.e., MFCV) was constant generally. But, the MFCVs that were detected by surface array electrodes showed different values at various positions. Surface EMG was interference wave due to a summation of many action potentials[3)]. In the motor end-plate zone, surface EMG was interference wave due to many action potentials which propagated in various directions. In the region apart from the motor end-plate zone, surface EMGs show identical waveforms. On the other hand, electrodes near the tendon could be detected the attenuated action potentials at nearly same timing. Therefore, MFCV, Rxy(Ts) and AMR could be rapidly changed in the region of the motor end-plate and the tendon.

CONCLUSION

Relations between MFCV and measurement locations in m. biceps brachii were declared by the experimental and the theoretical studies. In the theoretical studies, surface EMGs were simulated by tripole current model. MFCVs rapidly increased around the motor end-plate zone and the tendon and those showed constant values in the region other than the motor end-plate and the tendon. This result by the numerical model was in good agreement with the result by the experiment.

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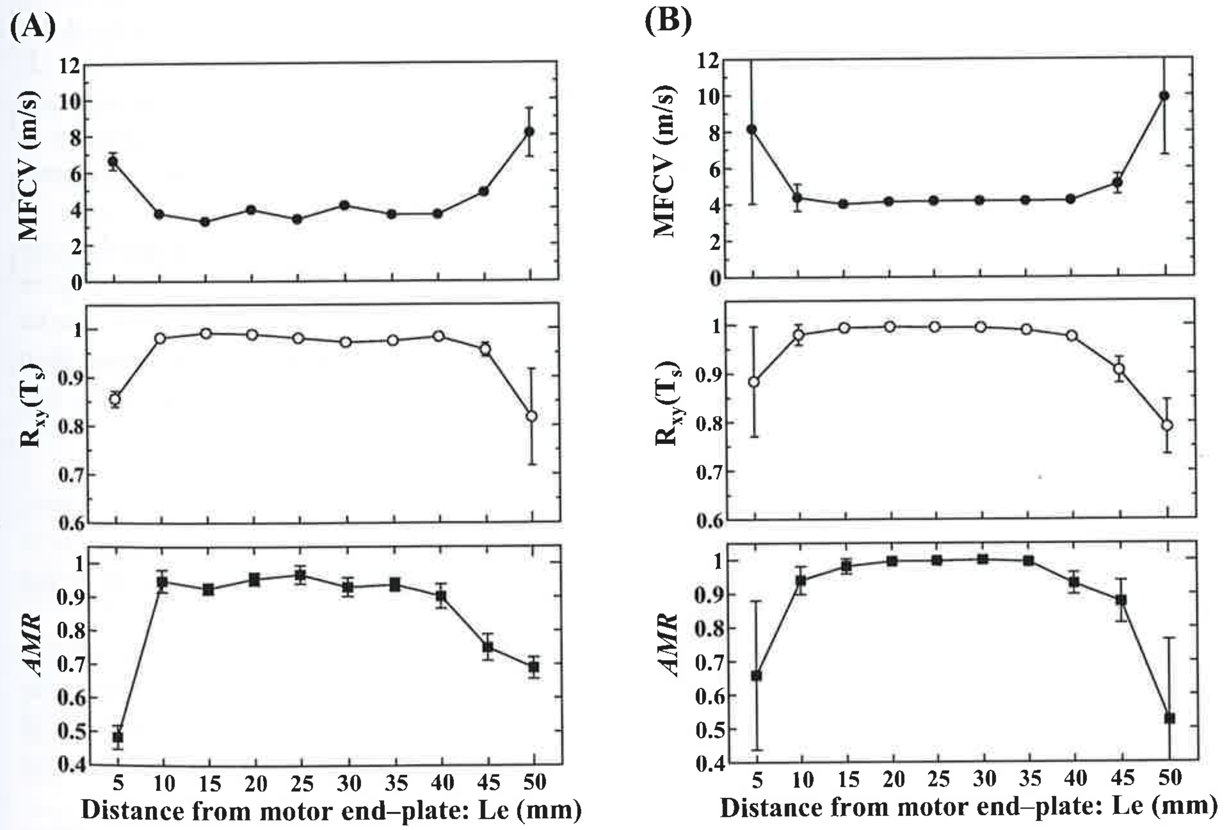

Fig.1 Distribution of MFCV, maximum correlation coefficient ($R_{xy}(T_s)$) and amplitude ratio (AMR) estimated from (A) detected surface EMGs and (B) simulated surface EMGs. L_e is the distance from the motor end-plate zone.

VALIDATION STUDY OF MUAP MODELS

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INTRODUCTION

Simulation is a very valuable tool for the understanding of the physiological and pathological aspects of the neuromuscular system. Simulation studies carried out so far start with the building of a generic muscle model from which different MUAPs can be synthesized. Then they focus on the relationship between anatomical and functional features of this muscle, on one side; and the parameters that characterize the shape of the generated MUAPs, on the other. The aim of this work was to test the validity and accuracy of state-of-the-art MUAP models. To this end, physiological and histological published data from real normal muscles [3] were used to estimate the values for the input parameters that were given to the models to simulate specific muscles. From these, MUAPs were generated extensively and characterized by standard waveform parameters. Finally these results were statistically compared with equivalent published data from real muscles [1].

METHODS

We constructed a MUAP simulation program assambling different parts: a model for muscle architecture [5], two alternative convolutional models for the generation of single fiber action potentials (SFAPs) [2][4], and a concentric non-punctual electrode model [5]. After theoretical and simulation analysis of both SFAP models, we concluded that Dimitrov's [2] was very superior to Nandedkar's [4]. Although they are quite similar in their structure and the mathematical functions involved, the later is unable to express the potential initiation at the end-plate and the potential extintion at the muscle-tendinous juntion. Therefore we chose Dimitrov's as the SFAP generator in our simulator.

For our study we used three different muscle types: tibialis anterioris (TA), biceps braquii (BB) and first interoseous dorsalis (FDI); from each of these three, 10 different muscles were simulated with an MVC of 30%, and MUAPs were synthesized considering 3 different needle insertion sites. Only MUAPs with an amplitude higher than 50 μV and a rise time lower than 1 ms were considered valid for the analysis. At the end we had 400 MUAPs from TA, 546 from FDI and 679 from BB available for our study.

RESULTS AND DISCUSSION

Statistical features of MUAP waveshape parameters (duration, amplitude, thickness, size index and phases) from real muscles [1] and those simulated in our study are given in the Table.

Results show the potential use of the simulator to facilitate the understanding of the neuromuscular system, mainly for its sensitivity to its input parameters which express the anatomy and physiology of the simulated muscles and their motor units. In accordance to real muscles behaviour, differences found in MUAP waveform parameters of simulated muscles of different type were statistically significant.

Moreover, simulated MUAPs were comparable to real ones: differences in their respective mean amplitudes and durations were statistically non-significant. The number of phases of the simulated MUAPs was only sligtly inferior to that number in real muscles. Still the range of their values was well within normality, and so was the percentage of polyphase potentials.

Obtained values for thickness and size index of the simulated MUAPs were quite inferior to real ones. TA presented the higher thickness values, followed by BB and FDI. Size index

values were comparable in TA and FDI, and very inferior to both of these in BB. However, the same tendency was found in real muscles. This proves that input parameters to the model are well conditioned in the sense that they exhibit a sensitivity to output (waveshape) parameters similar to what it is found in Nature. The later discrepancies are probably due to the simulated SFAPs that seem to be a bit too thin, originating MUAPs with too thin spikes as well.

CONCLUSION

We have tested the validity of a MUAP simulator comprised of models for muscle architecture [5], SFAP generation [2], [4] and a concentric needle electrode [5]. Dimitrov's SFAP model [2] has proven superior to Nandedkar's [4] with respect to the potential initiation and extintion phenomena. The simulation programs based on the studied models (only Dimitrov's SFAP model was included) are able to synthesize MUAPs similar to the ones recorded in real muscles, specially concerning their amplitude and duration. However minor discrepancies still appear between these signals, mainly due to the thinness of the SFAP spikes. Refined SFAP models are therefore required for simulating MUAPs still closer to real ones, accomodating features such as thickness or size index.

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	Real muscles						Simulated muscles				
TA	Duration	Amplitude	Phases	Thickness	Size index	**TA**	Duration	Amplitude	Phases	Thickness	Size index
Mean	11,4	0,666	3,2	1,670	1,170	Mean	11,4	0,617	2,9	1,093	0,432
SD	1,2	0,254	0,3	0,230	0,300	SD	1,0	0,412	0,5	0,229	0,577
Max	18,4	1,572		2,810	2,463	Max	14,5	2,104	5,0	1,683	1,672
Min	4,6	0,194		0,575	-0,397	Min	4,8	0,079	2,0	0,357	-1,312
PID						**PID**					
Mean	9,4	0,752	3,1	1,380	0,980	Mean	9,0	0,722	2,6	0,997	0,455
SD	1,3	0,247	0,4	0,220	0,380	SD	1,7	0,495	0,6	0,235	0,597
Max	18,0	2,301		2,610	2,281	Max	16,3	2,410	5,0	1,440	1,629
Min	4,0	0,188		0,485	-0,912	Min	4,4	0,087	2,0	0,260	-0,954
BB						**BB**					
Mean	9,9	0,436	2,6	1,460	0,650	Mean	9,8	0,424	2,9	1,043	0,140
SD	1,4	0,115	0,3	0,200	0,330	SD	1,1	0,235	0,6	0,239	0,446
Max	16,4	1,414		2,093	2,053	Max	13,9	1,389	6,0	1,689	0,951
Min	4,2	0,178		0,564	-0,539	Min	4,5	0,077	2,0	0,233	-1,414

Table. Statistics (mean, standard deviation, maximum and minimum values) of MUAP duration (ms), amplitude (mV), thickness (ms), size index and phases from real TA, PID and BB muscles [1] and equivalent mucles simulated in our study.

ELECTROMYOGRAPHIC STUDY OF STOMATOGNATHIC SYSTEM MUSCLES DURING CHEWING OF DIFFERENT MATERIALS

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INTRODUCTION

Clinical situations involving pain and sensibility in the chewing muscles, condylar sounds and limitation of the jaw movements characterize the craniomandibular disorders (CMD), also called temporomandibular dysfunction, syndrome of the pain and temporomandibular disorder or more simply TMJ dysfunction. The electromyographic equipments have become more and more accessible to the dentists, thus allowing the enhancement of studies on the mastication muscle electromyography.

Although the chewing gum is usually used for electromyographic studies on the mastication (CHRISTENSEN, 1996; PERRY, 1981; STOHLER, 1986), the muscular response to various commercial chewing gums can show different electric activities during the masticatory activity (KARKAZIS & KOSSIONI, 1997). Thus, due to the lack of scientific literature on the more suitable material for the EMG exam of the mastication muscles associated to a great demand of exams accomplished in the Laboratory of Electromyography, Department of Morphology, Faculdade de Odontologia de Piracicaba, a study was developed with the purpose of comparing the performance of several materials with respect to generation of the electromyographic trace, by using two commercial chewing gums (A and B) and two insipid materials, cotton ball and Parafilm M®.

METHODS

A total of 10 female young adult subjects, Graduation and Post-Graduation students of the Course of Dentistry of FOP/UNICAMP, aged from 18 to 27 years, with normal occlusion and no history of craniomandibular disorder was studied. In according to the ethical principles, the purposes of this study were carefully explained to the volunteers, who signed a formal consent. According to TURKER (1993), the surface electromyography recording contains some activity coming from another muscles that are being activated, synergistically or antagonistically to the studied muscle. This phenomenon is known as cross talk. Although the surface passive electrodes have been placed on the anterior face of the digastric muscle, the cross-talk in this region cannot be avoided (BASMAJIAN & De LUCA, 1985; De LUCA, 1997), that is, as these muscles are small and thin, the signal captation might be arisen from the adjacent suprahyoid muscle group rather than a single muscle (m. digastric).

A questionnaire was also supplied to the volunteers for better investigation on the palatability of the materials. The anterior portion of the temporalis muscle has been the more used in the electromyographic studies of the chewing muscles, since the interval between the myotendinous junction and the motor point does not present hair, thus representing the more suitable region for the placement of the electrodes.At the end of each assay, the EMG signals were examined in the computer screen "treatment of data" to guarantee the quality of the obtained data. At this time, the RMS (Root Mean Square) value of the respective signals was observed and scored in order to facilitate further analysis.

RESULTS

The results obtained in this study on the electric activity of the temporalis (anterior portion), masseter and suprahyoid muscles during the isotonic mastication of different materials – cotton ball, Parafilm, A and B – are demonstrated through coefficient of variation (CV) for each period of chewing activity for the different muscles and materials studied.

DISCUSSION

The study of EMG comprises not only the handling of an electromyograph and the analysis of captured signals but also the knowledge of the equipment, the recommended technique and the benefits that it offers. At first no comparison with the literature on the muscles and materials here studied was possible because there is no concern with the normalization of the signal and the standardization of the material in several reports (STOHLER et al., 1986; KARKARAZIS & KOSSOINI, 1997). For this reason the research about the standardization of a material that is indicated for the accomplishment of the EMG exam during the mastication was developed, allowing a more accurate analysis of the temporalis (anterior portion), masseter and suprahyoid muscles.

CONCLUSION

1. The best materials, statistically significant with the smallest variation coefficients, were the cotton and Parafilm, following its indications for the accomplishment of the EMG exam of the chewing muscles. 2. The Parafilm material showed a better palatability in relation to the cotton, according to a sensorial analysis of the materials through a questionnaire, thus choosing it among the studied materials as the more indicated for EMG exams.

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IMPEDANCE OF THE SKIN-ELECTRODE INTERFACE IN SURFACE EMG RECORDINGS

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INTRODUCTION

Surface EMG signal detection is strongly affected by the value of the Electrode-Skin contact Impedance ESI. The aim of this work was to evaluate the contact impedance for adhesive and silver electrodes in a number of conditions, including different skin treatments, frequency and amplitude of the injected current, and electrode surface.

METHODS

The measures were performed using the volt-amperometric technique. Thus, a sinusoidal current was injected between two electrodes placed on the skin and the voltage between them measured. The amplitude and frequency of the injected current were in the range 1-200 uA_{rms} and 4-200Hz, respectively.

All the measures were done between, one electrode (E1) with surface S_{E1} placed on a muscle and the other (E2) with surface S_{E2} made of a wet strip with metal inside (patient ground) applied to the wrist or to the armpit. It has been shown that, the measured impedance is a good approximation of the contact impedance between E1 and the skin ($S_{E1}<1/10S_{E2}$).

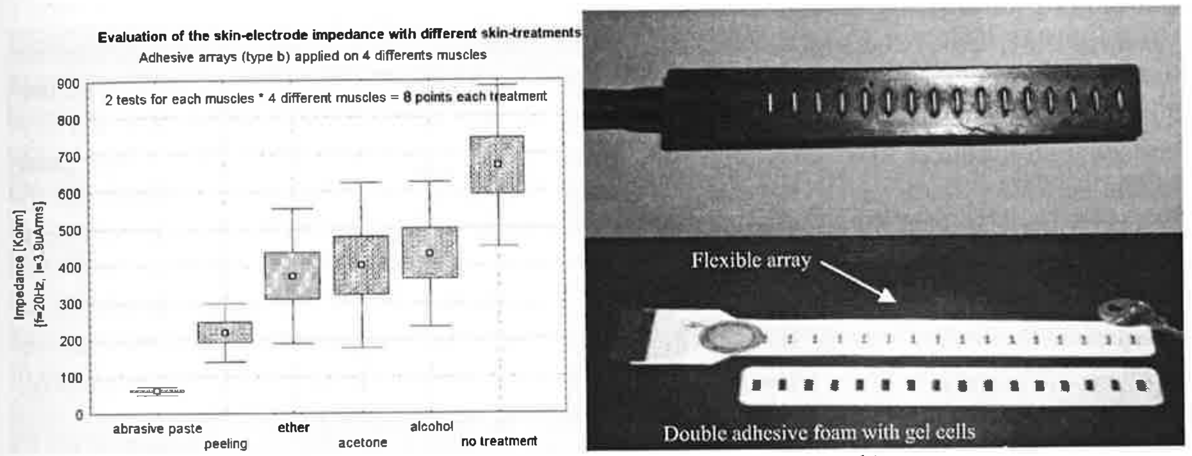
The measures were performed in order to address the following issues:

a) Comparison of ESI resulting with non adhesive dry electrodes for short-term use (bar Ag electrodes) and adhesive electrodes for long-term applications (Ag/AgCl electrodes with the skin contact represented by conductive gel). The comparison was done in a period of two hours. The electrode E1 was applied on the biceps brachii muscle and the electrode E2 to the wrist (S_{E1}=16 mm^2, S_{E2}=1600 mm^2). The injected current had an amplitude of 3.9 uA_{rms} and a frequency of 20 Hz. The skin was treated with abrasive paste.
b) Characterization of ESI for different muscles and adhesive electrodes. The muscles selected were vastus lateralis, tibialis anterior, deltoid, pectoralis major, palmaris longus and biceps brachii. Some condition as in a).
c) Evaluation of the ESI for six skin treatments and adhesive electrodes. The muscles selected were tibialis anterior, deltoid, pectoralis major and palmaris longus. The injected current was the same as in point a). The skin was treated with abrasive paste, peeling, ether, acetone, alcohol. Moreover, the condition of no treatment was evaluated.
d) Evaluation of the ESI as a function of the frequency of the injected current and of the area of the adhesive electrode. Electrode E1 was applied on the internal part of the forearm (S_{E1}=16-60-110 mm^2, S_{E2}=1600 mm^2). The injected current had amplitude 3.9 uA_{rms} while the frequency was 4-10-20-200 Hz. The skin was treated with abrasive paste.
e) Evaluation of the ESI as a function of the area of E2 with adhesive electrodes. Electrode E1 was applied on the biceps brachii muscle while E2 (S_{E1} = 16 mm^2, S_{E2} = 1600-3200-4800 mm^2) was applied to the wrist. Injected current as in point a), skin treated with abrasive paste.
f) Evaluation of the ESI for different locations of the electrode E2 with adhesive electrodes. E1 was applied on the biceps brachii muscle and E2 to the wrist or armpit (S_{E1}= 16mm^2, S_{E2}=1600mm^2). Injected current as in point a), skin treated with abrasive paste.
g) Evaluation of differences of the impedance values between adhesive electrodes closely spaced on the same muscle. Injected current as in point a), skin treated with abrasive paste.

RESULTS

The following results were obtained for each of the issues described in Methods:

a) The adhesive electrodes showed a significantly lower ESI (180 kΩ, after 10 minutes) than the bar Ag electrodes (310 kΩ, after 10 minutes). The impedance for adhesive electrodes significantly decreased with time (80 kΩ after 120 minutes) while that of the dry bar Ag electrodes increased.
b) The ESI was slightly and non significantly different for different muscles (the average impedance for the muscles investigated varied in the range 75-105 kΩ).
c) The Figure shows the influence of the skin treatment on ESI. Abrasive paste results in the lowest impedance (80 kΩ).
d) The ESI showed a twofold decrease with increasing the frequency of the injected current in the range 4-200 Hz and with increasing area of the gel skin-contact in the range 16-110 mm^2.
e) The reference electrode area does not significantly affect the estimate of ESI.
f) The location of the reference electrode does not statistically influence the estimate of ESI. Thus, increasing the distance between electrodes does not significantly modify the impedance.
g) In the measurement conditions described the difference of ESI between two closely placed electrodes (distance 10 mm) is always smaller than 15% of the average of the two ESI values.



a) b)

Figure 1. a) Skin-electrode impedance measured with six skin treatments in eight locations of four muscles of one subject. The investigated muscles are biceps brachii, vastus lateralis, tibialis anterior, and palmaris longus. b)Non adhesive array for short term use and flexible adhesive array for long term applications

DISCUSSION AND CONCLUSIONS

It is concluded that adhesive arrays with Ag-AgCl electrodes and conductive gel have a lower and more reproducible impedance than dry Ag bar electrode arrays. Skin rubbing with abrasive paste reduces electrode-skin impedance much more than cleaning the skin with a solvent. ESI is highly nonlinear and affected by current density J. The value affecting EMG recordings is the intercept of the regression curve Z=f(J) for J approaching zero.

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EMG STUDIES OF ARM MUSCLES UNDER INFLUENCE OF STATIC STRETCHING AFTER EXHAUSTIVE EXERCISES

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INTRODUCTION

The electromyography is likely the most relevant method for researchers interested in measuring the muscular tension through the muscle action potential (ALTER, 1999).
Few studies (DE VRIES, 1961, 1966; McGLYNN et al., 1979; JONES et al., 1987; LUND et al., 1991) have correlated the flexor muscles of the forearm to the degree of muscular tension post-exercise. Among these latter, divergent results are found with regards to electromyographic response of the muscle during the recovery, being considered as the main factor of divergence the degree of action potential generated by its motor units when using static stretching during the pause of the muscular work.
DE VRIES (1961, 1966) and McGLYNN et al. (1979) observed that an increase of the degree of muscular tension caused by an enhanced degree of activity of motor units occurs after the muscular exercise and that the static stretching promotes the reduction of this electric activity and consequently induces to a better muscular recovery and significant diminution of the pain post-exercise. Other studies (JONES et al., 1987 and LUND et al., 1991) demonstrated that in the injured muscle due to the exercise, its electric activity is diminished during the recovery even without the use of stretching.
The aim of this work was to study the degree of electric activity of the action potential of motor units of the biceps brachii and brachiorradialis muscles generated just after exhaustive exercise of forearm flexion in concentric and eccentric phases through electromyography, considering the influence of the static muscular stretching on this electric activity.

METHODS

This study was carried out in 10 male volunteers (age: 18-25 years old) selected among the students of the Institute of Biomedical Sciences of the Federal University of Uberlândia, to evaluate the electromyographic activity of the biceps brachii and brachiorradialis muscles in forearm flexion movements. The subjects were divided into 2 groups (experimental and control), containing 5 individuals per group, and received instructions about the exercises to be performed and signed a consent term agreeing with their participation in the study.
The electromyographic signals were recorded using a computerized electromyograph developed in the Biomedical Engineer Laboratory, Electric Engineer Faculty, Federal University of Uberlândia, Brazil, with the following characteristics: simultaneous acquisition of up to 8 differential channels with 10 GΩ input impedance, 93 dB at 60 Hz CMRR (common mode rejection ratio), ground electrode common to all channels, filters set at 20 Hz (low pass) and 5 kHz (high pass), and three amplification stages - the first supplying gains of 1, 2 or 5 times, the second gains of 82, 10, 22, or 39 times, and the third 10, 100 or 1000 times – thus making possible a gain of at least 100 times and at most 4960 times. In addition, the system possesses optical isolation – 1.5 KV (RMS) at 60 Hz – between the electronic circuit and the stage that is in contact with the volunteer. The gel-anointed ground electrode was used to eliminate external interferences to the electromyographic signs. Signals were captured using differential surface electrodes with 4.75 mm in diameter. The detection surface was standardized at 2.5 mm in diameter and the distance between the electrodes was 2 cm. An acquisition system of data (Alc EMG) was used to quantify the electromyographic sign amplitude, transforming the action potential in the root mean square (RMS), expressed in

microvolts (μV), thus representing the best parameter to evaluate the variables of the electromyographic sign, according to BASMAJIAN and DE LUCA (1985).

RESULTS

Statistically significant differences were not found for the electric activity of the studied muscles, with or without stretching post-exercise.

DISCUSSION

Our results are disagreeing with the findings of DE VRIES (1961, 1966) when studying on the theory of muscular spasm to justify the muscular pain post-exercise, who found a significant increased electric activity in the period of muscle recovery post-effort and significant decreased electric activity of the studied muscles when using static stretching during the muscular recovery. Similar results were also found by McGLYNN et al. (1979) who investigated the effect of static stretching in the period of recovery post-exercise, observing a decreased muscular electric activity with the use of static stretching.
These divergent results can be related to the adopted methodology, since both above mentioned authors captured the muscular electric activity after 24 hours of execution of the exercise, time in which the muscular pain appeared. According to the same authors, the pain induced to an increase of the muscular electric activity, which in turn, was decreased with the use of static stretching.
On the other hand, recent studies (JONES et al., 1987 and LUND et al. 1991) have shown that there is no any increase of the muscular electric activity in the recovery period after effort, therefore it would not be necessary the use of some stretching type with the purpose of decreasing the muscular electric activity. Thus, these studies are agreeing with our findings since no significant difference was found between the electric activity of the studied muscles in resting and after exhaustive exercise, whether with or without the use of static stretching.

CONCLUSION

Taken together, it can be concluded that once the stimulus of the effort to the muscle ceases, its electric activity returns at levels similar to resting position, with or without the use of stretching. Thus, the stretching has shown to be an efficient approach during the muscular recovery, having an important role in the removal of metabolic products generated during the effort (lactate, for example), but without any influence on the muscular electric activity.

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DECOMPOSITION METHOD FOR MOTONEURON FIRING RATE ESTIMATION FROM INDWELLING RECORDINGS

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INTRODUCTION

Electromyographic (EMG) signal constitutes an important source of information for the study of the neuromuscular system. For instance, the firing rates (FRs) of active motor units (MUs) may reveal details about the dynamics of muscle contraction. This can be helpful for neuromuscular diagnoses, rehabilitation therapies, control of functional prosthesis and sports medicine supervision. We are currently experimenting a deconvolution algorithm for the estimation of FRs of active MUs from surface EMG recordings. While this sort of techniques can be easily tested on synthetic signals elaborated from models of the musculo-skeletal system, where the FRs are perfectly known parameters, validation on experimental data requires obtaining the real FRs from some other independent and reliable methods. To this end, intramuscular and surface signals can be collected simultaneously. The decomposition of the intramuscular signals into their constituent motor unit action potential (MUAP) trains will permit a direct estimation of their FRs.

METHODS

We have developed a five phases decomposition algorithm:

-Phase 1. Signal preprocessing: a low-pass differentiator is used to attenuate the baseline drift and enhance the spikes [1].

- Phase2. Spike detection: an amplitude threshold is set proportional to the standard deviation (std) of the signal. Peaks larger than this threshold are detected [2]. Around these peaks, segments of 8.875 ms, presumably containing a potential or superposition of potentials, are stored.

- Phase 3. MUAP extraction: applying to the previous segments a self-similarity approach (as the one given in [3]) waveforms related to different trains of MUAPs are obtained. MUAP trains are split when two occurrences of the same MUAP take place within 15 ms (inconsistency) [2]. Similar MUAP trains are merged if no inconsistencies are introduced in the merging [2]. This extraction procedure is repeated twice to get a reliable classification. Mean and std curves are obtained from each set of waveforms corresponding to the same MUAP train.

- Phase 4. Peak identification: each detected peak is compared to the different MUAP mean curves and labelled to the most similar one. The matching criterion is the number of sampling points falling within the mean $\pm$ 1.5 std curves associated to a MUAP class. The "peel-off" approach [4] is used to identify superimposed potentials and a two-stage procedure is again used. Extracted potentials in the first stage with a signal/noise ratio (noise being the MUAP pattern subtracted from the signal) lower than a certain level, are retained for the second stage analysis. The best result from the two analyses is retained.

- Phase 5. Firing frequency estimation: the inter-impulse interval (IPI) sequence relative to each MUAP train is extracted. A 25-bins IPI histogram is obtained. Too large IPI samples (>187.5 ms) are removed. When IPI of double or triple value of a frequent one is found, it is considered that one or two potentials of that train were skipped: the interval is then divided by two or three, respectively. Bins with low counts (probably wrong detections) are emptied. The

histogram is smoothed by a 4 bins long averaging filter. The histogram average (within a certain interval around the histogram median) is taken as the final IPI mean estimator.

Synthetic data was used to evaluate the performance of the algorithm. Eight MUAP waveforms were manually identified from our EMG records. Three tests were performed with 4, 5 and 6 MUAP trains and with IPI mean and std as given in the upper part of Table 1. For each test, 5 20000 samples long signals (sampling frequency of 8 kHz) were synthesised, using a different combination of MUAP waveforms in each signal. White Gaussian noise was summed to all the signals yielding a 13-dB signal/noise ratio. A baseline drift (up to 50 Hz) was also added. This formed a realistic scenario with regard to the real recorded signals.

RESULTS

Results are shown in the lower part of Table 1. It can be seen that although a relatively large amount of peaks were not detected, wrong detections were quite few and the method was still able to obtain precise and consistent estimations of the IPIs.

Test	4 MUAPs	5 MUAPs	6 MUAPs
IPI: mean and Std. (ms).	60.4, 68.7, 54.1, 62.5 12.0, 13.7, 10.8, 12.5	60.4, 68.7, 63.4, 54.1, 62.5 12.0, 13.7, 12.7, 10.8, 12.5	60.4, 68.7, 63.4, 68.7, 54.1, 62.5 12.0, 13.7, 12.7, 13.7, 10.8, 12.5
# peaks	805	1018	1200
# detected peaks	623	702	851
# wrong detections	2	0	27
# undetected peaks	180	316	322
Estimated IPI: mean and std. (ms).	58.3, 71.3, 52.9, 65.2 0.96, 6.94, 2.70, 5.74	60.0, 66.9, 63.9, 54.3, 63.2 1.76, 5.57, 5.81, 2.28, 6.62	57.8, 68.0, 60.3, 68.0, 57.5, 64.2 0.99, 6.34, 2.21, 2.06, 7.17, 3.72

Table 1

DISCUSSION AND CONCLUSION

More tests have to be conducted, (especially with a larger number of trains in the signals) which implies improvement in the ability of the algorithms to distinguish between various MUAPs. However, in its present state, the method can already be applied to signals obtained at low contraction levels. The results can then be compared to other indirect FR estimation procedures, as the deconvolution algorithm previously mentioned.

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ANALYSIS ELECTROMYOGRAPHIC OF EXTENDING OF THE KNEE IN EXERCISES OF CLOSED KINETIC CHAIN IN INDIVIDUALS WITH PATELLOFEMORAL DISORDERS

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INTRODUCTION

The dysfunction of the articulation patellofemoral (DFP) it is a pathology of the articulation of the knee that attacks 20% to 35% of the general population[1,2]. Several factors are associated DFP, among them an unbalance among the muscles VMO (Vast Medialis Oblique) and VL (Vast Lateralis)[3,4,5]. However, the literature is controversial in relation to better degrees and to the best exercise type, if in Closed Kinetic Chain (CCF) or in Open Kinetic Chain (CCA), to prioritize the activity of the muscle VMO in relation to the long and oblique portions of the VL[6]. Several factors DFP they are associated, among them an unbalance among the muscles VMO (Vast MedialisOblique) and VL (Vast Lateralis)[3,4,5]. There is not in the literature, studies about the activity of the muscles VLO (Vast Lateralis Oblique) and VLL (Vast Lateralis Long) in CCF. Therefore, the objective of the work was it of analyzing the activity electromyographic of the muscles VMO, VLO and VLL in normal individuals and with DFP, in the 30 degrees and 60 degrees of the knee extension, during CIVM of extension of the leg in CCF in the equipment Leg Press to 45 degrees.

METHODS

They were appraised 20 volunteers of the feminine sex with age between 18 and 25 years (X = 23,6 ±3,53), being 10 healthy and 10 with patellofemural disorders (DFP). The inclusion Criterion for the group A and exclusion for the group B were: Not to present pains as well as more than two signs or positive tests of the articulation patellofemural, accomplished in the physical evaluation. The inclusion Criterion for the group B and exclusion for the group A were: Presence of excessive pronation subtalar, presence of torsion lateralis tibial, report of current pain or episodes previous of pain, angle larger Q than 14°, to present in the minimum three signs or positive tests in the previous evaluation. The volunteers accomplished in an aleatory way CIVM for 30 degrees, 60 degrees and 90 degrees of knee extension, in the equipment Leg Press to 45 degrees. The exam electromyographic, was accomplished with simple surface active differential electrodes. The electromyographic sign was quantified through the Root Mean Square (RMS) and normalized by extension CIVM of the knee to 90degrees.

RESULTS

The results revealed through the variance analysis (ANOVA) with repeated measures, that the muscles VMO, VLL and VLO didn't present significant difference in the electric activity in the angle of 30 degrees so much for volunteers healthy as for volunteers with DFP ($p>0,05$). In the same way, these muscles had the same behavior for 60degrees for volunteers with DFP ($p>0,05$). On the other hand, the muscle VLL was more active than the muscle VLO to 60 degrees for volunteers healthy ($p <0,05$). the application of the test Wilcoxon for the comparison among two groups, revealed that the electric activity of the muscles VMO, VLL and VLO didn't differ among the normal group and with DFP, in the angles of 30 degrees and 60 degrees of the knee extension in exercises of CCF ($p>0,05$).

DISCUSSION

Being like this, the muscles VLL and VLO presented different functions in volunteers healthy, once these muscles presented differences significant statically among 30 degrees and 60 degrees of the knee extension. In relation to the muscular unbalance told by several autors[3,4,5] as factors associated DFP, in this study this seems not to be a preponderant factor in DFP, since there was not it differentiates significant statically among the muscles VMO, VLL and VLO to 30 dgrees and 60 degrees of the knee extension in CCF between normal volunteers and DFP group.

CONCLUSION

The results of this research, in the used experimental conditions, allow to end that:

- In normal volunteers during the accomplishment of exercises in CCF, worked in the equipment Leg Press to 45 degrees, there was difference of electric activation among the muscles VLO and VLL, tends the muscle VLL larger activity that VLO to 60 degrees of the knee extension, demonstrating that these muscles present different functions.
- Doesn't have difference of electric activation among the muscles VMO, VLL and VLO to 30 degrees and 60 degrees of the knee extension between normal volunteers and DFP group in exercises of CCF accomplished in the equipment Leg Press to 45 degrees, therefore the muscular unbalance seems not to be a preponderant factor in DFP.
- In relation to the treatment of DFP the invigoration of the muscle VMO, should not be it considered as the principal used resource, other alterations biomechanics and neuromotors, they should also be emphasized in the same way that the invigoration of the muscle VMO.
- Once there is not difference of electric activation among the muscles VMO, VLL and VLO to 30 degrees and 60degrees of the knee extension between normal volunteers and DFP group in exercises of CCF accomplished in the equipment Leg Press to 45 degrees, the best angle to work in CCF with patient bearers of DFP is between 0 degrees and 30 degrees of the knee extension, once in this angle the stress patellofemoral is smaller.

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EMG ANALYSIS OF FOREARM MUSCLES ON THE ELBOW JOINT IN DIFFERENT POSITIONS

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INTRODUCTION

Electric stimulation and signal capturing emanated from muscles during the voluntary contraction have been considered as fundamental for kinesiologic and clinical studies of the muscular activity.[1] In this study, we have chosen the *flexor* and *extensor carpi ulnaris* muscles (FCU and ECU, respectively) because we believed that, based on its origins and insertions, they can be active in flexion movements of the forearm-elbow complex, since they cross the elbow joint.[4, 5, 6] Thus, the aim of this work was to study the action of the FCU and ECU muscles in flexion movements of the forearm through EMG in pronated, semipronated and supinated positions, with 50% maximum voluntary contraction (MVC).

METHODS

Ten untrained volunteers, 5 male and 5 female, 21-38 years old, with no history of diseases or other alterations that could influence the muscular activity were studied. Each volunteer performed 9 forearm flexion movements at free load with 3 repetitions at 50% MVC in semipronated, supinated and pronated positions. All movements were performed using an apparatus designed for muscular exercises called double-pulley.

EMG signals were captured using differential surface electrodes, with 100 times of gain, and recorded using a 16 channel-electromyograph (Lynx Tecnologia Eletrônica Ltda), 12 bits A/D converter board and the AqDados software, which provided numerical data as RMS (root mean square) values. The apparatus was calibrated for 500 μV/division and the recording speed was 200 ms/division, which resulted in a total response time of 4s.

RESULTS

Table 1 shows the RMS values of EMG activity of the FCU and ECU muscles in forearm flexion movements in the three studied positions with 50% MVC.

The mean RMS values of the FCU muscle in pronated position were significantly higher than those found in semipronated ($p < 0.01$) or supinated ($p < 0.01$) positions. Results obtained in supinated position were similar to those in semipronated position ($p > 0.05$). Regarding the ECU muscle, RMS values obtained only in pronated position were significantly higher than those found in supinated position ($p < 0.01$). There were no statistically significant differences between the RMS values obtained by the two muscles in the three studied positions (Figure 1).

DISCUSSION

In this study, a relatively high EMG activity was recorded for the FCU muscle in all movements, independent of the forearm position, agreeing with the findings of Cunninghan[2], Gray[3] and Sobotta[6] who reported a flexor action of the FCU muscle on the elbow joint. The highest activity of the FCU muscle was recorded in pronated position of the forearm. This value, associated with the position of the muscle in the forearm, suggests a mechanical disadvantage related to the previous shortening characteristic of the pronation. Meanwhile there would be mechanical advantage in the other positions, since in this position the muscle would be in relative lengthening state. Based on the smallest RMS value found in the supinated position, it can be concluded that this position would be that of larger mechanical advantage for the referred muscle.

EMG activity detected in the ECU muscle showed the highest RMS value in pronated position, followed by the semipronated and supinated positions. These values showed a pattern similar to the FCU muscle, as it would be expected, since both muscles situate almost

parallel in most of its extension, what would take to similar stretching changes when performing the supinated and pronated movements.

CONCLUSION

Based on the RMS values recorded, it can be concluded that (1) the FUC and EUC muscles are active in forearm flexion, independent of the position; (2) the FUC muscle is relatively less active than the ECU muscle in semipronated position; (3) the FCU muscle is relatively more active than the ECU muscle in supinated position; (4) both muscles show practically the same activity in pronated position; (5) the highest EMG activity in both muscles occurs in pronated position; (6) the smallest EMG activity occurs in supinated position.

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Table 1 – RMS values (μV) of the electromyographic activity of the flexor and extensor *carpi ulnaris* muscles in flexion movements of the forearm in semipronated, supinated, and pronated positions with 50% maximum voluntary contraction (MVC).

		Muscles					
		Flexor *carpi ulnaris*			Extensor *carpi ulnaris*		
Volunteer	50% MVC	Semipronated	Supinated	Pronated	Semipronated	Supinated	Pronated
1	18	130.8 ± 8.2	127.5 ± 4.7	228.0 ± 34.1	379.4 ± 118.1	129.8 ± 36.6	142.3 ± 24.4
2	21	94.7 ± 45.3	125.3 ± 10.3	145.2 ± 36.2	50.1 ± 9.9	101.4 ± 35.4	95.4 ± 17.5
3	10	48.5 ± 2.6	44.4 ± 1.4	59.6 ± 8.8	51.8 ± 2.6	48.9 ± 4.8	103.6 ± 6.8
4	18	59.4 ± 5.2	52.2 ± 9.3	70.8 ± 9.1	73.7 ± 9.0	37.5 ± 3.2	74.1 ± 9.4
5	18	157.2 ± 18.0	118.9 ± 10.2	261.4 ± 24.8	87.2 ± 7.1	84.5 ± 17.1	159.6 ± 73.2
6	10	50.0 ± 12.1	38.9 ± 10.8	55.3 ± 12.2	70.9 ± 6.7	74.3 ± 3.3	90.5 ± 11.6
7	07	47.8 ± 5.2	26.5 ± 5.4	61.7 ± 4.6	191.6 ± 10.2	82.8 ± 22.4	201.7 ± 17.9
8	18	77.5 ± 11.6	94.1 ± 8.3	163.6 ± 3.4	44.1 ± 4.5	50.9 ± 8.1	154.6 ± 2.9
9	07	33.1 ± 6.8	60.4 ± 4.2	112.3 ± 33.2	39.3 ± 1.6	61.9 ± 12.7	64.0 ± 19.1
10	07	79.1 ± 2.8	74.7 ± 0.4	77.3 ± 7.4	63.9 ± 11.0	62.3 ± 4.4	135.7 ± 6.7

The values are shown as the mean ± S.D. of a 3-movement series for each position.

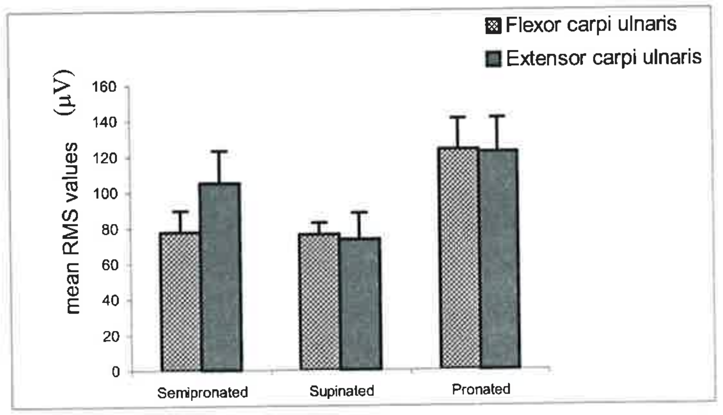

Figure 1. Mean RMS values of the flexor and extensor *carpi ulnaris* muscles obtained in flexion movements of the forearm in semipronated, supinated and pronated positions.

EMG STUDY OF THE FOREARM FLEXOR AND EXTENSOR MUSCLES AT DIFFERENT POSITIONS, LOADS AND ANGLES

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INTRODUCTION

The forearm flexor (brachii biceps, brachialis and brachiorradialis) and extensor muscles (brachii triceps) which act on the elbow joint have been studied separately in most cases for decades and almost always in isometric conditions. CAVANAGH (1974), already affirmed at that time that the limitation of comparisons between muscles was even more restrict in the dynamic situation, that is, in isotonic movements. The simultaneous action of mechanically agonist muscles and those considered antagonist and synergist of a certain movement needs more detailed studies, especially at the elbow joint. However, some authors investigated the behavior of forearm flexor muscles while others analyzed the action of extensor muscles. Few authors (BASMAJIAN & LATIF, 1957; BUCHANAN et al., 1989; FUNK et al., 1987), verified the simultaneous action of flexor and extensor muscles of the forearm.

According to WINTER (1979), biomechanics is based on the utilization of measurement procedures that allow the obtaining of different parameters of human movement. However, for ARAZJO & AMADIO (1996), the determination of internal forces is only possible with the development of models where their movement parameters are supplied by the basic measuring procedures: cinematography, dynamometry, anthropometry and electromyography. Considering this information, an electromyographic study of the simultaneous action of the brachii biceps (short and long heads), brachialis, brachiorradialis and brachii triceps (lateral long and medial heads) was made through isotonic forearm flexion and extension exercises. These exercises were carried out in pronated and supinated positions, free and with a load of 04Kg and 10Kg at angles of 30:, 60: and 90: for flexion, and 90:, 60: and 30: for extension utilizing an eletrogoniometer.

MATERIAL AND METHODS

The simultaneous action of the brachii biceps (long and short heads), brachialis, brachiorradialis and brachii triceps (lateral long and medial heads) muscles was studied electromyographically in 10 young adult male volunteers, right-handed, with no training and no history of articulation or neuromuscular disease which could influence the results, during free flexion and extension movements and with loads of 04Kg and 10Kg, in the pronated and supinated positions. The exercises were doneon a classic exercise machine called Double Pulley.

The results were recorded on an eighth channel Viking II computerized eletromyograph, calibrated for an amplitude of 500 microvolt/division and recording speed of 200 milliseconds/division resulting in a total recording time of 4 seconds. The filters were set at 10Hz (low frequency) and 10 KHz (high frequency). The signals were recorded using a pair of surface monopolar electrodes Beckman type, with an 11 mm diameter and 2 mm detection surface, placed under the skin of the studied muscles, after cleaning and tricotomy following BASMAJIAN & DeLUCA (1985) recommendations. An earthing electrode connected to one of the channels was also used to eliminate possible interferences in the signal. An electrogoniometer, placed on the right upper limb of the volunteer and connected to the eighth channel of the eletromyograph pre-amplifier, was used to determine forearm flexion (30:, 60: and 90:) and extension (90:, 60: and 30:) angles, showing the signal of the electrogoniometer on the monitor displaying the predetermined angles together with the signal produced by the muscles. For signal analysis, a computer program called SISDIN was used, which supplied

the numerical data in RMS (root mean square) of the electromyographic records expressed in microvolts (?V). The values in RMS were chosen for analysis, because according to BASMAJIAN & DeLUCA (1985), this is the value that gives the most information about the electromyographic signal.

RESULTS

The results of the research show that all muscles studied present electromyographic activity in the isotonic flexion and extension movements of the forearm, independent from the type of movement, load or forearm position. The values encountered in the supinated position were, in most cases, higher than those obtained from the pronated position.

DISCUSSION

The results of this research suggest, like BASMAJIAN & DeLUCA (1985), that the brachii biceps, brachialis, brachiorradialis and brachii triceps muscles must be better understood regarding their integrated functions. These results are also in concordance when they affirm that the brachialis muscle is active in all of the forarms positions, because its traction line is not altered with supination or pronation and that the brachiorradialis acts better in the pronated and semi-pronated positions. SOUSA (2000) is also confirmed by this research which considers the brachiorradialis muscle very active in the semi-pronated position. The results of this research are conflicting with these authors when they mention that the brachii biceps presents little or no activity in most cases when the forearm is pronated, and that it would be antagonistic during the extension, even with load. This research found evident potential in this muscle, especially in the pronated flexion with load, although presenting greater activity in the supinated position with load, with the short head more active in the pronated position and the long head more active in the supinated position, except in the supinated extension with load, where the long head presented the least activity.

CONCLUSIONS

Through the analysis of the results obtained in this study, according to the procedures employed, we can conclude that there are no positions or movements of the forearm where a single muscle is active alone. There is always simultaneous action of those muscles. The electrical activity of those muscles considered antagonists increased with greater loads applied. The forarms flexor and extensor muscles in supinated and pronated positions act simultaneously in the movements of flexion and extension, considering that electrical activity was always found in those muscles considered antagonists to the movement.

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ELECTROMYOGRAPHIC ANALYSIS OF THE PATELLA STABILIZIER IN SUBJECT WITH PATELLOFEMORAL DYSFUNCTION IN OPEN KINETIC CHAIN

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INTRODUCTION

Patellofemoral Dysfunction (PFD) is a pathology that affects 20 to 36% of the population in general[5], especially teenagers and young adults between 10 and 35 years old. It is more common among women and athletes[9]. The main symptom is diffuse pain on the anterior region of the knee, during or after activities[9]. The ethyology of PFD includes factors like: trauma, overuse, ostheocondral alteration, meniscal lesion, irritation of the sinovial plica, ligament weakness and bad patellar alignment[12]. Any insufficiency or instability between the medium (VMO) and lateral (VLO, VLL) dynamic stabilizers, can cause lateral dislocation or bad patellar alignment, leading to PFD[9]. The participation of the VMO as the support of the patellofemoral joint alterations conservative treatment, needs more research[7]. Little is known about the electric activity of the lateral components of the VLL and VLO quadriceps that are important patellar stabilizers. Only BEVILAQUA-GROSSO (1998) studied these muscles function in OKC and verified functional differences between these parts in OKC in normal subjects. MORRISH & WOLEDGE (1997), analyzed the activity of these muscles in subjects with PFD but only at 20° of flexion in OKC. Both of these studies revealed that the activation of these muscles was practically synchronic, suggesting that the VMO and VLO have reciprocal activity on the control of the patellar position. Therefore, the goal of this study is to verify, by the electromiographic activity, the functional differences of the VLL and VLO muscles at maximum voluntary isometric contraction (MVIC) in OKC in normal and PFD subjects in different angles of knee flexion.

MATERIAL AND METHOD

20 volunteers from the female sex between the ages of 18 and 25 years (x= 23,6; SD= 3,53) were selected by a physical evaluation. The volunteers were divided in two groups, 10 clinically healthy and 10 with PFD. To register the electromiographic activity, simple differentiated surface active electrodes were used with a 10 fold gain. The conditioner of the electromiographic signal was used to amplify and to filter the sign (gain: 4400, filter bandwidth,: 20 Hz-4KHz) and the rate of acquisition utilized was 2kHz rate with 12 bits of resolution as sample. The electromiographic signal was quantified through the root mean square (RMS) and normalized by MVIC extension of the knee at 90 degrees. The exercises were made on a flexo-extension table, providing 3 MVIC at the angles of 30°, 60° and 90° of knee flexion. ANOVA (Tukey test) and the Wilcoxon Signed Rank Test were used.

RESULTS

In normal subjects, the electric activity of the VMO muscle was higher at 30° than at 60° (p=0,01). In MVIC at 30°, the VLL and VMO muscles were more active than the VLO muscle (p<0,005). In MVIC at 60°, no significant statistical differences were observed between the muscles. The subjects with PFD did not present differences on the behavior of these muscles at the studied angles. The comparison among the groups revealed that there is no difference of activation between the normal and PFD subjects.

DISCUSSION

Some studies that found higher activity of the VMO in normal subjects at the last degrees outstand in the literature[2,6]. The electric activity of the VLL differs from the VLO at 30° of knee flexion being, therefore, functionally different, fact that is similar to the results of BEVILAQUA-GROSSO (1998), although this one was not realized at 30°, the functional difference was verified between these parts. In spite of these methodological differences, some authors did not observe differences in the electric activity of the VLO and VL muscles of normal and PFD subjects[4,10]. Nevertheless, these authors analyzed CKC and did not analyze VLO. ANDRADE (2000) analyzed the activity of VMO and VLO of normal and PFD subjects in CKC, not finding significant functional differences. MORRISH & WOLEDGE (1997), also did not find difference on the activation of the VMO and VLO muscles of normal and PFD subjects, despite analyzing only 20° of knee flexion.

CONCLUSION

The VLL muscle and the VLO muscle are functionally different in normal subjects, being able, therefore, to be considered two different parts of the thigh quadriceps muscle. The muscular instability does not seem to be a preponderant factor on subjects with PFD. This way, biomechanical and neuromotor alterations must be considered and corrected with the same emphasis given to the strengthening of VMO. Other etiologic factors must be analyzed in PFD, since a reduction on the activity of the VMO muscle in subjects with PFD was not found.

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VARIABILITY OF MOTION IN BACK-LIFTS VERSUS LEG-LIFTS

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INTRODUCTION

Lifting tasks have long been recognized as a source of low back injury and various musculoskeletal problems[1,2]. Bending forward in order to move/lift an object generates a high bending moment and high compression forces on the osteoligamentous structures of the lumbar spine[2]. Most biomechanical studies favored leg-lifting with the trunk erect and the load kept as close as possible to the trunk, when the object is lifted between the knees[3]. Contraction of the paraspinal muscles, the built-up of abdominal pressure and the simultaneous contraction of antagonistic muscle pairs have been identified as the major mechanisms that actively contribute to spinal stabilization. This protective ability of the trunk muscles is closely related to postural stability[4].Aim of the study is to compare back-lifting and leg-lifting in terms of postural stability.

METHODS

Nineteen healthy subjects (15 male, 4 female) participated in the study. The subjects performed two different tasks: 10 consecutive back-lifts and 10 consecutive leg-lifts, resting for at least five minutes between tasks. Subjects lifted and lowered the load sagittal symmetric in front of them between floor and desk level (height 70 cm). The box weighted 3.2 kg according to 1994 NIOSH – Recommended Weight Limits; lifting index of 1 in the defined environment[3,6]. We measured kinematics of segmental motion to identify joint specific range and variance. A videometric system was used (Motion Analysis Corp. High-Res., Santa Rosa, CA - USA). Subjects wore a set of 17 retroreflective markers located on both body sides according to the standard Cleveland Clinic marker set.

RESULTS

Coefficient of variation (CV) is defined as a quotient of standard deviation and mean.
Subjects age was 27.2 (SD 5.4) years; body weight was 70.3 (SD 14.1) kg; and body height was 171.3 (SD 9.2) cm. Two subjects reported on discomfort during back-lifting due to stooping limits. Eight subjects reported on fatigue during leg-lifts.
All subjects demonstrated well developed motion pattern in all measured joints (CV<50%). Variance of motion was distributed across the lifting cycle. When all data were pooled together, range of motion was positively correlated with variance of motion for back and leg-lifts coeff.cor$_{back}$ = 0.81; coeff.cor$_{leg}$ = 0.70). Large range of motion is correlated with large variance. The mean CV demonstrated the S.D. to be approximately 6.5% of range of motion.
Range of motion was effected by lifting technique in all joints. Back-lifting demonstrated significantly higher range of motion in trunk, pelvis and arm kinematics as compared to leg-lifting, while. leg-lifting showed significantly higher range of motion in leg joints.
CV values were effected by lifting technique in all joints, except for elbow joint. Back-lifting demonstrated significantly higher CV in leg and shoulder joints as compared to leg-lifting, while leg-lifting showed significantly higher CV in orientation of trunk and pelvis.

DISCUSSION

In back-lifting we found increased CV for leg motion. In leg-lifting we found increased CV for trunk motion. Corrective motion in inaccurate performance and balance stabilization seem to be increased at joints, which are less involved in the focal movement of lifting.
Our study shows that relative variance in trunk motion are larger during leg-lifts as compared to back-lifts. Inter trial CV of trunk motion was found to be higher in leg-lifts as compared to back-lifts in fast and slow pacing [7]. There are three possible reasons, contributing to the increased relative variance in trunk orientation during leg-lifts:
1) The complex interaction of hip, knee and ankle joint results in more variable pelvis position and orientation. Thus, there is a need for corrective motion in the upper body to perform the lifting task. Inter trial variance in arm motion is not increased, suggesting that most compensatory action is taken over by the trunk.
2) There is extreme dorsiflexion of the ankle. Some subjects even started to lift their heels during preparation trials for leg-lifting task. Such behavior reduces stabilizing capability at the ankle as postural responses are only efficient in one direction (plantarflexion). Reduced postural capability in the lower extremity may be compensated by trunk dynamics .
3) Several subjects reported on fatigue during one minute leg-lifting activity. These subjects may have altered their movement pattern during repetitive trials as response to fatigue[5]. In addition decreased postural stability has been reported to occur during fatiguing repetitive lifting. A significant decrease in postural stability suggests a higher risk of injury in the presence of unexpected perturbation such as slipping.
In contrast, during back-lifts we found smaller relative variance at the trunk levels. This may be due to the hip dominated lifting motion with knee and ankle remaining almost in neutral position. Postural ankle responses are thus possible in two directions. Increased CV at the ankle during back-lifts show, that this postural control capacity is used. In back-lifts there is increased range of trunk motion. Sagittal trunk moments are monotonically increased with trunk flexion[7]. This may lead to overuse back injury risk in back-lifts[3].

CONCLUSION

In self-selected manual lifting from floor subjects adopt intermediate posture. Our findings suggest that this is to avoid instability during leg-lifting and injury risk during back-lifting.

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A COMPUTER-BASED SYSTEM FOR ERGONOMICS AND KINESIOLOGIC EVALUATION

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INTRODUCTION

For the past few years, several computer-based systems have been developed for analysing biomechanic behaviour of humans, contributing for a better comprehension of anti-ergonomic situations. Such systems are used to detect complex problems in Ergonomics and to help professionals who work with pathological diagnostics related, for instance, to the work environment. However, such systems present some drawbacks with respect to statistic data generation, and are usually limited to the evaluation of very specific parts of human body (1,2). Based on these limitations, coupled with the need for a system able to store and manipulate data for different ergonomics situations, a computer-based system has been developed and is presented in this paper.

METHODS

The system, called Ergosoft, supports the synchronisation between electromyographic (EMG) signals and captured images of human movements. Such system was designed according to an analysis of requirements detected by interviewing some Kinesiologic and Ergonomics professionals. They agreed that it was very important to support the EMG signals along with respective images, during a patient movement. Therefore, a system was designed to trigger two processes at the same time: a) patient movements captured by a camera and b) an electromyograph device (3) to capture the EMG signals (through 5 surface electrodes) from the patient movements. Next, both the captured images and EMG signals were stored to be analysed by the Ergosoft.

RESULTS

Figure 1 shows a frame (in Portuguese) of the system with a captured image and one of the EMG signals (from biceps head muscle) captured during the movement carried out by the patient. In this picture, the patient´s image is related to the vertical bar on the EMG signal. The user is able to move this bar, and for each new position, its respective image is displayed by the system. The user can, in each frame, calculate the angle given by the arm movement, by only clicking three points on the image. Besides, it is possible to display all EMG signals captured by the 5 surface electrodes.

DISCUSSION

The system was presented to some experts and they have confirmed that all signals captured and presented by the system were satisfactory. Besides, being able to detect the exact moment where each muscle starts and ends its electric activity - followed by its related image - was a great achievement, since this has not been observed in other systems. Furthermore, the system is able to present detailed statistics and full visualisation of the EMG envelope and to print detailed reports of the case study in hand. All these features are used by professionals to give their diagnosis.

CONCLUSIONS

The main contribution of this work is the study of computer techniques to support ergonomics and kinesiologic evaluation in a more complete, efficient and interactive way. Through the case studies taken into account to evaluate the system it is possible to conclude that professionals, using the system, can perform their experiments with more precision for several parts of the human body.

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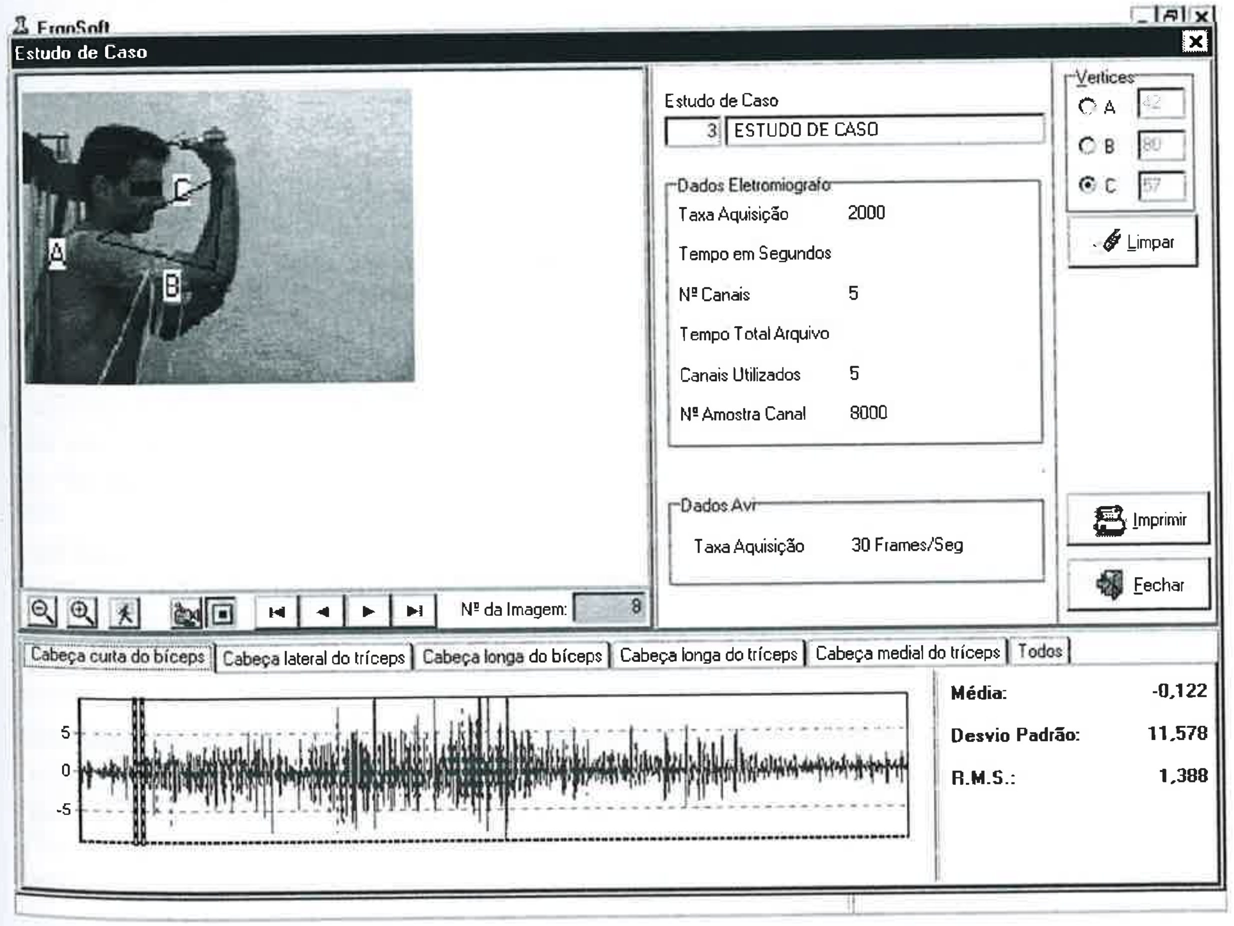

Figure 1: Ergosoft frame with a captured image and the EMG signal.

A PORTABLE MULTICHANNEL EMG ACQUISITION SYSTEM

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INTRODUCTION

The use of multichannel surface EMG techniques for the investigation of the neuromuscular system is finding growing interest in the fields of basic research, rehabilitation, sport and occupational medicine. Field studies require portable, long term recording equipment. Although some systems are commercially available, they only allow acquisitions from bipolar EMG electrodes and, usually, with a limited number of channels. The aim of this project was the design of a portable system that could perform long term EMG recordings from one- or two dimensional electrode arrays, which have several and well known advantages over simple bipolar EMG acquisition.

METHODS

To achieve the goal, a technical partnership has been signed between LISiN Bioengineering Centre (Dip. di Elettronica, Politecnico di Torino) and Sirio Automazione s.r.l., merging the know-how of the first and the technical facilities of the second; the project has been developed with grants from the European Union within the European Project NEW (QLRT 2000 00139), focused on the neuromuscular assessment on elderly workers.

RESULTS

A portable datalogger for long term recording from EMG electrode arrays (see figure) is now available and ready to be used in field studies. It features 64 input channels with a sampling rate of up to 2000 samples/s per channel. EMG probes with four, eight, or 16 channels each can be connected. The stability of the skin electrode contact over long term recordings is ensured by the use of patented adhesive arrays, which are available with different number of electrodes and interelectrode distances.

For each channel, the EMG signal is collected by a single differential input stage, band-pass filtered and then converted by 16-bits A/D converters; the CMRR (Common Mode Rejection Ratio) of the input stage exceeds 95 dB and the noise is less than 1 uV_{RMS}. Each probe has an additional input channel with embedded front-end for use with a mechanomyogram sensor or as a general purpose input.

All the signal conditioning, amplification and A/D conversion is performed directly into the probe and the samples are transmitted over a thin and flexible cable by means of a digital serial protocol; this architecture improves the wearability of the probe with respect to traditional analog types, and enhances the noise immunity even in case of long connection cables.

Several technical solutions have been implemented to obtain a user friendly equipment; among the others, an embedded impedance monitor warns the user about degradation of the skin electrode contact quality and an auto recognition feature reliefs the user from configuring the system after plugging the probes.

The user interface is composed of a graphical LCD display with touchscreen, where an intuitive icon based menu structure gives access to all the datalogger functions. A real time scope feature allows the user to check the signal quality (four channels at a time), directly on the LCD screen, before starting an acquisition.

Acquired data are stored on a standard PCMCIA type II memory card, that can be removed and plugged into any portable PC where a proprietary sofware allows the user to off line download, review and process EMG signals. Depending on the number of used channels and card size, acquisitions up to several hours are possible.

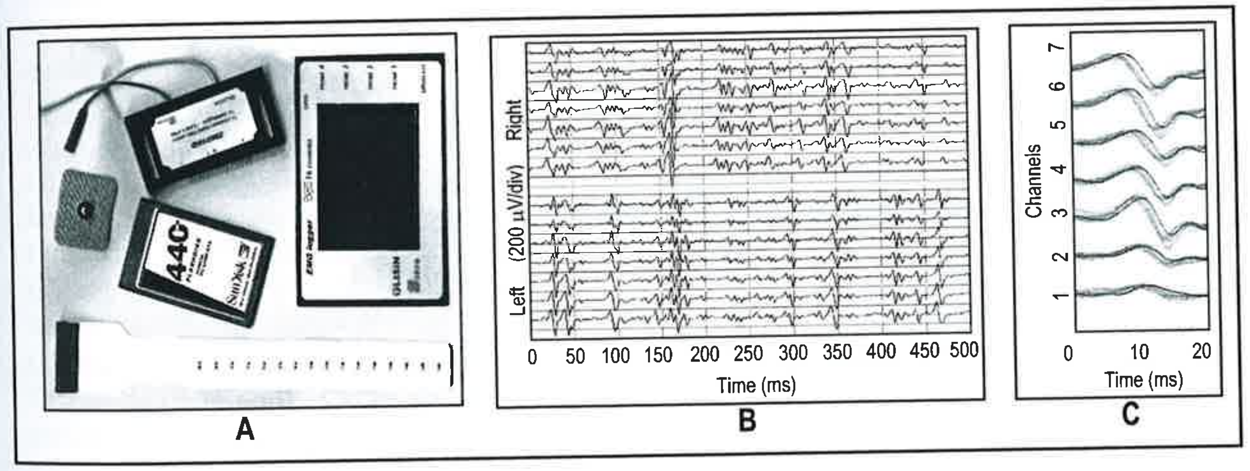
The datalogger is battery-powered and meets all the safety standards that apply to biomedical equipments.

DISCUSSION AND CONCLUSION

Some prototypes, that will be extensively used for investigations within the "NEW" European Research Project (QLRT 2000 00139), have been distributed for evaluation to several European research groups, who provided useful feedback for improvements.

An extensive test of the device has been recently performed at the ETH Centre of Hygiene and Applied Physiology in Zurich, acquiring EMG signals from left and right upper trapezius muscles of ten subjects, during a five minutes low level substained contractions (2% and 5% MVC) and during a 40 minutes long standardized computer task. For additional research purposes, intramuscular signals were also recorded with a Dantec electromyograph, whose data were synchronized with surface EMG signals by means of the additional input channel.

This pilot experiment, whose results are not reported here, showed that the quality of the signals was satisfactory over time, even at very low contraction levels, proving the reliability and the ease of use of the acquisition system.



A) the datalogger system, composed by main unit, a 16-channels probe, an adhesive array with 16 electrodes and 10 mm interelectrode distance and the PCMCIA memory card;
B) EMG signals acquired from left and right trapezius muscle during 5% MVC substained contraction; during the task, the subject was provided with visual force feedback;
C) averages of single motor unit action potentials during (grey) and at the end (black) of a fatiguing contraction (B), obtained with spike triggering averaging using intramuscular potentials as triggers. The datalogger and the intramuscular electromyograph were synchronized by means of the additional input channel.

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JOB AND ORGANIZATION ANALYSIS AND PERCEIVED RISK OF AGRESSION

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INTRODUCTION

Violence is one of the most important occupational risks in the mental health sector. We observe a high degree of complaints related to violence in the Mental Health Service, however the number of aggressions is low as well as the degree of physical harm, though the emotional harm is high. In order to develop a program to deal with violence in the Mental Health administration we conduced an evaluation of the job task and organization. Its management includes organizational and engineering measures.

MATERIAL AND METHODS

Information was collected by self-administered questionnaire, which included the performed tasks, the percentage of time in each task involving perceived violence, and the degree of the perceived risk, interpersonal relationships, task content, autonomy, and degree of task definition.
Some information was collected by observation method.

RESULTS

The study involved 10 centres and 187 workers. A sample of 106 occupations was taken, for a 100% response. The absence of control over the situations of violence at work is found to be associated to organisational factors and this is in turn a source of occupational stress and low level of satisfaction. The most relevant factors contributing to these results are:

1. Low level of task definition
2. Vague and imprecise orders
3. Conflicts in the line of command
4. Low level of information about results
5. Absence of written procedures to deal with violence
6. Low level of training to manage violent episodes and to perform the job at large
7. Absence of a policy to incorporate new workers
8. Low level of empathy among workers and between workers and managers.

DISCUSSION

As it usually occurs, the workers complaints about the level and frequency of an occupational risk hide complex organisational factors. In this study we found that the perceived risk of aggression is very high while the rate of aggressions is low. We think that much of this perception is modified by the conditions at work, which precipitates stress and inappropriate behaviour. Targeting only the risk factors for violence would result in a failure to control the situation. We think that we need to introduce a new job organisation, which includes a better definition of tasks and of line of command, introduction of feedback, and higher level of decision and accountability.
In order to accomplish this goal, we are working with the managers, trade unions and workers in a new organisational set.

CONCLUSIONS

1. Work organisational factors and interpersonal relationships act as modifiers of the perception and occurrence of violence. They should receive a priority attention in the design of interventions,
2. The main consequences of work injuries because of violence are not the physical harm, most of the times of mild severity, but the psychological and behavioural harm.
3. Workers, and co-workers, who suffer from violent injuries perceive a high degree of vulnerability against this risk. As a consequence we observe frequent problems of stress, role conflict and absence from work.
4. In the paradigm of the interaction agent-host-environment, we believe that the preventive measures should address the environment. This is the work organisation factors that we have detected as facilitators of violence and violence response.
5. Included in the improvement the organisational factors should be the best quality care of the patients, which we understand should contain the violent behaviours.
6. We believe that this should be completed with measures to increase the host ability to deal with violence, including, medical, psychological and social support of the victims.

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NEGATIVE AIR IONS IMPROVE EFFICIENCY OF ERGOMETRIC EXERCISE

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INTRODUCTION

Negative air ions have a purifying effect on hazardous substances, such as dust, bacteria. Many negative ions are detected in forests, at spas, and near waterfalls. Many reports have been published that the negative ions improve the subjective feeling, comfort, fatigue and occupational efficiency, but few evidence of the mechanism have been reported yet.

To investigate the physiological effect of negative air ion for humans, we analyzed the hormonal and immunological changes during the ergo metric exercise.

METHODS

In the artificial climate room(25oC, 50% humidity), ten young healthy volunteer were double-blindly exposed in negative-ion rich air (>10,000counts/cc: produced by ionizer equipment) or not. They were installed the caldiopulmonary testing instrument (Sensormedics Co, Vmax29c) and tried their limit by ergo metric exercise (gradually increased 20W/min). We sampled 3 times their bloods before the exercise, at the maximum point and at recovery point (15 minutes after the end of the exercise). We tested hormonal(adrenalin, noradrenalin, dopamine, cortisol and serotonin), immunological stress responses (white blood cell and lymphocyte subsets) and metabolic products (pyrubic acids and lactic acids). Paired t-test was used for statistical analysis.

RESULTS and DISCUSSION

Negative air ions significantly elevated the total amount of exercise(negative ion rich: 73.8±14.7kcal, negative ion lack: 70.3±22.3kcal). There were no statistical difference in O_2 consumption of maximum and of AT (anaerobic threshold) points.

Negative air ions decreased the hormonal stress response and decreased dopamine level at 15 min after exercise ($p<0.05$) and diminished the pyrubic acid and lactic acid at the max exercise ($p<0.05$). And negative air ions diminished the immunological stress response (total white blood cells, NK subsets).

This supported many reviews and suggested that the negative ions improved the metabolic efficiency of human. So we would need the further evidence of the ions effects.

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Table. Laboratory findings during and after exercise with negative air ions

*: p<0.05

Lymphocyte Subset(%)	PreExercise Mean ± SD	atMax Mean ± SD	Recovery(15min) Mean ± SD	Deference □ Max-pre Mean ± SD	Deference □ Recovery-pı Mean ± SD
Without Negative ions					
Lactic acids(3-17)mg/dl	13 ± 8.32	81.3 ± 21	55 ± 14.9	68.3 ± 21.3 *	42.1 ± 20.2
Pyrubic acids(0.3-0.9)mg/dl	1.13 ± 0.59	2.33 ± 0.6	2.69 ± 0.85	1.2 ± 0.48 *	1.56 ± 0.76
Cortsol(4-18)μg/dl	13.1 ± 3.73	15 ± 5.16	15.1 ± 5.65	1.89 ± 2.7	1.98 ± 3.03
Epinephrin(<100)pg/ml	41.4 ± 10.6	149 ± 109	65.4 ± 18.1	107 ± 107	24 ± 20.9
Norepinephrin(100-450)pg/ml	314 ± 118	1275 ± 822	522 ± 215	961 ± 777	208 ± 162 *
Dopamin(<20)pg/ml	8.75 ± 6.39	11.4 ± 8.14	13 ± 9.02	2.63 ± 10.2	4.25 ± 9.82 *
Serotonin(0.04-0.35)μg/ml	0.08 ± 0.02	0.1 ± 0.03	0.1 ± 0.05	0.03 ± 0.02	0.02 ± 0.03
WBC(/ μl)	5813 ± 1472	8213 ± 2215	6863 ± 1577	2400 ± 1301	1050 ± 1439
RBC(x104/ μl)	489 ± 43.2	511 ± 55.1	506 ± 41.4	22.3 ± 20.3	16.8 ± 13.9
Platelet(x104/ μl)	24.1 ± 2.53	24.7 ± 2.79	24.2 ± 2.42	0.66 ± 1.28	0.11 ± 1.27
NK(CD16)	25.5 ± 9.89	40 ± 15.6	27.6 ± 10.2	14.5 ± 9.86	2.13 ± 9.01
With Negative ions					
Lactic acids(3-17)mg/dl	12.7 ± 5.87	69.4 ± 18.6	48.3 ± 8.35	56.7 ± 16.8	35.6 ± 12.3
Pyrubic acids(0.3-0.9)mg/dl	1.16 ± 0.53	2.08 ± 0.44	2.57 ± 0.35	0.92 ± 0.45	1.41 ± 0.45
Cortsol(4-18)μg/dl	12.8 ± 5.97	13.9 ± 5.83	15.3 ± 4.78	1.06 ± 2.41	2.49 ± 5.45
Epinephrin(<100)pg/ml	31.5 ± 12.6	98.5 ± 52.2	48.3 ± 17.3	67 ± 54.4	16.8 ± 12
Norepinephrin(100-450)pg/ml	270 ± 69.6	1028 ± 489	384 ± 144	758 ± 465	115 ± 119
Dopamin(<20)pg/ml	11.3 ± 7.01	15.4 ± 5.01	9.25 ± 9.91	4.13 ± 4.73	-2 ± 7.45
Serotonin(0.04-0.35)μg/ml	0.05 ± 0.03	0.06 ± 0.03	0.05 ± 0.02	0.02 ± 0.02	0 ± 0.03
WBC(/ μl)	5300 ± 904	7638 ± 1346	5738 ± 1314	2338 ± 843	438 ± 825
RBC(x104/ μl)	480 ± 39.5	519 ± 26.4	492 ± 45.2	38.3 ± 32.7	11.6 ± 11.4
Platelet(x104/ μl)	23.4 ± 2.82	24.9 ± 3.04	23.7 ± 2.8	1.5 ± 0.55	0.33 ± 0.93
NK(CD16)	22.9 ± 9.49	38.9 ± 9.15	25.2 ± 12.4	15.9 ± 6.82	2.24 ± 4.48

TEST-RETEST RELIABILITY FOR RECORDING THE ERB'S POINT POTENTIAL WITH A CHANGE OF RECORDING POSTURE

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INTRODUCTION

In the field of physical therapy, we have tried to evaluate the central neural function on various types of the posture using the silent period[2)3)]. However, to clarify the central neural function for postural adjustment, it was difficult to evaluate the central neural function using the silent period on the dynamic posture: ambulation, running and so on, because typical condition for testing silent period was to keep a posture with making a voluntary muscle contraction to evoke a muscle response.

SEPs have emphasized changes in the amplitude and waveform of recorded potentials in patients with the cerebrum or spinal cord disorder and have focused on the evaluation of central neural conduction determined by the latencies of the SEPs recorded over the spine or scalp[1)]. As SEPs is useful to evaluate the central neural function and its recording is not needed to make a voluntary contraction of any muscles, we have considered to apply SEPs for the evaluation of the neural function on various types of the posture. However, it has not been seen any research reports about neural function using SEPs under a condition related to postural alteration. So, the present study was performed as the preliminary study to clarify the relationship between the median nerve SEPs and a postural alteration (i.e., lying, sitting, standing…). The purpose of this study was to clarify the reliability using a test-retest maneuver for median nerve SEPs focused to Erb's point potential, which was adequate for testing the reliability because it did not affect by spinal and/or cerebral function, with a change of recording posture using a tilt table and to compare the latency and amplitude on lying at various angles of tilt table.

METHODS

Subject was one healthy male volunteer with an age of 28 and a height of 176cm. He was informed about an outline of present study, and consented to participate in this study.

Median nerve SEPs was recorded in spine position on the bed with changing the tilting angle. Tilting angle of the bed was set at each of 0, 15, 30, 45, 60, 75, and 90 degree using Tilt-table SPR-3000 (Sakai Medical Co., Ltd., Japan). Median nerve SEPs with each tilting angle was recorded randomly with an incidence of once a day per a week with totaled three times. Electric stimulus were applied to the median nerve at the wrist under following conditions; intensity of 120% of motor threshold of thumb, duration of 0.2ms, frequency of 3.3Hz and number of 1000 times. An active electrode was put on the supraclavicular fossa (Erb's point) and a reference electrode was placed on the center of forehead (Fz) recommended the ten-twenty electrode system of International Federation of Clinical Neurophysiology. We used an averaging method with 1000 times. We determined the variation of the latency and amplitude of Erb's point potential at each tilting angle with calculating the a coefficient of variation (1) and compared the value of latency and amplitude at each angles (2).

RESULTS

(1) A coefficient of variation (%) of the latency and amplitude at from 0 to 90 degree was 10.0 and 19.0, 20.0 and 17.3, 10.0 and 19.6, 10.1 and 11.9, 10.1 and 12.2, 10.0 and 10.5, and 10.0 and 8.1, respectively. And the latency and amplitude of Erb's point potential at each angle was almost same in all trials. (2) There were not any significant differences of the latency of Erb's point potential at each tilting angle. The amplitude at 0 degree tended to decrease compared with those at other tilting angles.

DISCUSSION AND CONCLUSION

It was said that the amplitude of SEPs was the index of excitability of generator of potential and the latency indicated the conductive condition of pathways. The potentials acquired with SEPs were from volume conductors surrounding the generator resources[1)]. Therefore, a location and distance of electrodes, a proportion of volume conductor or alignment between generator and electrode affected waveforms[1)]. In the field of physical therapy, we have tried to evaluate the neural function on the various posture using evoked electromyography in upper[2)] and lower[3)] extremities. As the following study, we tried to apply the SEPs to evaluate the neural function on various postures and some dynamic situations. In the physical therapy evaluation, we should perform the SEPs recording on various posture or dynamic situation. So, it was needed to clarify the reliability of recording SEPs under such conditions. Moreover, for studying the reliability of SEPs in upper extremity, it was useful to compare the Erb's point potential because its generator was thought to be originated from a distal portion of the brachial plexus[1)] and it did not be affected the excitability of the spinal cord or cerebrum.

This study was the preliminary study for testing the median nerve SEPs with a postural alteration. And we investigated the test-retest reliability for the state of Erb's point potential in the median nerve SEPs with a change of recording posture. The findings in this study included: 1) A coefficient of variation of the latency and amplitude at each tilting angle was low value (8.1%-20.0%). 2) There was no significant difference of the latency at each tilting angle. 3) The amplitude at each angle was almost same in all trials. 4) The amplitude at 0 degree tended to decrease compared with those at other tilting angles. Erb's point potential reflected an abrupt geometric change of the volume conductor, anatomic orientation of the impulse, and branching of the nerve[1)], a brachial plexus injury or a disorder of sensory nerve conduction. A disorder of sensory nerve conduction was caused by ischemia[3)] or nerve traction due to postural alteration[4)]. From results of this study in a healthy subject, all Erb's point potentials of median nerve SEPs with a change of recording posture were recorded reproducibly and availably. Accordingly, it was expected that median nerve SEPs was able to record on various postures among lying, sitting and standing with reproducibility and availability. As to the decline of the amplitude at 0 degree compared with at other angles, because a subject in this study was healthy male, there was no symptom related to nervous disorder, an ischemic portion and an alteration of conductive condition. Consequently, the decline of the amplitude at 0 degree was thought to result from an alteration of the situation regulated by a distance and a direction between the active electrode and the generator of Erb's point potential via an angle changing. In conclusion from the results of Erb's point potential, median nerve SEPs was able to record on various postures in this study reproducibly and availably. But, in its recording, we have to pay an attention to the possibility of changing the size of potential between supine position and other positions.

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H-REFLEX FROM UNAFFECTED HAND MUSCLE IN HEMIPARESIS DURING STRAIGHT LEG RAISING

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INTRODUCTION

In physical therapy for patients with cerebrovasucular disease(CVD), it was generally thought that the function of movement on the unaffected side was not always normal. We experienced a lot of patients with CVD whereby coordination of movement in the upper extremity on the unaffected side was disturbed. And excitability of the spinal neural function on the unaffected side was increased in patients with CVD. We investigated the relationship between excitability of the spinal neural function in the upper extremity on the unaffected side and voluntary movement of the lower extremity in patients with CVD. This study was to examine the variation of H-reflex from unaffected hand muscle during straight leg raising in patients with CVD.

METHODS

Subjects were 4 patients with right hemiparesis caused by CVD and 4 healthy subjects. Healthy subjects did not have any neurological and orthopedic deficits.

We chose the supine position with keeping a straight leg raising (SLR) position bent the hip at 30 degrees as the testing task. And the task in this study was classified to 4 types as follows ; passive SLR position in the unaffected side (trial 1), passive SLR position in the affected side (trial 2), active SLR position in the unaffected side (trial 3), and active SLR position in the affected side (trial 4). In each trials and rest, the H-reflex under stimulus to the mediam nerve at the wrist was recorded from the abductor pollicis brevis muscle in the unaffected side. The stimulus intensity for evoking the H-reflex was 1.2 times stronger than that for threshold to evoke the M-wave. The duration, frequency and times of stimulation were 1ms, 0.3 Hz and 10 times, respectively. The H-reflex was analyzed and the amplitude ratio of H/M was determined. We compared the amplitude ratio of H/M during 4 testing task with that at rest. Statistical significance was tested using Student t-test.

RESULTS

The amplitude ratio of H/M in each trials was displayed on the figure. In healthy subjects, no significant difference was recognized in comparison among rest and all trials. In all patients with CVD, the amplitude ratio of H/M in trials 3 and 4 tended to increase than that at rest. In 2 patients, the amplitude ratio of H/M in trials 3 and 4 was significantly higher than that at rest ($p<0.05$).

DISCUSSION

The amplitude ratio of H/M is thought to be a index of excitability of spinal neural function. As a result of this study, it was suggested that excitability of spinal neural function in the unaffected arm might be increased during active SLR position in patients with CVD. Excitability of spinal neural function in the unaffected arm during passive SLR position was increase than that during active SLR position. Therefore, excitability of spinal neural function was influenced by muscle contraction of trunk and lower extremity during active SLR

position in patients with CVD. Hess et al.[1)] suggested that excitability of motor pathways by voluntary muscle contraction in the ipsilateral and contrateral side was facilitated using intracortical mechanism caused by imaging of motor task and increasing of excitability of homologous motor pathways. In this study, the facilitation of spinal neural function might be occurred by 2 mechanisms as following; 1) imaging of motor task with lower extremity facilitated excitability of upper extremity area on the brain, 2) muscles contraction of trunk and lower extremity facilitated excitability of upper extremity area on the brain.

Suzuki et al. [2)] suggested that the excitability of the spinal motor neuron function on the unaffected side in patients with CVD affected by the degree of motor dysfunction in the upper extremity and fingers on the affected side. In 2 patients with moderately hypertonus, the amplitude ratio of H/M in the unaffected hand during active SLR position was significantly higher than that at rest. In other patients with slightly hypertonus, the amplitude ratio of H/M in the unaffected hand during avtive SLR position tended to be higher than that at rest. From our results and Suzuki's report, it was suggested that excitability of spinal neural function in the unaffected arm during voluntary movement of the lower extremity might be influenced by the degree of muscle tone in affected side at the brain.

CONCLUSION

We investigated the relationship between excitability of the spinal neural function in the upper extremity on the unaffected side and holding the position of the lower extremity in 4 patients with right hemiparesis caused by CVD and 4 healthy subjects.

As a result of this study, it was suggested that excitability of spinal neural function in the unaffected arm might be increased during active SLR position in patients with CVD. And the excitability of the spinal neural function on the unaffected side in patients with CVD might affected by the degree of motor dysfunction in the upper extremity on the affected side.

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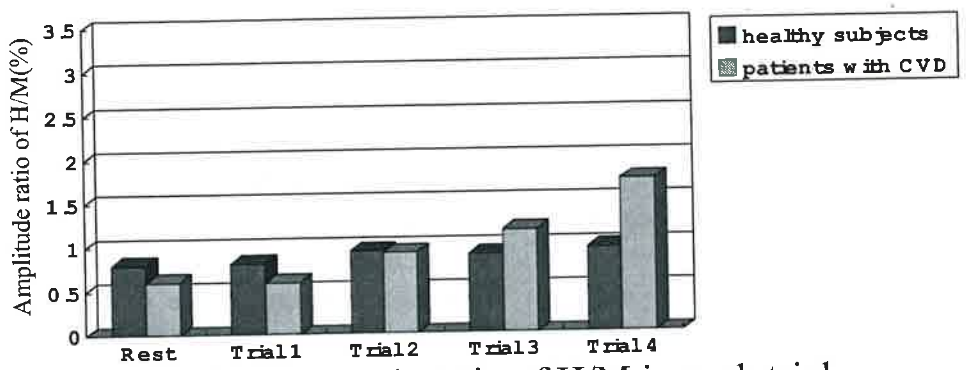

Figure. The amplitude ratio of H/M in each trial

Trial 1; Passive SLR position in the left side, Trial 2; passive SLR position in the right side, Trial 3; active SLR position in the left side, Trial 4; active SLR position in the right side.
In healthy subjects, no significant difference was recognized in comparison of during relaxation and all trials. In patients with CVD, the amplitude ratio of H/M in trials 3 and 4 was higher than that during relaxation.

MOVEMENT RELATED CORTICAL POTENTIALS ASSOCIATED WITH THREE KINDS OF BLINKS

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INTRODUCTION

There are three kinds of blinks; reflex, voluntary, and spontaneous blinks. The reflex blinks is evoked by the external stimulus which are the electrical stimulus, photic stimulus, or air-puffing. The voluntary blinks is caused by the intention of humans. The spontaneous blinks is caused unconsciously. Kaneko and Sakamoto (1) classified blinks into three types quantitatively by the use of parameters of EMG and EOG. Although the neurophysiological mechanisms of reflex blinks has been an object of study for a long time, little is known about the generation mechanism of the voluntary blinks and the spontaneous blinks in central nervous system.

The blinks is caused a reciprocal relationship between the contraction of m. orbicularis oculi related to the eyelid closer and the relaxion of m. levator palpebrae superioris which is responsible for keeping the eyelid open. We measured the movement related cortical potentials (MRCPs) associated with the contraction of m. orbicularis oculi related to the three kinds of blinks. These potentials of three kinds of blinks were compared each other to examine the participation of neural activities in the primary and/or the supplementary motor area in association with three kinds of blinks.

METHODS

Subjects were 12 healthy male students. Their mean age was 23.4 years old with a range of 22-28 years old. They all had a normal vision and did not wear glasses or contact lenses during the experiment. The subjects was sat in a comfortable chair in the experimental room that was maintained between 20 to 23 degree Celsius with relative humidity of 55-65%. The direction of eye was kept to the visual target. To begin with, the MRCPs in association with spontaneous blinks were measured at 100 times. It was required about 5-7 min to obtain the spontaneous blinks of 100 times. Next, the MRCPs related to the reflex blinks were recorded. The reflex blinks were evoked 12 times by the photic stimulus. Inter stimulus interval was 2 min to avoid the effect of habituation. Finally, the potential related to the voluntary blinks were measured 100 times. The blinks were repeated voluntarily at a self-paced interval exceeding 3 sec. The MRCPs were recorded using a Neuro-Pack Eight Computer (Nihon-Kohden, Japan). The time constant of the amplifier was taken to be 1.5 sec and the frequency of high-cut filter was set to be 100 Hz. The recording position of the MRCPs was Fz, Cz, and Pz (10-20 International System). The silver-silverchloride (Ag/AgCl) electrodes with 7 mm in diameter (Nihon-Koden, NE-121B) were used. The EMG activity related to blinks were recorded to obtain the trigger signal for use of "back-averaging" technique. The onset of blink (i.e., starting point of contraction of m. orbicularis oculi) was defined as time zero and triggered. The procedure of the recording of EMG was described in our previous paper (1). The resistance of all electrodes were less than 5 kΩ. Data were digitized on-line at 1000Hz and stored on hard disc. The waveform of MRCPs were averaged in the section for 2 sec before and 1 sec after the triggering point.

RESULTS AND DISCUSSION

The MRCPs preceding the voluntary blinks was evoked in all subjects. The lamp-like potential appeared in the range of about –400 to –700 msec with a mean value of -577 ± 151 msec in Cz

and -570 ± 153 msec in Pz. Though, it was difficult to classify the potentials into two components (i.e., Bereitschaftspontential: BP and negative slope: NS'). Kitamura studied the relationship between a complexity of movement and the amplitude of both the BP and the NS' (2). The amplitude of NS' was influenced stronger by the complexity of movements than that of BP (i.e., complex movements increased the amplitude of NS'/BP). They suggested that the complex movement involved the neural activation of larger areas in the motor cortex. In this study, the difference of the component of between the BP and the NS' was not clear. The main reason was that the voluntary blinks was simple movement. The neural activation related to the voluntary blinks was small. Thus, in this study, the MRCPs relation to the three kinds of blinks were estimated by the using of the summation of the BP and the NS'. The mean amplitude of MRCPs was 11.6 ± 3.8 μV in Cz and 9.4 ± 3.1 μV in Pz. The value of Cz is significantly larger than that of Pz by the t-test ($t=3.37$, $p<0.01$). There are neural participations of the supplementary and the primary motor area in the cerebrum cortex in generation of the voluntary blinks.

On the other hand, the MRCPs related to the reflex blinks and the spontaneous blinks were not occurred. Libet, Wright, and Gleason (3) examined about the MRCPs related to the spontaneous and the pre-planed voluntary acts that were abrupt flexion of finger. They reported that the MRCPs were evoked by the voluntary flexion of the finger, thought the MRCPs related to the spontaneous flexion of finger was not observed. In this study, we did not find the MRCPs association with the spontaneous blinks obviously. The results of Libet et al. supported our results. The mechanism of the spontaneous blinks should resemble the way of reflex blinks rather than of the voluntary blinks. In the future, we have to inquire to examine the waveform and frequency of the MRCPs associated with the spontaneous blinks more closely.

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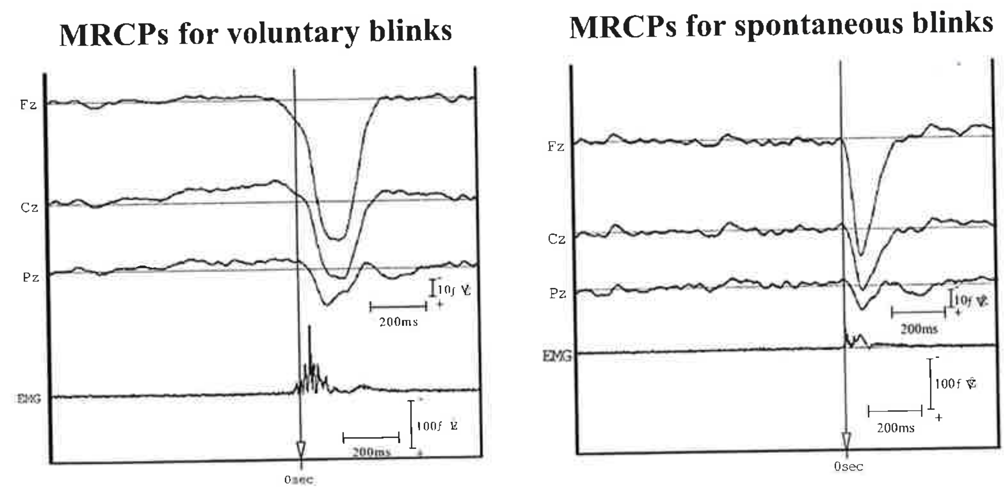

Fig.1 Typical waveforms of MRCPs association with voluntary and spontaneous blinks. In Cz and Pz, the negative lamp-like potentials appeared before onset of contraction of m. orbicularis oculi related to voluntary blinks.

P100 LATENCY OF VEP AND SACCADIC REACTION TIME DURING CONTRACTION OF SHOULDER GIRDLE ELEVATORS

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INTRODUCTION

We previously reported that a shortening of saccadic reaction time was observed while maintaining the neck flexion position, in which the neck extensors contracted (Fujiwara et al. 2000). The saccadic reaction time also shortened in association with isometric contraction of shoulder girdle elevators (Kunita and Fujiwara 1996; Fujiwara et al. 2001).

The neural pathway of saccadic eye movement is composed of a lateral geniculate body, occipital cortex, posterior parietal cortex, parietal eye field, frontal eye field, superior colliculus and reticular formation (Pierrot-Deseilligny 1995). Visual information processing time until the occipital cortex has been mainly examined by P100 latency of visual evoked potential (Skuse et al 1984; Seki et al. 1996). The present study investigates P100 latency of the visual evoked potential and the saccadic reaction time during the contraction of shoulder girdle elevators.

We hypothesized that the visual information processing system until the occipital cortex would be influenced by the activation regulating system in the brain associated with the activity of neck extensors, and subsequently P100 latency would shorten.

METHODS

Ten subjects took part in the experiments. The subject sat on a chair with his trunk stabilized by a restraining device. P100 latency and the saccadic reaction time were separately measured under contraction of the shoulder girdle elevators. Muscle force was set in 10% increments from 0% maximal voluntary contraction (MVC) and 50% MVC. The visual evoked potential was detected from the midline occipital electrode when the right retina was stimulated through the eyelid by LED with both eyes closed. The frequency of the stimulus was 1 Hz. The visual evoked potential was averaged for the 100s-epoch using the stimulus pulse as a trigger. P100 latency was defined as the time from the flash stimulus to the occurrence of the main positive peak. The saccadic reaction time was defined as the latency until beginning of eye movement toward the lateral target, which was moved at random time-intervals in jumps of 20° amplitude.

RESULTS AND DISCUSSION

A significant effect of exerted force on saccadic reaction time was observed (Fig 1A; $F_{5,59}=15.49$, $p<0.01$). The reaction times gradually decreased up to 30% MVC, and then began to increase again at 40% MVC. The time difference between 30% MVC and 0% MVC was 21.8 (SD=10.0) ms. A significant effect of exerted force on P100 latency was observed (Fig 1B; $F_{5,59}=6.12$, $p<0.01$). The changing pattern was essentially the same as that in the saccadic reaction time. The time difference between 30% MVC and 0% MVC was 3.6 (SD=3.1) ms. It was considered that the visual information processing time until the occipital cortex was shortened up to a certain muscle force of the shoulder girdle elevators, and then the processing time became lengthened. The neural pathway of saccadic eye movement and the visual information processing system may be influenced by the common regulating system in the activation of the brain associated with the activity of neck extensors.

The ratio in the shortening in P100 latency relative to that in the saccadic reaction time was obtained varied from 1.2% to 44.7%, and the mean value was 16.5 (SD=14.7) %. There was low degree of positive correlation between the T-scores of the saccadic reaction time and P100 latency (df=60, r=0.354, t=2.883, $p<0.05$). It was suggested that there was a great individual variation in the neural pathway of saccadic eye movement affected by the regulating system and/or in the degree of that affect.

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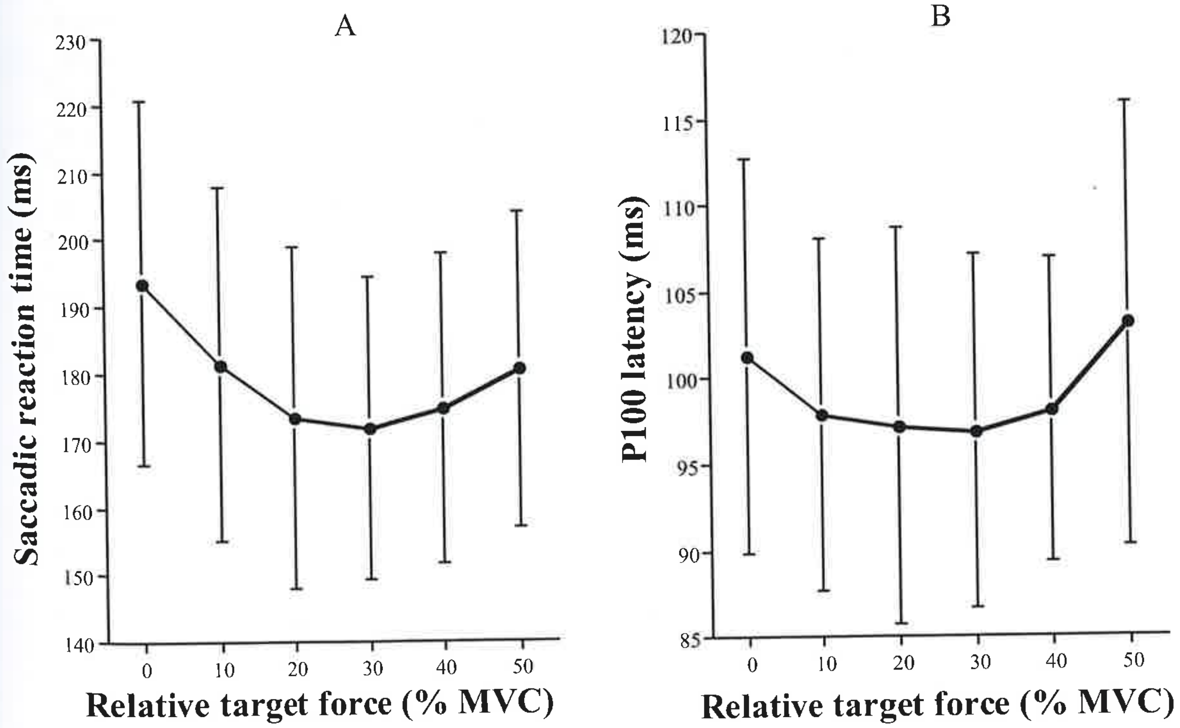

Fig. 1 Saccadic reaction time and P100 latency as a function of relative target force.
A: Saccadic reaction time B: P100 latency

EFFECTS OF WORKING MEMORY LOAD ON EVENT-RELATED EEG

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INTRODUCTION
Working memory is responsible for the short-term storage and online manipulation of information necessary for higher cognitive function, such as language comprehension, planning and solving problem. And, working memory is associated with activation of a prefrontal-posterior association cortex network[1]. Distributed cortical involvement in attention and memory processes are supported by PET[2] and event-related potential (ERP) studies. Electrophysiological evidence of posterior activation during memory processing is provided by P3b ERP components[3]. The purpose of present study is to examine the temporal cortical involvement by the effect of memory load in spatial and verbal working memory. Therefore, we examined ERPs of EEG while the subjects were performing working memory tasks.

METHODS
7 healthy subjects were seated facing a computer monitor at a distance of 57.3cm. They performed 6 versions of a continuous matching task (Fig.1). Matches occurred on 50% of the trials. Each stimulus item was drawn from a set of 12 capital letters. The letter on a given trial appeared at one of 12 positions, each of which was on one of 6 equidistant radii of an imaginary circular array 1 or 4 cm from the screen's center (the inner and outer circles of this array respectively subtended 1°and 4°of visual angle). Letter name and position were counterbalanced across conditions so that each letter and position occurred with equal probability. Each stimulus presented for 250 ms once every 4.5 s. A fixation point(Cue) appeared at the center of the screen 1.5 s before stimulus. In 6 conditions subjects were required to respond by pushing a microswitch with the right-hand index finger for non-matching stimuli (those that did not match the name or spatial position of the comparison stimulus) and a second microswithch with the right-hand middle finger for matching stimuli. Subjects were instructed to respond as quickly and accurately as possible. This task paradigm allowed us to examine the effects of changes in memory load on brain activity as a function of the following factors: stimulus (matching trial, non-matching trial), memory load (1, 2 and 3 back) and task type (verbal or spatial task). A repeated measures ANOVA was conducted for

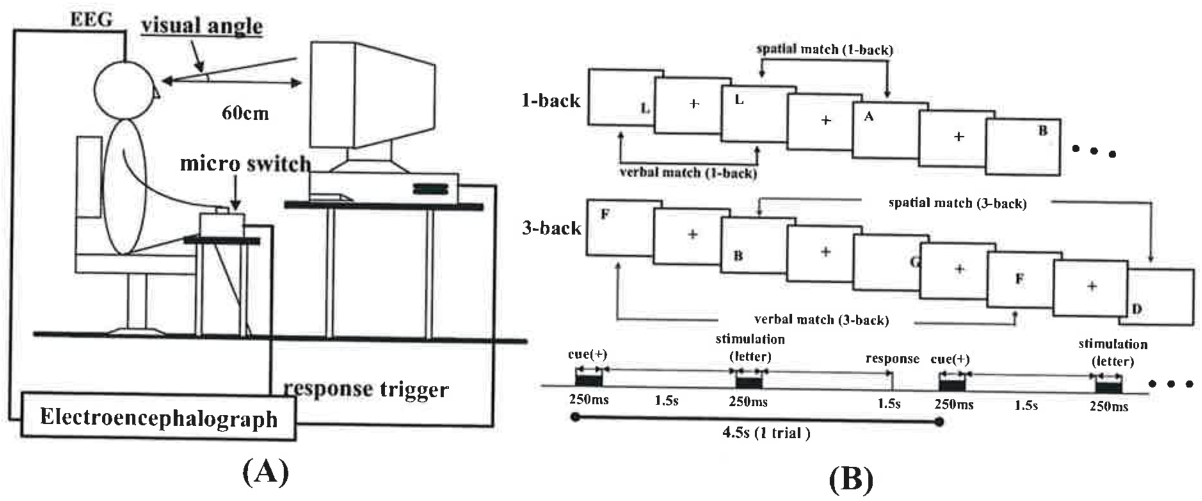

Fig.1. Schematic of the measurement system (A) and diagram of working memory task (B).

ERPs amplitude and latency.
ERPs were recorded from 19 scalp positions referenced to linked ear lobe with impedance below 5kΩ, and with a 0.16 - 30Hz bandpass filter. EEGs were degitized online with a sampling rate of 500 Hz. The analysis window was extended for 2 s following the onset of the stimulus, and a 100 ms pre-stimulus was used as the baseline.

RESULTS
Mean reaction time increased significantly as the task difficulty increased (1-back=0.722ms, 2-back=0.827ms, 3-back=0.839ms). Accuracy decreased significantly as the task difficulty increased (1-back=0.92, 2-back=0.84, 3-back=0.80). But, there was no significant effect to task-type (Accuracy: verbal=0.84, spatial=0.86, RT: verbal=804ms, spatial=788ms).
The following two ERPs were commonly observed, a positive peak component of P3b at 250-350 ms centered at the middle of the posterior scalp area (Pz). First, the P3b amplitude at Pz is smaller with increasing working memory loads. Secondly, in the case of non-matching trials, the peak amplitude of P3b at frontal area (Fz) was significantly lower in verbal tasks than in spatial tasks. In addition, the latency of P3b at the Fz was significantly shorter in spatial tasks than in verbal tasks. On the other hand, no significant difference in peak amplitude and latency for matching trail was observed by between the spatial and verbal tasks.

DISCUSSION
The amplitude of P3 measured at traditional posterior sites normally reduces as the task becomes more complex[4]. In our study, as working memory loads increased, amplitude of P3b reduced. Our results are thus consistent with precedent observations. Secondly, the difference of results in matching and non-matching trials was shown in frontal area. These findings suggest that matching and non-matching trials in working memory task differently modulate the components of the neural circuit in frontal area.

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APPLICATION OF ELECTRICAL STIMULATION PROGRAM IN DIFFERENT FREQUENCIES ON THE ANTERIOR TIBIAL MUSCLE

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INTRODUCTION

Nowadays, there isn't an attention concerning the election of the most adequate parameters in electrical stimulation for each specific type of fiber muscle. In this study, the effects of electrical stimulation protocols with specific features for tonic (20Hz) and phasic (50Hz) muscles were evaluated through strength and resistance analysis of the anterior Tibial muscle [1,5].

The purpose of this study was to evaluate two different frequencies of motor electrical stimulation in tonic and phasic muscles, in a training program for improving the strength and resistance of anterior tibial muscle, through analysis of fatigue test obtained from electromiographic signal and maximal voluntary muscular strength (MVC), providing a better understanding of the influence of these different stimulation parameters and their usage in the clinical practice.

METHODOLOGY

Fifteen healthy young male, without neuromuscular dysfunction in lower limbs, were classified randomly in three groups: Group I: Lower frequency stimulation (LFS) = 20Hz; Group II: High frequency stimulation (HFS) = 50Hz, and control group, who wasn't submitted to electrical stimulation. LFS and the HFS groups were submitted to electrical stimulation on dominant anterior Tibial muscle once a day, three times a week, during 2 months. Muscular conditions were evaluated before, during and after electrical stimulation, by means of muscular fatigue test, obtained by Median Frequency (MF) of myoelectrical signs and also through an isometric muscular strength test (load cell). The EMG signals were acquired with differential surface electrodes, with 10GΩ input impedance, 130 dB CMRR/2 ρfaraday, and 20 times of gain. The EMG signals were digitalized by a 12 bit A/D converter board, Butterworth filter, low-pass of 509 Hz, high-pass of 10,6 Hz, 100 times of gain, and 1000 Hz sample frequency. Electrodes were fixed on muscular belly of anterior Tibial M., just below the motor point, the load cell was fixed on the floor and on the line of the first metatarsus bone on the dominant lower limb. Each volunteer had contracted his anterior tibial muscle of dominant lower limb during 35 seconds at 80% of maximal voluntary muscular strength. Median frequencies were processed by specific routine to Matlab software, which gives the Median Frequency (MF) in eleven windows with three seconds each one. MF values were normalized in relation on first window value, considered as 100% of MF [2]. The MF and strength coefficients of inclination were studied through Non-parametric statistical model, to p=0,05%.

RESULTS

The resulting data showed no difference statistically significant between groups when the fatigue test was analyzed, however, the other test pointed increase of the muscular strength in HSF group ($p<0,05$).

DISCUSSION

The physical therapy applies frequently motor electrical stimulation tools to improve muscular conditions of their patients. However, there isn't a worry with parameters determination to specific types of fibers, tonic or phasic, for each treatment. It can be observed controversial data related to this subject on literature, probably due to the lack of studies about muscular strength and fatigue [3,4,6]. This study didn't find any difference about types of electrical stimulation on tibial muscle, which is considered a phasic muscle.

CONCLUSION

The resulting data showed no difference statistically significant between groups when the fatigue test was analyzed, however, the other test pointed increase of the muscular strength in HSF group. It can be concluded that, under this experimental conditions, the HSF protocol was able to improve strength muscular conditions and unable to do it for muscular resistance. It can be also stated that more studies about this subject are necessary in order to obtain definitive conclusions.

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ROLES OF UPPER EXTREMITIES AND BODY BALANCE WITH COMPLETE PARAPLEGIA IN STANDING FES

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INTRODUCTION

We have developed a functional electrical stimulation (FES) system using percutaneous electrodes , applied to spinal cord injury(SCI) patients, and restored standing function[1)2)3)]. Stability of standing with FES to SCI patients with complete paraplegia was given by efforts of their upper extremities, without feedback control or dynamic feed forward control. By the way, we developed a motion analyzing system for upper extremities in FES standing and gait of paraplegics, using eight force plates and three dimensional motion analyzer.

So, in this study, we measured reaction force of upper extremities from complete paraplegics in quite standing FES and analyzed their roles.

METHODS

a) Subjects: Five complete paraplegic patients (5 men) participated in this study . Their average age was 49 years old. The average time since spinal cord injury was 19 months and the average time of treatment with FES was 57.4 months .

b) Electrical stimulation: The stimulation data for stimulated Gluteus Maximus muscles, Gluteus Medius muscles, Adductor maximus muscles, and Femoral nerves were given by a portable multi-channel FES system (NEC Co. Ltd.) using about thirty percutaneous electrodes. The pulse trains consisted of a pulse width of 0.2 milliseconds, and a pulse interval of 50 msec. Each pulse amplitude was from 0 to -15 volt. These were fixed in standing.

c) Analysis: A motion analyzing system was consisted of eight force plates (Kistler Co. Ltd.), eight infrared-cameras and a computer (VICON). Patients were standing on two force plates and parallel bars were on four one used for measuring of reaction force of patient's upper extremities. Sampling frequency was 60 Hz, and 10Hz-low pass filtered. Reflective markers were placed over the lateral malleoulous, on the lateral side of knee joint line midway between the patella and the popliteal fold, over the greater trochanter of the femur , over the acromion of the shoulder, over the orecranon and over the wrist. Patients were quit standing with FES in parallel bars and measured with a motion analyzer in 2 seconds.

RESULTS

Reaction forces of upper extremities were small at vertical direction, and the average effort ratio of them was 10.9 %. But at anterior direction, reaction forces of bilateral lower extremities were negative and that of left upper extremities was positive.

Table 1 shows joint moments of hip, knee and ankle of five patients with paraplegia in FES standing. Hip joint moments were positive, but knee joint moments were negative.

When a patient elevated his arm to lateral or anterior direction, COP of bilateral lower extremities moved to remaining arm side. Though the change was small, when a patient elevated his arm to his mouse.

DISCUSSION

Main role of upper extremities was thought to be the effort for correction of body balance because hip extension moment evoked by FES were not enough. C posture is stable, and hip and knee joint moments become positive. It was reason why subjects pushed his arm backward by parallel bar that hip extension moment evoked by FES was small and they filled deficit. As that result, knee position was moved forward and knee flexion moment occurred. If knee flexion moment evoked by body gravity was larger than that by FES as muscle fatigue, knee flexion must occur. At that time, they moved his hip and knee joint position backward, but this posture was unstable and effort of upper extremities for body balance was large. Without feedback control in FES system, upper extremities tuned body balance in this way.

When a subject left his one arm from parallel bars, it was thought that many change occurred and remaining arm was the important role because body rotation moment evoked by elevation of arm must be controlled.

CONCLUSION

Without feedback control in FES system, upper extremities tuned body balance in FES standing like complete feedback control filling a deficit of feedforward FES control.

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Case	Side	Hip (Nm)	Knee (Nm)	Ankle (Nm)
1	R	17.3	-1.6	4.0
	L	10.2	-2.2	4.1
2	R	6.8	-9.0	8.9
	L	3.6	3.7	7.0
3	R	20.7	-21.8	21.5
	L	16.7	-24.6	20.2
4	R	2.2	0.6	14.7
	L	-0.7	-0.3	21.1
5	R	13.2	-19.3	12.0
	L	24.6	-14.7	10.2

Table 1: Joint Moments of five Palaplegia in FES standing. Positive values mean dorsal flexion in ankle joint and flexion in knee and hip joints.

ELECTROMYOGRAPHICALLY TRIGGERED ELECTRICAL MUSCLE STIMULATION FOR CHRONIC BRACHIAL PLEXUS PALSY

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INTRODUCTION

Chronic brachial plexus palsy leads to major impairment of everyday life.[1,2] Even if partial reinnervation occurs, muscle strength may be too weak for functional use. Conventional exercise therapy, even if performed regularly for a long period of time, fails to improve muscle strength. A further treatment modality to improve muscle weakness and function is motor re-education by a biofeedback technique. Retraining of central control might be an important factor for clinical improvement in lesions of the second motor neuron. The aim of the study was to assess whether EMG triggered electrical muscle stimulation could increase muscle strength, active range of motion and function, in chronic partially denervated muscles secondary to brachial plexus lesions.

METHODS

Six patients with a brachial plexus lesion caused by mechanical injury, dating back at least 3 years took part in the study. 4 patients were treated by EMG-triggered electrical stimulation, two served as controls. The lesion had to include the upper brachial plexus, either exclusively or as part of a global brachial plexus lesion. The strength of shoulder abduction and elbow flexion had to be between grade 2 and 4 according to the Medical Research Council (MRC) Scale [5], and the level of muscle strength had to have remained constant for at least half a year, which was assessed by history-taking.

Outcome measurements

Clinical muscle strength testing (according to an extended Medical Research Council Scale), measurement of range of motion (universal goniometer) and assessment of function (impairment for ADL and professional activities were rated on a VAS scale) were performed 2 months before, at the beginning and the end of the treatment period, and 2 months later. For the 2 controls, corresponding examination intervals were chosen. A senior physician and specialist in physical medicine and rehabilitation performed the clinical muscle strength testing, measurement of the range of motion and assessment of function at all follow-up examinations.

Intervention

EMG-triggered electrical muscle stimulation was performed with an electrical stimulator, which uses the surface EMG signal of the remaining voluntary muscle activity to trigger electrical muscle stimulation. Contraction time was two seconds. Electrical muscle stimulation was performed for 12 weeks, once a day for 20 minutes. The patients were instructed to carry out 15 to 25 contractions per minute, 6 days a week. The total daily stimulation time and the number of contractions per minute were entered in a logbook by the patient. Rectangular biphasic impulses were used. The pulse width was 0.5 ms for each phase of the biphasic impulse. Stimulation frequency was 50 Hz and stimulation intensity was as high as was tolerated by the patient, on average 65 V; the stimulation elicited a strong muscle contraction.

RESULTS

Muscle strength remained unchanged for control patients, and over the pre-treatment period of 2 months for all treated patients. Muscle strength of shoulder abduction and elbow flexion increased over the treatment period for all stimulated patients by 0.5 or 1 grade on the extended MRC scale, besides for shoulder abduction in patient 2. The increase in strength persisted for a further two months. The active range of motion for shoulder abduction increased by 30 degrees after the treatment period in patient 1; in patient 4 it increased by 20 degrees. In the remaining two patients as well as in controls, it remained unchanged.
Rating for functional impairment of ADL improved by 54 % for patient 1 and by 31% in patient 3. Patient 4 improved by 14 % on the VAS for ADL and by 20 % for his profession. Patient 2 rated no functional impairment at any time of examination, neither for ADL nor for profession. Four patients rated no functional impairment for their profession. The 2 controls rated no change in functional impairment.

DISCUSSION

EMG-triggered electrical muscle stimulation was found to improve muscle strength in all patients while active range of motion and function improved in 2 patients. The treatment included electrical stimulation and voluntary contractions, which were necessary to trigger the electrical stimulation. The effect of treatment is therefore a combination of electrical contractions, volitional contractions and re-education of motor function.

CONCLUSION

EMG-triggered electrical muscle stimulation may be regarded as a treatment option for patients with chronic partial plexus palsy, as it is liable to increases muscle strength, active range of motion and function of the affected arm.

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DETECTION OF MUSCLE ACTIVATION IN MOVEMENT ANALYSIS

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INTRODUCTION

In rehabilitation, there is a need to be able to quantify changes in muscle activation patterns during a long period of time. For example, to study the recovery of muscle activation patterns after stroke and to enable assessment of the impact of therapeutic interventions.
To assess timing aspects, the smooth rectified sEMG profiles are often compared to a threshold. However, this method introduces large systematic errors. Moreover, by averaging timing parameters over a number of steps, information about step to step co-ordination, consistency and variability is lost.

METHODS

The objective of this study was implement the approximated generalised likelihood ratio (AGLR) test (Staude, 1999) to detect the on- and offset times of muscle contraction for each step and compare the performance to the standard threshold method.
Methods: A computer simulation was conducted to compare the performances of a standard threshold criterion and the AGLR test. The simulation implied the detection of a discrete burst of SEMG activity, synthesised by applying a rectangular pulse with amplitude 20 in addition to a noise signal with a variance of 1 at known change times. For the threshold criterion, the SEMG was rectified and filtered recursively with a cut-off frequency of 25 Hz. The threshold was based on the primary noise level of the signal. For 250 realisations of the burst-modulated noise, the onset times were estimated for the two algorithms. The estimation error in onset detection time of the burst $\varepsilon = t_{exact} - t_{estimated}$ is computed for each method and realisation and plotted in a histogram.

RESULTS

The threshold criterion provides highly biased estimates if the threshold is not chosen properly. When the threshold is better tuned for the specific signal to noise ratio the results are more accurate but the estimates are still highly variable (std=4.2). The AGLR algorithm provides accurate estimates with a low estimation error ε (std=1.1).

DISCUSSION

Estimating change times with the AGLR test is much more accurate than with a threshold method. The on- and offset timing distribution gives relevant additional information, which cannot be obtained from the ensemble average profiles. Differences between the consecutive measurements can be observed and analysed statistically. From the estimated on- and offset times also other clinical relevant parameters, like co-contraction and stability of the gait pattern can be derived.

THE MODULATION OF THE SHORT LATENCY REFLEX LINKING THE ANKLE DORSIFLEXORS TO THE KNEE EXTENSOR MUSCLES DURING GAIT IN HUMANS: THE INHIBITION OF THE REFLEX DURING THE TRANSITION PERIOD FROM STANCE TO SWING PHASE

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INTRODUCTION

In man, stimulation of Ib afferents in the common peroneal nerve evokes non-monosynaptic excitation of quadriceps (CPQ)[1]. This reflex pathway could have a role in the functional situations when a co-ordinate contraction of the ankle dorsiflexors and the knee extensor are needed. During gait this co-ordinated contraction can be seen in the early stance phase. In this study the pattern and the nature of the modulation of this reflex during gait was investigated.

MATERIAL AND METHODS

The experiments included 2 studies of 10 volunteer subjects (7 male, mean age 33±8.5 years). Electrical stimuli were applied to the common peroneal nerve at caput fibulae every 2-5 gait cycles. Stimulus of an intensity adequate to evoke M wave of above 40% of M_{max} in anterior tibialis (TA) was used to evoke a maximum response in the quadriceps muscles (Q) [3]. Surface EMGs were sampled from TA, vastus medialis (VM), rectus femoris (RF), semitendinous and medial gastrocnemius of the right leg. Two small pressure switches were used to indicate the start and the end of the stance phase and to trigger the stimulator.
In the first experiment the maximum reflexes were evoked at different instants of the gait cycle. In the second experiment a modified knee orthesis was used to produce higher activity in the RF and VM during the periods when they were almost silent. Three periods were selected for the stimulation, midstance, terminal stance and terminal swing.

RESULTS

The CPQ reflex could be evoked in RF and VM from the terminal swing to the early stance phase with its maximum magnitude shortly after heel strike. During other parts of the gait cycle the Q muscle was remained areflexic. This was very similar to the pattern of the muscular activity in Q during gait.
The increased muscular activity during the silent period of the gait cycle was able to evoke reflexes during the midstance and terminal swing phase. This was increased proportionally with the increased in the muscular activity in RF and VM. Despite the similar increased level of muscular activity, no response was evoked during the terminal stance phase (figure1).

DISCUSSION AND CONCLUSIONS

During gait in humans the peak of activity in pretibial and Q muscles happens shortly after the heel strike which coincided with the appearance of maximum magnitude of the CPQ reflex. This may indicate a positive force feedback from the pretibial muscles to the Q muscles during the early stance phase of gait to help the Q to cope with the transferred body weight. The absence of this reflex in cats [2] and baboons [3] and its strong appearance in humans during the early stance phase are consistent with the unique characteristics of the plantigrade gait in humans. This specific characteristic of human locomotion necessitates the co-activation of ankle dorsiflexors and knee extensors during the early stance phase which can not be seen in the other animals.
The correlation during the early stance phase suggests an excitatory effect from the descending pathways onto the reflex. This correlation can not be extended to the whole gait

cycle since the Q muscle is almost silent during the major period of the gait cycle. One might wonder if the lack of reflex during this quiescent period is a result of the low muscular activity in the Q muscles or an active inhibition of the reflex. The increased muscular activity in the RF and VM provided by the modulated knee orthesis in the second experiment showed different responses during this quiescent period. Increased muscle activity did not reveal a CPQ reflex during the terminal stance phase that suggests an active inhibitory modulation of the reflex (figure1). The pattern of activity in the pretibial muscles during gait could indicate the functional role of this active inhibition. The TA activity happens during the early stance and terminal stance. The latter would help the clearance of the leg from the ground and the initiation of the swing phase. This is associated with the activity of knee flexor muscles, which is also a critical feature of this transition. The activity of this reflex during this period could result in the co-activation of the knee extensor muscles and this could hamper the transition from stance to swing. The active inhibition of this excitatory reflex linking the ankle dorsiflexors to the knee extensors could therefore facilitate the transition from stance to swing in humans.

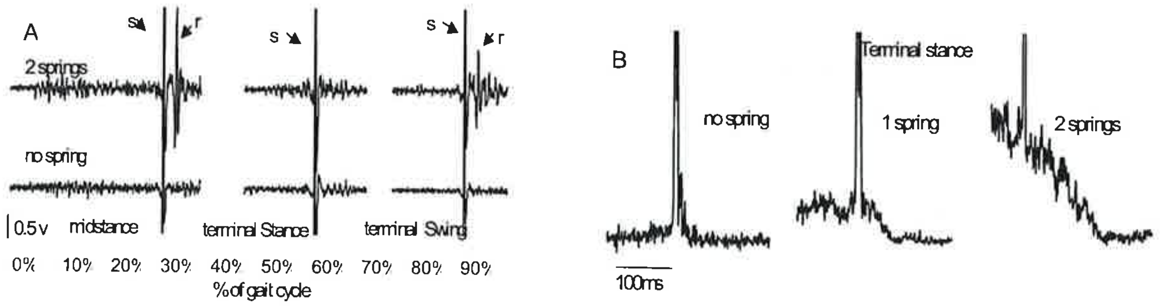

Figure1. The non-rectified averaged EMG of RF during the stimulation at three different periods of the gait cycle (A) and the rectified averaged EMG of RF during the terminal stance phase (B). No responses is evoked during the transition period (middle recording in A and whole recording in B) even with increased background EMG (r = reflex & s = stimulus).

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ELECTROMYOGRAPHICAL STUDY OF THE TERES MAJOR, LATISSIMUS DORSI, AND PECTORALIS MAJOR MUSCLE IN MEDIAL ARM ROTATION

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INTRODUCTION

In kinesiological investigation, electromyography is frequently used in studies of movement to obtain various data on muscle activity under different study conditions, but information from literature is generally controversial in respect to the function of different muscles, mainly with movements of the upper limb at the level of scapula-humerous and shoulder joint.

Although the *latissimus dorsi*, *teres major*, and *pectoralis major* muscles are considered responsible for medial arm rotation, DUCHENNE[3] questioned whether the *teres major* was responsible for this movement. SOUSA et al.[6,7] published two works stating that these muscles were not active in medial arm rotation. Nevertheless treatises on anatomy and kinesiology still has been attributed arm medial rotation to these muscles.

With these controversial kinesiological results and very few works on this aspect, the aim of our study was to evaluate the electromyographical activity of the *teres major*, *latissimus dorsi*, and the *pars sternalis and clavicularis* of the *pectoralis major* muscle in normal volunteers performing medial arm rotation in different positions with and without resistance.

MATERIAL AND METHODS

Twenty sedentary male volunteers were used of with 19 to 25 years-old. Electromyographical analysis were made of the *teres major*, *latissimus dorsi*, and the *pars clavicularis and sternocostalis* of the *pectoralis major* muscle with and without resistance medial rotation, with the arm hanging, 90° anterior flexion, and shoulder abduction at 90° and 180°. All readings were carried out using a 2 channel TECA TE-4 electromyograph with an 80dB common mode rejection differential amplifier, with band pass filter set at 10-10000 Hz, sensitivity 200 gV/cm, and analysis time of 500 ms/cm. Coaxial needle electrodes were directly inserted to a depth of 2 cm into the muscle belly of each muscle to measure the electrical potentials.

RESULTS

The *teres major* and *latissimus dorsi* was inactive in free medial arm rotation and was active in the medial rotation movements counter resistance. It was observed mainly in counter resistance medial rotation, that most volunteers showed a tendency to arm adduction causing the appearance of crosstalk. Alerted to this fact, the volunteers were encouraged to perform the action without adduction; this caused electromyographical silence.

Both the *pars clavicularis* and *sternocostalis* of the *pectoralis major* muscle showed similar behaviour. In free medial arm rotation both parts of the *pectoralis major* were inactive in most cases. When resistance to medial rotation was introduced, the number of active cases in both parts of the muscle increased.

DISCUSSION

The results is not in acordance with KENDALL[5] and BROOME & BASMAJIAN[2] and BASMAJIAN[1], who include the *teres major* and *latissimus dorsi* muscles in the arm rotation group, but agree with DUCHENNE[3] and SOUSA et al.[6,7], who do not attribute participation of this muscles in the medial rotation of the arm.

In free medial rotation, when resistance is introduced in the four tested positions, there is an increase in the number of active cases in both parts of the *pectoralis major* muscle. This suggests that both parts of the muscle systematically contribute to medial arm rotation when performed against resistance. These results are in total acordance with the electromyographical studies of KAMON[5], and FLINT et al[4].

CONCLUSION

The *teres major* and *latissimus dorsi* muscles do not participate in arm medial rotation, and the appearance of electrical potentials during movement was related to a concomitant adduction, showing that these muscles do not cause this rotation, in discordance with various authors from literature. The *pars clavicularis* and *sternocostalis* of the *pectoralis major* muscle participated in medial arm rotation more effectively against force than in free movement.

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ELECTROMYOGRAPHY STUDY IN UPPER AND LOWER PORTION OF THE M. RECTUS ABDOMINIS

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INTRODUCTION

There are controversies observed in literacy revision about the activation in superior and inferior portion of the M. rectus abdominis on variety types of exercises and none of those research have a relation between the postural changes and the activity of electromyography (EMG). Negrão Filho (1999) found mayor activity in upper portion of the M. rectus abdominis in raising trunk exercises and mayor activity in lower portion in leg lowering exercises in Gilleard & Brown (1994) study. LEHMAN & Mcgill (2001) didn't find any difference between the portions in isometric exercises of raising trunk and leg lowering. Based in this concepts the aim of this research is to analyze the activity of the EMG signs in superior and inferior portion, left and right sides of the M. rectus abdominis during the supine position in rest and orthostatic position, static and dynamic exercises purposed, farther on the EMG activity according to a incidence of altered postures observed in the volunteers.

METHOD

Participated in this study, 12 students of Physiotherapy and Physical Education courses randomized between female and male sex, therefore six athletes with age between 17 to 27 years old, 2 female and 4 male. Six sedentaries, between 21 and 23 years, 5 female and 1 male attended to the criteria of inclusion related below: no muskeletal disease and age between 17 to 27 years old. The exclusion criteria were: lumbago, brief history of spine, abdominal, hip or lower limbs surgeries, hernia, bone tumors and dermatology allergy in gel or adhesive tape. During the postural evaluation and digital camera Mavica FD85 (Sony) were used and to collect the EMG signals were used the portable electromyography with 16 channels and surface active electrodes and ground electrode (Lynx). It was proceed through the postural evaluation, photography registered and a collection of the EMG signals on the portions of the M. rectus abdominis as Negrão Filho (1999) and First, in rest with supine position and orthostatic position, and then during isometric and isotonic exercises of raising trunk and finally with isometric exercises in the leg lowering. The time to do the exercises were repeated 3 times of 15 seconds with interval of one minute. And analyzing the data comparing the values of the RMS (Root mean square) in each volunteer.

RESULTS

The maijority of volunteer even athletes and sedentary showed postural changes on left side. The analysis of electromyography resultaded that the superior portion of the bilateral muscles presented a higher potential than inferior portion. And the potential of superior left portion of the muscle were higher than the right proving the postural evaluation findings.

DISCUSSION

The exercises were realized with hip and knee in flexion to reduced the lombar lordosis related by Axler & Mcgill (2001) and Negrão Filho (1999). Isometric and isotonic exercises of raising trunk in both athletes and sedentaries showed major activity in superior portion than inferior, similar results from the Sarti et al (1996) study. The main activity of EMG in left superior portion even in athletes and in sedentaries could be correlated with altered postures observed in the postural evaluation, therefore Negrão Filho (1999) related the possibility that the anatomic asymmetric has relation in the portion of M. rectus abdominis. Norris (1993) found the major activity in inferior portion of the M. rectus abdominis in leg lowering exercises, different of this study that has occurred major activity in superior portion when in isometric exercises. Lehman & McGill (2001) verified no difference between raising trunk and the leg lowering using normalized EMG signals. The same study of Sarti et al (1996) verified that the isometric and isotonic exercises in EMG activity showed no significant difference in sedentaries but in athletes the major activity were observed in isotonic exercises.

CONCLUSION

No difference activity change in orthostatic position and supine position before and after the data collection, even in athletes and sedentaries. In isotonic and isometric trunk exercises, the EMG signals were similar in athletes and in sedentaries which superior portion were predominant over the inferior and the left superior portion were major than other portions. Due the postural changes in predominance on left side of the both groups, there was homogeneity in EMG activity.

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AN ELECTROMYOGRAPHIC BEHAVIOR ANALYSIS OF ANTERIOR AND POSTERIOR THIGH MUSCLE OF COMMON INDIVIDUALS IN THE JUMPING ACTIVITIES

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INTRODUCTION

One of the main responsible mechanisms for the articular stability is decurrent of the neuromuscular control system action by feed-forward and feedback. The control by feed-forward is responsible for the muscle preparatory activity, while the motor control by feedback plays the regulating role of the reactive muscular activity. The muscular activation level, either preparatory or reactive, influences in the muscular tension, thus promoting, a dynamic control on the joint that is being used (Swanik, et al., 1997).

In the healthy knee the static support is promoted by structures as the articular capsule, ligaments, meniscus and articular morphology, the capsulo-ligament tissues also possess a sensorial role: to detect the position and articular movement, thus mediating the motor control for the dynamic stability. On the other hand, the dynamic support is completed through the preparatory and reactive neuromuscular control where primary dynamic stabilizers are the muscles. Swanik (1997), concluded that the contraction of these muscular groups can increase the articular stability in up to 10 times, thus, the muscular tension level greatly increase the articular dynamic stability.

Therefore, the objective of this study was to analyze the electromyographic behavior of thigh anterior and posterior muscles in jumping activities in individuals without any knee articular muscle disturbance, with the purpose to establish a normality standard of knee articular protection dynamic system.

METHODS

Sixteen young and healthful individuals were analyzed, of both sexes, 7 men and 9 women, with age varying from 18 to 33 years old. For data collection, surface active electrodes were placed in the muscles vastus medialis oblique (VMO), vastus lateralis (VL), biceps femoris (BF) and semitendinous (ST) motor points for the pick up of muscles electric activity, one eletrogoniometer was coupled in the knee lateral face, to measure the angular moment of this joint during the jumping activity and a pressure sensor was placed in the ground to register the the beginning and end of the jumps. Each individual performed 10 vertical jumps and 10 horizontal jumps. For pick up of the electromyographics signs a module conditioner of signs, model MCS 1000 - V2 of the Lynx was used, with 16 input channels; a analog sign-to-digital conversion card (A/D) model CAD 12/36 of the Lynx, 12 resolution bytes, with DMA support (Direct Memory Access), a Aqdados software (in order to capture, treatment and storage) and Microcal Origin (in order to process the signals through elaborated routine with the purpose of full-wave rectification of the sign, linear envelope, amplitude normalization and time base). For the accomplishment of the statistical analyses were selected within jumping cycle, a representative interval jumping preparation and another representative interval one of jumping at landing moment .

RESULTS

For the descriptive analysis of the eletromyographic tracing of the jumping activity like vertical jump and horizontal jump, the cycle of these activities were divided in 3 periods.

In the vertical jump, the first period from 0% to 15% represents the ground contact loss; the second period from 15% to 90% represents the suspension phase and the third period from 90% to 100% the landing phase. We observed that in this activity the activation of the thigh's anterior muscles (VMO and VL) had been greater than the posterior muscles (BF and ST), as much in the beginning of the jump as in the landing. We also verify that moment before occurring the landing, thigh posterior muscles were active.

In the horizontal jump , the first period from 0% to 50% represents the beginning of knee flexion (±50°), preparing for the impulse; the second period from 50% to 70% represents the beginning of the knee extension until the ground contact loss and the third period from 70% to 100%, represents the suspension phase in air until the landing moment. In this activity, we observe great activity of VMO and VL muscles in the first period and after that, an activation of BF and ST is perceived in bigger amplitude approximately in 60%. In the third period, we observe a bigger and increasing activation of BF and ST until the landing moment, followed of a VMO and VL intense activation at the impact moment and deceleration.

In the statistical studies, the F values gotten by the variance analysis show us a significance level of 5%, therefore, the four studied muscles are not equally effective in the verification of the sign and exist significant difference between them.

DISCUSSION

We verify that the collected electromyographics signs had presented activation amplitude values upper of the ranging noise present in the environment, and can also verify the existence of a standard electromyographic activity of the muscles when compared between the individuals. Comparing the electromyographics tracings of the thigh anterior and posterior muscles during the jumps, we can observe that the posterior muscles activate themselves moment before the fall and decrease progressively, at the same time in which the anterior muscles increase of activity until the landing moment. This behavior is in accordance with the feed-forward mechanism (pre-activation) and feedback (reactive muscular activity), described by SWANIK (1997).

CONCLUSION

We can conclude after the accomplishment of the eletromyographic behavior analysis that the used acquisition system was capable to pick-up electromyographics signs of reliable way; that there is a standard electric behavior between VMO, VL, BF and ST muscles in the vertical and horizontal jumping activities.

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MOTION ANALYSIS DURING ONE-LEG STANDING; COMPARISON BETWEEN HEALTHY YOUNG AND ELDERLY SUBJECTS

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INTRODUCTION

Falls are a significant cause of morbidity and mortality in the elderly. Balance disturbance is one of the factors for falling. Postural control is divided into the quiet stance, perturbed stance and spontaneous perturbed stance (related to anticipatory postural control). Falls in the elderly often occurs in tripping over an object and stepping on a stair. We thought that there was a problem in the postural control of spontaneous perturbed stance in elderly, or a difference in that between young and elderly. So that, we focus to spontaneous perturbed stance and did this study that analyzed the motion for transitioning from bipedal to unipedal stance in healthy young and elderly. The purpose is to examine the organization of activation pattern of EMG, ground reaction force and postural changes for transitioning from bipedal to unipedal stance in healthy young and elderly subjects.

METHODS

The study included eight healthy young male volunteers (mean age 19.5, ±4.5 yr) and 11 healthy elderly male and female volunteers (mean age 80 ±4.8 yr). Elderly subjects have no problems that interfered with their abilities to maintain ADL independently, and they do not have the experience which fell within the past one year. The experimental procedure and its purpose were explained thoroughly to all subjects, and their consents ware obtained.

The experimental task was to raise one limb following the light (LED) and sound signal as fast as possible, and maintain its posture for 3-4 seconds. The raised limb was selected according to the instruction from experimenter.

EMG, kinetic and Kinematic recordings were collected synchronously using a motion analyzing system (TELEMG and ELITE PULASE, BTS, force plate, Kistler). EMG data was recorded bilaterally from four muscles; elector spinae lower part (ES), gluteus maximus (Gmax), gluteus medius (Gmed) and soleus (So). The COG (center of gravity) and COP (center of pressure) trajectory were computed based on the kinematic and kinetic parameters. Reflective markers were placed bilaterally on the following landmarks; (1) the acromion process, (2) the greater trochanter, (3) the lateral femoral condyle, (4)the lateral malleolus, (5)the head of the fifth metatarsal. The postural changes were measured by the trunk tilting angle calculated from the difference between the angles of the shoulder and pelvis. The shoulder angle was calculated by the angle of the line where both acromion processes were connected and horizontal line. The pelvic angle was calculated by the angle of the line where both greater trochanters were connected and horizontal line.

The significance of differences in group mean outcome measures was determined using an ANOVA and t-test.

RESULTS

The COP trajectory moved first toward raising limb side and then transferred to supporting limb side before the foot goes up from the floor in all subjects. On the other hand, the COG trajectory moved toward supporting limb side from start during transitioning from bipedal to unipedal stance in all subjects. On the EMG onsets, the raising limb Gmed occurred firstly, and the raising limb So and supporting limb SE were activated one by one next during the COP trajectory moved toward rising limb side in young subjects ($p<0.05$). On the other hand, the raising limb Gmed was not activated at that phase in elderly subjects. On kinetic data, the lateral component of force plate beneath the raising limb was increased significantly by young compared with elderly when the COP trajectory moved first toward raising limb side ($p<0.05$). The trunk tilting angle toward supporting limb in elderly was greater than in young at foot-off (fig.; $p<0.01$).

DISCUSSION

On one-leg standing task, the hip abductor muscles of raising limb side is the primary task muscles which activated preceding foot-off (1). That function is to move the pelvis toward supporting limb side for COG transition. On the results of this study, the hip abductor muscles of raising limb side were not activated in elderly. And the trunk tilting angle toward supporting limb in elderly was greater than in young at foot-off. These results suggest that the strategy concerning one-leg standing task is different in young and elderly.

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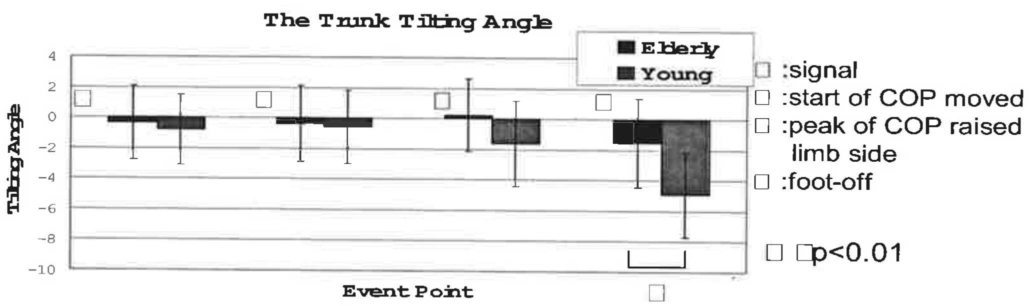
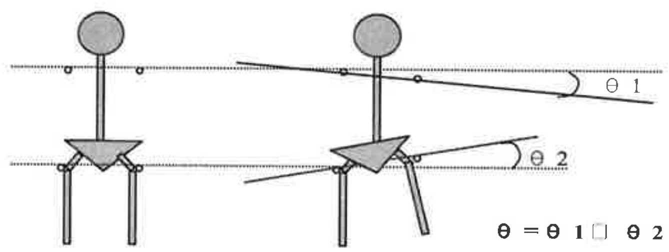


Definition of Trunk Tilting Angle: θ



SINGLE MOTOR UNIT MECHANOMYOGRAM: A SPIKE TRIGGERED AVERAGING APPROACH

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INTRODUCTION

From surface EMG signals it is possible to extract and classify single motor unit action potentials by means of automatic decomposition methods [2]. The present study aims to develop and test experimentally a non invasive method (based on surface EMG decomposition) to extract the mechanomyographic (MMG) response and the force twitch of single motor units during voluntary muscle contractions. The objective is to achieve a better understanding of motor unit mechanical and electrical properties, and of the possible correlation between them.

METHODS

Detection Modalities: EMG and MMG signals were recorded with an hybrid probe described in [1]. The probe consists of 4 silver bar electrodes for EMG detection and an integrated on chip accelerometer (Analog Devices, model ADXL202JE) for MMG detection. The probe is fixed on the skin with double adhesive tape. The EMG and MMG signals were amplified and filtered with dedicated amplifiers. The force signal was detected with 5 kg or 50 kg load cell, fixed to an isometric brace applied to the ankle or wrist of the subject; the DC offset of the signal was removed and its oscillations amplified. *Decomposition of the surface EMG signal:* The surface EMG signal was decomposed using a wavelet based segmentation and a neural-network based classification. Motor unit action potentials (MUAP) identification was based on a bank of filters matched to the MUAP shape at different scales. The extraction was performed by identifying the generation time for each MUAP and tracking its propagation using the two double differential signals acquired with the 4 electrode probe. Each identified MUAP was extracted in the spatio-temporal domain to preserve information useful for classification, such as the conduction velocity [2]. *Spike triggered averaging:* A sliding window averaging technique using the detected firings from surface EMG decomposition as triggers was used to extract single motor unit (MU) MMG response (from the MMG signal average) and force twitch (from the averaged joint torque signal). *Experimental modalities:* The biceps brachii and tibialis anterior muscles were investigated in 5 healthy male young subjects (age 26.0 ± 3.0 years). The hybrid probe was fixed on the skin between the most distal innervation zone and the tendon terminations. One minute voluntary contractions of the two muscles were performed at 2%, 5%, 8% ,10%, 20%, 40%, and 60% of the maximum voluntary contraction (MVC), with 5 minute rest between them.

RESULTS

In most cases it was possible to reliably detect single MUAPs and classify them as belonging to different MUs. The shimmer plots of the classification results revealed small shape differences between MUAPs classified as generated by the same MU. It was not possible to extract the complete firing pattern for the detected MUs but this was not a requirement for the averaging process. The presence of a deterministic single MU MMG and force twitch (see the Figure) has been revealed by the asymptotic analysis of the power of the averaged signal increasing the number of firings used for the averaging, for both muscles at all levels of voluntary contraction. The minimum number of firings required to obtain a reliable MMG response was around 30, while it was higher for the force twitch estimation. The duration of single MU MMG was between 40 and 80 ms.

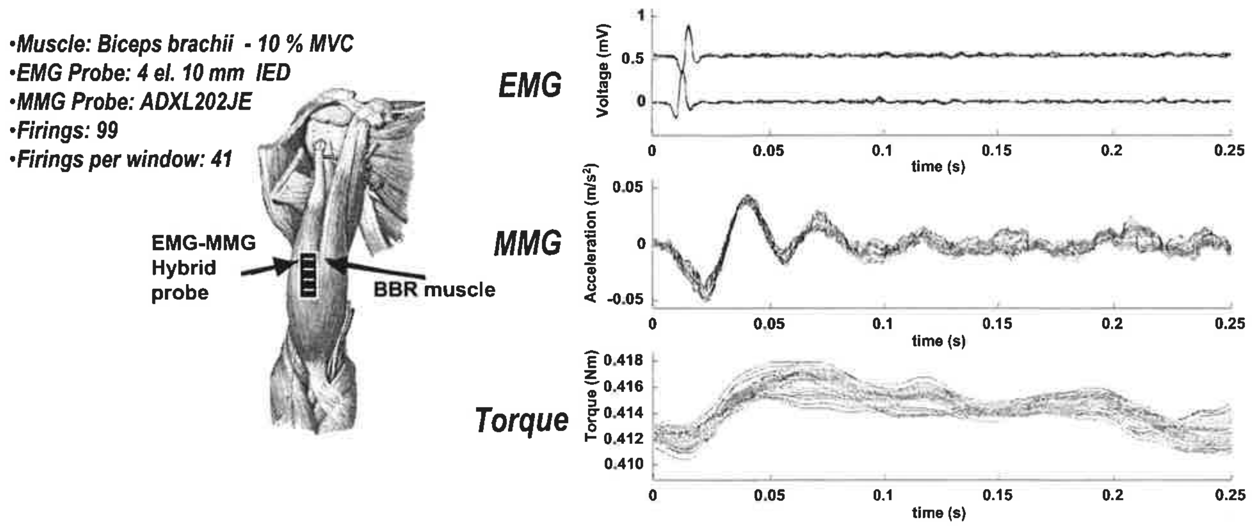

Figure. EMG (2 double differentials signals), MMG and force twitch contribution of a single motor unit, detected, as described in the text, from the biceps brachii muscle, during a one minute long contraction at 10%MVC. The responses are detected by a sliding window averaging technique, with a window shift of one detected MUAP. Each curve is the average of 41detected firings.

DISCUSSION AND CONCLUSION

A non invasive technique for the extraction of contractile and conduction properties of single MUs has been developed and tested experimentally. The knowledge of both electrical and mechanical properties of single MUs has potential usefulness in basic and applied physiological investigations.

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A PROTOTYPE OF HYBRID PROBE FOR SURFACE EMG AND MMG JOINT RECORDINGS

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INTRODUCTION

Many past studies report the joint recording of EMG and MMG signals. However, it is not yet clear which is the effect that a composite probe have on the recorded MMG signal. Since the MMG signal is generated by the vibration of the muscle during the contraction, a mass applied on the surface of the muscle may indeed affect the signal features. The aims of this study were:

1) to investigate the effects induced on the MMG signal features by the presence of a weight on the muscle, and 2) to design a light, compact, easy to place composite probe not affecting the muscle vibration, for the detection of surface EMG and MMG signals.

METHODS

Evaluation of the effect of a mass applied over the skin on the MMG signal features. The MMG signal was recorded with an on chip accelerometer sensor (ANALOG DEVICES model ADXL202JE), that was chosen because its light weight (1g), its small size (5 x 5 x 1.5 mm), its operating range (+/-2 gr) and bandwidth (0-100 Hz) fit the MMG signal features. An experimental protocol was designed to evaluate the maximum mass that can be applied on the muscle surface without significantly altering the recorded MMG signal. Different masses (from 0 gr to 62 gr) were placed under the accelerometer, over the belly of the biceps brachii muscle. The mass was increased from 0 gr to 16 gr in steps of 2 gr, then 32 gr and 62 gr weights were applied. The MMG signal was acquired from 7 healthy male volunteers (age, 26±3 SD). For each weight, a 10 s long contraction at 30% of the maximum voluntary contraction (MVC) was performed in two separate trials with 4 minute rest between them. The MMG signal features investigated were the mean power spectral frequency (MNF), the asymmetry of the spectrum (ASI) and the root mean square value (RMS).

Development of a hybrid probe for EMG and MMG detection. The designed hybrid probe includes, on the same printed circuit board, 4 silver electrodes with interelectrode distance 10 mm for EMG detection, the voltage followers for EMG buffering and the accelerometer chip for MMG detection. The global weight of the probe is 5 g and its dimensions are the same of the weights used in the described experiment.

RESULTS. No statistical difference was found between the first and the second repetition of the experimental session. There was a significant decrease of MNF of the MMG signal (see Figure) and a significant increase of ASI and of RMS with the increase of the mass on the surface of the muscle ($p<<0.05$, two-way ANOVA within subjects, independent factor MASS). The post-hoc Student-Newman-Keuls (SNK) test disclosed pair-wise differences of MNF for the couples 0-32g (with a 10% decrease) and 0-62g (with a 20% decrease), of ASI and RMS only for the couple 0-32g (with an increase of 40% and 25%, respectively). 16 g was considered the maximum mass applicable on the muscle, since no differences on the spectral and amplitude parameters are introduced on MMG signals with masses lighter than this value. The designed EMG-MMG probe weights 5 g and so it has no significant effect on the spectral and amplitude features of the recorded MMG signal.

DISCUSSION AND CONCLUSIONS

A study has been performed to evaluate the effect of a mass applied on the muscle on the MMG signal and an indication on the maximum mass which can be applied over the muscle without significant change of MMG features has been provided. A composite probe for the simultaneous detection of EMG and MMG signal that doesn't affect the MMG signal features was build and tested. The probe was used to study single motor unit MMG response with spike triggered averaging technique [1]. The results obtained in this study are limited to a large muscle, the biceps brachii. For smaller muscles, such as the hand muscles, these results may change.

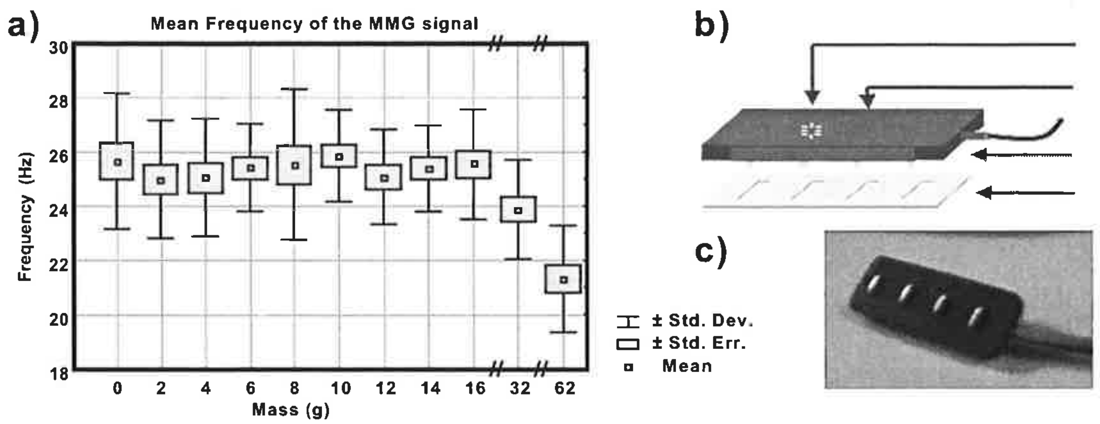

Figure a) Mean value, standard deviation and standard error of the mean of MMG MNF for the 7 investigated subjects; 30% MVC isometric contractions of the biceps brachii (10 s duration), with a mass on the surface of the muscle increasing from 0 to 62 gr. A clear decrease of MNF when a mass greater than 16 gr is applied can be observed. b) Schematic drawing of the hybrid EMG-MMG probe: 1- Analog Devices, model ADXL202JE on chip accelerometer, 2- Motorola, TL074 voltage followers for EMG buffering, 3- 4 silver bar electrodes with 10mm interelectrode distance, 4- double adhesive tape for fixation on the skin. c) Picture of the prototype EMG-MMG hybrid probe.

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RELATIONSHIP BETWEEN EMG AND MMG DURING ISOMETRIC MUSCLE CONTRACTION

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INTRODUCTION

EMG produces important information about muscle activity. We have used it to make the controlling data for functional electrical stimulation (FES). Muscles are activated in two manners such as recruitment and rate coding. It is difficult to know details of these manners by means of analysis of EMG data alone. Mechanomyogram (MMG) reflects vibration which is occur at contraction of the muscle. It showed different aspects of muscle contraction. MMG may produce the information of recruitment and rate coding (1). The purpose of this study was to know the difference between EMG and MMG. Our final goal is to utilize MMG data for FES control.

METHODS

Experiment 1. EMG and MMG were detected from the rectus femoris (RF), vastus lateralis (VL) and vastus medialis (VM) of nine normal subjects. EMG data were detected through bipolar surface electrodes. MMG data were detected through piezo-sensor (Hewlett Packard). All of these data were amplified, integrated and normalized. The condition of muscle contraction was isometric. The hip joint was fixed at 90 degrees and ankle joint at 0 degree. The knee joint was set to 60 and 90 degrees. Muscle contractile force was measured by KIN-COM (Chattanooga, TN). Based on the maximal voluntary contraction (MVC) of knee extensor, the session was arranged as follows; 20, 40, 50, 60, 70 and 80%MVC. External rotated position, mid position and internal rotated position of the knee were compared.
Experiment 2. EMG and MMG were detected from the rectus femoris of seven normal subjects. Data processing and experimental condition were same as the experiment 1. MMG data were processed by Fourier transformation

RESULTS

Normalized integrated EMG (NIEMG) and normalized integrated MMG (NIMMG) increased according to increase of %MVC at 60 degrees of knee joint. While NIEMG showed similar result at 90 degrees of knee joint as at 60 degrees, NIMMG showed peak at 70%MVC at 90 degrees. Both of NIEMG and NIMMG showed higher activity at 90 degrees than at 60 degrees of the knee joint. Only NIMMG of 80%MVC at 90 degrees showed decrease. VL showed more activity at external rotated position than at internal rotated position of the knee. VM showed more activity at internal rotated position than at internal rotated position. The differences among rotated positions of the knee were shown more critically by means of NIMMG than NIEMG.

Band of frequencies of EMG showed less than 200 Hz by Fourier transformation and that of MMG showed less than 100 Hz. According to increase of %MVC, amplitude of power spectrum of EMG was increased through whole of frequencies at both knee positions. On the other hand, MMG at 60 degrees of the knee showed shifts to higher frequencies in addition to increase of amplitude. Amplitude of power spectrum of MMG through 20 to 70%MVC at 90 degrees of the knee was increased, while that of 80%MVC was decreased.

DISCUSSION & CONCLUSION

These results indicate characteristic of EMG and MMG. MMG of 80%MVC MMG at 90 degrees of the knee decreased, Similar results were repoted by Orizio (1). Decrease of amplitude at high level of muscle contraction showed that the muscle did not vibrate enough in such a situation. It might be because inner pressure of muscle compartment was too high to vibrate. MMG showed different aspect of muscle activity. Fourier transformation produces valuable information about manner of muscle contraction. The characteristic of recruitment and rate coding may be clear by means of the coupling study of EMG and MMG. The information detected from MMG in additon to EMG can be useful for making strategy of FES control.

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ELECTRICAL AND MECHANICAL ACTIVITY OF AGONISTS AND ANTAGONIST DURING RELAXATION FROM MVC AT A DIFFERENT MUSCLE JOINT ANGLE.

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INTRODUCTION

Since Gurfinkel et al. [1992] showed that relaxation from unfused tetanus of human muscle stimulated electrically has a slow and fast phases, similar to those recorded on an isolated muscle, a different indices of slow (early) and fast (late) relaxation were separated in voluntary contraction [Jaskólska and Jaskólski, 1997]. It was found that these phases responded differently to muscle fatigue [Jaskólska and Jaskólski, 1997] and are differently affected by human age [Adach at al., 2001]. Relaxation from MVC can be affected by antagonists activity because a different changes in agonist and antagonist muscles activity in relation to joint angle (muscle length) occur. As a result of decreasing muscle electrical activity during relaxation, there are dimensional changes of muscle structures recorded as a mechanomyogram (MMG) [Orizio, 1993]. A big MMG signals were observed during relaxation from unfused tetanus of isolated motor units in rats and it was noted that the MMG amplitude was dependent upon the amplitude and velocity of tension changes [Bichler and Celichowski, 2001]. As during voluntary movements there is unfused muscle contraction, the MMG amplitude (RMS MMG) during relaxation from MVC should also depend on relaxation speed (rate, time).

When the influence of joint angle on MMG amplitude at MVC level was tested in humans, it was found that at an angle smaller (Ls) than the optimal (Lo), RMS MMG decreased (Ebersole et al., 1999). In animal's experiments, Fragioni et al. (1987) found that shortening and lengthening caused a decrease of the amplitude.

The questions were 1/ if the BB muscle has a bigger effect on relaxation rate at an elbow joint angle different than that for the BR muscle, 2/ which relaxation phase is affected by antagonist activity depending on a elbow joint angle, and 3/ if increased amplitude of the muscle oscillation (RMS MMG) during relaxation was smaller at a bigger and smaller muscle length, compared to the optimal.

The aim of the study was to estimate the electrical (EMG) and mechanical (MMG) activity of agonists and anatgonists during relaxation from MVC at an optimal elbow join angle (Lo) as well at the angles that were smaller (Ls) and bigger (Ll).

METHODS

Twenty-two young, physical education female students were tested four times. The first and second sessions were done to establish optimal angle (Lo). The third and fourth sessions were done to measure the relaxation indices at Lo, as well as at the angles that were smaller (Ls = Lo - 30°) and bigger (Ll = Lo + 30°). All testing sessions consisted of five trials of 2 or 3-sec MVC at each angle with simultaneous recording of EMG and MMG signals from biceps brachii (BB), brachioradialis (BR), and triceps brachii (TB) muscles. To assess the speed of relaxation, the following relaxation indices were measured: early, late and most late relaxation rate (ERR, LRR, and MLRR, respectively; %F/5ms). The BIODYNA dynamometer was used to measure torque versus time for right elbow flexor muscles. To assess the possible differences across BB, BR and TB muscles, joint angles, and across relaxation phases, root mean square (RMS) of EMG and MMG signals were calculated for the early, late, and most late relaxation, as well as for the 50-ms time period of MVC, measured just before the emission of the signal ending the contraction. The custom-made EMG/MMG recording device was used to record EMG and MMG signals.

RESULTS

The changes of the early relaxation rate with elbow joint angle were different than the changes in late and most late relaxation rates, and were accompanied by changes in EMG and MMG amplitudes, which usually did not follow the same pattern as the relaxation indices changes.

The MMG amplitude increased during relaxation compared to MVC, and the biggest increased was during the late and most late relaxation, and at the short muscle length independent on the muscle function (agonist, antagonist).

The correlation coefficients between a percent decrease of EMG-RMS and the early, late and most late relaxation rates were similar in BB and BR at Ls, Lo and Ll. However, when correlation coefficient between an amplitude of EMG and relaxation rates were analyzed, the early relaxation rate at Ls showed a bigger correlation with EMG-RMS of the BR muscle than for BB muscle, but the late relaxation showed a bigger correlation with EMG-RMS of the BB muscle than for BR muscle.

DISCUSSION

1/ When percentage decrease of EMG-RMS was considered, the results did not show an angle dependent difference in an effect of the BB and BR muscle on relaxation rate, independent on relaxation phase. However, when interrelationship between an EMG-RMS and relaxation rates were analyzed, there was an angle difference in an effect of the BB and BR for early relaxation rate.

2/ The present results showed that the bigger electrical activity (and a smaller decrease of the EMG-RMS) of the TB muscle (antagonist) during late relaxation the faster late relaxation rate (for most late relaxation rate the respective correlation coefficients were smaller). The early relaxation rate was not affected by electrical activity of the antagonists muscle (TB). However, it showed a significant dependence on a decrease of EMG-RMS of the BB and BR muscles (agonists), independent of the joint angle.

3/ As was expected, the increased amplitude of the muscle oscillation (MMG-RMS) during relaxation was smaller at a bigger muscle length independent on the muscle function: for TB it was at Ll while for BB and BR muscles at Ls.

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SURFACE MECHANOMYOGRAM AND FORCE RELATIONSHIP DURING SKELETAL MUSCLE TETANIC CONTRACTION

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INTRODUCTION

During muscle activity its surface movement, taking place simultaneously to the tension development at the tendon level, can be tracked by position transducers. The resulting electrical signal is named surface mechanomyogram (MMG).The aim of the study was to investigate the relationship between the force generation process and the surface mechanomyogram during the on (from rest to the contraction state) and off (from the contraction to the relaxation state) transient phases of the tetanic response of the skeletal muscle.

METHODS

The medial gastrocnemius of four cats was exposed, the distal tendon was secured to a force transducer linear from 0 to 100 N. The MMG was detected by means of a laser beam optical distance sensor with a ± 10 mm range of measurement and a resolution of about 6 μm. After instrumentation a supramaximal stimulation was administered at 30,40 and 50 Hz for 9 s for each frequency. The force and the MMG were normalised to their maximum attained during each stimulation rate. For each stimulation rate the normalised signals were averaged. The time spent to reach the 50% of the delta (0-100% and 100-0% for the on and the off phase, respectively) was calculated.

RESULTS

In the figure the average MMG and force signals during tetanic stimulation at 50 Hz are reported. The results at 30 and 40 Hz rates are overlapping the ones here reported. MMG always anticipated and trailed the force signal during the on and off phases, respectively. Independently of the stimulation rate the time spent to reach the 50% of the variation was: on-phase about 76 ms for force and 33 ms for MMG; off-phase about 83 ms for force and 132 ms for MMG. A clear counter clockwise hysteretic cycle can be found when the sequence of the on and off phases force (Y axis) and MMG (X axis) values are plotted (see the figure).

DISCUSSION

MMG and force during the on-phase: the activity of the contractile elements could yield first to a muscle belly deformation with low tension in the in series elastic elements, as the stretching of these last increases the simultaneous muscle dimensional changes level off and the increment of the tension at the tendon becomes larger. This could explain the non linear relationship between the MMG and force during the on phase of the tetani. *MMG and force during the off-phase:* MMG recovery was much slower than the force one. This provided a clear hysteretical cycle when the dynamic behaviour of the two signals was compared. To this may contribute: a) the slow recovery of the intra-muscular pressure (Ameredes & Provenzano, 1997), increased during the tetanus, that may delay the recovery of a normal muscle perfusion and hence of a normal muscle geometry. b) the time spent by interstitial

fluid to reoccupy the inter-fibres space after having been squeezed out during the tetani (Tsuchiya et al., 1993).
Throughout the tetani other mechanical quantities presented a time relationship with the tension at the tendon similar to the one we found in our work between MMG and force. Indeed Hatta et al.(1988) showed, in tetanic contraction of the frog muscle, that the muscle transverse stiffness changes were faster and slower than the force ones during contraction and relaxation phases, respectively. On these basis we can hypothesise that the reduction in the transverse stiffness may facilitate the muscle transverse diameter changes monitored by the laser distance sensor as muscle surface displacement. Conclusion: the MMG/force comparison during the transient phases of the tetanic contraction may provide information about the dynamic responses of the muscle elements transmitting the tension at the tendon level and of the muscle elements influencing the muscular transverse dimensional changes as well as about their possible interaction.

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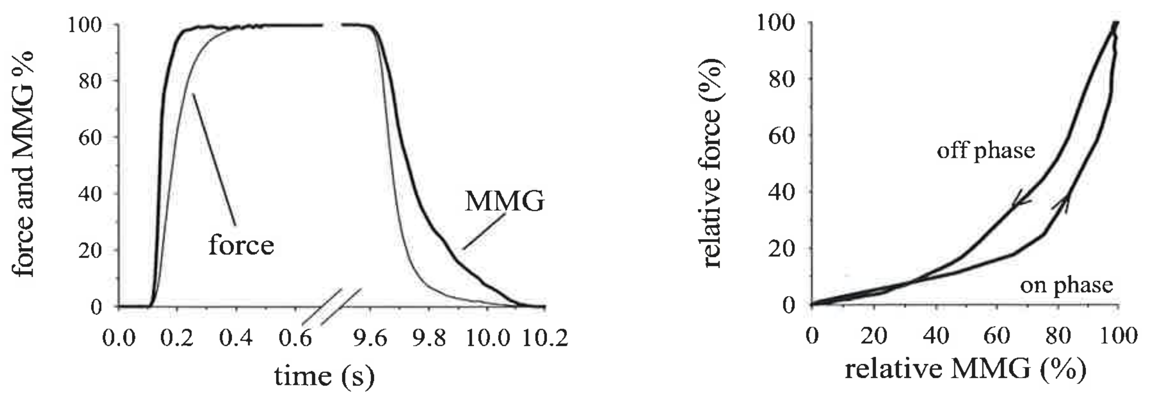

Figure 1. The average normalised force and MMG signals from the four cat gastrocnemius have been reported. The 30 and 40 Hz tetani presented the same behaviour. The right panel show the hysteresis due to the much slower MMG recovery during the relaxation phase.

THE EFFECT OF A PRACTICE MODALITY ON BILATERAL MOTOR TRANSFER

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INTRODUCTION

The learning of a skill with the dominant limb enhances learning the same skill with the non-dominant limb. The motor control explanation for the bilateral transfer phenomenon is based on the schema theory (Schmidt, 1975). Action is controlled by a generalized motor program (GMP) mechanism and can be produced by different muscles (e.g. either limb). This explanation is supported by EMG studies (Hicks et al., 1983). The recall schema, another mechanism responsible for the selection of muscles, can be enhanced by practice variability. Lai et al. (2000) suggested that a stable GMP developed previously by constant practice is requisite to developing an effective recall schema by variable practice. We sought to determine which practice conditions most reinforce the common elements of the task and thereby facilitate learning in the non-active limb. Our assumption was that, during acquisition of a novel tracking task by the dominant extremity, a practice modality in which the speed of tracking varies improves bilateral transfer more than a practice modality at constant speed or a combination of speeds.

METHODS

Upon verifying via a questionnaire that they were unskilled in the use of a joystick, we selected 36 volunteers for the study (17 male, 19 females; ages 18-23). All the subjects were undergraduate Physical Education students. The subjects were randomly divided into 3 practice groups: 1) Constant speed (C), 2) Variable speed (V), and 3) Mixed speed (M). Each group member was required to perform a tracking task with his/her dominant hand; specifically, the subject moved a joystick to maintain the image of a moving car on its track, as displayed on a CRT. The researchers kept the speed of the car fixed (70 km/h) for the C group and varied it randomly (60-90 km/h) for the V group. The M group performed the first half of the practice trials at fixed speed (70 km/h) and the second half at randomly varied speeds, like the V group. Each group received 12 practice trials over 3 weeks and was tested before (pre) and after (post) the practice period at a constant speed that was identical for all groups. The transfer test on the contra-lateral hand was performed on the dominant hand on the day of the post-test.

For the performance-behavioral variable, we measured the absolute tracking error (TE). TE was defined as the number of exits from the track. For the physiological variables, we recorded the surface EMGs of a wrist pronator (Palmaris longus-PL) and wrist supinator (Extensor digitorum-ED), so as to assess the level of transfer that took place due to the practice modality. Measures of particular interest to the researchers were the degree of co-contraction (CO) and the total amount of activation (TAC). The signal was integrated (IEMG) and normalized for each test. We marked the subject's skin so as to ensure consistent placement of the electrodes from test to test.

For statistical analysis purposes, we chose a one-way ANOVA with a rejection level of 0.05.

RESULTS

TE. Despite a lack of significance in the one-way ANOVA, the results showed a clear advantage for the V practice group, with an overall decrease in TE. The C group demonstrated the highest overall TE (Fig. 1).

TAC. All the subjects, regardless of the practice modality, activated the ED stronger than the PL (ratio 3:1). The V group showed the least activation of both muscles (Fig. 1). The PL muscle activity during the transfer task differed significantly ($P<0.05$) among the practice groups and especially among the C group.

CO. The V group showed significantly increased CO in comparison to the other groups. The C and M groups had some interaction but did not differ significantly.

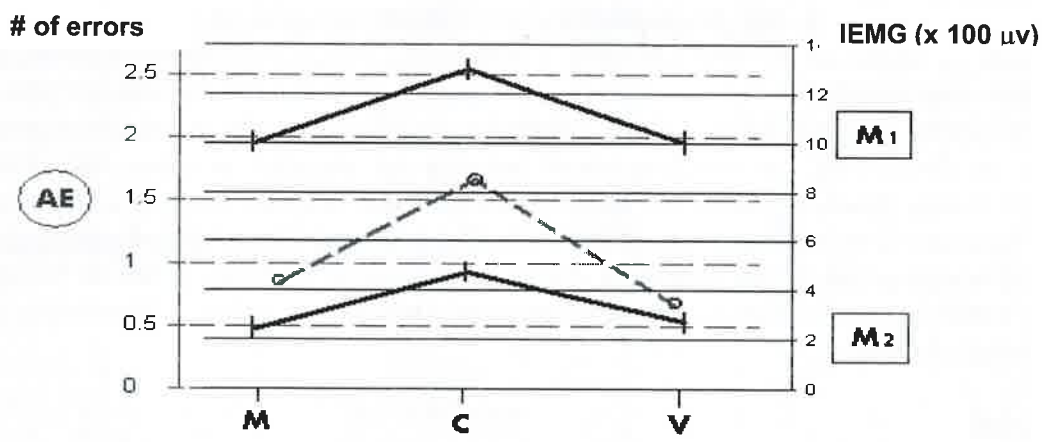

Fig. 1. Summary of results: IEMG (blue) of ED and PL against absolute error (green) during the transfer test by the three practice groups (M, C, and V)

DISCUSSION

Both the TE and TAC results supported our assumption that the V practice enhanced learning on the non-dominant side, as evidenced from the transfer task. However, the increased CO in this group during the transfer test might indicate that the GMP developed on the dominant side is not "stable" enough. The C group showed greater hesitation during the transfer task, as evidenced by the excessive PL activity throughout the task. However, the C group showed less CO than the V group. The M group performed the transfer task in a manner that was closer to the V group in TE and TAC; the M group also had the least CO. It can be argued that the M group developed a more "stable" GMP and, for this reason, their recall schema was better than that of the other groups.

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EFFECT OF 7 DAYS IMMOBILIZATION ON THE ABILITY OF FORCE REGULATION DURING ISOMETRIC ABDUCTION OF INDEX FINGER

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INTRODUCTION

Although a large number of studies have been conducted into negative influences on motor function after long-term disuse, little is known about the affect of short-term disuse and its mechanism [1, 2, 3]. It is particularly important to clarify the background of the disorder in regulation movements after short-term disuse to understand several processes causing the sudden decline of motor function after a few days rest. This study examined the effect of 7 days immobilization on the ability of force regulation during the isometric abduction of the index finger.

METHODS

Healthy subjects (12 males, mean 25.9 years) participated in the present study. Their left hand was immobilized with a plaster cast for 7 days. The following measurements were taken before and after the immobilization period and again at the recovery period. Subjects were regulated at the constant force level of 500 g, 1000 g, 1500 g and 2000 g during isometric abduction of the index finger. Standard deviation of the force trajectory (F-S.D.) was calculated at each force level as an index of stability of the force regulation. EMG activities were recorded with surface electrodes placed over the first dorsal interosseous (FDI). The root mean square (RMS) of the EMG amplitude was calculated at each force level. The RMS at each force level was normalized with the RMS at 2000 g force level. Changes in the cross-sectional area (CSA) of the FDI were evaluated with a magnetic resonance imaging system and a National Institute of Health (NIH) image program.

RESULTS AND DISCUSSION

The F-S.D. during force regulation increased after immobilization compared to that before immobilization. The rate of increase was 5.6%, 29.0% and 34.8% at a force level of 500 g, 1000 g and 1500 g respectively, being significant in the 1000 g and 1500 g level ($P<0.01$). This result shows that it was more difficult to regulate constant force levels after short-term disuse. The RMS at the same force level also increased after immobilization. The rate of increase was 27.2%, 35.5% and 36.4% at a force level of 500 g, 1000 g and 1500 g respectively, being significant at all force levels ($P<0.01$). This result shows that the compensatory increase of recruitment of the motor units occurred after short-term disuse to produce the same muscular strength. The value of F-S.D. and RMS in the recovery period almost returned to that before immobilization. In contrast, little change was observed in the CSA (less than 0.3%). In conclusion, the decline of submaximal motor function after short-term disuse is due to the decline in neuromuscular function, not to the decrease in muscle volume.

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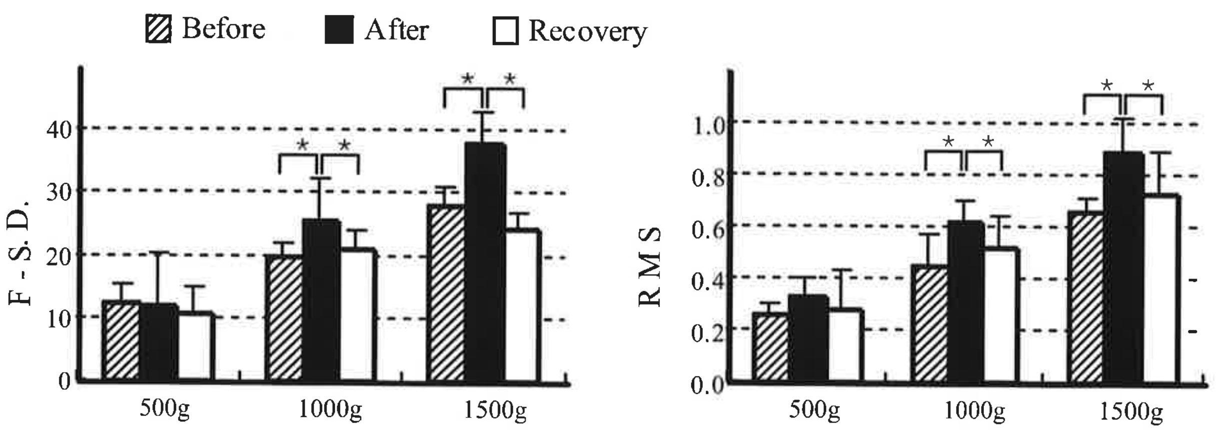

Fig. Comparison of standard deviation of the force trajectory (F-S.D.) and root mean square of the EMG amplitude (RMS) at each force level before and after the immobilization period and in the recovery period

EVALUATION OF THE EFFECT OF VIM-THALAMOTOMY AND ITS TREMOR CONTROL MECHANISM

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INTRODUCTION

The main characteristic symptom of patients with Parkinson disease (PD) and essential tremor (ET) is tremor. Cells in the thalamic ventral nuclear group fire at frequencies close to that of tremor for patients with PD and ET (Lenz et al., 1994). Tremor can be abolished by localized lesion of the ventral intermediate nucleus (Vim nucleus) of the thalamus (Narabayashi, 1989). The aims of this study are to evaluate the effect of Vim-thalamotomy by analyzing the power spectrum of postural tremor (PT) and resting tremor (RT) and to elucidate tremor control mechanism of Vim-thalamotomy.

METHODS

Ten patients with drug resistant severe postural and/or resting tremor, 7 patients with PD and 3 patients with ET were treated by Vim-thalamotomy after giving informed consent. Tremor was detected by accelerator sensor (MT-3T, Nihon Kohden, Japan), and assessed before and after Vim-thalamotomy. Tremor was evaluated under two conditions: (1) maintaining the index finger in a horizontal position using visual feedback with or without a weight load of 50g (postural tremor = PD), (2) resting the index finger with eye closed (resting tremor = RT). The power spectrum was calculated by auto-regressive model (AR model). The peak frequency (PF) and its peak power (PP) were evaluated under the two conditions. We also evaluated the total power (TP) (i.e., the summation of the power spectrum in the frequency range 0.5 to 50 Hz) of the power spectrum calculated by FFT. The statistical differences of these parameters between before and after Vim-thalamotomy were analyzed by t-test for paired data.

RESULTS

The patients with severe tremor demonstrated a periodic waveform. Therefore, the power spectrum had several peak frequency components including a fundamental component (the lowest frequency component) and some harmonic components. Then, we evaluated both PF and PP for the lowest frequency component. After Vim-thalamotomy, PF for both PT and RT significantly increased. On the other hand, PP and TP for both PT and RT significantly decreased.

DISCUSSION

Physiological tremor in normal person has the 8- to 12- Hz component. It is considered that this component originates from central nervous system. The 4- to 6- Hz component as seen in PD and ET before Vim-thalamotomy can be considered as the 8- to 12- Hz component shift to the lower frequency component. Indeed, the 4- to 6- Hz component originates from central nervous system. The neurons of the external globus pallidus (GPe) and the subthalamic nucleus (STN) are oscillating in PD. This oscillation is transmitted to thalamo-cortical projection neuron via thalamus. Vim-thalamotomy can suppress the tremor for PD and ET by blocking the oscillation of thalamo-cortical projection neuron.

CONCLUSION

PF, PP and TP for both PT and RT are useful indices for evaluating the effect of Vim-thalamotomy. Vim-thalamotomy can suppress the tremor for PD and ET by blocking the oscillation of thalamo-cortical projection neuron.

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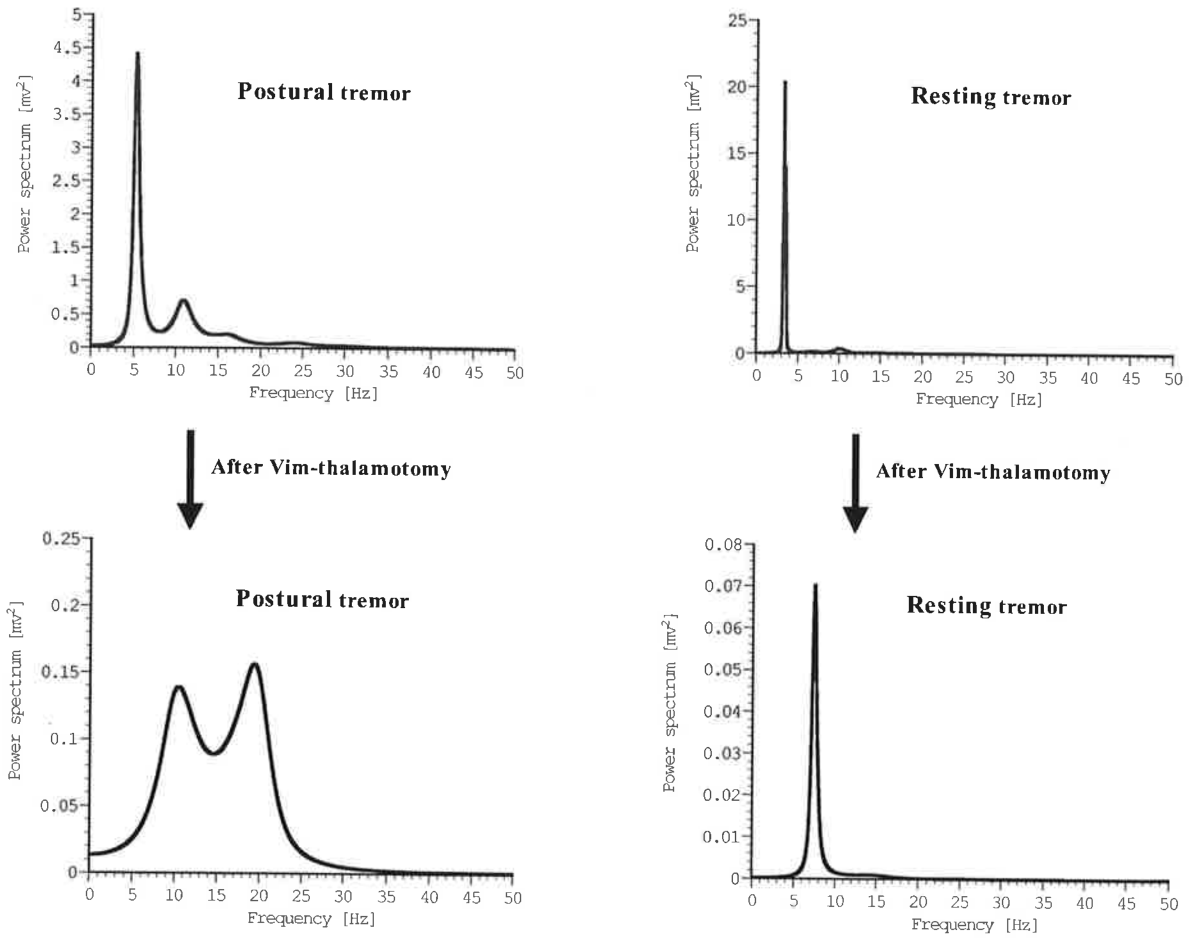

Fig. 1: Power spectrum calculated by AR-model before and after Vim-thalamotomy

CLINICAL NEUROPHYSIOLOGICAL ASSESSMENT OF RESIDUAL MOTOR CONTROL IN POST-SPINAL-CORD-INJURY PARALYSIS

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INTRODUCTION - Post-mortem human and animal studies have established that the spinal cord is rarely completely severed after blunt trauma.[1,2] Motor function after spinal cord injury (SCI) can be assessed in daily clinical practice, using the International Standards for Neurological and Functional Classification of Spinal Cord Injury.[3] In the absence of demonstrable voluntary movement, suprasegmental control is reduced to at most, the level of only being able to "influence" ongoing segmental neural activity. Sherwood and coworkers[4] adopted the term "discomplete," to differentiate SCI subjects who, in the presence of clinically complete paralysis, showed such subclinical, neurophysiological function.

METHODS - With approval of the local review board for human research, 67 spinal cord injured subjects were clinically evaluated using the American Spinal Injury Association (ASIA) Impairment Scale (AIS), and chosen for this study because they exhibited complete loss of motor function caudal to their spinal cord lesion (41 AIS-A and 26 AIS-B). They included 65 males and two females ranging in age from 21 to 80 years (mean age 43.1 $\pm$ 13 years). SCI levels ranged from C3 to T12 with time post-injury of from 1 to 458 months (38 years) with a mean duration of 104 $\pm$ 92 months. The neurological level and clinical completeness of their injury were obtained using the ASIA Impairment Scale and spasticity was assessed using the Ashworth Spasticity Scale. Brain Motor Control Assessment (BMCA), a protocol which records surface EMG (sEMG) from trunk and lower limb muscles[5] was carried out by technologists trained to instruct subjects to perform selected, specific voluntary and reflex motor tasks in a highly repeatable fashion for the three trials of each maneuver. The protocol sequence is relaxation, reinforcement maneuvers, voluntary and passive movements, tendon taps, manual clonus elicitation, vibration, and plantar surface stimulation, without and then with volitional suppression. Following validation of signal quality, the envelope of sEMG activity as an estimate of the corresponding spinal motor neuron pool activity over time, was calculated using a root mean square (RMS) algorithm. An average envelope was then calculated for the three repetitions of each maneuver. ANOVA, Turkey's (HSD) test, Fisher's exact test were used to analyze the relationship between individual markers of discompleteness and the clinical measure of spasticity, Ashworth Scale scores.

RESULTS - Of the 67 clinically complete , 43 were discomplete (64% overall), 32 subjects (74% of discomplete group), met criteria for presence of the tonic vibration response (TVR marker), 23 subjects (53% of discomplete group) could volitionally suppress the plantar reflex (PRS marker), and 13 subjects (30% of discomplete group) showed repeatable activity coinciding with at least one of the reinforcement maneuvers (RMR marker). Spasticity measured by the Ashworth scale was compared to sEMG activity recorded during BMCA movements. There was a significant relationship between the sEMG, Ashworth scale scores ($p<.001$) and the TVR ($p=0.02$). The RMR and PRS markers were not related to the Ashworth score.

DISCUSSION - From work done in a decerebrate cat model we learned "the maintained reflex [response to vibration] was abolished by making the preparation spinal or by anaesthetizing it, but it persisted after removing the cerebellum.[6] TVR in human subjects would utilize spindle afferents and their oligosynaptic segmental reflex arcs in the same way as passive stretch. Other than knowing that at least brainstem structures must be intact in the animal model.[16] The second most common marker, PRS, indicates descending inhibitory function is present to some degree and can be used volitionally. The least common marker was the RMR which is largely a postural response to strong efforts to contract uninvolved muscles. Ashworth Scale scores correlated strongly with EMG recorded during passive movement and significantly with the presence of the TVR marker. This could be expected because both passive movement and vibration utilize muscle spindle inputs to the spinal cord and probably share other neural circuitry within the CNS. These findings point again to the widely used definition of spasticity provided by Lance in 1980,[7] in which he says that spasticity is more than increased tone or resistence to passive stretch. He wrote that spasticity is also "...one component of the upper motor neurone syndrome." The upper motor neuron syndrome would include differing degrees of descending excitatory and inhibitory 'control,' when intact, or 'influence,' when weakened. Coupled with the passive stretch results, the discomplete markers provided a more complete and objective view of spasticity "as one component of the upper motor neuron syndrome"[7] and discomplete markers are some of the other components of the upper motor neuron syndrome.

CONCLUSION - It would be naive to think that the treatment of SCI, in a living system as complex as the human nervous system could be condensed into a single modality or technique. It is more likely that subtle changes in function provided by each member of a group of interventions would have a combined effect and the monitoring of such physiological events as the markers for discomplete, will allow proper selection of treatments and demonstrate their effects on function more sensitively than current clinical assessment. Much work needs to be done to identify and characterize the physiological markers for other neurological syndromes of motor control and to bring new intervention strategies targeting the neural mechanisms made apparent by those markers and using those markers to evaluate the effects of an intervention on that system.

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THE EFFECT OF DIFFERENT JOINT ANGLE ON STRETCH REFLEX SENSITIVITY IN TRICEPS SURAE MUSCLE

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INTRODUCTION

The stretch reflex response is affected by joint angle and muscle length (Yamamoto et al. 2000a). It is suggested that the angle-dependency of stretch reflex is different in the elbow flexor synergists, brachioradialis and biceps brachii. On the other hand, it is well known that the triceps surae muscles in the ankle extensor synergists, i.e., soleus (SOL) and gastrocnemius (GAS), have quite different functional characteristics. In this study, therefore, the aim is to examine the effect of different joint angle on the sensitivity of short- and long-latency reflexes in triceps surae muscles. Moreover, our novel method was used for the estimation of reflex sensitivity (Yamamoto et al. 2000b, Nakazawa et al., 2001). In general, it was difficult to directly compare the sensitivity of stretch reflex responses among different muscles, because the stretch stimulus, i.e., the change of muscle and muscle spindle length, is different in the different muscles. Therefore, we standardized the muscle fiber length measured by ultrasonography in each muscle, and then we can compare the reflex sensitivity as the input-output property in stretch reflex pathways among different muscles and/or different subjects.

METHODS

Five healthy male subjects aged 22 - 23 years participated in this study. Mechanical stretch stimuli were given to the triceps surae muscles, SOL and GAS, by torque motor system during weak isometric contraction ranging from 3 % to 12% of maximal voluntary contraction (MVC). In this study, the relationship of muscle length between SOL and GAS was changed by the combination of ankle and knee joint angles. The initial ankle position was three angles; 0, -10, -20 deg, and the initial knee position was also three angles; 0, 45, 90 deg. The stretch stimuli were applied to the subject's ankle in dorsiflexion direction 40 times at velocities for a range of about 20 - 250 deg/s in each joint angle combination. The EMG signal was full-wave rectified and averaged over a period from 100 ms before, to 300 ms after, the onset of the stretch stimulus. The stretch-induced reflex EMG activities were divided into the short-latency reflex response (M1) and the long-latency reflex response (M2) according to pervious studies.

The stimulus intensities of each muscle were measured by B-mode ultrasound system with 7.5 MHz linear-array transducer. The passive muscle stretching was applied to each muscle at various velocities to ensure immobilization of the probe. The actual change of muscle fiber length was possible to calculate from a planimetric model. Sensitivity of each reflex response was evaluated by the multiple linear regression analysis with two independent variables, muscular stretching velocity (MSV) and background EMG activity (BGA). The partial regression coefficient was defined as a reflex gain of each independent variable, and the intercept for an axis in each independent variable (x-axis intercept) was defined as a reflex threshold of each independent variable.

Statistical differences among mean values were tested by multi-factorial ANOVA.

RESULTS and DISCUSSION

The onset latency of M1 response were generally within the range (40 - 55 ms in SOL, 35 – 50 ms in GAS) reported by previous studies. The onset latency of M2 response (65 – 80 ms in SOL, 60 – 75 ms in GAS) was also agreed well with values reported in the previous studies. This indicates the correct definition of short- and long-latency responses in this study. And also, it found that the onset latency of M1 and M2 in both muscles depended on the stimulus intensity of muscle stretching and the ankle joint angle; the longer muscle length tends to the shorter onset latency. This suggests that the onset latency of stretch reflex is affected by the initial muscle length and stretch speed, and also it may be cause by the slack of muscle fiber and the visco-elastic property of muscle-tendon complex and so on.

Figure 1 shows the mean value of the reflex gain of soleus and gastrocnemius for the MSV indipendent variable of multiple regression analysis. The MSV reflex gain of M1 was siginificantly higher in SOL than that in GAS. This suggests that the SOL has high reflex sensitivity as compared to GAS, because of the anti-gravity function. And also, the reflex gain of SOL depended on the ankle joint angle. This indicates the muscle length dependency . The reflex gain of GAS and SOL also depended on the knee joint angle. However, this angle pedendencies are not related to the muscle length chnage. This suggestes that the angle dependency of reflex gain of GASand SOL may be related to other facters; pressure of foot sole, pre-synaptic influences, and so on.

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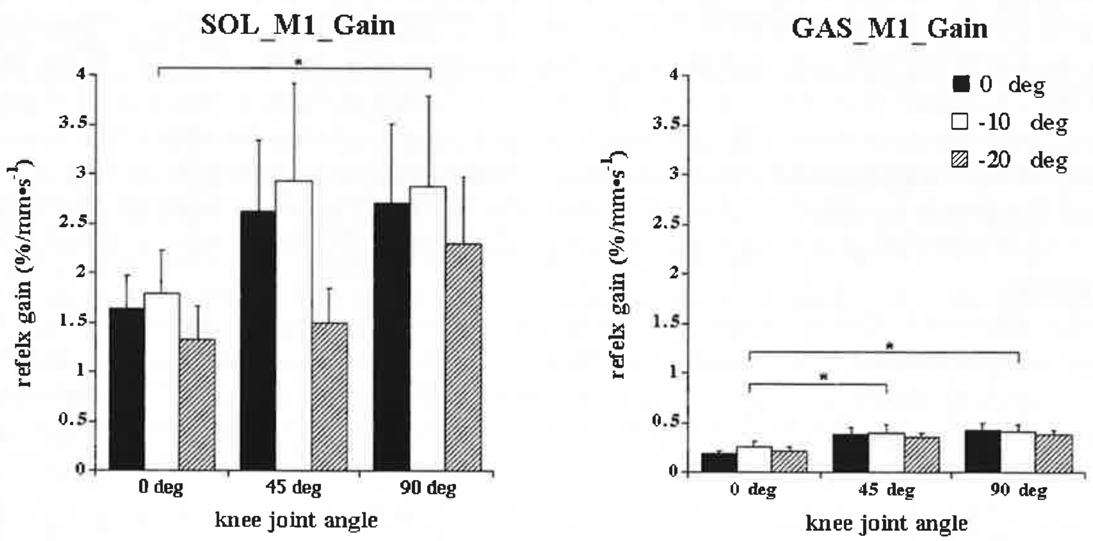

Figure 1 The mean and SEs of reflex gain for the MSV independent variable of multiple regression analysis in triceps surae muscles, soleus and gastrocnemius. The asterisk represents the significant difference among the knee joint angle ($p<0.05$).

BASIC STUDY ON ESTIMATING NUMBER OF ACTIVE MOTOR UNITS BY PROCESSING EMGS IN VOLUNTARY ISOMETRIC CONTRACTION

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INTRODUCTION

There were a few reports concerning to the size of active motor units (MUs) and the number in voluntary isometric contraction [1-2]. A new method of estimating the number and the size with mass electromyogram (EMG) and firing rates of each MU is developed and applied to the human muscles in voluntary isometric contraction.

METHODS

A) Principle: A model for generation of mass EMG is proposed, based on the knowledge of behaviors of active MUs: (1) mass EMG is the sum of motor unit action potential (MUAP) of all the active MUs. (2) An input of each MU is a statistically independent random pulse train. (3) The firing rate of each MU is a function of isometric muscle force. (4) Normalized waveform of MUAP, h(t) is assumed to be identical for all the MUs [2]. (5) We divide the force range into segments. Suppose N_j is the number of MUs recruited in segment j, K_j the mean size, and $f_j(P_i)$ the firing rate at the muscle force level P_i. Finally, one can obtain following relations.

$$\tilde{N}_j = \frac{N_j}{\alpha} = \frac{\{m_2(P_j)-R_j\}^2}{\{m_4(P_j)-3m_2^{\,2}(P_j)-Q_j\}f_j(P_j)} \tag{1}$$

$$\tilde{K}_j = \sqrt{\alpha}K_j = \sqrt{\frac{m_4(P_j)-3m_2^{\,2}(P_j)-Q_j}{m_2(P_j)-R_j}} \tag{2}$$

where

$$Q_j = \sum_{l=1}^{j-1} N_l K_l^{\,4} f_l(P_j)\alpha = \sum_{l=1}^{j-1} \tilde{N}_l \tilde{K}_l^{\,4} f_l(P_j) \tag{3}$$

$$R_j = \sum_{l=1}^{j-1} N_l K_l^{\,2} f_l(P_j) = \sum_{l=1}^{j-1} \tilde{N}_l \tilde{K}_l^{\,2} f_l(P_j) \tag{4}$$

Namely, one can estimate N_j and K_j successively from j=1 by using the second moment $m_2(P_j)$ and fourth moment $m_4(P_j)$ of mass EMG and the firing rates.

B) Experimental procedure: Mass EMG and isometric force in voluntary isometric contraction of three healthy male adults were recorded from the brachialis and extensor digitorum communis (EDC) muscle. Two electrodes were inserted into the muscle tissue of brachialis apart approximately 2cm longitudinally to measure mass EMG, and a pair of surface electrodes were used for EDC.

RESULTS

The firing rate - force relation measured for brachialis muscle [3] and that for EDC [1] were used. We calculated the second and fourth moments of EMGS at various muscle force levels. Results are shown in Fig. 1. The MU of lower threshold force was small in size; the size principle holds in these relations. Number of MUs with the higher threshold force was smaller; this feature agrees with the result of FDI muscle [2].

CONCLUSION

We developed a new method of estimating the number of active MUs and their size with EMG and applied it to the brachialis and EDC muscles in isometric contraction. The estimated results agreed with the physiological findings.

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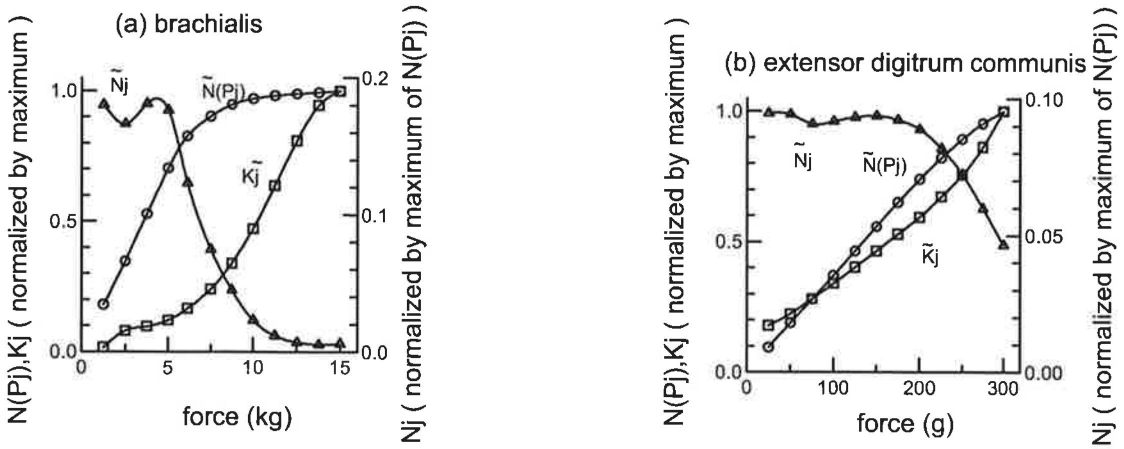

Fig.1 Estimated results of number Nj (triangles) and the size Kj (squares) of MUs recruited in segment j and the total number of MUs (circles), j=1,2, ... , 12. The maximum force for (a) is 26kg, and that for (b) is 700g.

NEGATIVE ELECTROTONIC POTENTIAL PRECEDED VOLUNTARY ACTIVATED MOTOR UNIT ACTION POTENTIAL RECORDED AT END-PLATE REGION IN HUMAN VASTUS MEDIALIS MUSCLE

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INTRODUCTION

When we recorded single motor unit action potential (MUAP) with monopolar surface electrode from just on and in the vicinity of end-plate in human m. vastus medialis, the monophasic negative electrotonic potential was found in the first negative phase of MUAP waveform.

In the present study, we would propose that the negative electrotonic potential (Early Negative Deflection: END) was originated from the volume conducted current when motor nerve action potential arrived to the terminal of motor axon.

METHODS

Electrical signals (MUAPs) were recorded from right m. vastus medialis by custom-made surface array electrode, which consisted of 10 platinum wires (each 1 mm in diameter, 2 mm in center-to-center separation). The wire electrode (probe electrode) was connected to the positive input of a differential amplifier. An Ag/AgCl disc electrode (disc diameter: 5 mm; reference electrode) was connected to the negative input. The disc electrode was placed on the skin surface overlying the patella to which the distal end of the objective muscle fibers was attached. The subject was requested to increase the isometric tension of knee extension gradually ($10N{\cdot}s^{-1}$) to the recruitment threshold tension of the objective motor unit or a little above. Immediately after the objective motor unit fired, the subject relaxed the tension to obtain only one electrical response of the unit (twitching).

RESULTS

1) Monophasic negative potential (END; indicated by arrows in figure 1) appeared in the first negative phase of MUAP waveform recorded in the vicinity of the end-plate. The END amplitude decayed in an exponential manner with the distance from the end-plate to electrode and the peak appeared almost in phase with each other.

2) Onset of END preceded 0.23 ms (S.D. = 0.02, N = 11 units) from the timing of generation of MUAP at end-plate (or muscle fiber membrane).

3) The muscle temperature decreased to 15.3 °C when the ice was applied. Removed the ice, the muscle temperature returned to 30.5 °C for 45 min. During this recovery period, the interval time shortened 1.14 ms to 0.36 ms.

DISCUSSION

END is always accompanied with the subsequent voluntarily activated MUAP. The END spreads to surrounding tissue electrotonically with an apparent time interval from the onset timing of MUAP at the end-plate. The present interval time is good agreement with the synaptic delay at end-plate in rat diaphragm muscle (0.217 ms; Hubbard & Schmidt, 1963). Muscle temperature dependency of the interval time gives Q_{10} of 3.4 from 15.3 °C to 20.9 °C. Katz & Miledi (1965) reported the transmission delay with Q_{10} of 3.14 in frog sartorius muscles at the temperature range between 2 °C and 19 °C. The present temperature dependency is, also, consistent with the results from frog sartorius muscle.

From the negative polarity and electrotonic property, it considered that the source of END is end-plate potential (EPP). However, EPP could not distinguish from action potential in physiological condition.

We propose that END originates the current flow from nerve action potentials arrived to the nerve terminal.

CONCLUSION

Measurement on the interval time from the END to the onset of the muscle action potential would make possible to us for detecting transmission delay at the motor end-plate in intact human muscle.

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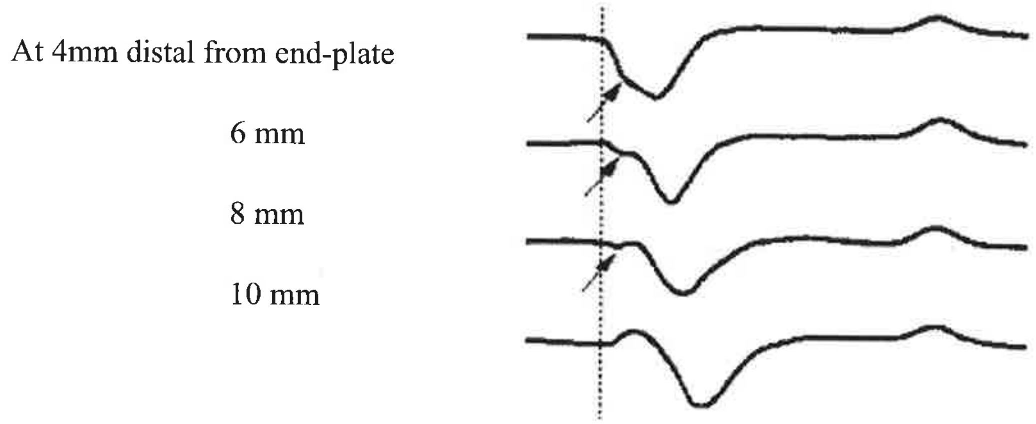

Figure 1: Typical electrical signals from a twitching motor unit. Peak of the transient negative phase appeared with a latency time proportional to the distance from end-plate, but END (indicated by arrows) appeared in phase each other.

MOTOR UNIT FIRING PATTERN DURING DOUBLE AND SINGLE CLICKING ON A COMPUTER MOUSE

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INTRODUCTION

Self-controlled (SC) rhythmic double clicking on a computer mouse has been shown to provoke anticipatory activity in the trapezius muscle and high peak rates of motor unit (MU) firing in the finger extensor digitorum communis muscle (EDC) (1, 2, 3). In this study it is hypothesised that an anticipatory activity is also present in the EDC muscle and that it is more pronounced during SC clicking compared to external command (EC) clicking. Further, that due to the preparation of the second click the activation strategy during the first click in a double click imposes higher peak firing rates and involves new recruitment to a larger extent than a single click.

METHODS

Three subjects performed four sequences in random order consisting either of ten single or ten double clicks as a SC rhythmic task with a frequency of approximately 0.5 Hz or as a response to a visual EC random in time, respectively. All mechanical clicks were recorded from a trigger in the mouse.

During all four sequences intra muscular EMG was recorded from quadripolar wire electrodes in EDC. The signal was decomposed into individual MU action potential trains using a computer algorithm based on waveform shape recognition (4). A MU firing pattern was classified as continuous when activity was identified before, during and after clicking, thus allowing for calculation of average baseline to peak firing rate in all four sequences. MU firing pattern was classified as intermittent, when the MU was only recruited in relation to the performed click. Firing pattern for each MU during 10 consecutive clicks in each of the four sequences were averaged after alignment of the first bottom push down

RESULTS

In total 23 MUs were identified, 7 were classified as continuous with a total of 10152 firings and 16 as intermittent with 505 firings. For the continuous MUs a baseline mean firing rate was calculated for the time periods from 1 to 0.6 sec before the click and from 0.6 to 1 sec after. The time for onset of increase in firing rate were defined as the time when firing rate exceeded baseline + 2 SD. Firing rate in all four situations increased from a baseline of approx 13 (12.3-13.1) to a peak of 20 (18.2-21.8) firings per second with no difference between the 4 situations. However, onset of rise in firing rate was as a mean 619 and 554 ms prior to the first click in the SC situations and only 311 and 326 ms prior in the EC situations. Further, the contribution of intermittent MUs was larger during the EC situations with a total number of 362 firings compared to 143 during the SC.

CONCLUSION

In conclusion both double and single click resulted in increased central drive evoking firing rate modulation as well as new recruitment. SC clicking was preceded by longer anticipatory activity causing a higher accumulated number of firings, while EC clicking caused a higher peak load with a steeper rate of rise of firing rate and a larger contribution from new recruitment.

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CORRELATION OF MECHANICAL ACTION WITH DIRECTIONAL TUNING IN THE FIRST DORSAL INTEROSSEOUS (FDI)

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INTRODUCTION
Motor tasks are accomplished via the recruitment of appropriate combinations of motor units to produce desired forces. It has been demonstrated in muscles such as biceps that many motor units are differentially recruited depending on the direction of external force produced isometrically (Herrmann & Flanders, 1998). It has also been demonstrated in the first dorsal interosseus that motor units can exhibit differential mechanical action (Thomas et al., 1986). This suggests that recruitment could be associated with mechanical action at the motor unit level. In an ongoing study, our data clearly show evidence of differential recruitment in the first dorsal interosseous (Suresh et. al., 2001). The purpose of the present study was to determine whether there is a correlation between the mechanical action and recruitment pattern of a single motor unit.

METHODS
All participants were consented via protocol approved by Northwestern University Institutional Review Board. Five neurologically normal participants were seated upright in a Biodex chair with their right upper arm comfortably resting on a plastic support positioned in 45° of abduction, 60° of elbow flexion, and 45° of wrist pronation. The 2^{nd} digit was carefully placed in line with the 2^{nd} metacarpal and the long axis of the forearm creating a 0° or neutral (abduction/adduction) metacarpalphalengeal (MCP) joint angle and inserted in a plastic vise attached to a six degrees-of-freedom load cell .(JR3 Inc.). An oscilloscope screen provided visual feedback about the direction and magnitude of force production at the MCP joint. Fine wire electrodes were inserted in two maximally separate locations of the FDI. The recorded single motor unit (SMU) data was bandpass filtered from 20-2000 Hz and digitized at a rate of 10 kHz. The x (abduction-adduction) and y (flexion-extension) coordinates of force generated by the MCP joint were filtered (DC-200 Hz) and digitized at a sampling rate of 1kHz.
Participants were instructed to generate slow ramped forces in a series of radially directed isometric combinations of flexion, abduction, extension and adduction at the MCP joint. When a single motor unit was identified, the subject was instructed to develop force in multiple random radial directions to establish the recruitment pattern of the same motor unit. Each direction was completed twice and contained a 1-2 minute duration of sustained SMU activity at a firing rate of between 8-11 Hz in order to allow for spike-triggered analysis of the force data. Data Analysis: On-line and off-line template matching was accomplished using Spike2 software (CED). Each single motor unit was discriminated with a template that was matched across every trial for a given subject. Plots of the x-y force coordinates during which the SMU was recruited was superimposed upon plots of the x-y force coordinates that the subject generated during the entire experiment. In order to determine the mechanical action of a single motor unit the resolved single motor unit spikes were used to spike-trigger average the force data in all three (x,y,z) resolved directions.

RESULTS

For each subject force twitches were calculated for 2-3 different motor units. The force twitches of all the motor units displayed stable contraction times as well as consistent force magnitudes regardless of the direction of force production. In every subject more units were recruited that displayed preferential recruitment for abduction. For these units, the magnitude of the calculated force twitch of the SMU was greater for abduction than for flexion. In three subjects, units were also recruited that displayed preferential flexion recruitment. For these units the magnitude of the force twitch was greater in the flexion as compared to the abduction direction. The SMUs also demonstrated differential recruitment properties where the recruitment threshold was greater in the non-preferred direction.

DISCUSSION AND CONCLUSION

While many motor units were discriminated and analyzed for differential recruitment, only 2-3 motor units were available per subject for determination of mechanical action. This is due to the necessary stipulation that the unit's firing rate is maintained in the 8-10 Hz range in all force directions. As a consequence we were only able to determine a general correlation between mechanical action and the recruitment pattern of a SMU.

Preliminary analysis of the data does seem to suggest that the recruitment threshold for a single motor unit is possibly determined by the mechanical action of the motor unit within the constraints of desired force output. However more data is necessary in order to statistically quantify the correlation as well as to establish the extent of directional specificity.

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THE EFFECTS OF ELBOW ANGLE ON FATIGUE OF THE BICEPS BRACHII AND BRACHIORADIALIS MUSCLES

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INTRODUCTION

A differential effect of fatigue on the electromyographic (EMG) signals of biceps brachii and brachioradialis has been reported for sustained low force isometric contractions. However, the observed differences between the two muscles are not consistent across studies [2, 4]. Because the elbow angle at which the optimal muscle length occurs is not the same for these two muscles [1], it appears that this factor might explain some of the discrepancies found across previous studies. Therefore, the purpose of this study was to assess the effects of elbow angle (50 degrees versus 110 degrees of flexion) on changes in EMG signal amplitude and frequency content of the biceps brachii and brachioradialis muscles during a sustained low-force isometric fatigue task.

METHODS

Six healthy subjects (age range: 25 to 35 yrs) volunteered for the study. Surface EMG signals were recorded from the short and long heads of the biceps brachii and from the brachioradialis muscles with bipolar surface electrodes (8 mm electrode diameter, fixed inter-electrode distance of 20 mm). Signals were pre-amplified at the electrode site (X35) and fed into a differential amplifier with adjustable gain setting (X100 to X100,000; 10-4000 Hz; CMRR: 87 dB at 60 Hz). EMG signals were stored on computer with a sampling frequency of 2000 Hz.. Elbow flexion torque was measured with a transducer aligned with the elbow, which was positioned at about shoulder height. The subject's wrist was secured in a rigid cuff attached to a metal arm extending from the transducer. The metal arm was positioned to form either a 50-degree or a 110-degree angle between the forearm and upper arm (elbow angle; 0 degrees being full extension).

Two separate sessions were performed for the 50 and 110 degrees conditions, at least one week apart. For a given session, the fatigue task consisted of sustaining an isometric elbow flexion effort at 25% of the maximum voluntary contraction (MVC) torque (obtained at the corresponding angle) until exhaustion. Exhaustion was defined as the point in time when force decreased below 25% MVC for a 5-s period.

The Root Mean Square (RMS) value and the mean power frequency (MPF) and median frequency (MF) of the EMG power spectrum (2048 points, raised cosine window, Fast Fourier Transform) from each muscle were calculated for every 5-s period of the fatigue task. Differences between *muscles* and *angles* during fatigue were evaluated with repeated-measures ANOVAs. A level of significance of 0.05 was chosen for all analyses.

RESULTS

Generally, no significant differences in the behavior of EMG variables with fatigue were found between the two muscles. In contrast, a more pronounced increase in EMG RMS amplitude was found for both muscles at 50 degrees compared to 110 degrees (Figure). No such angle-related differences were found for the MPF or MF (Figure).

DISCUSSION AND CONCLUSION

A differential effect of fatigue between biceps brachii and brachioradialis was observed for some subjects in the present study. However, the pattern of differences was not consistent across subjects. Also, varying elbow angle had a consistent effect on the increase in RMS amplitude with fatigue, but not on the frequency content of the EMG signal (MF or MPF). Consequently, it appears that the different elbow angles used in previous studies looking at biceps brachii versus brachioradialis fatigue at low forces [2, 4] cannot explain the inconsistent results reported with regards to a differential fatigue effect on the two muscles. Our present data indicate that the strategy used by subjects, in terms of the relative involvement of BB and BR for a given fatigue task, is not consistent but rather varies across individuals [3], and could be a significant factor explaining the inconsistent results of previous studies.

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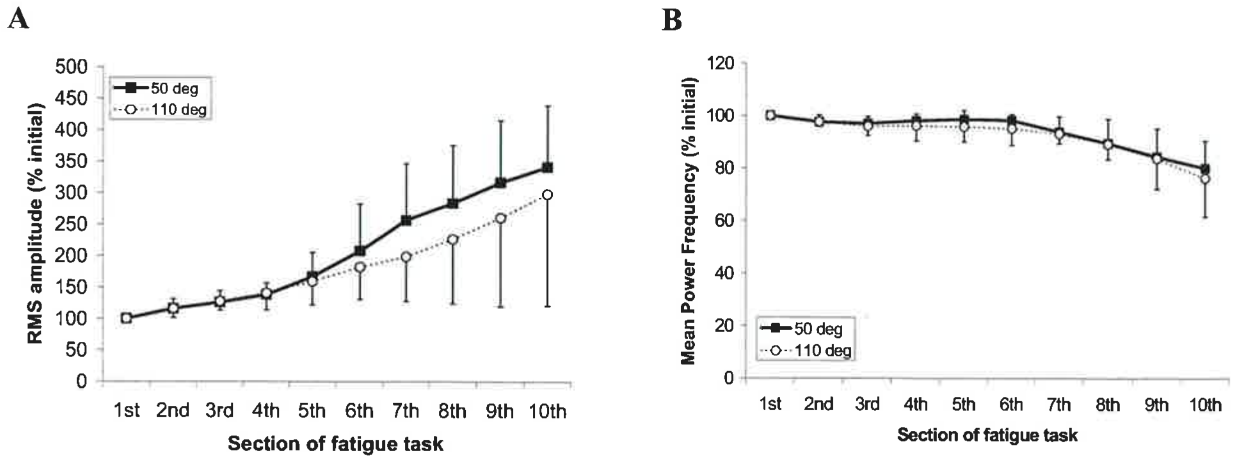

FIGURE. Group mean (+ or – standard deviation, n = 6) for the Root Mean Square (RMS) value (A) and the mean power frequency (MPF) (B) of the brachioradialis muscle during both fatigue tasks (50° and 110°). Because of the different times to exhaustion between the two angle conditions and between subjects, time as been expressed as a section of the fatigue task (tenth).

EFFECTS OF FORCE AND FATIGUE ON EMG SIGNALS OF THE QUADRICEPS OF MEN AND WOMEN

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INTRODUCTION

Differences in the frequency content of electromyographic (EMG) signals between the muscles comprising the quadriceps femoris complex have been reported [2, 6]. Similarly, the effect of force on the EMG median frequency (MF) is not identical across all muscles of the quadriceps [6]. These differences have been explained by potential differences in the fiber type content between the muscles [1, 7]. In addition, significant differences between men and women in the EMG frequency content of the vastus lateralis (VL) muscle (and its changes with force or fatigue) have been shown by some [6], but not others [3, 4]. The gender-related differences would also possibly reflect differences in fiber composition between men and women for this muscle group [5, 8].

Therefore, the purpose of the present study was to determine, in a single group of subjects, the effect of gender on the changes in EMG amplitude and MF or mean power frequency (MPF) with increasing force and with fatigue in the quadriceps muscle group.

METHODS

Fourteen healthy adults (7 women, 7 men; age ranging between 22 and 43 yrs) volunteered for this study. The muscle group of interest was the quadriceps femoris. Subjects were seated upright, with the hip and knee at about a 90° angle. The force produced during knee extension efforts performed with the dominant (right) leg was measured with a load cell (range of 0-500 N). EMG signals from the rectus femoris (RF), vastus medialis (VM) and VL muscles were recorded with bipolar surface electrodes (8 mm electrode diameter, fixed inter-electrode distance of 20 mm), pre-amplified at the electrode site (X35) and fed into a differential amplifier with adjustable gain setting (X100 to X100,000; 15-4000 Hz; CMRR: 87 dB at 60 Hz). EMG signals were stored on computer with a sampling frequency of 2000 Hz.

Subjects had to perform two main isometric tasks during a single experimental session: two to five ramp contractions and a fatigue task. For the ramp contractions, subjects had to gradually increase (linearly) their force from 0 to 100% of their maximal voluntary contraction (MVC) force in a 6 s period. The fatigue task consisted of a sustained maximal effort until force decreased below 50% MVC for a 5 s period. Immediately after the fatigue task, subjects were asked to perform another ramp contraction.

For the ramps, the Root Mean Square (RMS) value and MF and MPF of the power spectrum (512 points, raised cosine window processing, Fast Fourier Transform) were calculated from a 0.5-s window at each of the following force levels: 10, 20, 30, 40, 50, 60, 70, 80 and 90% of maximum. For the fatigue task, RMS amplitude and MF and MPF were calculated for consecutive 5-s windows throughout the whole fatigue task. Mixed-design two-way ANOVAs were used to determine the effect of *gender* (independent factor) and *force* or *fatigue* (repeated measures factors) on EMG RMS values and MF or MPF. Differences across the three muscles were also evaluated using three-way ANOVAs where in addition to the *gender* and *force* or *fatigue* factors, a *muscle* factor was included in the model. The level of significance chosen for all tests was 0.05. Pearson-product moment correlation analyses were also performed to test the association between selected variables.

RESULTS

Generally, no effect of the *muscle* factor was found on the MF or MPF values. Also, few differences were observed between men and women with regards to the EMG variables studied. This was true for all three muscles investigated. Amongst the significant differences found were a greater relative increase in EMG RMS amplitude with increasing force for all three muscles of the quadriceps of men compared with women, and a more pronounced increase in MPF with force for the VL muscle of men compared with women (Figure). Also, no differences between the three muscles and between men and women were observed concerning the effect of fatigue on the EMG variables, except for the VL, where the effects of the *force* factor and the *gender* X *force* interaction were no longer significant for the post-fatigue ramp.

DISCUSSION AND CONCLUSION

The greater increase in RMS amplitude (for the three muscles) as well as in MPF (for the VL) with increasing force in men compared with women, could be explained by a greater proportion and/or size of type II fibers in men. However, the magnitude of the gender-related difference in the MPF increase was relatively small (8%). This, combined with the observation of no difference between men and women for changes in EMG signal amplitude or frequency content with fatigue for the present muscle group suggests that only a modest difference exists in fiber type content and/or area in the quadriceps femoris of men compared with that of women.

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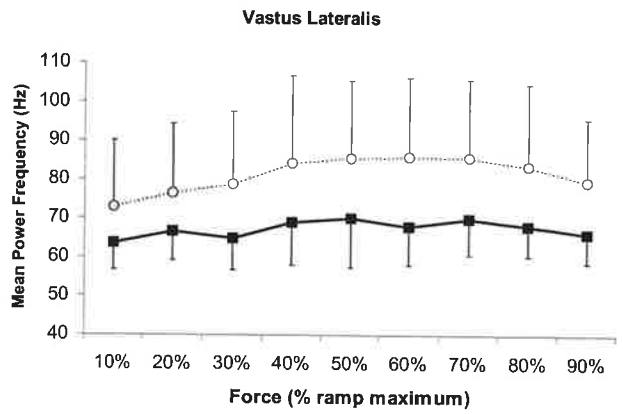

FIGURE: Group mean (+ or – standard deviation) of the mean power frequency (MPF) of the vastus lateralis muscle is shown with increasing force for both men and women. Note the more pronounced increase in MPF with force for the men compared with the women.

ELECTROMYOGRAPHIC ANALYSIS OF THE MASTICATORY MUSCLES DURING CHEWING IN PATIENTS WITH TEMPOROMANDIBULAR DISORDERS

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INTRODUCTION

Acording to CLARK et al. (1987) and CARLSON et al. (1998) the eletromyography (EMG) is a usefull tool to demonstraded the muscular disorders, like as the muscular hyperactivity, by the amplitude and spectral analysis. The EMG frequency analysis could indicate the muscular fatigue, the type of activated muscular fiber, the recruitment rate, the conduction velocity of the muscular fiber, motor units (MU) synchronization and the discharge rate. Considering the advantages of the EMG as diagnosis method, we evaluated the activation amplitude and pattern, the median frequency (MF) and the inclination coefficient of linear regression of the spectrogram (SLOPE) of the registers from masseter muscle and the anterior portion of the temporal muscle of the volunteers with temporamdibular disorders and clinicaly normal.

METHODS

The experimental sample consisted of 20 female volunteers, divided in two groups. The TMD group was to constituted with 10 volunteers with miogenic TMD (mean age 24 ±4,19) and the CONTROL group constituted with clinicaly normal volunteers (mean age 23 ±1,68). The EMG signals were acquired by differential surface electrodes, with 10GΩ input impedance, 130 dB CMRR/2 picofaraday, and 50 times of gain. The raw EMG signals were digitized by a 12 bit A/D coverter board, Butterworth filter, band-pass of 10,6-509 Hz, 20 times of gain, and 1KHz sample frequency. Three time sets were selected from full EMG signal registers, always including 3 masticatories cycles completely. The values of Root Mean Square (RMS), the MF and SLOPE the were evaluated by the *Student* t test to compared the studied groups, and the Kruskal-Wallis test was used to variance analysis of the 3 time sets, both with the significance level set as $p<0.05$.

RESULTS

No significant statistic differences were found by Kruskal-Wallis test between mean normalized RMS values of the each muscle, in any of the time sets ($p>0.05$), for both groups. Also to the mean normalized RMS and MF values no significant statistic differences were found when its were compared between groups ($p>0.05$) to any interval of the studied time sets. The CONTROL group SLOPE values (figure 1) showed a regular behaviour for all the muscles, being negative in the first time set, positive in the second and negative in the thrid. The TMD group showed negative values of SLOPE in all the analyzed time sets (figure 1).

DISCUSSION

Based on obtained results it was not possible to verify fatigue by MF and RMS analysis in the three analysed time sets. This finding may be attributed to the short period of mastigatory activity register, which favors the muscle oxidative activity, preventing accumulation of hydrogen ions and metabolits on muscle, as well as the shift in the sodium and potassium concentrations to the maintenance of the MU discharge activity. A different strategy of MU discharge between groups was evidenced in the analysis of the SLOPE. In the CONTROL group the slope values in the first time set are negative for all the studied muscles, indicating a trend of the MF to shift to lower frequencies of the UM discharged, prioritizing the activation of the type I MUs. All the values of slope are positives in the second analyzed time set, demonstrating the trend higher frequency of discharged (type II MU). Once the MUs of type II are fatigade quickly during the contraction, the neuromuscular system again depends on the type I MUs, that are more fatigue resistant. In the third time set the values of SLOPE are negative for all the muscles, indicating the trend of the MF to shifted to the lower dischange frequencies again. However, the SLOPE found in the TMD group were never positive suggesting, like as thought over by KRAEMER, FLECK & EVANS (1996), the existence of a less ratio of type II fibres in the mastigatory muscles in TMD volunteers, or still, smaller cross-sectional area of these fibres related to the type I muscle fibres and alteration in the MU recruting.

CONCLUSION

No changes are found to the values RMS and the MF during 15 seconds of masticatory activity in the TMD or CONTROL groups. The SLOPE values suggest that TMD and CONTROL groups show different trends on MU recruitement of the mastigatory muscles, or an alteration in the ratio of MU type I and II.

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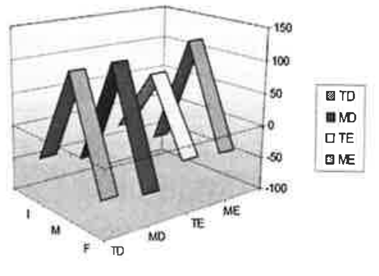
CONTROL



TMD

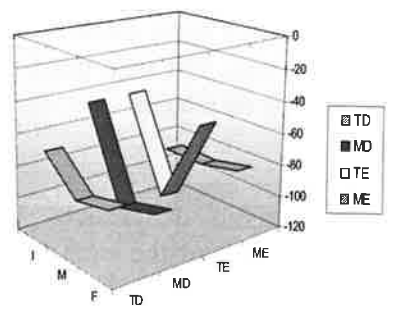


Figure 1- TMD and CONTROL mean values the inclination coefficient of linear regression of the spectrogram in the registers from rigth (MD) and left (ME) masseter muscle and the rigth (TD) and left (TE) anterior portion of the temporal muscle in the three studied time sets. Values in Hertz/minute.

THE EFFECTS OF DIFFERENT PROTOCOLS FOR ELECTRICAL STIMULATION IN THE FATIGUE RESPONSE OF THE WRIST FLEXOR MUSCLE GROUPS

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INTRODUCTION

Besides being a very useful therapeutic tool to restore sensory and motor functions (1), neuromuscular electrostimulation (NMES) has been widely used to promote muscular strengthening and hypertrophy. However, the frequency of the stimulating current should vary and is normally based on the therapeutic target. According to (2) the ideal frequency to increase muscle power and reduce fatigue should be around 50Hz. On the other hand (3) suggest that a 2500Hz frequency modulated at 50Hz is more pleasant and produces stronger contractions. Nevertheless, for applications involving high and low frequencies, the major limitation for NMES is muscle fatigue (4). Electromyography has been used to functionally evaluate muscle activity by means of methods such as the median frequency (MDF) (5). However, few authors used MDF to analyse fatigue associated to the immediate effect of NMES protocols (6; 7). Therefore, this paper aims to investigate the effect of two different NMES protocols in the fatigue of the wrist flexor muscle groups by means of MDF analyses during the stimulating program.

METHODS

The NMES protocols were applied for the wrist flexor muscles groups of the non dominant limb of 12 healthy volunteers (male and female) aged 18 to 29 ($X = 23 \pm 2{,}6$). The subjects were randomly divided into two groups of six, for analysis. The EMS program consisted of a single 30 minutes section using a electrostimulator (DUALPEX 961 - QUARK). One of the groups (1G – MF) was stimulated using pulsating current (T=100μs) at a frequency of 2000Hz modulated at 50Hz . The other group (2G – BF) was also stimulated by pulses of current (T=250μs), but at a lower frequency (50 Hz). Because the EMS response can never be exactly the same for any two people, it was necessary to adjust the EMS amplitude for each person. To do so, the intensity of the current was steadily increased until the desired contraction was obtained, without causing pain or discomfort to the subject. The EMG signal was collected by a pair of surface Beckman electrodes (20mm from each other) longitudinally positioned on the ventral portion of the muscles. An EMG signal conditioner was used to amplify and filter the signal (gain: 4400, filter bandwidth: 20Hz-4KHz), which was then sampled at a rate of 2kHz with 12 bits resolution per sample. In order to evaluate the effects of those two protocols, the subjects were asked to perform three maximum isometric contractions (4 seconds each), at different stages of the NMES program: a) 10 minutes before NMES; b) every 10 minutes during EMS; c) 15 minutes after NMES; d) 24 hours after NMES. The EMG signals were collected during those contractions and the MDF calculated.

RESULTS

The t student test ($p<0.05$) showed that, for the experimental conditions used in this work, group (2G –BF) presented median frequencies considerably lower ($p=0.0022$) when compared to group (1G – MF) (Figure 1), indicating higher fatigue response.

DISCUSSION

The results show that there was a significant difference between the two groups, with higher fatigue response for the group stimulated with lower frequencies. A possible explanation to that result may be that the muscular fatigue can be caused by the decreasing of the action potential propagation, which reduces the contractile capacity of the muscle fibres (8). Besides, there is an alteration on the velocity of conduction of the muscle fibre, depicted by the MDF. This alteration, according to (5) is the best parameter to recognise muscle fatigue. The works found by the authors analysing muscle fatigue by MDF study its behaviour during muscular training programs (6; 7), but do not evaluate the immediate effect of NMES, making it difficult to compared our results with theirs.

CONCLUSION

Within the experimental conditions, the results allow us to conclude that NMES of lower frequencies generates higher fatigue then medium frequencies stimulation. This suggest that, as a whole, if the therapeutic objective is to avoid muscular fatigue, one should use medium stimulation frequencies for the therapeutic protocol.

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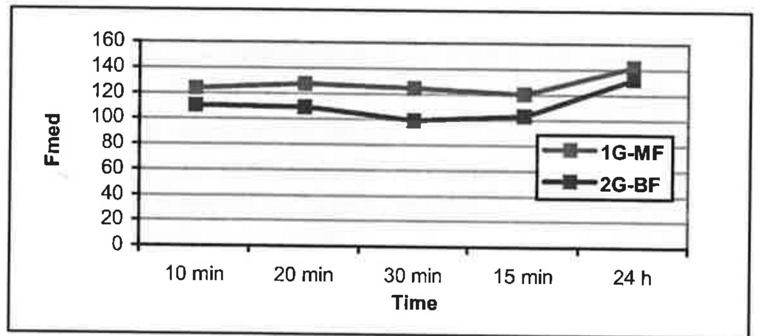

Figure 1 – MDF values for contractions of the wrist flexors during NMES program.

A METHOD TO DISTINGUISH THE EFFECTS OF FORCE AND MUSCLE FATIGUE UPON THE SHIFT OF THE MEAN FREQUENCY OF THE SURFACE EMG DURING MAXIMUM ISOKINETIC KNEE EXTENSIONS

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INTRODUCTION

Frequency analysis of EMG signals, using the mean or the median frequency of the power spectrum (MNF or MDF), have been widely used to characterize peripheral muscle fatigue during isometric contractions assuming constant force, i.e., static contractions. In the fields of rehabilitation medicine, sports medicine, and ergonomics it is more "natural" for the subject to perform functional tasks similar to daily life activities, i.e., dynamic contractions. The effect of the force (or torque) level, joint angle, and velocity on the MNF of the quadriceps muscles in the unfatigued state during dynamic contractions have not been sufficiently investigated. In addition, spectral analysis during these conditions must be carefully evaluated because there are changes in: the number of active motor units, the firing rates, the geometric relation between the active muscle fibers and the electrode, the geometric relation to the innervation zone and tendon, and the muscle fiber lengths (1). All such factors contribute to the nonstationarity of the EMG signal. Time-frequency methods have been introduced in the analysis of EMG signals to handle such time-varying effects (2).

During repetitive, isokinetic contractions performed with maximum effort, peak force will decrease markedly during the initial 40 to 60 contractions followed by a phase with little or no change. Our group and others have reported that MNF shows a similar pattern. Therefore, part of the decrease in MNF may be related to the decrease in force during dynamic contractions, i.e., an overestimate of the muscle fatigue.

METHODS

Surface EMG signals were obtained from the right vastus lateralis, vastus medialis and rectus femoris muscles. Instantaneous mean frequency (IMNF) based on the continuous wavelet transform (CWT) of the surface EMG signals was calculated off-line using MATLAB® (2). Each subject performed two types of tests using an isokinetic dynamometer. First, subjects performed five gradually increasing static (ramp) knee extensions at 5 different joint positions (knee flexed 120°, 100°, 80°, 60°, 40°) with duration of approximately 2-3 seconds. Then each subject performed 100 repetitive maximum isokinetic contractions of the right knee extensors from 90° of flexion to 0° (full extension). As the arm of the dynamometer moved up from 90° to 0°, subjects were encouraged to perform maximally for each contraction throughout the full range of motion (ROM). The subjects relaxed in the return movements.

RESULTS AND DISCUSSION

Figure 1a shows one individual's ramp knee extensions up to maximum voluntary contraction (MVC) in different parts of the investigated ROM. Averaging the five ramps (same angle) with respect to force reduced the estimation variance of MNF. It is obvious that MNF increased with increasing force. In addition, MNF varied considerably with angle. This is partly (or mostly) due to the moment arm change in the musculoskeletal system, i.e., more muscle force must be produced with the increased degree of extension in order to produce the same external force. Even though most subjects, according to our previous study (2), have significant positive correlations between MNF and force, there were also subjects who showed no change or an increase only during the initial part of the ramp. Therefore, normalization was performed with respect to each individual's force-MNF relation. If, as indicated in Figure 1a, MNF is force-dependent, the decrease in MNF during repetitive maximum contractions will be dependent upon both the force decrease and the peripheral muscle fatigue. In perspective of fatigue we will tend to overestimate the muscle fatigue during dynamic contractions. In Figure 1b both the absolute MNF changes and the MNF changes compensated for the force decrease are shown. Because MNF varies considerably with respect to angle (cf. Fig. 1a), we used a cyclostationary technique (3) in the present study when determining the relationship between force and MNF. To reduce errors, we determined the force-MNF relationship in the part of the ROM where peak force occurred. One can argue against our method of compensating for the force dependency of MNF because we used a static contraction when we actually were dealing with dynamic contractions. An alternative could have been to use single dynamic contractions at different percentages of MVC level. However, that would have been considerably more time consuming, and it would have been difficult to gather enough data points for the estimation of the force-MNF relationship. A high number of single dynamic contractions would also increase the risk that the force-MNF relationship would be contaminated by fatigue.

CONCLUSION
The estimation of peripheral muscle fatigue during dynamic contractions is complicated due to the nonstationarity of the EMG signal. Biomechanical factors such as position in ROM and force level are associated with considerable variability in MNF. In the present study we have presented a method that takes the individual force-MNF relationship in consideration when estimating the muscle fatigue during dynamic contractions.

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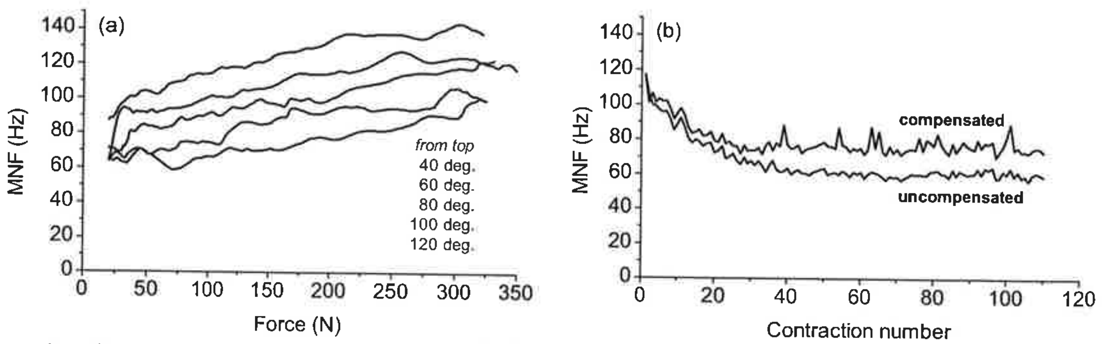

Figure 1: a) MNF vs. force for five different joint positions during ramp contractions. b) MNF during repetitive submaximal isokinetic contractions, without compensation and with compensation based on individual force-MNF relationship.

ELECTROMYOGRAPHY, TENSION AND SPECTRUM OF FREQUENCY OF TRAPEZIUS MUSCLE BEFORE AND AFTER APPLICATION OF TENS

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INTRODUCTION
The transcutaneous electrical stimulation (TENS) has been used in physical therapy as a device to relieve symptoms of pain and sensitivity. It is a non-invasive technique that, through electrodes fixed in the painful points, promotes a soft and safe stimulation, leading to a block in the passage of painful message from this area to the Central Nervous System [4,5,7,9]. Some authors has observed changes in the values of amplitude and frequency of electromyographic (EMG) signal after TENS use, when they verified an evident relaxation, however the mechanisms for what this changes act, modifying the EMG signal, is still unknown. In the present work, we propose to analyze the differences in the value of amplitude and frequency in the EMG signal obtained before and after TENS, showing, this way, the changes occurred and contributing to clarify the action mechanism of TENS on the muscular tension and pain.

METHODS
It was evaluated, by electromyography, the upper fibers of trapezius muscle, bilaterally, in 20 individuals, of female sex, from 30 to 50 years old, with complain of pain and/or tension in cervical area, scapular waist and headache. This EMG analysis had been done before and after TENS use, during 20 minutes with 100 HZ of frequency, 30 seconds of contraction time and one second of relaxation time, with intensity according to the patient's tolerance. The signals were acquired with gain of 5000 and the samples were digitized by a 10 bits A/D converter board with sample frequency of 4000 Hz and stored in a hard disc of a personal computer.
The records were obtained through disposable surface electrodes (DUOTRODE).
The data processing and analysis was carried out through the routines of Matlab software, specially developed to EMG signal. This data were statistically analyzed.

RESULTS
In rest, we observed that TENS application on trapezius muscle, determined increase of MdF (median frequency), at the same time that were verified a significant decrease of EMG activity, for both sides. In the maximal contraction, we observed significant decrease of MdF and increase of amplitude after TENS.

DISCUSSION
Median frequency is linearly related to the speed conduction of muscle fibers. So that, the median frequency increase is an indicative of speed conduction enlargement, consequently, improving the muscular contraction capacity.

The amplitude, on the other hand, is related to the muscle strength and its decrease reflects muscular tension and strength capacity reduction. The results obtained of median frequency and amplitude allow us to suggest that there was a muscular relaxation after TENS, decreasing the postural activity of these elements and increasing their contraction capacity. It's important to highlight that the volunteers, who took part of sample related relaxation sensation and tension and pain relief. The results of this research are in accordance with the studies that revealed that the MdF decrease is an indicative of physiological fatigue and reflects the speed conduction of action potential decrease in the muscle fiber, often with a consistent increase of EMG signal amplitude [6,1,3,8,2]. Analyzing the obtained results , showed that a high amplitude should be responsible for the fatigue and the median frequency deslocation towards low frequencies should be interpreted as a compensation.

CONCLUSION

The TENS was effective in the tension muscular relaxation and relief of trapezius muscle, bilaterally, increasing its contraction capacity. The effect caused by larger contraction capacity generated a consistent increase of amplitude observed during maximum contraction and the MdF deslocation that was interpreted as compensation.

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EVALUATION OF MUSCULAR FATIGUE AND THE RECOVERY WITH USE OF PHYSIOLOGICAL TREMOR

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INTRODUCTION

Physiological tremor (hereafter tremor) is involuntary and invisible mechanical vibration. The characteristic of tremor has two main frequencies for any body part. One frequency component is 10 Hz called as α rhythm component and the other one depends on the mass of body part called as mechanical reflex component. It was reported that the origin of the generation was the central nervous system and the system of mechanical factor of muscle and reflex system (1). In the study, the muscular fatigue during muscular load and in the recovery period was evaluated with use of tremor, comparing with EMG. The aim is to prove that tremor is more useful physiological quantity than EMG in the evaluation of muscular fatigue.

METHODS

(2-1) Items of experiment: The experiment of circadian rhythm of tremor and the experiment of muscular fatigue during weight load and the recovery period were carried out.

(2-2) Measurement: In the measurement of tremor, ten subjects attended to the experiment. The index finger is held horizontally in space and other fingers were fixed on board. The sublect held the finger horizontal position by looking the marker in the tip of the finger. The acceleration component was detected for one minute by piezoelectric sensor (9G111BW, NEC Sanei). In the circadian rhythm experiment, the measurement was carried out at three times, i.e., 10:00, 16:00, and 22:00 for ten days. In the fatigue experiment, weight load of 200g and 400g on the finger was used. The light load and heavy load were adopted to produce accumulative fatigue and acute fatigue, respectively. The subject continued to hold the load till exhaustion. Before and after the load, the tremor was measured maintaining the finger horizontally without weight load.
In the measurement of EMG, electrodes (NT-613, Nihon Kohden) were stuck on belly of m. extensor digitorum, which played a role on maintenance of the index finger horizontally. The distance between the electrodes was 1 cm. The signals of tremor and EMG were amplified by charge amplifier (6D07,NEC Sanei) and bio-amplifier (AB-601J, Nihon Kohden), respectively. The signals were sampled with sampling frequency of 1KHz.

(2-3) Analysis: The power spectrum of tremor was calculated with sampling points 1024. The total power (TP) was obtained for 1 Hz to 50 Hz, and the total powers for lower and higher frequencies, which divided by intermediate of two peak frequencies, were also calculated. The peak frequencies (PF) were obtained by AR model. In the analysis of EMG, TP and mean power frequency (MPF) were evaluated for 1 to 500 Hz.

RESULTS

(3-1) Circadian Rhythm: The value of TP of tremor did not significant change during measurement. That is, circadian change of tremor was not recognized. It is not necessary to take care the time measuring the tremor.

(3-2) Fatigue Experiment: The result of relative expression of TP for load 200g is shown in Fig.1. The result of EMG denoted the similar one, but the change was small. The results of TP in tremor and EMG for load 400g gave similar results to the load 200g, and the time till exhaustion was shorter than that for 200g.

DISCUSSION

(4-1) Evaluation of TP by tremor and EMG: During holding the load, the relative value of TP and EMG based on the value before load increased. The muscle to hold weight load indicated the recruitment and synchronization till the exhaustion, since the TP of EMG increased and the MFP of EMG denoted lower value. At the time of the exhaustion, the relative value of TP of tremor was much larger than that of EMG as shown in Fig. 1. In the recovery period, the change of TP of the tremor was evident than that of EMG. The tremor showed the useful item compared with EMG to detect the functional change of muscle.

(4-2) Evaluation of TP in frequency bands: In the process of accumulative fatigue due to weight load 200g, TP for α rhythm component of tremor denoted larger change than that of mechanical reflex component, so that α rhythm component played a role in the accumulative fatigue. On the other hand, in the process of acute fatigue due to load 400g, both components gave similar value of TP, so that both components contributed to the muscular fatigue. In the evaluation of EMG, it was difficult to discriminate two kinds of fatigue stated above.

(4-3) Recovery in different kinds of fatigue: The light load of 200g and heavy load of 400g led accumulative fatigue and acute fatigue, respectively. The change of TP of tremor during load and in the recovery denoted the remarkable change in the accumulative fatigue. The tremor was possible to discriminate the kind of muscular fatigue in the recovery period.

CONCLUSION

In state of muscular exhaustion and in the recovery period, tremor gave the lager change compared with that of EMG, so tremor proved the effective physiological quantity in the evaluation of muscular fatigue.

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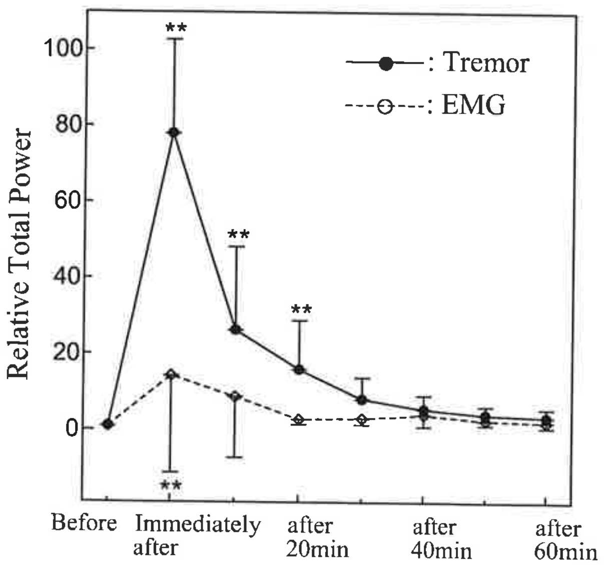

Fig. 1: Relative value of TP for tremor and EMG before and after exhaustion in weight load 200g Marks and are significant levels of 5 %and 1 %, respectively.

THE USE OF MEDIAN FREQUENCY ANALYSIS AS A PARAMETER OF FATIGUE INDUCTION IN TRAINED AND NON-TRAINED SUBJECTS

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INTRODUCTION

Muscular fatigue refers to a class of effects that damages the motor activitie performance during exercises. The last 50 years of research, however, showed that the fatigue is not caused by the involvement of a single system, indeed; mechanisms of fatigue are changeable by enviroment conditions. The electromyography as a tool to muscular fatigue evaluation has been proposed by many authors, and the Median Frequency (MdF) analysis as a study parameter is the most accepted actually. The aim of this study was to investigate MdF changes in order to evaluate sedentary and athletes' subjects after induced muscular fatigue, supposing that the training improves performance, by decreasing the fatigability.

MATERIALS AND METHODS

Twenty four male subjects (mean age= 21 years old, standart deviation=1.89) without osteoarticular pain history or lower limbs disfunctions were evaluated, divided in two groups (n=12) of sedentary (S group) and athletes (soccer players) (A Group) subjects. The EMG signal were acquired with differencial surface eletrodes (Lynx Eletronics), with 10GΩ input impedance, 130 dB CMMR/2 pfaraday, and 20 times of gain. The raw EMG signals were digitalized by a 12 bit A/D converter board, Butterworth filter, low-pass of 509Hz, high-pass10, 6Hz, 100 times of gain, and 1000Hz sample frequency (Lynx Eletronics). Other components also used were: a universal model of load cell with capacity until 100 Kg (Kratos Dinamometers), reference electrode and a rest bench. After detection of anterior tibialis muscle motor point, the electrodes were fixed on their muscular belly just below motor point, and the load cell was fixed on the floor and on the line of the first metatharsus bone on the dominant lower limb.To determine the maximal voluntary contaction (MVC) correlated to the value of the effort measured by load cell, three contractions (three seconds each) were performed. After 10 minutes in resting, each subject achieved the fatigue induction during 35 seconds on 80% MVC, followed by visual feedback and stimulated verbally to maintain the task. Median frequencies of 3 initial and final seconds were processed by Matlab software and statistically compared by groups using Welch's unpaired t test ($p<0.05$).

RESULTS

The resulting data showed significant differences between groups on 3 initial seconds ($p=0.0234$) and extremely significant on 3 final seconds ($p=0.00031$) of the task.

DISCUSSION

The word *fatigue* is considered as a too ample and less scientific concept to design a phenomenon that can be defined by many ways. Basmajian & De Luca (1985) suggest a division of fatigue in three categories: 1) subjective fatigue, characterized by the reduction of mental concentration, motivation and other psychological factors; 2) objective fatigue, showed by decrease on work ability; 3) physiological fatigue, observed by changes in physiological processes, and that happens before the objective fatigue and studied by electromyography (Basmajian & De Luca, 1985). The technological advances allow the spectral components analysis of the electromyographic signs, and a lot of researchers observed a gradual compression of EMG spectrum towards to lower frequencies on isometric muscular contractions (Komi, 1979; Hagberg, 1981; Kilbom et al., 1983). Median frequency analysis of Power Spectral Density (PSD) is the most accurate way to analyze towards to lower frequencies because it is the geometrical point of PSD graphic and shows the frequency variations with more accuracy. Physiologist researchers usually evaluate muscle power production as a muscular fatigue index, in special, the failure point. However, this concept has some practical disadvantages, because during the fatigue will be studied after be happened, and so, the failure point has a small use at ergonomic and clinical researches, because those studies require fatigue indications before the failure muscular point (De Luca, 1997). Although, Median Frequency shows the physiologic fatigue beginning before objective and subjective fatigue can be related by subjects.

CONCLUSION

It can be concluded that, under this experimental conditions, that MdF can be elected as a valuable parameter to distinguish between trained and non- trained subjects (sedentary and soccer athletes).

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ELECTROMYOGRAPHIC ANALYSIS OF MUSCULAR FATIGUE IN BRUXISM SUBJECTS

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INTRODUCTION

A muscle hyperactivity generated by bruxism has been considered an important compound of Temporomandibular Disorders (TMD) because it can cause pain and fatigue. Laskin (1969) made a theory about the etiology of muscular pain, in which emotional stress and anxiety improve frequency and intensity of muscular hyperactivity. So, the hyperactivity causes muscular spasms and pain, triggering a vicious circle – the more the stress and anxiety increases, more will be the pain and hyperactivity, producing more stress, closing the circle. Lund et al. (1991), however, considered which that theory doesn't have any foundation, because the pain can interrupt the vicious circle. Observing the motor function in five pain musculoskeletal conditions, the authors concluded that the resulting data don't show a vicious circle created by stress, pain and muscular hyperactivity. Otherwise, on hyperactivities conditions, the agonists muscles activities were reduced by pain, even when the pain is referred, besides that, antagonists muscles showed a little improve on this activation. The authors suggest that muscular dysfunction is a normal protective adaptation, but not the pain reason, calling the bruxism pain as post exercise soreness, connected with chronic muscular trauma.

Although both hypotheses don't have confirmation yet, the interesting point of view is the alteration of muscular behavior that causes the pain. The analysis of the electromyographic (EMG) sign frequency spectrum is useful for the understanding of the muscular physiological alterations, mainly fatigue (De Luca, 1997). So, the aim of this study was to investigate the eventual fatigue of the masticatory muscles (anterior portion of temporal and masseter muscles) by the amplitude and frequency of myoelectric alterations in bruxism subjects.

METHODS

The EMG signals were acquired with differential surface electrodes, with 10GΩ input impedance, 130 dB CMRR/2 ρfaraday, and 100 times of gain. The raw EMG signals were digitalized by a 12 bit A/D coverter board, Butterworth filter, low-pass of 509 Hz, high-pass of 10,6 Hz, 100 times of gain, and 1000 Hz sample frequency. EMG signs were collected bilaterally during five seconds of isometric movements, which was in 20 female volunteers classified in Control (C) and Bruxism (B) groups (n=10), processed by the non – normalized amplitude envoltory (NNA) and by the median frequency (MdF) – Matlab software, statistically analysed using Wilcoxon and Signal tests.

RESULTS

The NNA was higher in C than in B group, and the mean of MdF was significantly lower in every muscles of B group, excepting the right Masseter muscle.

DISCUSSION

Temporal muscle has more slow fibers, while masseter muscle is compound by fast glycolic fibers (Van Boxtel, 1983). Thus, the fatigue should be occur on masseter muscles, however, this research doesn't revel it. All studied muscles showed fatigue, mainly in B group, suggesting a type of fiber alteration.

It has been established that temporal muscle should be more fatigue resistant, because they have more slow fibers. Is known also that fast glycolic fibers presents on masseter muscle are recruited mainly during powerful tasks, however they are more liable to fatigue. We can presume, although this mechanism doesn't be totally explained, the muscular alterations on muscular hyperactivity can modify the recruited type of muscular fiber. So, due to imbalances caused by muscular hyperactivities in bruxism subjects (Lyons, et al., 1993;Hori et al, 1995; Ormeño et al., 1999), temporal muscles (anterior portion) carry out masseter tasks, may changing they fiber recruitment from slow to fast glycolic fibers, became more fatigable.

CONCLUSIONS

It can be concluded that, under this experimental conditions, temporal muscles (anterior portion) and left Masseter from B group presented fatigue, which suggests that the muscular hyperactivity caused by bruxism can be considered an etiological factor extremely important in TMD.

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ELECTROMYOGRAPHIC EVALUATION OF MUSCLE MASTICATION IN HEALTHY SUBJECTS DURING CONTINUOUS CHEWING

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INTRODUCTION

The pain in masticatory muscles (anterior portion of Temporalis M. and Masseter M.) has been related by several researchers in cases where the patients performs a large amount of tasks, as in bruxism and the lengthened chew mastication, and then associated to muscular fatigue.[1]

Physiological fatigue, which can be observed by changes in physiological processes, is studied by spectrum frequencies changes of electromyographic signal, for example, the spectrum compression towards low frequencies during isometric muscular contractions, but not much studied at isotonic situations (Kogi & Hakomada, 1962; Magnussom & Petersen, 1970; Komi, 1979; Petrofski & Lin, 1980; Hagberg, 1981; Kilbom et al, 1983 *apud* Semeghini, 2000)[2]. In other hand, pain can be described as an unpleasant sensorial and emotional experience associated potential or tissular damage[3], in accordance with International Association to Study of Pain.

The aim of this study was to contribute to electromyographic (EMG) evaluations over masticatory muscles fatigue occurring in continuous isotonic contractions in healthy subjects, in order to analyze their Median Frequencies (MdF).

METHODS

Eight volunteers were selected (age =19 to 27 years), free of Temporomandibular Dysfunction (TMD) signals and symptoms, verified through anamnesis and clinical examination. The experimental fatigue induction was realized by chewing bilaterally wet cotton rolls, asking each subject to chew for a period of 40 seconds in order to reduce the initial cotton consistency. Afterwards, the subjects were conducted to remain in rest for 80 seconds. Next, following the operator's command, the subject initiated the chewing and kept it till the moment he felt muscular tiredness. The EMG signals were acquired with differential surface electrodes, with 10GΩ input impedance, 130 dB CMRR/2 ρfaraday, and 100 times of gain. The raw EMG signals were digitalized by a 12 bit A/D converter board, Butterworth filter, low-pass of 509 Hz, high-pass of 10.6 Hz, 100 times of gain, and 1000 Hz sample frequency. EMG 10 seconds signals were taken in the beginning of chewing and immediately after the subject pointed the sensation of fatigue. Median Frequencies were obtained by Matlab software, with sets of 5 selected cycles of 100 milliseconds each (selected considering the most homogenous signals of isotonic contractions), this method applied to each muscle on both situations (before and after fatigue induction).

RESULTS

The resulting data were statistically faced to unpaired T test ($p< 0,05$) and showed that, excepting left Temporal muscle, all muscles had lower MF after fatigue induction ($p<0,0001$).

DISCUSSION

Muscular fatigue can be defined as a moment which a singular muscular group is unfit for maintain a constant power level. Depending on studied approach, this is a phenomenon which is connected to a damaged motor performance or to a decrease on EMG spectrum frequencies[4].

The resulting data showed on this research are in accordance to reviewed literature, because the MdF of right temporal (anterior portion) and masseter muscles had a significant decrease after the subjective fatigue sensation, which was established after 500 seconds, on average, of continuous chewing.

These results confirm that the study of most homogenous active portion of chewing cycle (continuous isotonic contractions) allows EMG signal frequencies and so, the induced fatigue during chewing, reiterating data showed in isometric conditions, may point to faithfully of frequency spectrum variations on fatigue determination (Palla & Ash, 1981; Malton et al., 1992; Lyons et al., 1993; Shi et al., 1993; De Luca, 1997).

Masseter muscle can be characterized as a strongest muscle on biting (Sicher, 1981; Figún & Garino, 1989; Madeira, 2001). However, Li et al. (1995) has related that pain improvement on masseter muscle recruits pre-motor neurons which realize inhibitor synapses on trigeminal motor nucleus. This inhibitory synapse has a function of decrease a functional overload to limit the tissular damage and provides muscle fiber cicatrization. This protect inhibitory mechanism of masseter muscle was observed on this research and suggests that the improvement of temporal muscle activity during continuous chewing is compensatory, because clinically is known that, even under painful conditions, the stomatognathic system doesn't stop your functions. Therefore, it can be probable that the improvement of temporal muscle (anterior portion) happens to contribute with the function ability of stomatognathic system[5].

CONCLUSION

It can be concluded that, under this experimental conditions, continuous isotonic contractions are able to induce fatigue in masticatory muscles of healthy subjects, becoming a valid protocol to study muscular hyperactivity during chewing.

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M WAVE AND H-REFLEX OF SOLEUS DURING MUSCLE FATIGUE BY ELECTRICAL STIMULATION IN HEALTHY SUBJECTS

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INTRODUCTION

The muscle tear often occurs caused by quick stretch and eccentric contraction during sports activity. The cause of muscle tear was thought as poor flexibility, inadequate muscle strength and/or endurance, dyssynergic muscle contraction, insufficient warm-up and stretching to prior to exercise and so on[1)]. In addition, the muscle fatigue is also important factor of muscle tear[2)]. It has been said that there are not only muscle fiber but also central nervous system as the potential fatigue site[3)]. Therefore, in order to prevent the muscle tear caused by muscle fatigue, we have supposed that there is a necessity for studying not only muscle function but also neural function corresponding with fatigued muscle. We will consider that we should apply the evaluation of muscle fatigue using evoked electromyography (eEMG) to prevention of muscle tear caused by muscle fatigue. In this study, we evaluate the muscle function and the spinal neural function innervating fatigued muscle using the M wave and H-reflex.

M wave has been used for evaluation of muscle fatigue due to reflection of the excitability of muscle membrane related to the muscle contraction[4)]. And it has been said that the H-reflex is reflected by the excitability due to alpha motoneuron pool[5)]. To apply the evaluation of muscle fatigue using eEMG to prevention of muscle tear caused by muscle fatigue, we studied the M wave and H-reflex of soleus muscle on fatigue by electrical stimulation (ES) in healthy subjects as the preliminary.

METHODS

Subjects were five healthy and non-athletic male, with a mean age of 26.6+/-4.6 (21-30) years and a mean height of 171.4+/-4.7 (164-176) cm. Every subject's dominant leg was right side. As to the task to induce muscle fatigue in this study, the intermittent ES was administered to the soleus muscle of the dominant leg for 10 minutes in the sitting position with the knee bent at 50 degrees and ankle kept at 0 degree. The task to induce muscle fatigue was consisted by following conditions of electrical stimulation; turning on of 10 seconds, turning off of 5 seconds, duration of 0.3ms and frequency of 30Hz in the biphasic square-wave pulses.

The M wave and H-reflex were recorded from the soleus muscle of the right side using Viking II*e* (Nicolet) before and after the task to induce muscle fatigue. The stimulus intensity for evoking the M wave was supra-maximum and that for evoking the H-reflex was 1.2 times for the threshold to appear the M wave. The duration of stimulation was 1.0ms and the frequency of stimulus was 0.3Hz. The active electrode was placed on the soleus muscle and a reference electrode was put on the lateral malleolus of fibula and we set up the band pass filter from 20 to 2000Hz. The number of recording the M wave and H-reflex were 3 times and 20 times, respectively. We analyzed the peak-to-peak amplitude of M wave and the amplitude ratio of H/M in this study. The reason why we used the amplitude ratio of H/M as the parameter in this study that the amplitude ratio of H/M should not have been affected any changes in the excitability of muscle fibers consequent to muscle fatigue[6)].

RESULTS

The amplitude of M wave after the task significantly decreased compared with that before the task ($p<0.05$). However, results of amplitude ratio of H/M varied as follows; decrease in two subjects, increase in one subject and unchanged results in two subjects.

DISCUSSION AND CONCLUSION

The muscle fatigue is important factor of occurrence of muscle tear[2)]. And it has been said that there are not only muscle fiber but also central nervous system as the potential fatigue site[3)]. Therefore, in order to prevent muscle tear caused by muscle fatigue, we have supposed that there is a necessity for studying not only muscle function but also neural function corresponding with fatigued muscle. In this study, we evaluate the spinal neural function innervating fatigued muscle using the M wave and H-reflex. M wave has been extensively used in the experiment of muscle fatigue because M wave reflects the excitability of muscle membrane properties related to the muscle contraction[4)]. It has been generally said that the amplitude of M wave decrease because the excitability of muscle membrane declines during muscle fatigue[4)]. In this study, we considered that the muscle fatigue had induced because the amplitude of M wave after the task significantly decreased compared with that before the task. The amplitude ratio of H/M has been considered as the index of the relative excitability of spinal neural function as against the muscle membrane[6)]. In the studies of relationship between task using ES and H-reflex, there is a report of increased and unchanged in the amplitude of H-reflex[7)]. In the study of transcutaneous electrical nerve stimulation (TENS), it has been reported that amplitude of H-reflex decreases after TENS[8)]. And also, it has been reported that amplitude of H-reflex does not change significantly although the result of modified Ashworth test improves after TENS in plantarflexor spasticity[9)]. From the mention above, it is considered that ES have a large influence to peripheral muscle function than the spinal function. In this way, about result of this study, we supposed the results of amplitude of H-reflex varied although the amplitude of M wave was declined by ES in this study.
In conclusion, it was suggested that there were not consistent change in the amplitude ratio of H/M, although the amplitude of M wave significant decreased by muscle fatigue caused by only muscle contraction due to ES in this study.

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AGE-RELATED EMG VARIABLES DURING FATIGUING ISOMETRIC CONTRACTIONS

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INTRODUCTION

Clarifying the mechanisms of age-related deterioration in neuromuscular function is one of the most important issues in human ergonomics, physical education, and rehabilitation because this deterioration may decrease activities in daily life (ADL) and cause stumbling or falling. Recent advances in surface EMG studies have introduced potential utility of the spectral profile and conduction velocity of myoelectric signals noninvasively derived from active muscles, for detecting their metabolic conditions (Okada et al., 1998; Merletti et al., 2002). The purpose of this study was to quantify the neuromuscular adaptation that occurred with aging by comparing changes in surface EMG variables from the tibialis anterior muscle in young and older subjects.

METHODS

Subjects were 27 healthy older women (65.0±2.7 years old) and 24 healthy younger women (22.7±2.6 years old) informed of the experimental procedure. Subjects conducted isometric dorsi-flexion of the ankle with a maximal voluntary contraction (MVC) for 5 seconds and fatiguing isometric contraction with 60% MVC for 1 minute. Surface EMG was picked up from the tibialis anterior muscle (TA) using a surface electrode array consisting of 4 stainless steel contacts spaced at 10 mm intervals. EMG variables such as the root mean squared value (RMS), median frequency (MDF), and muscle fiber conduction velocity (MFCV) were calculated during MVC contraction and 60% MVC for 1 min fatiguing isometric contractions. We evaluated the behavior of EMG variables during fatigue using the slope of regression line.

RESULTS AND DISCUSSION

Dorsi-flexion force and MFCV during MVC were significantly greater in the younger than in the older group ($p<0.05$). Surface EMG amplitude increased and the waveform slowed in all subjects during fatiguing contraction. As fatigue progressed, the RMS increased and the MDF and MFCV decreased significantly ($p<0.01$). Changes in the MFCV were significantly smaller in older than in younger subjects ($p<0.05$), but no change were seen in the ARV and MDF (Fig. 1). It has well known that the value of MFCV and its behavior during fatiguing contraction depend on the fiber composition of muscles (Komi and Tesch, 1979; Sadoyama et al., 1988). These results suggested that the neuromuscular system in older subjects was affected by the decline of motor unit activation and selective atrophy of fast twitch (FT) fibers (Kimura, 1996).

The EMG behavior during submaximal fatiguing contractions revealed the mechanism underlying neuromuscular deterioration in older subjects, i.e., the deterioration of neuromuscular function during submaximal exercises may influence control strategies for maintaining balance and walking in ADL. Our results indicate the utility of applying the EMG observed during fatiguing contraction to the noninvasive evaluation of neuromuscular function in the elderly.

CONCLUSION

We could evaluate the neuromuscular adaptation that occurs with aging by comparing changes in surface EMG variables from the tibialis anterior muscle in young and older subjects.

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Fig.1 Slope of regression line for the MFCV plots during 60%MVC fatiguing contractions.

SUCTION MUSCLES ELECTROMYOGRAPHIC STUDY OF ANTERIOR OPEN BITE CHILDREN

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INTRODUCTION

The prolonged nutritive and non-nutritive sucking habit, can determine several kinds of malocclusions, as the open bite (PROFFIT, 1978, BLACK *et al.*, 1990, MOYERS, 1991, SCHWARTZ & SCHWARTZ, 1992, MORESCA & FERES, 1992) and damages in stomatognatic system (BLACK *et al.*, 1990). Some characteristics found in children that use pacifier sucking habit are: anterior open bite, lip incompetence, tongue alterations and protrusion of the superior incisives (BLACK *et al.*, 1990). The goal of this study was to analyze the electromyography (EMG) signal by muscular activation pattern in children aged between forty-eight to fifty-nine months presenting nutritive and/or non-nutritive sucking habits and anterior open bite malocclusion.

METHODOS

Twenty-two children, with anterior open bite, primary dentition, increased muscular force, presenting pacifier and/or bottle habits, participated on this research. The values of the variability coefficients (VC) of the rectified full-wave, low-pass filtered and normalized by the average amplitude EMG signal were analyzed. The studied muscles were the orbicularis oris (upper), mentalis, and left and right buccinators. The tested movement was the sucking of different consistency drink, with straw. The drinks were water, yogurt, and chocolate cream. The Beckman surface electrodes acquired the EMG signals with one centimeter between their detection points. The raw EMG signals were digitalized by a 12-bit-A/D converter board, *Butterworth* filter, low-pass of 509Hz, high-pass of 10.6Hz, gain of 100 times, and 1,000Hz sample frequency.

RESULTS

The VC values found were not higher than 32% in the activation patterns of the studied muscles. The results of VC were: 17.62%, 19.43%, 19.33%, and 13.23% during water suction, respectively for the orbicularis oris, left buccinators, right buccinators, and mentalis muscles. During yogurt suction, for the same muscle sequence, the results of VC were: 20.61%, 16.46%, 17.23% and 12.26%. For the chocolate cream suction, the results were: 24.61%, 17.07%, 21.04% and 13.70%.

DISCUSSION

Nowadays, the use of the electromyography data obtained for study of the movement comes suffering several alterations mainly as a result of two factors. The first of them is the technological progress, the introduction of the computer science in the acquisition means, registration and processing of the electromyography sign. The second, it is the necessity of the standardization of the techniques used by the researchers in the different stages of its studies, to create groups of data to compare and reproduce the results, since the lack of details on the protocol, the registration equipment and the techniques of the sign processing make difficult the validation and comparison of the obtained results. This concern with the establishment of similar methodology to they be followed for collection, registration, analysis and interpretation of the electromyography signs has been discussed by several authors (BASMANJIAN & DE LUCA, 1985, TÜRKER, 1993).

Like this, if on one side several authors (BARIL & MOYERS, 1960, GUSTAFSSON & ALHGREN, 1975, ESSENFELDER & VITTI, 1977, YAMAGUCHI *et al*, 1996, STORMER & PANCHERZ, 2000), tried to describe the characteristics of the electric activation of the muscles orbicular oris, mentalis, and left and right buccinators in different contexts, on the other hand the direct comparison with our results demands caution and special considerations. In this research the values of CV are below 32% and, as they are related to a biological event, they are considered low and, therefore, indicative of the confirmation of an activation pattern described in relation to the medium width of the electromyography sign surface for the four muscles studied in the suctions of the different liquids, reaffirming its likeness in the activation pattern along the movement, independent of the consistency of the sucked drinks.

CONCLUSION

This shows there is an activation pattern of those muscles in the studied sample. In agreement with the data found in the present experimental situation, differences were not found in the activation quality of the orbicularis oris (upper), mentalis, and left and right buccinators muscles during different consistency drink suctions in children with open bite malocclusion and sucking habits.

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PERIORAL MUSCLES ELECTROMYOGRAPHIC STUDY OF DECIDUOUS DENTURE CHILDREN

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INTRODUCTION

It is observed in the literature that researches of perioral muscles influence in teeth position. These muscles and teeth unbalance can determine individual morphologic, functional, and aesthetic damage. (KYDD, 1957; GOULD & PICTON, 1962; VIG & COHEN, 1979; PRAHBU & MUNSHI, 1995). This projects the importance of studying the orbicular oris (upper) and mentalis muscles electric pattern. It is known that efficiency and reaching preventive action are better when knowledge is acquired and absorbed by people. Thus, Paediatric Dentistry has the opportunity of early action assuring adequate masticatory system development, going along with its dynamics (PASTOR & MONTANHA, 1994; CARVALHO, 1995; CARVALHO, 1996; CORRÊA, TOLLARA & AMAR, 1996). Based on this, the goal of this study was to analyze the electromyographic (EMG) signal by the values of (root means square) RMS and the muscular activation pattern by the normalized envoltory of the EMG sign in twelve children with primary dentition.

METHOD

The analyses were made from the comparison of the electrical activity amplitudes of the orbicularis oris (upper) M. and mentalis M.. Twelve children with primary dentition and good health and oral conditions participated on the research. The elected movement was sucking of different drinks (water, yogurt, and chocolate cream) with a straw. The EMG signals were acquired with the *Beckman* surface electrodes, with 1 cm. distance between the detection points. The raw EMG signals were digitalized by a 12 bit A/D converter board, *Butterworth* filter, low-pass of 509 Hz, high-pass of 10,6 Hz, 100 times of gain, and 1,000 Hz sample frequency. In order to obtain muscular activation, the first EMG signal process step was its complete rectification to obtain the curve absolute value, when the negative deflections were converted to absolute values. The next step was the high fluctuations suppression of the sign amplitude, smoothing it. The result is known as linear cover. Normalization based on brutal signal RMS values was determined to investigate muscles patterns. Besides it, each studied muscle curve variation coefficient (VC) was calculated. VC is a quantitative representation of cover repetition, that means, the values incidence that formed it. In last instance, VC indicates an activation pattern among studied volunteers. Registered brute EMG sign processing was calculated by Matlab software routines, specially created by *Aqdados 4.18* software – *Lynx Eletronica Ltda.* to processed sign. *EMG 126* routine was used to construct a time and amplitude normalized curve of an archive group. This curve represents a subject group normalized cover medium graphic. Data analyses of RMS values were carried out by the unpaired *Student test,by Graph Pad Instat®, 3.01 version, Free Demo, Graph Pad Software Inc.*

RESULTS

The results showed that, for values of RMS, there was not significative statistical difference (p<0,05) among the activation amplitude of mentalis and orbicularis oris (upper) muscles in sucking of the three different drinks. The orbicularis oris (upper) M. was 143.63, 156.89 and 144.06μV and, the mentalis M., 159.06, 160.95 and 164.79μV, respectively for water, yogurt and chocolate cream suction. For the normalized amplitude envoltory, it was not found variation coefficients higher than 17% in the activation patterns of the muscles during the different consistency drinks suction, showing that there is an activation pattern of that muscles in the studied sample.

DISCUSSION

Nowadays, EMG data used to movement study, have been passed through alterations as a result of two mainly factors. Technologic advance is the first. This is informatic acquiring methods introduction, EMG registering and processing. The second factor is the used technic padronization need by researchers in their different stages studies. This need exists in order to create data set that is possible of results comparison and reproducibility, once protocol detail lack and problems with registering equipment and signal processing techniques exists and difficulties obtained results validation and comparison. This common rules establishment worth to be followed during EMG sign collect, register, analyze and interpretation has been discussed (BASMANJIAN & DE LUCA, 1985, TURKER, 1993).Then, if several authors have tried to describe orbicularis oris (upper) and mentalis muscles electric activation characteristics in different contexts (SALES & VITTI, 1979, GUSTAFSSON & AHLGREN, 1979, ESSENFELDER, 1992), in the other side the direct comparison with our results require special considerations and care. In the present study, VC values were lower than 17% and, as they are related to biologic event, they are considered low and can indicate and activation pattern confirmation described related to superficial EMG sign meddium amplitude to both studied muscles in sucking of different drinks. This reaffirms their likeness in activation pattern during the movement, independent on sucked drink.

CONCLUSION:

In agreement with the found data in both analyses, in the present experimental situations, it was not found differences in the electric activities and in the activation quality of the orbicularis oris (upper) M. and mentalis M. during different consistency drinks suctions in children with complete primary dentition

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ATLAS OF THE INNERVATION ZONES OF UPPER AND LOWER EXTREMITY MUSCLES

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INTRODUCTION

The standardisation of electrode placement is an important issue in surface EMG muscle assessment. Many research efforts were devoted to this problem in the past, but no common agreement has yet been reached. The objective of this work was to assess the quality of the surface EMG signals recorded from the superficial muscles of the upper and lower limb, the back, the abdomen and the neck regions and provide indications on the correct electrode placement for these muscles. For this purpose, the variability of the innervation zone (IZ) location among subjects was evaluated. It is indeed well known that electrode placement on the IZ can determine misleading results in EMG analysis.

METHODS

EMG signals were recorded for each of the 55 muscles, from 9 healthy male subjects (age: 27 ± 4 years). The muscles investigated are listed and divided in groups, according to their anatomical position, in the caption of the Figure. In total, 28 muscles of the upper limb, 15 of the back, abdomen and neck, and 12 of the lower limb were analysed.

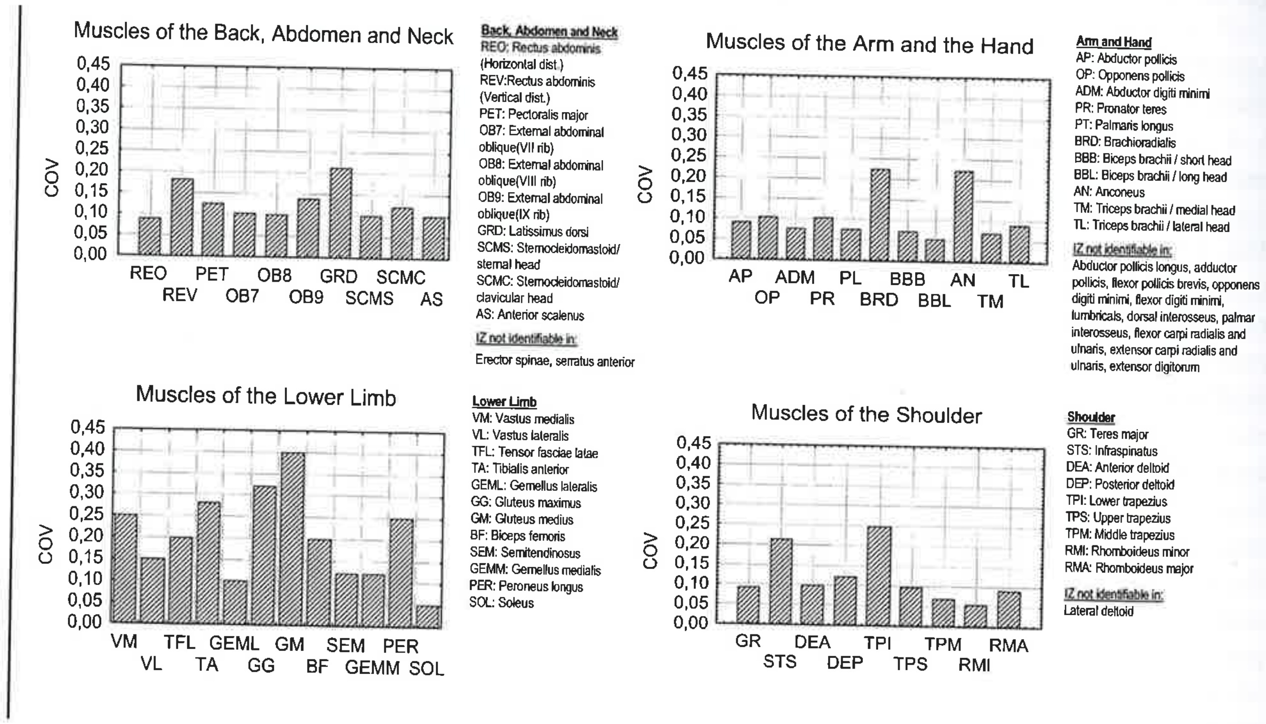
The EMG signals were detected using linear arrays of 16 electrodes with inter-electrode distance 2.5 mm, 5 mm or 10 mm, according to the length of the muscle, in single differential configuration. For each muscle, the subject was asked to perform two levels of contractions: a maximal voluntary contraction (MVC) and a submaximal contraction (< 30% MVC). The IZ was identified for each muscle and subject by a visual inspection of the signals, under the electrode pair corresponding to a minimum of signal amplitude and phase reversal. The location of this electrode pair was marked on the skin and its position defined in relation to anatomical landmarks previously defined. Whenever it was possible, the position of the IZ was expressed as a percentage of the distance between two anatomical landmarks; otherwise, the position was defined as the absolute distance from an anatomical landmark in a specified direction. The quality of the signals acquired from each muscle was judged in terms of the presence or absence of a clear MUAP propagation and in terms of the possibility of identifying the IZ location. Except for isolated cases, only one innervation zone was found in all the muscles analyzed; no significant difference was found between the positions of the innervation zone at the two levels of contraction, suggesting that all detectable MU's are innervated in the same location.

RESULTS

From 16 out of 55 muscles (14 of the upper limb and 2 of the back and abdomen) it was not possible to record signals from which the IZ could be identified and propagation observed. In the remaining muscles, the variability of the IZ location among subjects was assessed from the coefficient of variation (COV, defined as the ratio between standard deviation and mean of the IZ coordinate with respect to the selected anatomical reference). 19 out of 39 muscles had a COV<0.1 showing a small variability; 10 out of 39 had a COV in the range 0.1-0.2; 8 out of 39 had a COV in the range 0.2-0.3 and 2 out of 39 had a COV>0.3, showing high inter-subject variability. In all the 39 out of 55 muscles examined, an electrode location not overlapping with the innervation zone could always be clearly identified.

DISCUSSION AND CONCLUSIONS

Multi-channel surface EMG signals were recorded from the superficial muscles of the upper and lower limb, the neck, the back and the abdomen for a number of subjects. The quality of the signals was visually assessed and resulted very poor in less than one-third of the investigated muscles. Suggestions for proper electrode placement in a region between the innervation zone and tendon terminations were given, in order to increase standardisation and repeatability of EMG measures and allow comparisons of results between different laboratories and countries. However, whenever available, the use of multi-electrode detection systems, remains the best procedure for proper electrode placement.



Coefficient of variation (COV) of innervation zone location among different subjects in 42 muscles out of the 55 studied. A COV<0.1 was assumed as index of small inter-subject variability.

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ANALYSIS OF ELECTROMYOGRAPHY ACTIVITY OF THE DELTOIDE MUSCLE: ALTERATIONS RELATED TO THE AGING AND TO THE IMPINGEMENT SYNDROME

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INTRODUCTION

Although don't exist the epidemic data for Brazilian population, the literature about shoulder pathology suggest that females, with more than 40 years old, are the most affect by rotator cuff dysfunction like the impingement syndrome in the III stage (LIESDESK et al., 1997). However, in spite of the age be a factor considered in the diagnosis and determination of the stage of impingement syndrome (NEER, 1983) any electromyography study, present in the consulted literature, considered a specific age group or even researched differences among the electromyography signal of volunteers of different age groups involving the scapular girdle and shoulder abduction. Thus, the purpose of this study was to present an electromyography investigation of the three different portions of the deltoid muscle, in volunteers with diferent age groups, performing isotonic movement of shoulder abduction accomplished in the scapular plane for evidencing possible alterations of the electrical activity and of the pattern of muscular activation, related to the aging.

MATERIAL AND METHODS

The experimental sample consisted of 20 female volunteers, sedentary and without any history of neurologic and neuromuscular disease, instability or restriction of movements of the scapular girdle. The group Girls was to constituted with 10 volunteers aged from 21 to 30 years (24±3 years), the group Ladies, with 10 volunteers, had ages from 42 to 71 years (57±10 years), and the group Impingement, also with 10 volunteers, had ages from 43 to 77 years (60±9 years). All volunteers gave informed consent before participation.

The double differential (DD) surface electrodes with 10GΩ input impedance, CMRR minimum of 84 dB and gain of 10 times (Delsys EMG Systems Inc) and a reference electrode were used in this study.The signals were digitized by a 12 bit A/D converter board with a sample frequency of 1KHz, Butterworth filter with band-pass of 10,6-509 Hz and, for this experiment, the internal gain was fixed at 300 times. Measurements of the articular position during the shoulder isotonic abduction were taken with the use of an electrogoniometer of cylindrical siliconed rubber rods, conjugated to the acquisition system for syncronic registers. The volunteers performed actively the shoulder abduction in its whole amplitude (0-180°) starting with the arm in medial rotation, with the dominant upper limb only and during 5 seconds. The volunteers were also trained to rotate the arm externally from 90° of abduction and complete the elevation with this positioning, in a regular and continuous manner. Furthermore, the volunteers were instructed to avoid any compensatory movement.

RESULTS

The values of RMS indicated a greater activity of the anterior portion in relation of the others portions to Girls. This distinction in the electric activity of the anterior portion was not found for the Ladies and it was suggested as an age-dependent alteration. In the Impingement the electric activity of the anterior portion was greater than the midle and posterior portions. The normalized smooth (figure 1) revealed similar patterns of electric activation to Girls and Ladies.. At Impingement the abduction in the scapular plan showed an inversion of the pattern, the anterior and midle portions was more activated in medial rotation and the posterior, in lateral humerus rotation.

DISCUSSION

The results of the Girls and Impingement groups to RMS values analysis showed a greater activation amplitude of the anterior portion in relationship of the middle and posterior and were acording to MCCAN et al. (1993). However, in the Ladies group no differences among the electric activity of the anterior portions, middle and posterior were found, it is suggested a loss of distinction of the electric activity of the anterior portion in relationship to the other portions that it is aged-dependent. The normalized smooth of the anterior portion in the Girls group demonstrated that your electromyography activity always increased in relation to of the middle portion above 90° of shoulder abduction with the lateral humerus rotation. According to KAPANDJI (1990), the shoulder abduction associated to medial humerus rotation became the posterior deltoid portion activated since beginning of the movement, as was observed in smoths of the Girls and Ladies groups, although in this last group the differences are subtler.

CONCLUSION

The results of this study suggest that, in these experimental conditions, volunteers of age superior to 40 years presented decrease of the myoelectrical activity amplitude and change in the activation pattern of the three portions of deltoid muscle during the isotonic shoulder abduction, without load, performed in the scapular plane. Thus, investigations that contemplate the study of the shoulder pathologies and your rehabilitation by exercises should consider differentiation, when the results be extrapolated for groups with more than 40 years old.

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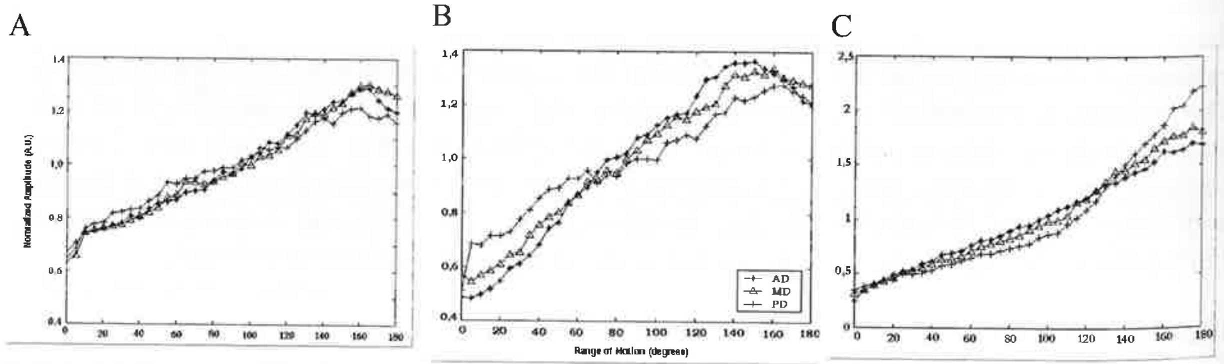

Figure 1. The normalized smooth of anterior (AD), midle (MD) and posterior (PD) portions of the deltoid muscle during the shoulder abduction movement performed to Ladies (L) Girls (G) and Impingement (I) groups.

STUDY OF THE MEDIAL SUPERIOR REGION OF THE ORBICULAR MUSCLE IN CHILDREN WITH ANGLE CLASS II, DIVISION 1 WITH PREDOMINANTLY NASAL OR ORAL RESPIRATORY MANNER

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INTRODUCTION

Mouth breathing has long been stated to cause serious effects on the development of craniofacial skeleton and occlusion of the teeth, on account of the imbalanced relationship between external and internal muscles forces of the stomatognathic system (SONG, H.; PAE, E.).

Therefore, for the precocious diagnosis, and to the correct elaboration of the treatment plan, it is very important to known if in individuals with Angle Class II, division 1 malocclusion described by (ANGLE, E. H.), the peribucal musculature could suffer environmental influences, such as oral breathing, to the point of changing its physiology.

PROPOSITION AND METHODS

The purpose in this electromyoghaphic study is the comparison of the medial superior region of the orbicular muscle of the mouth in two groups: G1 (respiratory manner predominantly nasal) and G2 (respiratory manner predominantly oral).

50 Brazilian children, aged six to nine, 25 males and 25 females, with Angle Class II, division 1 malocclusion were studied with a 16 channels electromyograph with amplification gain of 1000 times, with a 12 bits resolution analog/digital converter and a specific software for data acquisition and processing made by EMG System do Brasil Ltda, with sampling frequency of 1000 Hz per channel, that supplied the data (in RMS) for each discussed child. Surface electrodes with 10 millimeter diameter, placed as described by (BARIL & MOYERS) were used to the study, and the electromyographic evaluation was accomplished during the rest situation and for 21 mandibular-lip movements, that simulate the daily activities.

RESULTS AND DISCUSSION

Test T (student) and test F for electromyographic variables averages comparisons among the two groups G1 (nasal breathing) and G2(oral breathing).Var = variable (there were 21 variables, representing rest position; deglutition; facial expression and mastication).

These results pointed out that there was not statistically significant difference among G1 and G2 **for this evaluated sample**. Basically for 3 reasons: The variable nature of the sample (infantile behavior); the subjectivity with which the actions were executed (for example, it was not possible to standardize the intensity with the lips were compressed), and the use of p=95%. These are the reasons why we suggest more research for confirmation. It would be especially interesting to perform the same experiment with older children, to detect the age were a differentiation may occur (from the moment where other factors install, like buccal habits, impeding maxila correct development) resulting the labial incompetence.

Var.	RMS mean G2	RMS mean G1	SD G2	SD G1	test F variance	p variance	test T	p
REP	15,299	15,150	10,344	8,421	1,509	0,320	0,056	0,956
A	21,792	26,538	13,443	16,323	1,474	0,348	-1,122	0,267
B	36,788	44,859	19,777	41,506	4,405	0,001	-0,878	0,384
C	34,604	33,884	16,514	18,008	1,189	0,675	0,147	0,883
D	78,100	66,354	24,922	24,317	1,050	0,905	1,687	0,098
E	80,629	79,864	38,304	31,059	1,521	0,311	0,078	0,938
F	62,345	62,151	41,767	40,007	1,090	0,835	0,017	0,987
G	106,605	90,074	71,082	33,773	4,430	0,001	1,050	0,299
H	56,987	50,844	57,760	32,995	3,064	0,008	0,462	0,646
I	47,654	52,275	21,176	26,996	1,625	0,241	-0,673	0,504
J	64,531	50,493	73,918	24,431	9,154	0,000	0,902	0,372
L	65,125	55,138	63,363	22,030	8,272	0,000	0,744	0,460
M	46,158	57,677	19,540	46,223	5,596	0,000	-1,148	0,257
N	47,379	53,274	21,380	22,682	1,126	0,774	-0,946	0,349
O	43,022	37,628	37,115	22,434	2,737	0,017	0,622	0,537
P	28,709	33,349	17,950	22,522	1,574	0,273	-0,805	0,425
Q	32,973	36,212	20,868	22,398	1,152	0,732	-0,529	0,599
R	47,786	49,927	31,451	27,547	1,304	0,521	-0,256	0,799
S	59,708	64,171	51,402	37,099	1,920	0,117	-0,352	0,726
T	33,307	33,224	25,681	40,064	2,434	0,034	0,009	0,993
U	35,558	35,875	35,971	40,295	1,255	0,583	-0,029	0,977
V	59,740	38,669	47,731	32,979	2,095	0,076	1,816	0,076

CONCLUSION

The muscular dynamics profound knowledge enables the bases of a correct therapy. Thus, eletromyography constitutes a great exploration field with undeniable contributions not only for the Dentistry, but also for Physiotherapy.

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THE MORPHOLOGICAL STUDY OF THE MUSCLE SPINDLE OF THE EDL MUSCLE IN TWO NORMAL PIGS

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INTRODUCTION

At the last Congress of ISEK in Sapporo, Japan, we demonstrated that there were three types of muscle spindles in the extensor digitorum longus (EDL) muscle and suggested that there were several types of distribution patterns of the muscle spindles in the mouse EDL muscle. Although EDL muscle of the mouse is separated in four tendonated fibers, however, this of the pig EDL is separated in three tendonated peaces. In this study, we further examined muscle spindles in four peaces of one third tendonated in the pig EDL muscle. (Fig. 1)

METHODS

Four tendonated peaces of one third of EDL muscles from two normal pigs (6 months old, male) were used in this study. Each EDL muscle was cut into 15 sections under a stereomicroscope. Each was the largest peace in three tendoneated peaces. From each section, muscle spindles were isolated and were morphologically analyzed.

RESULTS

Muscle spindles in pig EDL muscles were divided into three types by comparing the ratio of diameter to longitudinal length of each spindle. We found standard (62%), long (30%) and short types (8%).

DISCUSSION

These results coincide well with our previous data shown in the mouse EDL muscle. Further studies will be necessary to compare the muscle spindles in EDL muscle in some other animals. On the other hand, the density of the muscle spindles in the pig EDL muscle was lower than the one of mouse EDL muscle. This would be because the pig EDL muscles do not move so dynamically as the mouse EDL muscles do. The pig has only three tendonated parts in the EDL muscle, while the mouse has four parts. In addition, the pig is kept under limited movements as a domestic animal, while the mouse is permitted to move freely as an experimental animal. These natural and circumstantial differences may explain the obtained differences in muscle spindles between the pig and the mouse EDL muscles.

CONCLUSION

According the ratio between maximum diameter of equatorial versus whole longitudinal length were calculated, and muscle spindles were arranged in three types, which were the standard, long, and short type as we previously reported at the last ISEK meeting Sapporo.

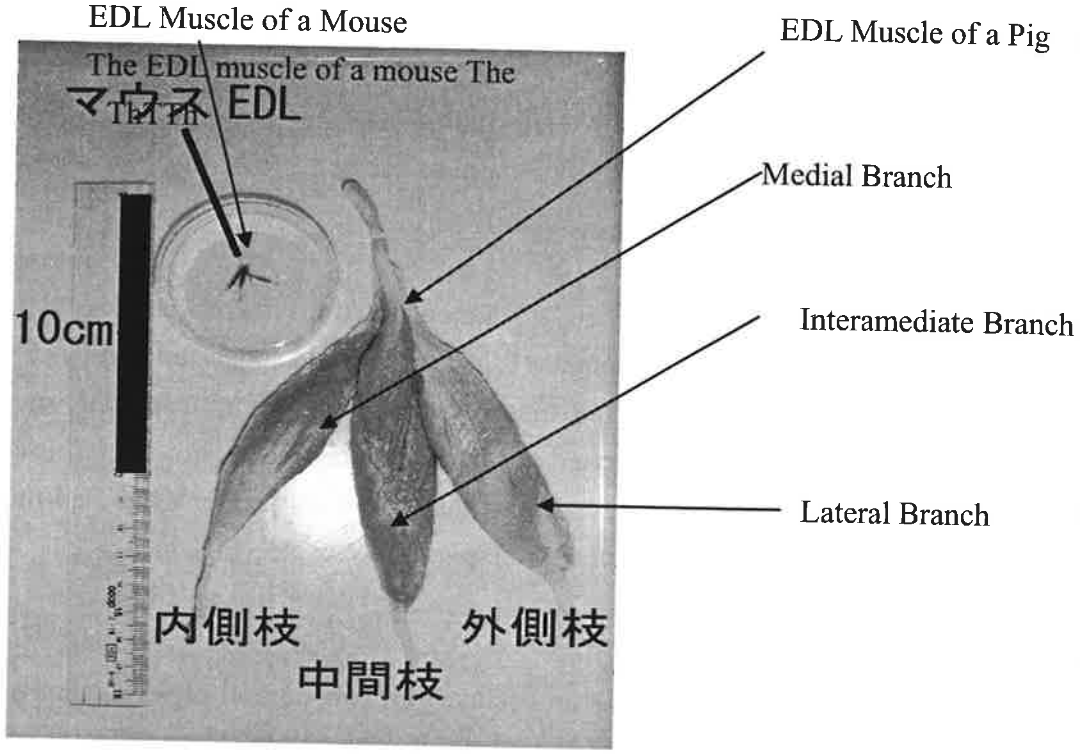

Fig. 1 EDL Muscles of a mouse and of a pig. Muscle Spindles were picked up from Intermedi ate Branch

Type of Muscle Spindles	Standard Type	Long Type	Short Type
Numbers of Muscle Spindles	48 (62%)	22 (30%)	6 (8%)
Whole Longitudinal Length (mm)	3.64 +/- 0.79	8.02 +/- 0.93	2.20 +/- 0.27
Longitudinal Length of Equat orial Area (mm)	1.04 +/- 0.20	1.29 +/- 0.15	0.48 +/- 0.10
Maximum Diameter (mm)	0.29 +/- 0.06	0.32 +/- 0.07	0.16 +/- 0.01

Table 1: Number of Muscle Spindles, Whole Longitudinal Length, Longitudinal Length of Eq uatiorial Area, and Maximum Diameter of Equatorial Area

EFFECT OF ONE-WEEK IMMOBILIZATION ON THE EXCITABILITY OF THE CENTRAL AND PERIPHERAL NERVOUS SYSTEMS

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INTRODUCTION

After muscle immobilization, activity is lowered not only in muscles but also in the nervous system. These functional disruptions in each anatomical area would systematically affect a motor control. Electromyography (EMG) amplitude is known to be increased during lower level voluntary contraction following immobilization. The present study was executed to investigate the mechanism, which is originated in the central nervous system, of disrupted skillful motor control during lower level force production with tracing a trajectory at a constant force after short-term immobilization.

METHODS

Four healthy male adults aged 21-25 years participated in the present study. Their right leg was immobilized from the distal thigh with a plaster cast for 7 days. Measurements were taken before and after the immobilization. Surface EMG was recorded from the soleus. First the maximal voluntary contraction (MVC) force of ankle plantar flexion was measured. Then the subject maintained a constant 5% force of the MVC (5% MVC) measured before the immobilization. The absolute value of the 5%MVC before immobilization was adopted also after immobilization. Root mean square (RMS) amplitude of the EMG during 5%MVC was calculated. While the subjects maintaining this low level contraction, either transcranial magnetic stimulation was applied to the left motor cortex to elicit motor evoked potentials (MEP), or the tibial nerve was electrically stimulated in the popliteal fossa to elicit H-reflex. H-reflex amplitude was measured as ratio of peak H-reflexes amplitude to peak M-wave amplitude. To explore an alteration of the muscle mechanical potential of force production, a twitch force during eliciting of supra-maximum M-wave was measured (F-twitch).

RESULTS

In the comparison between pre- and post-immobilization, MVC was lower in 3 out of 4 subjects after immobilization, but not in all the subjects. RMS during 5% MVC was higher in all the subjects after immobilization than that before immobilization (Fig. 1). Common changes in all the subjects found after immobilization also included the increase of MEP amplitude (Fig. 2). However, there was no common tendency to all the subjects on the H-reflex amplitude. The H-reflex amplitude after the immobilization was lower in 3 subjects and higher in 1 subject. F-twitch was lower in 2 subjects after immobilization, whereas higher in the 2 other subjects.

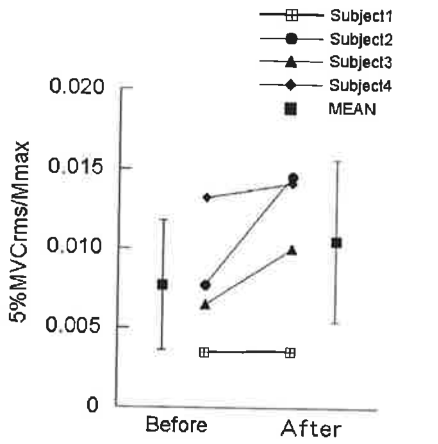

Fig.1 The comparison of 5%MVCrms between before and after immobilization.

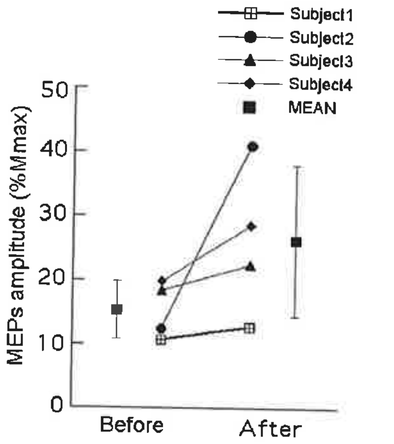

Fig.2 The comparison of MEP amplitude between before and after immobilization.

DISCUSSION

The parameters altered commonly in all the subjects after immobilization were RMS during 5% MVC and MEP amplitude. Increased RMS during 5% MVC could reflect the evidence that more high-threshold motor units were recruited in the post-immobilization stage during muscle contraction at the 5% level of MVC, although that level was absolute value and common to pre-/post-immobilization. Our interpretation is that more central command is needed to produce a constant submaximal force in compensation for decreased muscle fiber contractile tension [1] and/or decreased spinal motoneurone excitability in some cases. The result on the MEP amplitude suggests that the excitability originating from the primary motor cortex affect the increased 5% MVCrms as compensatory muscle activation. The absence of distinct common tendency on the alteration of H-reflex amplitude in all the subjects, did not contradict our previous study [2].

CONCLUSION

In conclusion, it was indicated in the present study that the cortico-spinal tract predominantly contributes to the compensatory muscle activity during voluntary force production at submaximal constant level after immobilization.

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RETROGRADE ANALYSIS OF LESION OF PERIPHERAL NERVES CAUSED BY WAR TRAUMA

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We present the research results of 239 patients with war injuries of the peripheral nervous system in the period from 1991-1999. All subjects were examined and followed electroneurographically in the Laboratory for EMNG Hospital Osijek; control examinations, McGill Questionnaire Test, Numerical Rating Scale and Verbal Description Scale were done in the years 1998 and 1999.

The injuries were caused by bullets in 107 (44,7%) subjects and at 132 (55,2%) by bomb-shells and mines. In 101 (42,5%) patients there were injuries of nerves on upper limbs, in 129 (53,9%) on lower limbs and in 9 (3,7%) there were injuries of other nerves (pr. facialis). In 70% of cases there were injuries of nerves, but also injuries of muscular mass, bigger blood vessels and vital organs; in 20% of subjects there were fractures, as well. Very clear time concordance of clinical recovery of injury nerve and electromyographic recovery was more than obvious – in 167 (69%) subjects the state was definitive and in 72 (30,1%) not.
The rehabilitation was implemented in all subjects; in 113 (47,2%) it was a combination of in/and outpatient clinic and in 126 (52,7%) only out-patients clinic.

The testing was done in 103 (143,1%) subjects, who were willing to take part in control examination. The results of the testing dealing with the scale of pain showed a high degree of pain – between 8 and 10 (taking into consideration that the values on the scale were from 0 to 10). In 90% subjects there was a high degree of pain at the moment of wounding, but also at the rehabilitation – the changes in the character of pain were present and there were: the unbearable pain, glowing, burning and blunt pain. The indicator of physical status is also typical – each unstable physical state leads to the feeling of having more and more intensive pain. There is a very interesting fact that the pain was, in the beginning, objectively stronger, but according to the statements of subjects there was a feeling of lower intensity because of very strong feeling of surviving. Later on, the intensity of pain was objectively decreased, but the feeling was subjectively stronger. In the case of affection of peripheral nerves there was, most commonly, the causalgiae syndrome which oscillates in intensitiy depending on the location; it was provoked by touch, moving of a limb, but also by emotions.

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PARAMETERS OF POSTURAL SWAY FOR YOUNG AND ELDERLY PEOPLE

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INTRODUCTION

In the last time, many authors have directed their attention to the postural control system. This system consists of three sensory modalities: visual, vestibular and somatosensory. The sway of the centre of pressure (COP) is the result of an activity of the postural control system. The force platform has been used to measure the COP sway. The outcome, obtained in this way, has the form of two time series. The plot of the data on the plane is called stabillogram. Collins and De Luca [1] showed that postural sway can be considered as the result of a stochastic mechanism. To model the COP trajectory they analysed mean squared COP displacement vs time interval. Time-scales can be recognised from the stabillogram-diffusion plot.

In many investigations the Newton equation with stochastic force was considered as a model of the body dynamics. Chow et al. [2] proposed a pinned polymer model. Alonso-Sanchez and Hochberg [3] extended this model, adding non-linear element and renormalisation group analysis. Frank et al. [4] modelled the postural sway by means of multivariate Ornstein-Uhlenbeck processes. The outline of models of the postural sway is presented by Rosenblum et al. [5]

METHODS

Langevin equation is adopt in this paper as the model of the postural sway, The solution of this equation permits to find the matrix of the mean squared displacement as a function of time

$$\mathbf{C}(\tau) = 2\mathbf{D}\left(\tau + \gamma^{-1}\left(e^{-\gamma\,\tau} - 1\right)\right), \tag{1}$$

where **D** is the diffusion matrix, and γ is the friction coefficient. C(τ) is a 2×2 matrix and, in general, this matrix has a nondiagonal form. For $\tau \ll \gamma^{-1}$, the mean squared displacement is a square function of time interval τ. For $\tau \gg \gamma^{-1}$, this displacement is a linear functions of that interval. It permits to distinguish Fokker-Planck time-scale and diffusive time-scale respectively.

Postural control system of twenty young healthy people and twenty elderly people with various illnesses was studied. In the experiment the force platform, produced by PROMED – Janusz Olton, Poland, was used. COP displacements in anteroposterior direction, and in mediolateral direction were measured. The data were collected for open eyes and next for closed eyes. Subjects, during trials, stood on the platform in upright stance with joined heels, with an angle between the feet of 30° and loosely lowered arms. Stabilograms were registered for a period 32 sec. and were sampled with 32 Hz frequency. The experiment was carried out with the agreement of the Bioethical Commission at The Ludwik Rydygier Medical University in Bydgoszcz.

RESULTS

From experimental data mean squared displacements were obtained and they were well characterised by theoretical curves up to 0.5 sec. In most cases the reciprocal of the friction coefficient γ^{1} was of the order 0.1 sec. Below this value, mean squareed COP displacement is approximately a square function of the interval τ and, in this way, the Fokker-Planck time-

scale is specified. The diffusion time-scale is not fully attained for the investigated populations.

The diffusion matrix D and the friction coefficient γ describe the state of the postural control system. The state of the postural control system was specified by diagonal and off-diagonal elements of the diffusion matrix when the coordinate system defined by anteroposterior and mediolateral directions was used. In every case, for simplicity, the diagonal form of the matrix C(τ) was calculated and two diagonal elements D_u and D_v of the diffusion matrix are used as a measure of the state of the postural control system. In Table 1 the results of the investigations are presented. The diffusion coefficient for the population of young subjects is almost three times greater, when they have closed eyes, in comparison to open eyes.

Two populations in the group of elderly subjects were distinguished. For the first population (15 subjects), the ratio D_{ueo}/D_{uec} and D_{veo}/D_{vec}, where indicator (eo) denotes open eyes and (ec) is for closed eyes, is comparable to the ratio obtained for young people. The value of this ratio, for the second population (5 subjects), differs from the results obtained for young people.

For the population of elderly subjects diffusion coefficients D_u and D_v, are significantly greater from the coeficients obtained for young people. For the second population of elderly subjects, the friction coefficient under eyes-closed conditions is about three times greater than in all other cases.

DISCUSSION

We have shown that the state of the postural control system can be specified with the use of the diffusion coefficient and the friction coefficient. The measures permit to distinguish different states of the system. The results, shown here, suggest that this approach to the analysis of human stabilogram can be very useful in the diagnostics of various illnesses.

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coef.	Type	Young people	Elderly people	
			First population	Second population
D_u (mm^2 s^{-1})	eo	3.52 ± 0.67	13.12 ± 2.74	40.16 ± 16.71
	ec	8.84 ± 1.67	43.20 ± 21.34	22.77 ± 10.07
D_v (mm^2 s^{-1})	eo	1.64 ± 0.26	5.06 ± 1.21	23.33 ± 11.28
	ec	4.16 ± 1.09	11.19 ± 4.68	11.61 ± 7.37
γ (s^{-1})	eo	8.05 ± 0.46	10.32 ± 3.65	6.55 ± 1.04
	ec	8.78 ± 0.70	10.13 ± 1.81	15.91 ± 1.46

Table 1. Group means and standard deviations for D_u, D_v and γ coefficients, for population of young subjects and for two populations of elderly subjects, for open-eyes (eo) and closed-eyes (ec) conditions.

THE EFFECTS OF EXERCISE ON BODY COMPOSITION, POSTURE AND PREVENTION OF LOW BACK PAIN DURING PREGNANCY

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INTRODUCTION

During pregnancy, body undergoes tremendous changes hormonally and anatomically which may cause different problems such as low back pain. Studies reveal that the incidence of pregnancy related low back pain is approximately between 50% to 70% (Bushnell 1949, Heckman 1994). Women who have a previous history of back pain are at more risk than the others. Ostgard and Andersson (1991) found increased incidence of back pain in younger women, whereas Mantle et al (1977) found an increased incidence of back pain in older women. There seems to be no relationship between height, weight, weight gain during pregnancy or birth weight and pregnancy related back pain (Mantle 1977). As the pregnancy progresses, weight and size of the uterus together with the mother's weight increase. Therefore, the body center of mass is shifted anteriorly which results in an increased torque at the lumbar motion segments. On the other hand, the increased level of relaxin during the last months of the pregnancy increases the laxity of the joints. Therefore, during the upright standing posture the workload of the lower back muscles is significantly increased which may facilitate the occurrence of the low back pain (Jacobson 1991). The benefits of exercise for pregnancy have been addressed with many authors and are reviewed by Ireland and Ott (2000). Most of the available studies refer to the pregnancy of athletic women. There are only a limited number of relevant studies related to the effects of exercise on pregnancy of the women without regular exercise. However, the conclusions on the impact of exercise on the mother, the fetus and the course of pregnancy are contradictory. Hall and Kaufmann (1987) evaluated the effects of aerobic and strengthening exercise on pregnancy outcomes and found more favorable outcomes in the women who continued to exercise during their pregnancy. But, Dale et al (1982) in a similar study indicated no significant differences between women who continued their training and women who did not continue their exercise during pregnancy. Although various exercise guidelines and recommendations are available, they are usually conservative and are frequently based on controversial opinion (ACOG 1994). Therefore, there may be some uncertainty about the safety and advantages of exercise during pregnancy among most of the physicians and pregnant women. This study was conducted to examine the effects of exercise during pregnancy on back pain, body composition and posture in women without having any regular exercise before their pregnancy.

METHODS

Subjects and tasks: Forty pregnant women were studied. They were between the ages of 21 and 28 years and collected from two clinics after an initial meeting with the first author. The subjects were randomly subdivided into a control and an experimental group equally. Experimental group participated in a moderate aerobic exercise program during pregnancy, three times per week and 30 to 45 minuets on each session. Control group did not have any exercise during their pregnancy. Subjects in both groups did not have any history of previous pregnancy, back pain, surgery of any kind or any known disease which could affect the results of this study. All subjects were volunteers. Before participation, a recommendation from their physician was required and the subject and her husband gave their written informed consent. The study was approved by the medical ethical committee of the Gynecology hospital. Those who missed any measurement appointment, were absent in three consequent or total of four training sessions, or did not have a natural delivery (had cesarean) were

excluded from statistical analysis. Therefore, the data of 15 subject in experimental group and 12 subjects in control group were included in this study.

Procedure: Height, weight, low back pain (LBP), lumbar lordosis and body fat percent (BFP) based on the skin folds of triceps, abdomen, iliac crest and thigh were measured during pregnancy at 13, 25 and 36 weeks gestation. Skin fold were measured with a caliper. Quebec back pain disability scale was used to quantify LBP (Fritz and Irrgang 2001). This scale quantifies the back pain between zero and 100. Using flexible ruler the shape of the spine in sagittal plane was simulated and transferred into a page. Then the obtained angle between two consequent right angle line drown at L1 and S1 area determined the amplitude of lumbar lordosis. The data were then analyzed using multivariate analysis of variance (MANOVA - Repeated Measure). Differences were considered significant if p value was below 0.05 (α=0.05).

RESULTS

The overall incidence of LBP among subjects was 80% and began during the second trimester with the mean of 29.7 (±11.2) and 40 (±21.5) which were significantly increased at 36 weeks gestation to 47.5 (±15.2) and 50.6 (±19.7) for experimental and control groups respectively. These differences between groups were not significant due to the high standard deviations. The weight gain during pregnancy was very similar in both groups (average = $11.2_{kg} \pm 2.2_{kg}$). Height, lumbar lordosis, BFP of both groups in different months of pregnancy are shown in table 1. In general, pregnancy resulted in an increase of the BFP. This increase was distributed differently on experimental and control groups, 11.8% & 20.9% respectively. Lumbar lordosis did not changed significantly in any group.

CONCLUSION

Exercise during pregnancy was beneficial for pregnancy in different ways. The presence of serious LBP in exercised group was delayed having a different pattern from that of the control group. Exercise decreased the BFP leading to a better body composition without affecting the necessary weight gain during pregnancy.

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ANTICIPATORY ACTION OF POSTURAL MUSCLE WHILE ARM FLEXION IN SUBJECT WITH DIFFERENT STANDING POSITION

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INTRODUCTION

Many previous studies have shown that the activation onset in the postural muscles of the legs and trunk that control standing postures precedes that in the focal muscles that rapidly move the arm [1]. The preceding activations of the postural muscles are adjusted by the program selected in advance in order to moderate the effect of disturbances of posture and equilibrium caused by the arm movement [3,5,9]. Many studies have suggested that the two main factors affecting the postural set are the state of equilibrium [2,3,5] and the initial standing position just before arm movement [6,7,8]. The steadiness of the standing posture varies according to the center of foot pressure (CFP) in the anteroposterior direction, and thus, the equilibrium has presumably a close correlation with the initial standing position. We investigated changes in activation timing and magnitude of the postural muscles associated with initial standing positions when subjects with various quiet standing positions flexed their arms bilaterally. The working hypotheses were as follows: (1) The changing pattern in activation of the postural muscles according to initial CFP positions will differ among subjects with various quiet standing positions. (2) No differences in activation of the postural muscles will be found between initial CFP positions located near quiet standing.

METHODS

The subjects were divided into 3 groups of 10, we will call backward, middle, and forward groups, in which the position of the center of foot pressure (CFP) in quiet standing posture was less than 38%, from 38% to 48%, and more than 48%, respectively.

All measurements were performed while subjects were standing on a force platform (WAMI, WA1001) composed of three load-cells. The platform was used to record CFP position in the anteroposterior direction. The CFP position was defined as the distance of CFP from the heel relative to total foot length. Subjects maintained standing postures at various CFP positions in the anteroposterior direction, and then started bilateral arm movement at their own pace. They were instructed to move their arms at maximum speed and to stop them at the frontal horizontal level. The CFP position was randomly changed in 10% increments from 20% to 80%. Electromyograms were recorded using bipolar surface electrodes (spaced 3 cm apart) placed over the following muscles: the soleus (Sol), tibialis anterior (TA), biceps femoris (BF), rectus femoris (RF), erector spinae (ES) at the level of the iliac crest, rectus abdominis (RA) at the level of the navel, and anterior deltoid (AD) on the left side. The integrated EMG was computed for a period of -300 ms to -150 ms with respect to the AD activation and that value was adopted as background activity. The activation magnitude of the postural muscles was evaluated by the integration value during the first 50 ms in EMG burst. Time differences in the first EMG burst starts of the postural muscles to AD were analyzed as the start time by visual inspection on a computer screen.

RESULTS

The changing pattern of background activity accompanying the initial CFP position was different among subject-groups in the rectus femoris and biceps femoris. The CFP position generating the alternation in the background activity between those muscles was located more forward in the following order: backward group < middle group < forward group. In the trunk and crus, the changing patterns of background activity of the front and rear muscles showed no significant difference among subject-groups.

Activation magnitude of the biceps femoris and erector spinae was not different among every initial CFP position.

Fig. 1 shows the start time in the biceps femoris and erector spinae. At more backward CFP positions, the biceps femoris and erector spinae did not begin activating before the AD, while at more forward CFP positions these muscles began activating earlier than the AD. In only the biceps femoris, the preceding action to the AD was clearly observed at more forward CFP positions in the order of the forward, middle and backward groups. Between initial CFP positions adjacent to quiet standing posture, no significant difference was observed in the preceding activation time of the biceps femoris. This phenomenon was not observed in the erector spinae.

DISCUSSIONS

Sensory information according to muscle activity in the femur is thought to be the essential information source for perception in the relative positional change from quiet standing. The activation of the erector spinae may be adjusted by a mechanism different from the biceps femoris. It has been considered that muscle sensory information from the trunk, in every subject, correlated with sensory information from the neck muscles and the vestibular organs, and perhaps took part in the perception of the gravity direction [4]. Massion [9] divided postural adjustment into the balance control over the entire body and the positional control of specific body parts in order to provide a reference axis for action.

From the start time in the biceps femoris, it is concluded that disturbance of body balance caused by arm movement is anticipated in relation to the CFP position in quiet standing posture, and based on this anticipation the action timing of the postural muscles was adjusted.

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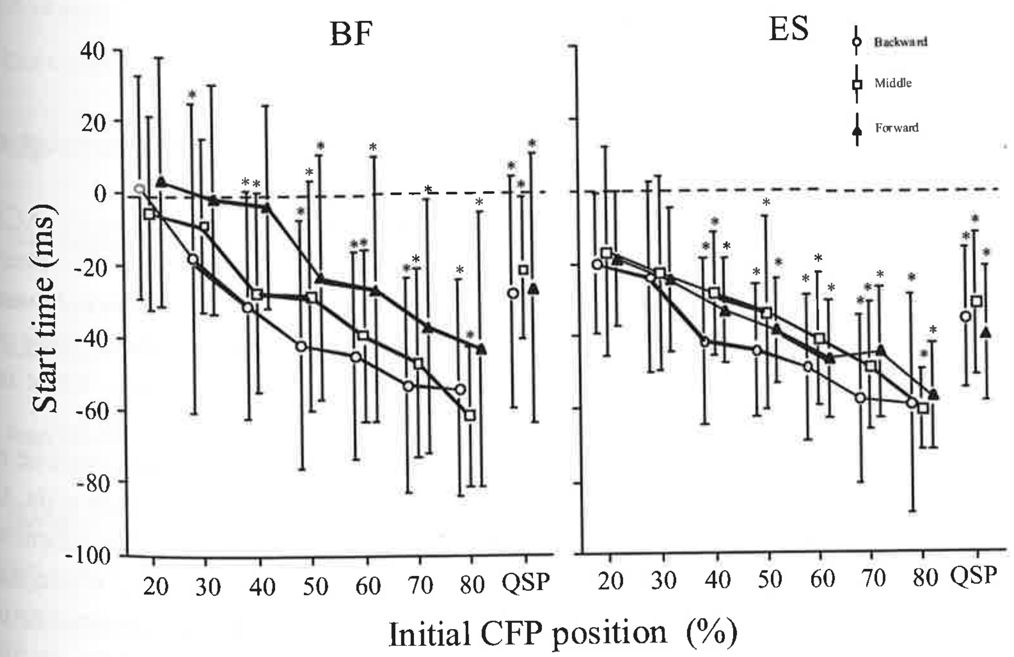

Fig. 1 The start time in the biceps femoris and erector spinae.

DEVELOPMENT OF THE ADAPTABILITY OF ANTICIPATORY POSTURAL CONTROL WHILE FLOOR OSCILLATION IN CHILDREN

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INTRODUCTION

The development of postural control has been examined by using various postural perturbations, such as platform translation [5], moving room [1], release of trunk support [4]. We have obtained some evidence that the posture is controlled anticipatively while imposing the horizontal floor oscillation in upright stance, and we have been able to evaluate the adaptability of anticipatory postural control in a short time [2]. The body is a multi layer structure connected by a lot of joint, and it will be important to examine the postural control from the sway characteristic of various body portions. Therefore, we analyzed the body sway characteristic while the floor oscillation, and investigated on the development of the adaptability of anticipatory postural control.

METHODS

Subjects were 52 healthy younger children (5 and 6 years old) and 38 adults. The subjects maintained a standing posture with their eyes closed on the oscillation table with force platform. The table was sinusoidaly oscillated with 0.5 Hz frequency and 2.5 cm amplitude in anteroposterior direction. The oscillation for 60 seconds was repeated five times. Data analysis was performed for first and fifth trial. The center of foot pressure (CFP) was detected by a force platform and, and we calculated the mean speed of the CFP fluctuation. To measure the sway of various body portions, the position sensors were placed over the external malleolus, knee, trochanter major and acromion. Amplitude spectrums of their displacement in the anteroposterior direction were analyzed. In the frequency component of floor oscillation, an amplitude change ratio (ACR) and a phase difference of displacement between adjacent body portions were calculated, in which the adaptive change were analyzed.

RESULTS

CFP mean speed in children was larger than that in adults. The adaptive improvement of postural control was observed for 6 years in boys and for 5 years in girls.

In adults, body sway decreased in every segment with repeating the trial, and every ACR showed negative value in 5th trial. The adaptive decrement of ACR was observed for every segment in men and women; particularly, the largest decrease was observed in the trunk. No sex difference was shown in the phase characteristic. The adaptive increment of phase delay of higher portions relative to lower portions was observed for all position in women, while no adaptive increment of the phase delay between trochanter-acromion was shown in men.

ACRs in children were definitely larger than those in adults, and the value was not negative in 5th trial. In 5 years, the ACR of the crus segment tended to decrease adaptively in only girls. In 6 years, the sex differences were shown in the segment with the adaptive change. The adaptive decrement of the ACR was observed for the trunk and thigh segments in girls, while the decrement was observed for only the trunk segment in boys. The phase delay was smaller in children than in adults. The phase delay in the children was smaller than that in adults. In 5th trial, significant differences between children and adults were observed among the adjacent portions except for trochanter-shoulder. The adaptive increment of the phase delay was observed between knee-trochanter for only 6 years girls.

DISCUSSION

In our previous study with regard to adaptive development of postural control, the adaptive decrement in CFP mean speed was 1 year earlier in girls than in boys [3], The result of the present study accords with that result. From the result of ACR, it is conceivable that the ability to decrease the fluctuation in each body segment develops 1 year earlier in girls than in boys. The developmental change happened markedly in the trunk. The phase difference shows possibly to be smaller in children than in adult, particularly in the leg, and that the postural control in female makes efficient use of the higher flexibility.

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Table. Mean and Standard deviation of CFP mean speed, ACR and Phase delay

		Sex											
		Male						Female					
		5 years		6 years		Adult		5 years		6 years		Adult	
		1st	5th	1st	5th	1st	5th	1st	5th	1st	5th	1st	5th
CFP mean speed (mm/s)	Mean	180.4	172.0	172.6	146.3 *	90.5	52.1 *	185.9	161.4 *	163.5	132.0 *	81.4	56.4 *
	(SD)	(32.5)	(37.6)	(27.8)	(38.6)	(41.7)	(19.8)	(40.9)	(31.3)	(36.6)	(28.0)	(23.5)	(15.5)
ACR (degree)													
Crus segment	Mean	0.98	1.14	1.12	1.05	0.33	-0.05 *	1.21	0.84	1.23	1.04	0.18	-0.10 *
	(SD)	(0.85)	(0.39)	(0.70)	(0.51)	(0.39)	(0.41)	(0.76)	(0.39)	(0.63)	(0.34)	(0.39)	(0.56)
Thigh segment	Mean	1.05	0.95	0.77	0.58	0.06	-0.22 *	0.93	0.88	0.99	0.48 *	0.15	-0.15 *
	(SD)	(0.52)	(0.90)	(0.94)	(0.67)	(0.39)	(0.46)	(0.77)	(0.48)	(0.66)	(0.51)	(0.40)	(0.42)
Trunk segment	Mean	1.01	0.46	1.08	0.48 *	-0.10	-0.41 *	1.14	1.04	1.11	0.36 *	-0.16	-0.39 *
	(SD)	(0.92)	(0.75)	(1.06)	(0.65)	(0.56)	(0.34	(0.69)	(0.65)	(0.65)	(0.85)	(0.37)	(0.31)
Phase delay (degree)													
Ankle-knee	Mean	1.41	0.40	1.43	2.70	3.25	8.92 *	2.97	1.42	1.80	3.21	4.86	8.65 *
	(SD)	(6.34)	(6.74)	(6.11)	(5.42)	(4.47)	(2.20)	(4.01)	(5.66)	(5.95)	(6.89)	(3.68)	(3.07)
Knee-Trochanter	Mean	8.29	6.79	6.57	6.00	7.40	13.77 *	5.93	6.26	4.03	6.44 *	7.79	12.71 *
	(SD)	(5.57)	(3.06)	(4.80)	(4.60)	(4.91)	(6.47)	(4.25)	(4.62)	(4.36)	(3.25)	(4.29)	(6.55)
Trochanter-Shoulder	Mean	21.35	24.40	24.85	21.36	23.60	24.96	17.18	16.63	20.59	20.84	12.70	17.52 *
	(SD)	(7.35)	(8.12)	(7.81)	(8.78)	(22.65)	(15.69)	(7.98)	(5.74)	(4.46)	(5.04)	(7.07)	(10.07)

NOTE CFP, Center of foot pressure; ACR, Amplitude change ratio

* denote a significant difference between 1st and 5th trial (*P*< *.05*).

SURFACE EMG SUPPORTS THE USE OF AN ERGONOMIC MICROSCOPE WORKSTATION

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INTRODUCTION

Numerous studies in a variety of occupational groups report the prevalence of work-related musculoskeletal complaints and discuss the pathophysiological basis of cumulative trauma disorders. We found, however, only little information about work-related physical disorders of professionals operating conventional microscopes for prolonged time. This is in striking contrast to the large number of professional microscope users who complain about headache, neck, shoulder and back pain, and eye strain. One of the authors (A.K.) suffered from progressive tension neck syndrome with subsequent tension headache over a period of seven years. In an endeavor to improve his condition, he designed a prototype ergonomic microscope workstation according to his personal ideas and experience. In this study we set out to record systematically surface EMG activity in order to document that operating an ergonomic versus a conventional microscope results in markedly reduced muscle activity and less muscle strain, and thus, may prevent the development of cumulative musculoskeletal disorders during prolonged microscope-related work.

METHODS

Twelve healthy volunteers (8 males, 4 females; age 24 to 50 years, mean 35 years; body height 160 to 192 cm) served as subjects. We recorded muscle activity using a standard 8-channel electrodiagnostic system (Nicolet Viking IV) while sitting on a standard office chair and operating a standard microscope ("standard condition"), and on a special ergonomic chair with support for the lower back operating a prototype ergonomic microscope ("ergonomic condition"). Surface EMG was recorded unilaterally over the paramedian upper and lower neck (third and seventh vertebral level), mid and lower back (eighth thoracic and third lumbar level), biceps brachii, brachioradialis, flexor carpi radialis, extensor digitorum communis, trapezius (upper part), and sternocleidomastoid muscles, throughout a 40 minutes recording period (20 minutes for each microscope workstation). There were no rest periods other than the time needed to take position at the other microscope.

Seven subjects were studied in the standard condition first, five subjects started with the ergonomic condition. Data were stored on floppy disk and analyzed off-line. Artifact-free windows of ten seconds duration were selected and the area under the curve was measured. Mean and standard deviations were calculated for each muscle in each condition. To analyze the effect of prolonged sitting and operating the microscope, we compared the first two with the last two recordings in each condition.

RESULTS

Sitting on a standard office chair and operating a conventional microscope resulted in extensive EMG activity in all recorded muscles. Group average EMG activity was largest in finger extensors, trapezius and mid-thoracic paraspinal muscles, and subsequently less in lumbar and upper cervical paraspinal muscles, biceps brachii, brachioradialis, sternocleidomastoid, lower cervical paraspinal, and wrist flexor muscles. In the ergonomic condition, a profound reduction of mean EMG activity was noted in all recorded muscles (Fig. 1). Across all subjects, most attenuation was observed in trapezius, biceps brachii, upper

and lower cervical paraspinal muscles. EMG activity was even largely reduced in forearm muscles involved in operating knobs and handles of the mechanical stage, due to the support of the forearms on slanting wings. All twelve subjects showed statistically significant attenuation of EMG activity in upper and lower cervical paraspinal and trapezius muscles, ten subjects in wrist flexor, nine subjects in sternocleidomastoid, biceps brachii, and brachioradialis, eight subjects in mid-thoracic, and seven subjects in lumbar paraspinal and finger extensor muscles. Although there were different individual patterns of the most strained muscles, there was no correlation with individual age, body height or microscoping experience. There was also no difference whether subjects were first enrolled in the standard or the ergonomic condition. Individual muscles displaying most extensive activity while operating a conventional microscope were apt to manifest the largest EMG signal reduction, hence the highest degree of improvement, when operating the ergonomic microscope workstation. With sustained work, EMG activity decreased further while operating the ergonomic microscope ($p < 0.001$ in all muscles combined; $p < 0.05$ in upper cervical paraspinal and trapezius muscles, each, tested with analysis of variance for repeated measures). In contrast, EMG activity did not significantly change over time in the "non-physiological" posture in front of the conventional workstation when combining all muscles, but rather tended to increase in sternocleidomastoid, trapezius, thoracic and lumbar paraspinal, and wrist flexor muscles.

DISCUSSION

To our knowledge, this is the first neurophysiological study which demonstrates in a larger number of subjects the advantage of an ergonomic versus a conventional microscope workstation concerning musculoskeletal symptoms. We document that a more relaxed and "physiological" posture while operating an ergonomic microscope results in much lower levels of continuous muscle activity as compared with a standard microscope. The microscope workstation presented here was not specially designed for this study, but has rather been in constant use by one of the authors (A.K.) since several years. Although our near-to-ergonomic microscope workstation did not sufficiently accommodate all individual anthropometric measures in different subjects, EMG activity was substantially reduced in all muscles, in particular in trapezius, biceps brachii, upper cervical and lumbar paraspinal muscles, as compared to the standard microscope. This reduction of EMG activity should in theory prevent the development of cumulative musculoskeletal disorders during prolonged microscope-related work. In fact, upper cervical syndrome and tension headache have completely resolved in A.K. since exclusively operating the ergonomic microscope. Noteworthy, despite differences in the individual pattern of muscle activity, the most strained muscles were usually those which experienced the largest benefit in the ergonomic condition. Finally, we confirm the feasibility of surface EMG recordings in neurophysiological research issues related to occupational medicine.

ACKNOWLEDGMENT

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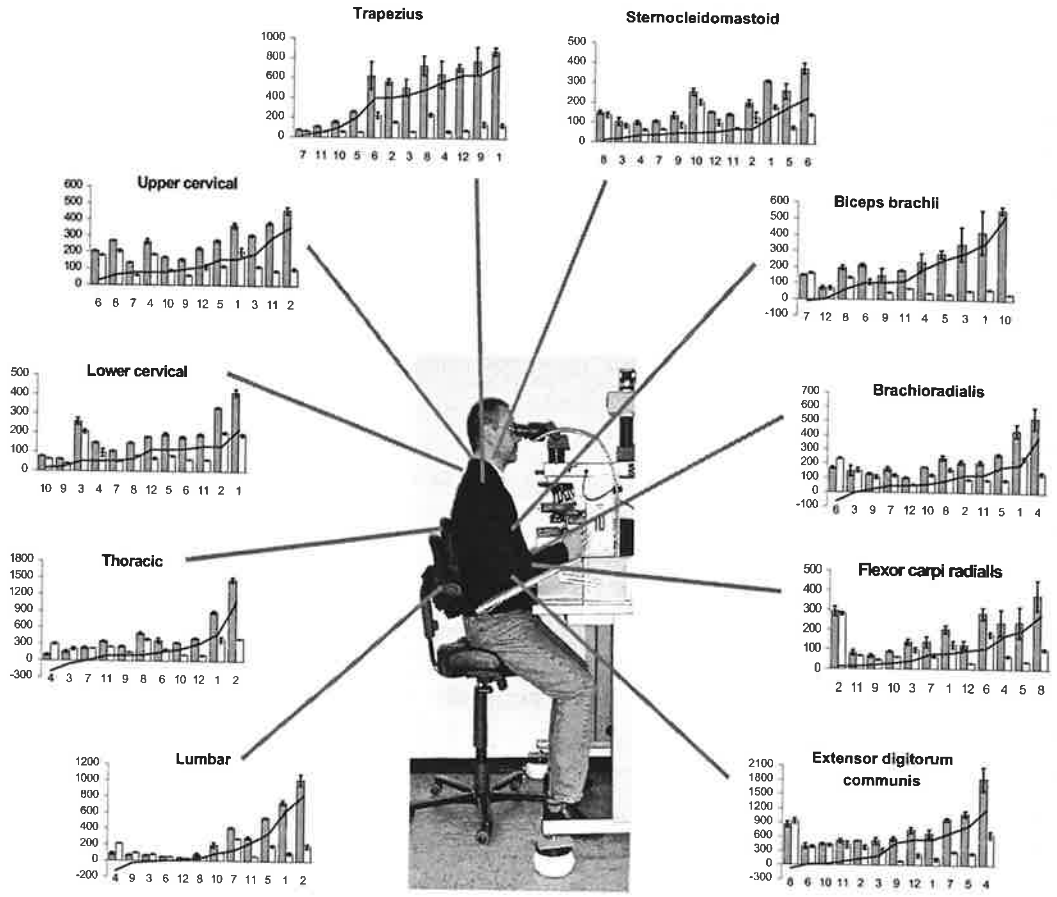

FIGURE
Photograph showing one subject working in front of the prototype near-to-ergonomic microscope workstation. Surface electromyographic area [mVms] recorded in individual muscles of twelve subjects while operating a standard microscope (condition 1 - gray columns) and a prototype ergonomic microscope (condition 2 - white columns). The numbers on the x-axis represent individual subjects. For each muscle, subjects are arranged according to incremental improvement (black line) from standard to ergonomic condition. Note the larger "gain" in those muscles with larger electromyographic activity in the standard condition.

MUSCULAR SYNERGY BETWEEN POSTURAL AND FOCAL MUSCLES DURING FATIGUE IN ISOMETRIC PUSH EFFORTS

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INTRODUCTION
When a maximal push effort is exerted until exhaustion, a decrease in reaction forces and in the displacement of the centre of pressure is associated with a decrease in maximum force (Le Bozec et al., 1999). Since muscular fatigue alters muscle contractile efficacy (Bigland-Ritchie and Woods, 1984) and also proprioceptive information and cortical control (Gandevia, 1998), it is likely to affect movement and postural control.
The purpose of this study was to explore whether the muscular synergy was the same during fatigue as without fatigue, where postural EMG has been shown to precede focal EMG following a well-defined chronology (Le Bozec et al., 2001).

METHODS
Subjects were seated; the contact between the seat and the thighs covered 100% of the ischio-femoral length. Subjects were instructed to exert bilateral isometric pushes with their upper limbs on a dynamometric bar. They performed pushes from zero up to 70% of their maximal voluntary force (70% MVC) as rapidly as possible, and to maintain this level until exhaustion. In addition to force measurement, electromyographic signals (EMGs) were recorded using bipolar surface electrodes from nine muscles : a) the shoulder and the elbow muscles, that is focal muscles (the primum movens : Serratus Anterior, SA and Deltoideus Anterior, DA; Pectoralis Major, PM; Trapezius Superior, TS, in addition to Biceps Brachii, BB and Triceps Brachii, TB); b) muscles crossing the lower trunk and pelvis, that is postural muscles (Rectus Femoris, RF; Rectus Abdominis, RA; Erectores Spinae, ES). Mean power frequency (MPF) and root mean square (RMS) were calculated over two-second intervals. Six right-handed subjects participated in the experiments.

RESULTS
The results showed that the push force started to decrease 3 minutes after the onset of the 70% MVC effort (Figure 1). The push force reached 45-50% MVC when exhaustion occurred, about 12 minutes later. The muscular synergy remained the same during the 12 minute fatigue period, as in the absence of fatigue, in both focal and postural muscles. But, fatigue did not induce the same effects in these muscles (table 1). During the fatigue process, MPF and RMS decreased in the focal muscles. The postural muscles did not yield identical features. While MPF decreased in RF and RA, as in the focal muscles, there was no difference in ES. Also, RMS was increased in RF and RA, contrary to ES for which it diminished. In other words, the modifications in the lower trunk and pelvis flexors EMG parameters were different from those displayed by the extensors.

DISCUSSION
The results showed that: 1) the muscular synergy included the same postural and focal muscles during the fatigue process; 2) the EMG patterns were modified in focal as compared to postural muscles during the fatigue process : fatigue does not intervene in an identical way in focal or postural muscles.

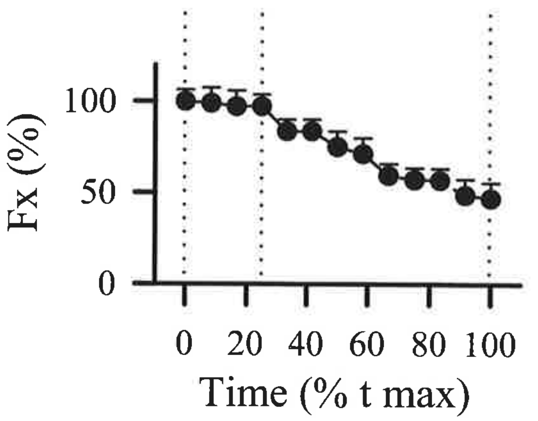

Figure 1 : Variations of force (Fx) obtained during a push effort exerted until exhaustion.

	MPF	RMS
Focal muscles	DOWN	DOWN
RF, RA	DOWN	UP
ES	EQUAL	DOWN

Table 1 : Variations of the MPF and the RMS obtained during a push effort exerted until exhaustion, during the decrease of Fx.
Focal muscles : SA, DA, PM, TS, BB, TB.
Postural muscles : RF, RA, ES.

Indeed, fatigue occurs in all the upper limb and shoulder muscles, which is shown by MPF decrease towards low frequencies and RMS decrease. However for postural muscles, it is likely to depend on the level of force invested in the effort. If it is superior to 15-20 % MVF, taken as the fatigue threshold (Monod and Scherrer, 1965), no fatigue effect would be expected and the RMS decreases as the force diminishes. This is the case for the ES. When it is superior to the threshold, fatigue occurs and RMS increases, so as to adapt to the reduction of muscle contractile efficacy.

CONCLUSION
It appears that, while the muscular synergy is not different during fatigue, postural muscle adaptation contributes to create optimal conditions to reduce the peripheral effects of the fatigue process in relation to their fatigue threshold.

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CHANGES OF POSTURAL ADJUSTMENT ASSOCIATED WITH BILATERAL ARM FLEXION DURING ACHILLES' TENDON VIBRATION

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INTRODUCTION
Many previous studies have shown that the activation onset in the postural muscles of the legs and trunk that control standing postures precedes that in the focal muscles that rapidly move the arm [1]. The activation timing of postural muscle is probably affected by the perception of the initial standing position [2]. Eklund [3] reported that the vibration on the Achilles' tendon generated the reflex action in the gastrocnemius, and then the body leaned backward. It was found that when the leaning was intercepted, the subject illusorily perceived as the body leaned forward [4, 5]. This study examines what kind of change in the anticipatory postural control is caused by the illusionary perception of standing position, which was generated by the vibration on the Achilles' tendon. The working hypotheses were as follows: (1) the activation of the postural muscles will change based on the illusionary perception of standing position, (2) the activation will adaptively change based on the knowledge of control result.

METHODS
Subjects (n=12) performed rapid arm flexion bilaterally. The arm flexion was performed 5 times by the following three conditions: the quiet standing condition, the condition that vibrate on the Achilles' tendon in 100 Hz frequency with the trunk fixed by a stopper in quiet standing, the condition that reproduced a standing position perceived in the vibration condition. The standing position was represented by the pressure center of the foot, which was shown by the relative distance (%) from the heel to the length of the foot. Electromyograms were recorded using bipolar surface electrodes (spaced 2 cm apart) placed over the following muscles: the anterior deltoid, rectus abdominis at the level of the navel, erector spinae at the level of the iliac crest, rectus femoris, long head of biceps femoris, tibialis anterior, and soleus on the left side. The mean amplitude of EMG was computed for a period of -2300 ms to -300 ms with respect to the action onset of the anterior deltoid and that value was adopted as background activity. The activation magnitude of the postural muscles was evaluated by the integration value during the first 50 ms in EMG burst. Time differences in the first EMG burst starts of the postural muscles to AD were analyzed as the start time by visual inspection on a computer screen.

RESULTS AND DISCUSSION

The reproduced positions were located forward by about 20% of the foot length compared with the quiet standing position, and they showed no significant differences among all the five trials. It means that the illusion that the body leans forward is generated in every trial with the vibration stimulation. In the first trial of arm flexion under the vibration condition, the biceps femoris started to activate before about 40 ms of the anterior deltoid. The preceding time was same as that in the reproduction condition. Such a change in the muscle activation timing by the vibration stimulation was not observed in the erector spinae. In both muscles of the biceps femoris and erector spinae, the mean amplitude of electromyogram for the first 50 ms after the start of activation showed no differences among the three conditions. The preceding time of the biceps femoris in the vibration condition became near the value in the quiet standing condition as the trials were repeated. It has been definitely shown that the activation timing of the biceps femoris is mainly regulated based on the perception of initial standing position, and that the postural control is adjusted adaptively as becoming advantageous for the equilibrium maintenance based on the knowledge of the control result.

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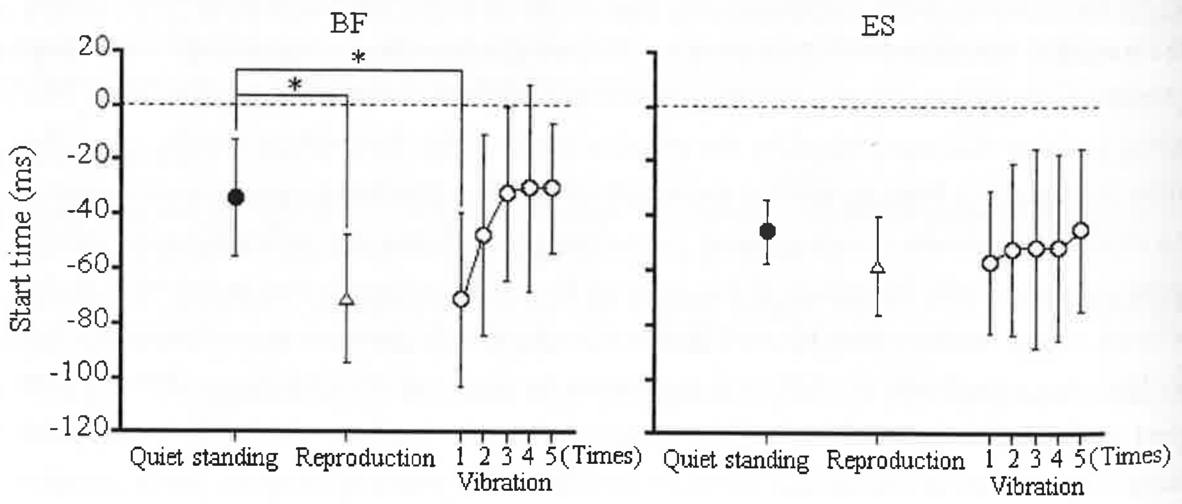

Fig. 1. Mean and Standard deviation of start time in the biceps femoris and the erector spinae. An asterisk indicates a significant difference (p<0.001) relative to quiet standing condition.

INFLUENCES OF LIGHT FINGERTIP CONTACT ON BALANCE DURING BIDEDAL STANCE IN YOUNGER AND OLDER ADULTS

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INTRODUCTION

When a standing subject lightly touches a stable object with a fingertip, body sway is reduced, even when the forces applied by the fingertip are mechanically inadequate to stabilize the body [1,2]. The stabilization is thought to be achieved via sensory cues arising from the fingertip providing information about overall body orientation and motion [3]. This information then can be used to produce compensatory reactions to attenuate sway during quiet standing. Such complex sensory-motor integration requires faithful transmission and processing of signals coming from the fingertips and so the question remains as to whether haptic stabilization of posture can be achieved when tactile acuity is less reliable, as it is the case in human aging [4]. In the present study, we addressed this question by comparing the ability of younger and older adults to use light touch contact as a balance aid during quiet bipedal stance.

METHODS

Preliminary data were obtained in 27 healthy participants (young, n=17, 24±4 yrs; older, n=10, 68±8 yrs). Participants stood on dual-force platforms (AMTI) in their normal standing position, feet comfortably apart and weight evenly distributed. Experimental trials (duration, 1 min) included two visual conditions (vision; no vision), three fingertip contact conditions (no touch; touch smooth; touch rough) and two support surface conditions (stable surface, unstable, foam surface). In trials with fingertip contact, participants were instructed to maintain a light contact with their dominant index fingertip on a touch-plate placed slightly lateral and anterior to the subject. The touch plate was instrumented with force transducers to measure horizontal (F_x and F_y) and vertical forces (F_z). Mean sway amplitudes (cm) of the centre of foot pressure (CoP) in the antero-posterior (AP) and medio-lateral (ML) planes were computed. Pressure sensitivity at the tip of the index finger was also assessed in participants using calibrated Semmes-Weinstein monofilaments. The smallest monofilament that the subject could reliably detect (70% level) provided an estimate of the detection threshold.

RESULTS

In both younger and older adults, light touch contact reduced postural sway significantly in the AP direction (repeated measures ANOVA, p<0.001) but not in the ML direction under all sensory conditions (vision and surface). As shown in Fig 1A, the level of stabilization achieved by older adults in the light touch-no vision condition was comparable to that seen in younger adults. However, older adults exerted significantly greater contact forces (t=7.08, p<0.001, Fig1 B) in order to achieve the same level of stabilization and this across all conditions (vision, no vision, stable, unstable). In the younger adults, contact forces never exceeded 0.75 N (mean across conditions 0.33±0.11 N) whereas forces up to 2 N were recorded in older adults (mean 0.81±0.47 N). As expected, older adults exhibited significantly higher detection threshold for pressure (mean 0.20±0.11g, t=3.85, p<0.01) as compared to young adults (mean 0.06± 0.03g). Pearson's moment correlations between contact forces generated during standing and detection thresholds for pressure measured at rest revealed that both were highly associated (Pearson's r range, 0.51-0.67, p<0.001). Thus, individuals with higher thresholds tended to use higher contact forces to stabilize their posture during quiet standing and vice-versa.

DISCUSSION

The present results provide further evidence of the stabilizing effect of light touch on balance during normal stance [2]. In both younger and older adults, body sway in AP direction was reduced by ~45 % when fingertip contact was allowed both for the stable and unstable surfaces. While the level of sway reduction achieved by fingertip contact was comparable in the two groups, older adults consistently used higher contact forces. The range of contact forces (0.33- 2N) produced by the older adults, however, remains well below that necessary to get effective mechanical body stabilization. To get 40% reduction in sway, contact forces in the order of 5-8 N would have been necessary [5]. Rather, the higher contact forces used by older participants seemed to have been dictated by the necessity of applying slightly greater pressure in order to optimize sensory cues from the fingertip because of a reduced tactile sensibility. The fact that contact forces during standing were highly correlated with detection thresholds for pressure supports this interpretation. Thus, age-related decline in touch sensations may influence the ability to use hand contact for postural stabilization.

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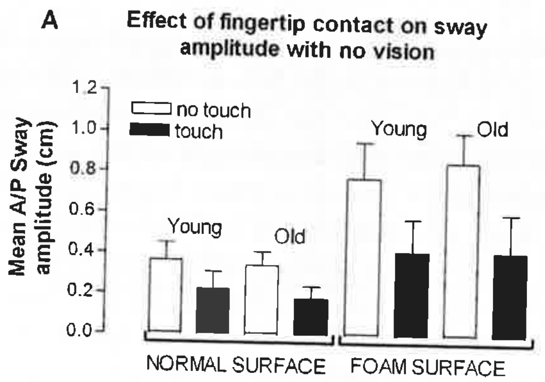

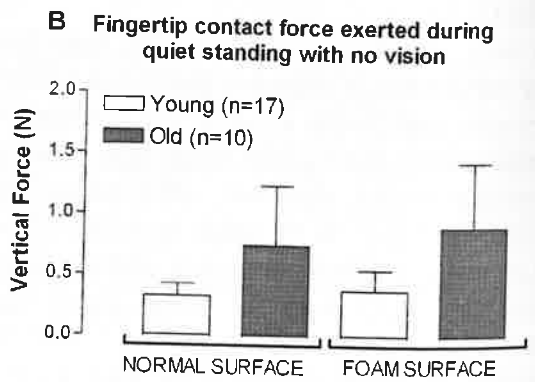

POSTURAL SWAY IN PATIENTS AFTER ACL REPAIR

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INTRODUCTION

Tears of the anterior cruciate ligament (ACL) are common injuries to the ligaments of the knee. Apart from biomechanical changes a proprioceptive deficit occurs after ACL rupture.[1] Impairments of proprioception may lead to increased postural sway which has been documented in patients with chronic ACL insufficiency.[2] Shiraishi et al[3] suggest that knee function after ACL-reconstruction should be closely related with knee proprioception and indicate the usefulness of stabilometric assessment in the evaluation of the function of ACL-reconstructed knees. Denti et al[4] demonstrated a clear motor control deficit in ACL-reconstructed patients compared to normal control subjects. Strength and coordination training can improve standing balance of patients with or without ACL reconstruction.[5,6]
The aim of this study was to assess postural sway in patients after ACL reconstruction by means of computerized dynamic posturography.

METHODS

After informed consent was obtained twenty subjects (11 women and 9 men; median age 30 years, range 17 - 39) who had undergone ACL-reconstruction and a standard rehabilitation protocol and twenty healthy age- and sex-matched controls (11 women and 9 men, median age 30 years, range 17 - 38) were included in this study. Time since surgery had to be at least one year. Exclusion criteria were balance disorders, medications affecting the CNS, symptoms of dizziness or lightheadedness, known neurologic disorders, abnormal vision (without glasses). Postural sway was measured using a commercially available computerized dynamic posturography device (Pro Balance Master System, Neurocom International, Inc, Clackamas, OR, USA). This device consists of a moveable dual forceplate with two 24 x 46 cm footplates connected by a pin joint. The two footplates are supported by four force transducers (strain gauges) mounted symmetrically on a supporting center plate. A fifth transducer is bracketed to the center plate directly beneath the pin joint.[7] The measurements included four subtests. Subtest 1: Fixed support and normal vision (EO). Subtest 2: Fixed support and absent vision (EC). Subtest 3: Sway referenced platform and normal vision (EO/SS). Subtest 4: Sway referenced platform and absent vision (EC/SS). These four subtests are the conditions 1, 2, 4, and 5 of the sensory organization test and are called the modified sensory organization test. Each subtest lasted 20 seconds and was performed three times consecutively. For analysis the mean of the three repetitions was calculated. During testing the subjects had to wear a safety harness fixed to an overhead bar to prevent injury in case of a fall. Measurements were made with subjects dressed in shorts and without shoes. The subjects were instructed to stand on both feet as still as possible and to look straight ahead.
Statistical analysis: The Mann-Whitney U Test was used for statistical analysis. A p value of less than 0.05 was considered as statistically significant.

RESULTS

Detailed posturographic data of patients after ACL-reconstruction and controls are given in table 1. Sway velocity was increased statistically significant in patients after ACL-reconstruction compared to the control group in subtests with sway referenced platform either with normal (0.40 deg/s vs. 0.33 deg/s, $p < 0.04$) or absent vision (1.08 deg/s vs. 0.87 deg/s, $p < 0.03$). The differences in subtests with fixed platform did not reach statistical significance. There was no influence of age and gender.

DISCUSSION

Postural sway is statistically significant increased in patients after ACL-reconstruction in conditions four and five of the sensory organization test. Corresponding with our data on static posturography O'Connell et al[8] reported no significant differences in postural sway parameters between ACL-deficient and control subjects using a postural sway meter. During conditions with sway referenced platform postural information from the feet and legs is unreliable and vestibular inputs may play a greater role in postural responses.[9] As documented with this study knee proprioception seems to have influence on balance during these conditions. Subjects with one ACL-reconstructed knee showed more postural sway than healthy age- and sex-matched controls under challenging conditions like standing on a moveable sway referenced platform (with either normal or absent vision). A possible explanation for these results is that somatosensory information from ACL-reconstructed knees is incomplete, unreliable or missing and therefore the complex response patterns to maintain equilibrium are negatively influenced. Computerized dynamic posturography seems to be more sensitive to detect differences in postural sway than a static system.

CONCLUSION

Reduced postural stability should be taken into consideration when planning a rehabilitation program after ACL repair. Balance and proprioception retraining should be included in rehabilitation early in the postoperative period.

	EO	EC	EO/SS	EC/SS
ACL repair	0.20 (0.15; 0.22)	0.28 (0.23; 0.35)	0.40 (0.33; 0.48)	1.08 (0.83; 1.32)
Control	0.17 (0.13; 0.20)	0.23 (0.20; 0.27)	0.33 (0.30; 0.38)	0.87 (0.77; 0.93)
	NS	NS	$p < 0.04$	$p < 0.03$

Table 1. Postural sway recorded with computerized dynamic posturography.(Sway velocity (degrees/second), data are given as median (25^{th} percentile; 75^{th} percentile))
ACL repair, patients after ACL reconstruction; EO, fixed support and normal vision; EC, fixed support and absent vision; EO/SS, sway referenced platform and normal vision; EC/SS, sway referenced platform and absent vision; NS, not significant.

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NEURO-MUSCULAR ADAPTATION FOLLOWING A BELOW KNEE AMPUTATION

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INTRODUCTION

A transtibial amputation (TTA) can be considered as a major perturbation to the person affected. The loss of a major part of the body clearly has a psychological impact, but also huge physical changes occur that are reflected as perturbations both on a neurophysiological and biomechanical level. A TTA involves that the foot and major distal part of the shank is removed, m.soleus is thinned out or removed and m.gastrocnemius is removed from the ankle and fixed on the remaining part of the tibia. In order to get back to his or her regular life, the amputee has to cope with these changes by adapting to the new situation. The objective of this study was to increase the understanding of adapting principles in response to an altered biomechanical and neurophysiological condition following TTA, especially with respect to joint coordination and muscle coordination patterns.

METHOD

Nine unilateral below knee amputees and nine age- and activity level matched non-amputated right-footed control subjects (all males) participated in this study. The amputees used a Variflex as prosthetic foot. All subjects performed six maximal vertical countermovement jumps.

Kinematic and kinetic information was obtained using five ProReflex cameras (200 Hz) and one Kistler force plate (1000 Hz), respectively. ProReflex markers were placed in a standard manner (or at similar places on the prosthesis) on both legs to estimate the angles and the angular velocity in the ankle and knee joints. Minimum and maximum angles during the jump (prior to take off) were determined as well as the timing of maximal velocity relative to take off. The ground reaction force was used to determine take-off.

In all subjects, electromyography (EMG) was used to study the timing of activation of five leg muscles in both legs. Electrode pairs (20 mm inter-electrode distance) were placed over the m.gastrocnemius (GAS), m.rectus femoris (RF), m.vastus medialis (VM), m.biceps femoris (BF), and m.gluteus maximus (GLU). A 32-channel Mark 5 system (BioSemi) was used for this purpose (2000 Hz/channel, filter settings 5-400 Hz). A digital high pass filter (20Hz) was used to remove movement artefacts. EMG power was used to study the timing of muscle activation relative to take off.

The data obtained from the averaged leg from the non-amputees (CS) were compared with the amputated (AL) and non-amputated legs (CL) of the amputees using Mann-Whitney U tests (SPSS). Left-right and nonamputated-amputated side differences were tested using a Wilcoxon signed rank test.

RESULTS

The profile of the ground reaction force curves of the control subjects and amputees were very similar. Between these two subject groups, no differences were found in their ability to jump straight up. The results of the kinematic and EMG data are presented in Table 1. AL still has some movement in the prosthetic ankle joint (range 10 degrees) although much less than in the ankle joint of CS (range 60 degrees) and CL (range 54 degrees). Minimal and maximal ankle angles and minimal knee angles were asymmetric in amputees while the minimal ankle angle was asymmetric in the control subjects. The maximal angular velocity of the ankle joint was found earlier in AL compared to CS and CL, while no differences in timing of maximal angular velocity in the knee was found between the 3 groups. In CS and CL the maximal angular velocity in the knee is about 8 ms earlier than in the ankle, while in AL about 11 ms later. The EMG results showed that the onset of GAS, RF and VM was delayed in AL compared to CS. No differences were found between AL and CL and CS and CL.

DISCUSSION AND CONCLUSIONS

Despite the observation that the amputees have two different legs and showed more asymmetry in leg kinematics than the control subject group, we did not find any differences in the symmetry of jump performance. This indicates that the amputees have adapted to their new leg situation, since the same neuromuscular control with a different mechanical system would result in an asymmetric performance. This is supported by the observation of a difference in muscle activation between the amputated leg and the control subjects. However, it is also possible to jump straight on one leg. Additional recordings with two force platforms could examine this.

	CS	AL	CL
Kinematic data			
Minimum ankle angle (degree)	85 (5)**	100 (15)*	84 (17)**
Maximum ankle angle (degree)	146 (17)	110 (15)*	138 (11)**
Minimum knee angle (degree)	101 (9)	102 (12)	112 (11)**
Maximum knee angle (degree)	175 (2)	169 (7)	176 (3)
Timing max ang vel ankle (ms)	-65 (6)	-87 (18)*	-61 (13)**
Timing max ang vel knee (ms)	-72 (7)	-76 (17)	-68 (12)
Timing max ang vel ankle-knee (ms)	8 (8)	-11 (16)*	8 (6)**
EMG data			
Onset GAS (ms)	-573 (127)	-438 (115)*	-506 (164)
Onset BF (ms)	-584 (97)	-551 (134)	-557 (161)
Onset GLUT (ms)	-612 (80)	-546 (101)	-554 (158)
Onset RF (ms)	-562 (92)	-354 (87)*	-467 (179)
Onset VM (ms)	-644 (98)	-441 (141)*	-497 (126)

Table 1: Results from the kinematic and EMG data presented as mean (standard deviation) for the average leg of the control subjects (CS), the amputated leg (AL) and control leg (CL) of the amputees. For the kinematic data, number of subjects (N) is 8 for CS and 7 for AL and CL. For the EMG data, N=9 in all groups. The timing variablesof maximal angular velocity (max ang vel) and EMG onset are relative to take-off. * Significant difference between AL and CS (Mann-Whitney U test). ** Significant differences between left and right in the control subjects or amputated and control leg in amputees (Wilcoxon signed rank test).

ELECTROMYOGRAPHY: EVALUATION OF ORBICULARIS ORIS MUSCLES IN REST AND MAXIMUM CONTRACTION IN MOUTH BREATHERS CHILDREN, BEFORE AND AFTER MYOTHERAPY

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INTRODUCTION

This research had been idealized in sense to contribute with the knowledge on mouth breathers children's evaluation and treatment and it proposes to study, through electromyography, the evaluation of superior and inferior *orbicular oris* muscles in rest situation and maximum contraction, in children with mouth breathing, before and after myofunctional therapy and to compare them with nasal breathing children.

METHOD

The sample was composed by 30 children, Brazilian, of both sexes, from 8 to 12 years old , being 15 children with mouth breathing and 15 children with nasal breathing. All children were submitted to an electromyographic evaluation (EMG) of superior (OSB) and inferior (OIB) *orbicularis oris* muscles. EMG recordings of superior and inferior *orbicularis oris* muscles were carried out during rest and during maximum activity of lips, named here as Tests 1 (child stayed for 3 seconds with musculature of mouth relaxed in spontaneous situation of rest) and 2 (.child stayed with lips pressed strongly one against the another for 3 seconds). The equipment used was an Signal Aquision System with four channels, a digital-analogic conversion board, model CAD 10 /26 and a data acquisition program and mathematical processing developed by Lynx Electronic Technology Ltda.
The muscular activity was captured by double surface electrodes, disposable (DUO TRODE), properly fastened 2 mm above free margin from superior lip and 2 mm below free margin from inferior lip, after previous cleaning of skin of the region with ethyl alcohol to 96° GL.
The results of EMG exam were quantified in RMS (rooth mean square) for own data acquisition program and expressed in microvolts (μV).
After EMG evaluation, 15 mouth breathing children were submitted to treatment of breath alterations by a otorhinolaryngologist. Soon after we accomplished myofunctional therapy, two sessions per week and total duration of 4 months, according the time of treatment established for each stage proposed by Hanson and Barret (1995).
For comparative analysis of results, we divided muscular activities recordings in the Groups:
Group NB - muscular activities recordings of 15 nasal breathing children;
Group MB-1 - muscular activities recordings of 15 mouth breathing children in the first electromyographic evaluation (pre myofunctional therapy) and,
Group MB-2 - muscular activities recordings of 15 mouth breathing children in the second electromyographic evaluation (post myofunctional therapy).
To establish comparisons among studied groups, we used no-parametric tests (Steel & Torrie, 1960): Kruskal-Wallis test to analyze independent data and Wilcoxon test to analyze dependent data.

RESULTS

During rest, OSB and OIB showed higher activity in mouth breathers in relation to the nasal breathers children and during maximum contraction, there was higher activity in OSB and

OIB muscles in nasal breathers children in relation to same muscles in mouth breathers children. The results also showed smaller activity in rest and higher activity during maximum contraction after myofunctional therapy, statistically significant, comparing mean values of EMG recordings of studied muscles, between mouth breathers children pre and post treatment We attributed this reduction of muscular activity to myofunctional therapy, that improved function and, in consequence, there was smaller muscular effort to maintain the posture, recruiting a minimum number of motor units.

DISCUSSION

In reviewed literature on electomyographic evaluation , we verified that there are controversies as the presence or not of electric activity during rest position. Some authors (5,6,7) had found no EMG activity in a relaxed muscle, during the rest.

Moyers (3) explained that the mouth breathing patient constantly stays with the half-open lips in position of rest mandibular with difficulty of approaching them due to the hipotonicity that is developed in orbiculars oris muscles, consequence of altered breath. With myofunctional treatment, the tendency is that the breathing way is transferred from mouth to nose, rest posture becomes appropriate with a re-adaptation of this musculature, leading the patient to present conditions of maintaining lips together, effortlessly, and, as consequence, the muscular activity will decrease.The results showed lower level of activity of studied muscles in mouth breathing children and it increases after myofuncional therapy. This increase can be explained because myofunctional therapy looked for adequating functions and to develop new muscular patterns through specific exercises (4,2,1).

CONCLUSION

There is an increase of muscular activity in mouth breathing children when the maintenance of posture is demanded, demonstrated by higher activity of the studied muscles in mouth breathers in relation to nasal breathing children, during rest. Mouth breathing children present smaller contraction force, when effort is demanded, generating smaller activity, demonstrated by smaller activity of the studied muscles in mouth breathers in relation to nasal breathing children, during maximum contraction. The reduction in muscular activity in rest and increase in muscular activity during maximum contraction after myofunctional therapy, that approached activity levels to the obtained in nasal breathers children confirm that it ,indeed, acts in function's re-establishment.

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MODULATION OF SOLEUS EMG ACTIVITY BY FEMORAL NERVE STIMULATION IS MODIFIED IN HEMIPARESIS FOLLOWING STROKE

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INTRODUCTION

Spinal pathways are thought to assist muscular coordination in human bipedal stance and gait (Meunier et al., 1994). Sensory afferents could contribute to this coordination by modulating the excitability of heteronymous muscles. In human, stimulation of femoral nerve (FN) induces a short-latency inhibition of soleus (SOL) reflex (Meunier et al., 1990) and voluntary activities (Forget et al., 1998). This spinal mechanism may contribute to basic coordination between knee and ankle extensor muscles in functional tasks. Muscular incoordination is a major impairment following a cerebrovascular accident (Bourbonnais et al., 1992). Since incoordination of extensor muscles results in abnormal synergies at the paretic lower limb, the questions then arise whether this spinal mechanism of regulation of soleus excitability by femoral nerve afferent 1) is modified in hemiparetic subjects as compared to healthy subjects and if so, 2) to what extent this dysfunction is related to leg motor impairment?

METHODS

The FN was stimulated (pulse duration = 0.5 ms & intensity = H max/2 of rectus femoris) on the right side of 14 control subjects (41 ± 14 y. old) without orthopaedic or neurological deficit and on the affected side of 11 hemiparetic subjects (45 ± 14 y. old). The subjects were seated and instructed to produce a voluntary isometric contraction of SOL at two levels of activation (20% and 40% of maximal EMG). The EMG signal was first amplified (5000 x), filtered (30 Hz to1 kHz) and digitized at a sampling rate of 3000 Hz. The effect of FN stimulation on SOL integrated EMG activity was assessed from 22 to 99 ms after FN stimulation within 6 consecutive time windows of 12 ms duration. Surfaces of SOL integrated EMG (iEMG) recorded during these time frames were compared to the baseline EMG activity before FN stimulation, at the two levels of SOL activation and between the hemiparetic and the control subjects. The Chedoke-McMaster Stroke Assessment (CMSA) was used to assess motor performance of the hemiparetic subjects at the leg and the foot. A Composite Spasticity Index (CSI) evaluated levels of spasticity and the walking time at comfortable and maximal speed over a 5-meter distance was used to measure gait velocity. Spearman rank-order statistics were used to correlate motor impairment measures and EMG modulation of SOL induced by FN stimulation.

RESULTS

Within both experimental groups, the contraction level of voluntary EMG activity (i.e.: 20% or 40% of maximal EMG at Sol) did not affect the pattern of modulation of SOL EMG activity by FN stimulation. However, at a comparable level of baseline voluntary EMG activity, patterns of SOL modulations were different between the two groups (see Figure). The hemiparetic subjects did not show, on average, the typical inhibition observed in the control subjects.

In 40% of the control subjects, a short-latency (26 ± 1ms) and short-duration (8 ± 2ms) facilitation (25 ± 44% increase of baseline values) was observed after FN stimulation. In all control subjects, the FN stimulation produced a marked inhibition (48 ± 22% decrease of baseline values) of SOL voluntary EMG activity at a mean latency of 34 ± 4 ms and duration of 59 ± 34 ms.

A higher percentage of the hemiparetic subjects (64%) showed the FN-induced facilitation. This facilitation was also of greater amplitude (111 ± 99 %) compared to control values (p =

0.016) but the latency (26 ± 3ms) and duration (13±7 ms) characteristics were not significantly different between the two groups. In contrast to the control subjects, the majority of hemiparetic subjects did not show the FN-induced inhibition at latencies corresponding to the pattern observed in healthy subjects. At both voluntary contraction levels, baseline EMG activity of hemiparetic subjects were correlated with the spasticity index (CSI) scores ($r = -0.760$; $p = 0.007$ at 20% of SOL max EMG level and $r = -0.836$; $p < 0.005$ at 40% level) and also with the motor performance (CMSA) scores at the leg ($r = 0.81$; $p = 0.003$ at 20% level and $r = 0.81$; $p = 0.009$ at 40% level). Facilitation of SOL at short latencies (within 22 and 34 ms post-stimulation) was correlated with CSI scores at 20% ($r = 0.88$; $p<0.001$) and 40% ($r = 0.66$, $p=0.038$) contraction levels. Facilitation was also correlated to CMSA scores at the foot ($r = -0.651$; $p = 0.030$) at 20% contraction level whereas only a tendency ($r = -0.51$; $p = 0.13$) was observed between these last two variables at 40% contraction level

DISCUSSION

The short-latency modulation of SOL EMG activity by FN stimulation reveals strong intersegmental influences of FN onto SOL motoneuronal pool (Meunier et al., 1990). The consistent finding is a marked inhibition of SOL voluntary EMG activity observed in all healthy subjects tested. A short-duration (8 ms) facilitation may precede the heteronymous inhibition in 40% of the cases. However, hemiparetic subjects show, on average, non-significant inhibition after FN stimulation. The results suggest an alteration of this heteronymous inhibition. On the contrary, facilitation was the preponderant influence of FN afferents onto SOL voluntary EMG activity at short-terms latencies.

Dysfunction of some spinal mechanisms is well documented in spastic hemiparesis. Our results confirm that these dysfunctions usually result in a global facilitation of spinal pathways (Artieda et al., 1991). The functional significance of these spinal mechanisms is still to be established. The correlations analysis revealed that, in the hemiparetic subjects, the amount of EMG activity that can be generated (i.e. baseline levels) and the modulation capability by sensory afferents are related to the level of spasticity and motor impairment.

CONCLUSION

Intersegmental inhibition of SOL EMG triggered by FN stimulation is modified in hemiparesis following stroke. The modulation observed, which is largely facilitatory, is correlated with the clinical measures of motor impairments.

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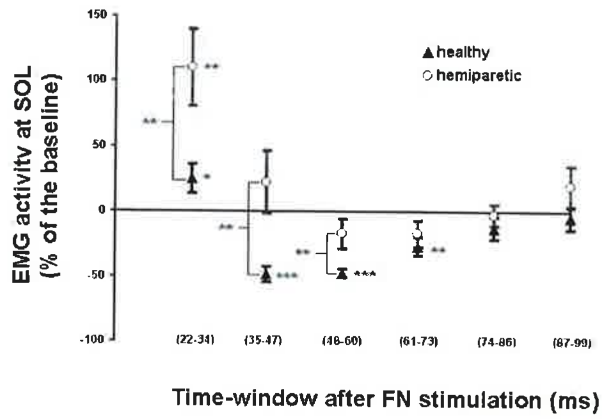

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INTERRELATIONSHIP BETWEEN ASYMMETRIC BODY POSTURE AND RESTING ELECTROMYOGRAPHIC ACTIVITY OF THE MASSETER MUSCLE

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INTRODUCTION

The aim of this study was to determine whether an interrelationship exists between asymmetric body posture and the resting electromyographic activity of the masseter muscle.

METHODS

Eight postural parameter (rotation of the cervical spine, shoulder height, scapular abduction, iliac crest height, shortinnig of trapezius, quadratus lumbarum, iliopsoas and hip adductor muscles) were evaluated for asymmetry in 30 patients and were combined in a score. These parameters were compared with asymmetry of the resting electromyographic activity of the masseter muscle.

RESULTS

There was no significant difference between mean resting activity of right and left masseter muscle. 20 patients had 2 or 3 and 8 patients had 1 asymmetric body posture parameter. In asymmetry score, 9 patients had negative and 13 patients had positive results. Pelvic obliquity correlated with a contralateral electromyographic activity of the masseter muscle, scapular abduction with an ipsilateral increased electromyographic activity of the masseter. The whole score corralated with the asymmetry of the surface electromyography of the masseter.

CONCLUSION

We conclude that an interrelationship exists between body posture and the craniomandibular system.

CHANGE IN MUSCLE ACTIVITY FOR KNEE FLEXION ANGLE DURING CONTINUOUS WEIGHT SHIFT IN STANDING

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INTRODUCTION

The center of gravity movement to either side is the basics of the locomotion including the walking. It often uses continuous weight shift to either side as the important practice problem about physical therapy. However, it is difficult to evaluate the coordination of muscle activity which is necessary for the center of gravity to move to either side with electromyography (EMG). If the coordination of muscle activity becomes able to be evaluated, the purpose of physical therapy can be made clear. Therefore, in this study, it examined about muscle coordination during continuous weight shift by EMG and the possibility as the evaluation..

METHODS

The change in muscle activity for knee flexion angle during continuous weight shift in standing was examined in 16 healthy subjects (1.71±0.04 m, 66.2±6.3 kg, age 23.3±5.4). Periodic sound signal of 1.0 Hz was given successively in the order and the subjects were requested to move from side to side synchronously to the signal for 30 seconds. The frequency of the periodic movement was 0.5 Hz. The angle of knee flexion during continuous weight shift was kept at the three positions: 0 degrees, 30 degrees and 60 degrees. The width of stance was 0.3 m from left caput ossis metatarsalis V to that of right. The subjects were maintained upright position by the experimental equipment that was made with steel. A slide board was put to the experiment equipment and an electric displacement meter (NEC, 9E08-D3) was fixed there. The subjects were instructed so as not to separate their sole from the floor and their chest was fixed to the slide board. Horizontal position data and EMG (NEC, Syna Act MT11) were recorded synchronously using personal computer at sampling frequency of 1 kHz. EMG was measured from seven muscles of right side; gluteus maximus (GM), gluteus medius (GMed), tensor fasciae latae (TF), vastus medialis (VM), biceps femoris (BF), tibialis anterior (TA) and soleus (So). Muscle activities during continuous weight shift were ranked from the mean of full-wave rectification for EMG signal with 10 periods. The average rectified signals of EMG signal were also calculated and the each of signals was normalized by the maximum value of 0 degree knee flexion. In addition, the spectrogram was calculated for EMG signal during the periodic movement to confirm muscle contraction. The fluctuation of periodic time and peak position data was calculated from left peak to left peak with position data. Statistical analysis was used Bonferroni multiple comparisons procedure for position data and Kendall's coefficient of concordance for EMG data.

RESULTS

There was not a difference in each condition in the movement distance to either side. The variation of the movement period was larger in 0 degrees of knee flexion than 30 degrees and 60

degrees. The activities of GMed, TF, VM, TA and So significantly changed dependent on knee joint angle. The activities of GMed and TF were large in order of 0 degrees, 60 degrees, 30 degrees of knee flexion. On the other hand, the activities of VM, TA and So were large in order of 60 degrees, 30 degrees, 0 degrees of knee flexion. As for a lot of muscle activity, it indicated a circle orbit about the scatter diagram with position data (fig.1). Spectrogram was effective to confirm the existence or non-existence of muscle contraction during continuous weight shift.

DISCUSSION

The activities of GMed, TF, VM, TA and So significantly changed dependent on knee joint angle though there was not a difference in each condition in the movement distance to either side. Under the condition of closed kinetic chain, as for the relation between joint angle and the muscle activities, examination isn't accomplished. In this study, the activities of GMed and TF were large in order of 0 degrees, 60 degrees, 30 degrees of knee flexion. It was thought that hip abductor muscles were important to control continuous weight shift in the position of knee extension. Moreover, it was required that more lower extremity muscles contract cooperatively in the position of knee flexion.

We evaluated the activity pattern by the scatter diagram in addition to the size of the muscle activities. The muscle activity increased proportionally to the displacement and the scatter diagram between position data and muscle activity was similar to the hysteresis phenomenon. As for these things, the timing of contraction and relaxation was important in continuous weight shift. It was thought that the scatter diagram between the position data and EMG signals could be used for the evaluation of dysfunction. In addition, when there is abnormality of muscle tone, a correct evaluation will be made in using spectrogram.

CONCLUSION

These results suggest the angle of knee joint must be chosen according to the purpose of physical therapy. Moreover, it was thought that the scatter diagram between position data and muscle activity could be used for the evaluation of the dysfunction.

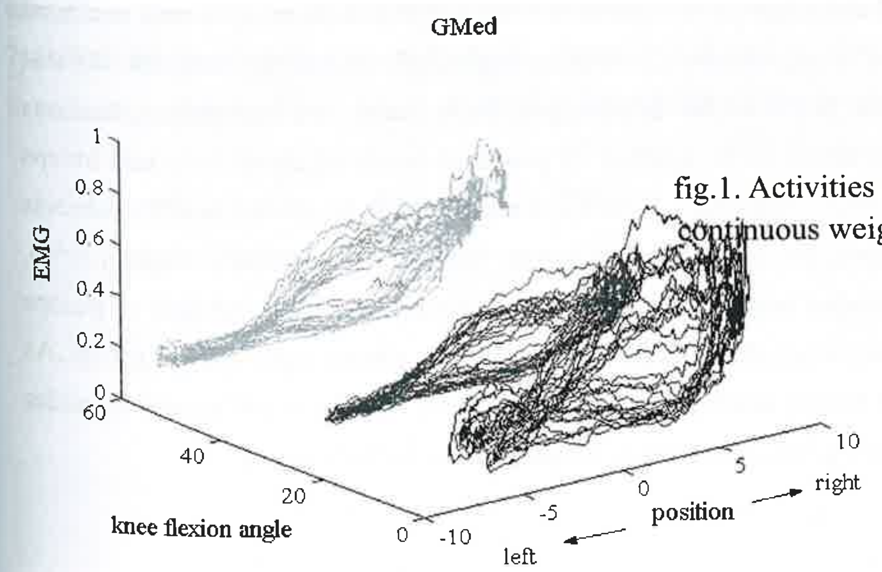

fig.1. Activities of gluteus medius during continuous weight shift at each position.

SURFACE ELECTROMYOGRAPHIC FINDINGS OF THE SHOULDER GIRDLE AND UPPER EXTREMITIES MUSCLES DURING PUSH UP MOTION

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INTRODUCTION

For the spinal cord injury patients (SCI), establishing push-up and transfer motion is essential for independence of ADL. Analyzing these motions with surface electromyography from the beginning of training, we want to assess the effect of rehabilitation program objectively. The purpose of this study was to define the basic data of these motions. So that, we measured the motor activities of upper extremities and shoulder girdles with surface electromyography.

METHODS

The subjects were 5 healthy subjects (3 males and 2 females ,average of age:20.4±2.6 years) and 5 SCI patients with paraplegia(4 males and 1 females , average of age:32.2±10.1 years). Assessed muscles were anterior and posterior heads of the deltoid, biceps and triceps brachii, pectoralis major, infraspinatus, latissimus dorsi and lower trapezius. Push-up in their wheelchair and transfer motion between wheelchair and bed were analyzed. In order to regulate motor potential, %MMT was calculated. It is the ratio of active motor potential during motion to maximal active motor potential during the manual muscle test served as a normalization value of 100%.

RESULTS

During push-up, %MMT of each muscle compared healthy subjects with SCI patients is almost same. Especially %MMT of triceps brachii is relative high (30% at healthy subjects, 20% at patients). %MMT of anterior heads of the deltoid, pectoralis major, infraspinatus, latissimus dorsi and lower trapezius is about 15%. %MMT of posterior heads of the deltoid and biceps brachii is about 5%. During transfer motion, %MMT of each muscle compared healthy subjects with SCI patients is different. As for healthy subjects, %MMT of pectoralis major (50%), biceps brachii (25%) and anterior heads of the deltoid (25%) is relative high. %MMT of triceps bradhii (17%) is relative low. Other shoulder girdle muscles are almost same in both group .As for SCI patients, %MMT of biceps brachii (40%), anterior head of deltoid (40%) and shoulder girdles (30%) is high. %MMT of triceps bradhii is almost same in both groups.

DISCUSSION

Many authors have suggested that the force of contraction of triceps brachii, pectoralis major and latissimus dorsi is important for push-up motion. In our study, almost same findings were considered. However, for SCI patients, trunk extension system is essential factor for independence of ADL. In this study, it is suggested that activity of not only shoulder girdles, but also anterior heads of the deltoid, biceps brachii and pectoralis major is especially important for trunk extension system.

CONCLUSION

In this study, it is considered that activity of anterior heads of the deltoid, biceps brachii and scapular muscles is essential for SCI patients. Because of these muscles are necessary for push-up, extension of trunk and keeping this position. For SCI patients, not only exercise for push up motion, but also balance exercise is important.

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KNEE EXTENSION EXERCISES ASSOCIATED TO HIP ADDUCTION IN THE EMG ACTIVITY OF THE VMO- VLO MUSCLES

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INTRODUCTION

Patellofemoral pain syndrome is the one of the most common musculoskeletal disorders found in the physical therapy and orthopedic practice. As the quadriceps femoral muscle, specially it's oblique portion, and the Vastus Medialis Obliquos (VMO) and Vastus Lateralis Obliquos (VLO) muscles are critical for the dynamic stabilization of the patella, the imbalance of the VMO and VLO muscles can lead to patellar malalignment (1). Selective strengthening of the VMO muscle associated to exercises of hip adduction has been suggested for conservative knee rehabilitation (3; 5). However, very few works studing the electrical activity of the VMO and VLO muscles in closed kinetic chain exercises (CKC) can be found (2). Therefore, the aim of this work is to investigate the electrical activity of the VMO and VLO muscles during maximum voluntary isometric contraction (MVIC), for exercises of knee extension, associated to hip adduction, in CKC.

METHODS

The exercises were performed on a horizontal Leg Press equipment (Vitally) by 10 healthy women, aged 20 to 30 years (22,1±1,91) with no history of knee, hip or ankle pathologies. All subjects read and signed an institutionally approved Consent Form, prior to the experiment (Resolution 196/96 of the National Health Committee). The electromyographic (EMG) signals were collected by surface active differential electrodes (gain = 100) fixed on the midline of the muscle belly and further amplified and filtered (gain = 10; band pass filter = 10Hz – 450Hz) for data acquisition (sample frequency = 1000Hz; resolution = 12 bits).
The MVIC exercises were performed with the knee and hip joints flexed at 90°, the tibia at neutral position and the hip abducted at 30°. A resistance device was placed on the distal portion of the thigh to prevent adduction movements. The EMG signals were recorded during 4 seconds, and normalised by the percentage of the MVIC for knee extension.

RESULTS

The ANOVA, with repeated measures, and the contrast variables analyses revealed that the electrical activity of the VMO muscle was significantly higher (p=0.0441) than the electrical activity of the VLO muscle in the exercise of isometric knee extension associated to isometric hip adduction.

DISCUSSION

Our results show that the proposed exercise, on a leg-press equipment, selectively activates the VMO muscle. The majority of the studies found in the literature, evaluating VMO and VLO muscles, performed this exercise in open kinetic chain (3; 5). According to (1) and (4), the activity of those muscles should remain approximately synchronous, suggesting that they have a reciprocal action in controlling patellar position. If it is so, and according to the finds of this work, it is clear that more studies must be performed to better evaluate the activity of those muscles in order to develop better techniques for patient treatment.

CONCLUSION

This work suggests that exercises of knee extension, associated to hip adduction, in CKC on horizontal Leg Press equipment, selectively activates the VMO muscle and could be indicated for the treatment of dysfunctions in the medial compartment of the knee (ligaments, meniscus and PPS).

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STUDY OF TEMPOROMANDIBULAR DISORDERS IN PATIENTS WITH DIFFERENT CLASSES OF DENTAL OCCLUSION

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INTRODUCTION

The purpose of this research was to study the relationship between dental occlusion and head posture, with anteriorization of the head. This research used the Angle´s classification (1889) to determine the patient´s class of dental occlusion. There are three classes: I, II, III. The first class is characterized by the sort key, cusp mesiovestibular of the first molar occlude in the sulco mesiovestibular of the first lower molar, which always presents bad dental positions, for example: crowding of teeth, bite open, dental inclination and others. in the class II we can observe distalização of the sulco mesiovestibular of the first molar inferior in relation with the mesiovestibular cusp of the first upper molar. This class can be shared in 1°, when the superior incisive teeth are vestibulary; 2° when the central incisive are retroinclinados and the laterals incisive are shows or not vestibulo-mesio-version. The class III is characterized by the mesial of the sulco mesiovestibular of the first molar inferior in relation with the cusp mesiovestibular of the first molar superior. This class and the class II present left and right subdivisions according to the predominance (class III about the class II and this about class I).

According to Kendall (1995) the normal curve of the cervical spine is front convex. The correct position of the head and neck is when the head is in a very balanced position, and can be positioned with the minimum muscle strong. In a lateral view the reference point is the ear and the neck presents a normal front curve. In a posterior view the reference is the medium line of the head with the cervical spine process. The head is not inclined or rotated, and the chin is not restricted. The anteriorization of the head is when the neck extension muscles are in a shorten and strong position, in this situation the muscle can develop an adaptive shorten position.

Rocabado (1983) and Saboya (1987) affirm that there are relationships between postural alteration and occlusion alteration, for example: when the head is front and down inclined, the occlusion contacts are in front, by the other hand, when the head is extend and behind inclined the occlusion contacts are in a posterior position.

The function of the cervical musculature is keep the balance between head position and all musculature of the stomatognathic system, this fact explain why a simple alteration in any structure can change all the system. Typical alterations in patients who presents craniomandibular dysfunction is the muscle tension of the cervical spine and head posture (Huggare et al, 1992).

We cant affirm that dysfunction of occlusion will be always a cause of postural alteration. A lot of studies are trying to prove that occlusion modification can cause different effects in each person, like postural alteration. Many results shows that through a modification of the dental occlusion, it is possible to change the postural dynamic of the person too (Milani et al, 2000).

METHOD

To elaborate this study were evaluated 48 female volunteers, ranged from 17 to 35 years of age.Through a questionary elaborated by the authors of this research, were evaluated the class of dental occlusion with a clinic exam and head posture with photograph analysis.

RESULTS

Through the evaluation were verified that 2.04% of the patients did not present an occlusion classification and 4 (8.16%) presented normal occlusion without any alteration. According to Angle´s classification, 19 (38.77%) of the patients were in class I, 11 patients were in class II, in this case 10.20% of 48 patients were in the 1° division and 12.24% were in 2° division. The subjects in class III added up 28.57% of patients.

In the evaluation of the craniomandibular dysfunction, 75.51% of the subjects showed head anteriorization, 22.4% showed discreet anteriorization of the head and 1 (2.04%) patient did not present any alteration in head posture.

In the comparative evaluation between occlusion and head posture, among the patients who presented head anteriorization , 13 subjects (35.13%) had occlusion classification I, 18 (91%) were class II and 35.13% presented class III, 8.10% were normal and 1 patient did not present occlusion classification.

In the group which the patients presented discreet head anteriorization (22.44%), 1 patient presented normal occlusion, 45.45% presented class I of Angle, 36.36% were class II and 1 patient class III.The patient who did not present anteriorization, presented occlusion class I.

DISCUSSION AND CONCLUSION

In our sample were verified that 97.95% of the patients presented head anteriorization and only 1 subject (2.05%) did not present any kind of alteration in the head posture.

Between the patients who presented head anteriorization the class III was the predominant, plus with class I, while that subjects who presented discreet head anteriorization the majority were occlusion class I, and after class III.

This research can´t show the real clinical condition of craniomandibular dysfunction because it is just a statistic evaluation of a small group. So it is necessary an evaluation with a biggest casuistic to determine the most important craniomandibular alterations and occlusion alterations.

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ELECTROMYOGRAPHY ACTIVITY AND SEVERITY OF THE TEMPOROMANDIBULAR DISORDER

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INTRODUCTION
Acording to CLARK et al. (1987) and CARLSON et al. (1998) the eletromyography (EMG) is a usefull tool to demonstraded the muscular disorders, like as the muscular hyperactivity present in pacients with myogenous temporomandibular disorders (TMD). The diagnostic index are used to classified pacients with TMD, however no relationship between the severity and electromyography (EMG) amplitude was studied. Thus, the purpose of this study was to analyse the electric activity of the masticatory muscles of pacients with myogenous TMD classified as severe (n=8) and moderate (n=5) according with FONSECA (1992).

MATERIAL AND METHODS
All volunteers related orofacial pain for at least 6 moths. Thriteen volunteers (2 males and 11 females) participated of this experiment. The study was approved by institution's human ethics comitte and written informed consent was gained from all subjects. The masseter and temporalis EMG signals were acquired with differencial surface electrodes, with 10GΩ input impedance, 130 dB CMRR/2 picofaraday, and 50 times of gain. The raw EMG signals were digitized by a 12 bit A/D coverter board, Butterworth filter, low-pass of 509 Hz, high-pass of 10,6 Hz, 20 times of gain, and 1000 Hz sample frenquency. The RMS values normalized by de maximal interscuspidation were compared by the Mann-Whitney test with the significance level set as $p<0.05$.

RESULTS
The mean RMS values, during rest mandibular position, to the severe TMDgroup were: 0,16 ±0,16 to rigth masseter (RM); 0,13 ±0,11 to left masseter (LM); 0,33 ±0,22 rigth temporalis (RT); and 0,19 ±0,14 to left temporalis (LT). The mean RMS values, during bilateral mastication to the severe TMDgroup were: 0,59 ±0,26 to RM; 0,58 ±0,26 to LM; 0,56 ±0,16 to RT; and 0,56 ±0,23 to LT. The mean RMS values, during rest mandibular position, to the moderate TMDgroup were: 0,09 ±0,05 to RM; 0,11 ±0,12 to LM; 0,19 ±0,11 to RT; e 0,10 ±0,05 to LT. The mean RMS values, during bilateral mastication, to the moderate TMDgroup were: 0,58 ±0,25 to RM 0,61 ±0,36 to LM; 0,46 ±0,19 to RT; and 0,52 ±0,19 to LT. Comparing this results, no significance differences were found in the muscles activity ($p>0.05$).

DISCUSSION
Based on obtained results it was not possible to verify relationship between the severe and moderare diagnostic index proposed by FONSECA (1992) and normalized EMG amplitude. However, this statement do not invalidate the FONSECA diagnostic index aplication since the majority of the questions that performe this classification criteria are about pain aspects of the TMD and there are conclusive founds about the relationship between pain and EMG signals.

CONCLUSION
Thus, in these experimental conditions, we concluded that the classification by FONSECA severity index did not present relationship with the electromyography activity registered in theses TMD pacients. This conclusion must be related with the multifactorial etiology of these disorder.

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ELECTROMYOGRAPHY EVALUATION OF THE SHOULDER GIRDLE MUSCLES DURING OPENED AND CLOSED KINETIC CHAIN EXERCISES TO UPPER LIMB

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INTRODUCTION

The most actual studies of the rehabilitation protocols to the shoulder girdle, paticularly to the impingent syndrome, are classifing the exercises acordding to myoelectrical activitation level (MACCAN et al., 1993). Thus, the aim of this study was to evaluate the electromyography activity of shoulder girdle muscles during closed (CKC) and opened (OKC) kynetic chain exercises.

In this study, we are considering that in CKC exercise the distal extremity, the hand, is lying at a fixed resistency with axial compression thougth the long axis of the upper limb. On the another hand, the OKC exercise must be performed without load forward the glenoumeral joint and the distal portion of the upper extremity staying free and no-supported.

MATERIAL AND METHODS

Ten sedentary females (20 years old ±1,6) volunteered to participate in this experiment. The study was approved by the institution's human ethics comitte and written informed consent was gained from all subjects.

The biceps, triceps, deltoid, pectoral major and trapezius EMG signals were acquired by differential surface electrodes, with 10GΩ input impedance, 130 dB CMRR/2 picofaraday, and 50 times of gain. The raw EMG signals were digitized by a 12 bit A/D coverter board, Butterworth filter, band-pass of 10,6-509 Hz, 20 times of gain, and 1KHz sample frenquency.

The CKC exercise was performed by the volunteer in the ortostatic position, elbow full extended, resting on the body weigth into dominant upper limb with 90 degrees shoulder abduction on scapular plane and medial rotation. The OKC exercise was performed by volunteer in the same position, however, keeping the dominant upper limb no supported at 90 degrees shoulder abduction with the maximal resistency previously tested during the register of the data. The RMS values normalized by de maximal voluntary contraction were compared by the Mann-Whitney test with the significance level set as $p<0.05$.

RESULTS

The RMS values obtained during the CKC and OKC exercises and normalized by de maximal voluntary contraction are in the Table 1.

During OKC exercises the deltoid and trapezius had the greather electric activity ($p<0,05$) when compared with the others muscles. During CKC exercises only the trapezius muscle showed greather activity ($p<0,05$). No significant statistic differences were showed by the Mann-Whitney test ($p>0.05$) to mean RMS values of the triceps, trapezius, pectoral major muscles when compared the OKC and CKC exercises. However, comparing OKC and CKC exercises the mean RMS values of the deltoid and biceps muscles were higher in the OKC exercise ($p<0.05$). Otherwise, the triceps, trapezius and pectoral major showed no siginificant difference with the studied exercises ($p>0.05$).

DISCUSSION

Comparing the myoelectric ativity of the studied exercises we found more similar between the activation amplitudes of the muscles during the CKC exercise. These greather co-contration for the CKC exercises must be guaranteed for biomechanic and neurophysilogic aspects. In the CKC exercises the contraction and the abduction are assistanced by the support of the upper limb together with the abdutor musculature action, how is showed in the OKC exercise studied. Moreover, the CKC exercises stimulate the neuromuscular spindle, Golgi tendineous organ and capsular receptors by the axial compression. This stimulation increase the co-contraction and, consequently, decrease the superior translation of the humeral head if the rotator cuff is weak (KIBLER, 1998).

CONCLUSION

The results of this studied suggested, in these experimental conditions, the OKC exercise produced greather activation in the deltoid and trapezius muscles. However, the CKC exercise promoted muscular co-contraction and could be recommended in reabilitation programs when the stability and decrease of the superior translation of the humeral head are targeted.

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TABLE 1. The RMS values obtained during the closed (CKC) and opened (OKC) kynetic chain exercises and normalized by de maximal voluntary contraction. (n=10)

	Biceps	Triceps	Deltoid	Pectoral	Trapezius
OKC	0,44 ±0,22*	0,23 ±0,16	0,98 ±0,57*	0,32 ±0,18	1,36 ±0,61
CKC	0,23 ±0,14	0,37 ±0,20	0,57 ±0,26	0,54 ±048	1,08 ±0,51

* $p>0.05$ values

READINESS TO ADOPT PHYSICAL ACTIVITY: AN APPLICATION OF THE TRANSTHEORETICAL MODEL IN REHABILITANTS WITH CLBP

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INTRODUCTION
The Transtheoretical Model (TTM) postulates an integration between motivation and cognitions and is based on the premise that people move through a series of stages in the attempt to change a behaviour. Five stages of change have been reliably identified: precontemplation, contemplation, preparation, action and maintenance. The major cognitions are self-efficacy and decisional balance. Self-efficacy and perceived pros for physical activity should increase across the stages, while the cons should decrease. The aim of the study was to examine the predication of the model.

METHODS
Our sample consisted of 87 rehabilitants with cLBP (mean age: 45.9 years; SD: 9.97; range: 22-62 years; 16% female). The model was tested by a questionnaire which the patients got 3 weeks before admission to a reha-center.

RESULTS
The perceived cons for physical activity decreased significantly across the stages, while the perceived pros and self-efficacy increased.

DISCUSSION
The results of our study corroborate the beliefs of the TTM for physical activity.
Keywords: behaviour chance – chronic Low Back Pain – decisional balance – motivation – physical activity – rehabilitation – self-efficacy – Transtheoretical Model

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PREVALENCE OF POSTURAL ALTERATIONS IN TOTALLY BLIND INDIVIDUALS

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INTRODUCTION
The visually impaired have their basic needs – such as moving around, security, physical integrity, self-esteem, freedom, space and orientation – affected, provoking disturbs in the formation of the body schema which consequently bring postural alterations. The eye is at the same time an interoceptive and exteroceptive receptor of the tonic-postural system and will help in the seeing individual's dynamic and static postural adjustments and repercussions (BRICOT[2]). When information from the eye is asymmetric or pathologic, it translates an adaptation reaction, leading to a new pathological or adaptive postural adjustment (BIENFAIT[1]; BRICOT[2]). Therefore, the objective of this research was to evaluate postural alterations prevailing in total blind individuals.

METHODOLOGY
Ten totally blind individuals from the Center for the Support of Impaired Individuals in the São Paulo City University (UNICID) were evaluated. We included in this research 8 male and 2 female volunteers aged 23 to 58 (x=33) with total manifestation time, in the case of acquired blindness, for more that 6 years of visual impairment. As to the aetiology of the visual impairment, 7 of the volunteers had congenital visual impairment and 3 had traumatic impairment. Individuals undergoing physiotherapeutic treatment or with other neurological pathologies (aphasia, ataxia, motoneural traumas in top or bottom members) were excluded. The static postural evaluation preconized by KENDALL[3] was done. For the postural evaluation, we used a form with personal data, anamnese and physical exam (anterior, posterior, left and right static visual inspection). For the physical exam, the volunteers were positioned 20 centimeters ahead of the acrylic symmetrograph by Carci. All participants signed the term of formal participation in researches, written in Braille and approved by the UNICID's Committee of Ethics in Research. For data analysis, the frequency distribution of data analysis was used.

RESULTS
In the postural evaluation, the main postural alterations found were: for anterior eye segment, neck deviation to the left (60%), more elevated left shoulder (90%), valgus knee (70%) and more elevated internal maleolus (60%). For the posterior eye segment, the elevated left shoulder was confirmed (90%). We also found dexterous convex thoracic-lumbar scoliosis (70%) and corresponding dorsal convexity. For the left and right lateral eye segments, we observed, head forward (100%), protruded shoulder (90%), cervical hyperlordosis (70%), thoracic hyperkyphosis (60%) and lumbar hyperlordosis (90%).

DISCUSSION

The prevailing postural alteration found in the visual impaired evaluated was the kyphosis-lordosis posture, described by KENDALL[3]. The data of this work are in accordance with the findings of MOSQUERA[4], who described the kyphosis-lordosis posture as a characteristic of the visually impaired. Such posture is probably due to constant unbalance situations suffered by the visually impaired, making them concentrate their gravity center at the bottom and forwardly, causing a posture of pelvic antiversion and protruded head and shoulder and cervical hyperlordosis. Another postural change observed was the difference between the shoulders height, which may be related to the use of the walking stick, for the volunteers who use the stick in the dominant hand, present elevation of the counter-lateral shoulder and scoliosis. This shows why all right-handed volunteers had their left shoulder more elevated. The scoliosis in the volunteers was another common characteristic related to the height of the shoulders which provoked a broad structural desaligment involving the vertebral column. Considering that the visual impaired presents disturbs in the formation of the body schema, and, consequently, postural alterations, during physiotherapic treatment, it is important to make the blind individual aware of the body schema before postural rehabilitation.

CONCLUSION

In experimental conditions used, we suggest that the postural alteration prevailing in visual impaired individuals was kyphosis-lordosis and that this is probably due to alterations in the formation of the body schema and by daily life conditions of visually impaired individuals.

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ANALYSIS ELECTROMYOGRAPHIC OF TRAPEZIUS AND STERNOCLEIDOMASTOIDEUS MUSCLES DURING NASAL AND ORAL INSPIRATION IN NASAL AND MOUTH BREATHER CHILDREN

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INTRODUCTION

Patients with stomatognatic system dysfunction, according to Rocabado[8], present anterior head and neck position with loss of physiological lordosis and occipital extension over the atlas bone, due to bilateral hyperactivity of sternocleidomastoideus muscles.It is probable that the level of cervical muscles activity will be abnormal in mouth breather children because they have to change their head and neck position in order to reduce the narrowing of airways.This study aims to analyze the pattern activity of sternocleidomastoideus (SCM) and trapezius muscles during nasal and oral inspiration in mouth breather children, and compare the activity of these muscles with the same ones in nasal breathers.

METHOD

46 children ranging from 08 to 12 years old, from both sexes were evaluated through electromyography . The selected children were divided in two groups; Group I, constituted by 26 mouth breather children and Group II, constituted by 20 nasal breather children. Electromyographyc recordings were obtained through surface electrodes in the SCM and trapezius muscles, bilaterally , during oral and nasal inspiration. The signals were acquired with gain of 5000 and the samples were digitized by a 10 bits A/D converter board with sample frequency of 4000 Hz and stored in a hard disc of a personal computer. Data in root-mean-square(RMS) expressed in microvolts (μV) were analyzed through Kruskall-Wallis statistical test.

RESULTS

There was significant difference in the electric activity of the studied muscles which showed an increase during the nasal inspiration test in oral breathers. However, statistically, in oral inspiration test there was not a significant difference among the mouth and nasal breather children. Within the groups, only mouth breathers group showed higher activity during nasal inspiration.

DISCUSSION

The results of our study confirm Campbell[2] findings, where accessory muscle activity during breathing (scalene and SCM) became important when breathing level was increased, and sternocleidomastoideus muscle action arose when an individual needed to exercise with an inspiratory pressure larger than 20 cm of H_2O. Sousa[11] has highlighted the activity of SCM in the respiratory mechanics, where it was observed an action potential of this muscle during maximum inspiration. These results confirm the opinions of some authors[6,7], who consider the SCM as an inspiration accessory muscle.In situations of breathing action effort a more vigorous diaphragm contraction happens, preceded by inspiratory accessory muscle action[10] This is demonstrated through the highest activity of SCM muscle , in mouth breather children with larger effort in the nasal inspiration. Nasal obstruction increases posterior nasal resistance, and leads to mouth breathing habit This obstruction could turn into an abnormal breathing habit which can persist even when original cause has already been eliminated, because there is an important change posture and muscle shortening, that should be also treated[1]. Costa[3], verified an increase of the activity of SCM in situations of breathing effort.The small electric activity of the studied muscles observed during mouth breathing, in both groups, is in accordance with Sartor[9] ,who considered the nasal resistance twice as large as the oral resistance, having double increase of breathing effort compared to the effort required for mouth breathing. Gross[5] stated that mouth breathing can be a normal development stage, related to environmental factors, and many young people and adults show this reaction to the absence of significant alteration of the airways. During nasal inspiration we did not observe a larger effort in relation to the oral inspiration in the nasal breather group, which did not happen in the mouth breather group.The results obtained by Costa et al[4] do not support the present study because the authors observed a small increase of SCM activity during free deep inspiration movement through mouth, comparing to breathing through the nose. Such observation can be related to the need of altering the head position to accomplish oral inspiration which was demonstrated by Vig[12] ,who observed that, after two hours of induced mouth breathing, the measure of head extension reached 5 degrees. It is also important to highlight that the mentioned authors accomplished their study with healthy individuals.

CONCLUSION

With these results it can be concluded that the studied muscles develop hyperactivity during nasal inspiration in mouth breathers due to a larger breathing effort, caused by a larger resistance of the airways, which demands the action of the accessory breathing musculature . The importance of this conclusion is that mouth breather therapeutic should not be addressed only for orofacial changes , because , as it was demonstrated, they also present change in cervical muscles. We also suggested that further studies related to the level of cervical muscle activity after postural and respiratory treatment should be done.

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MEDIAN FREQUENCY OF THE SURFACE ELECTROMYOGRAPHIC SIGNAL BEFORE AND AFTER TENS TREATMENT IN SUBJECTS WITH TMD

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INTRODUCTION

Muscular pain and fatigue are symptoms that are frequently reported by individuals with temporomandibular disorder (TMD). In this matter, TENS has called the attention of professionals and researchers as a procedure that has a significant effect on pain, and, according to THOMAS, FRUCHT, JONAS & KAPPERT and EBLE, JONAS & KAPPERT, acts on muscular fatigue as well. This can be evaluated through surface electromyography (EMG). The objective of the present work was to investigate the effect of TENS on pain and the median frequency of the EMG signal on the masseter of both normal individuals and individuals with TMD.

METHODOLOGY

The right and left masseter of twenty female volunteers (aged 20-33, x=24,26) – ten with TMD and 10 without TMD – were analyzed through EMG before and after the application of TENS. All volunteers signed the Formal Participation Term approved by the FOP/UNICAMP Committee of Ethics in Research Involving Human Beings. The electromyographic signal was collected in Faraday's electrostatic cage through a Signal Conditioner module (model MCS-V2 by Lynx Electronics Ltd.) with 12 bits of dynamic range resolution, Butterworth filter of 509-Hz lowpass and 10.6-Hz highpass, 100-times gains, and 1000-Hz sampling frequency. Lynx Electronics Ltd. differential active surface electrodes (10-GΩ input impedance, 130-dB CMRR and 2 faraday peak, 100-times gain, 20-Hz highpass and 500-Hz lowpass) were positioned in the middle of the masseter. The electromyographic signal was captured in isometric contraction, kept for 5 seconds with 2-minute intervals between contractions. The TENS (symmetric bi-phasic quadric pulsing, 150 Hz, 20 μs, sensorial level stimulation, 50-% frequency modulation – Dualpex 961/Quark) was applied once on both groups for 45 minutes through self-adhesive electrodes applied on the anterior portion of the M. temporal and M. masseter bilaterally. VAS was applied immediately after TENS. The average frequency was obtained through FFT, using the hanning window. The frequency values were compared through the Mann-Whitney values with significance value $p<0.05$.

RESULTS

The results showed that there was a difference ($p=0.0346$) in the VAS values, which were observed before and after the TENS. After the treatment, there was a significant pain reduction in individuals with TMD. As to the median frequency values, a decrease in the averages of the median frequency of M. masseter after TENS application in both groups ($p=0.05$) was observed bilaterally. The groups were compared before and after the treatment and, before the TENS application, the average of the median frequency in the group with TMD was found to be lower than that of the group without TMD ($p=0.05$). This situation was not reverted after the TENS application.

DISCUSSION

In the present study, we observed a reduction in the values of median frequency of M. masseter bilaterally after TENS application in both groups. This suggests fatigue of the M. masseter in the individuals with or without TMD. The results contradict the findings of THOMAS[1], FRUCHT, JONAS & KAPPERT[2] and ELBE, JONAS & KAPPERT[3], for, according to the authors, TENS reduces the fatigue of skeletal muscles due to the increase of the blood circulation on the area, consequently improving oxygenation and removing the metabolites resulting from muscular contraction.

It is believed that the difference in the results is due to the stimulation time and the current parameters used by FRUCHT, JONAS & KAPPERT[2]. The authors applied TENS for 20 minutes in burst mode and, in the present work, 45 minutes of application in sensorial mode were used.

Electric stimulation in sensorial mode is a hard-to-control variable and, there may have been some muscular contraction that was imperceptible for the human eye, especially when the current intensity was increased in order to reduce the accommodation effect. These muscular contractions possibly promoted an accumulation of metabolites and, consequently, a reduction in the speed of conduction of the muscular fiber that caused a higher concentration of electromyographic signal amplitude in low frequencies, being interpreted as muscular fatigue. These conclusions are only speculations for the blood analysis was not performed in this work.

CONCLUSION

In the experimental conditions used, we can conclude that a 45-minute TENS application promoted significant relief in the pain of individuals with TMD. However, in the spectral analysis of the electromyographic signal the application of the resource promoted a higher concentration of amplitude of electromyographic signal in low frequencies, suggesting the fatigue of the M. masseter bilaterally in both studied groups, being higher in the group with TMD. These results were possibly obtained due to the range of excitability used during the TENS application. Therefore, new works considering several application times and different excitability ranges are necessary.

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BALANCE AND GAIT: ARTHRITIS VERSUS AMPUTEE PATIENTS

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INTRODUCTION

Clinically, it seems that there is some similitude in the pattern gait of the amputee and arthritis patients. The aim of this study is to evaluate if there is any real difference in balance and/or gait analysis in amputee and arthritis patients. We didn't find any refference in literature about this comparison.

METHODS

We chose two groups: one with amputee patients and other with arthritis patients.

- One man, 72 years old, farmer, with left transtibial amputation (traumatic ethiology) prothesised since 1999.
- One man, 35 years old, driver, with right transfemoral (traumatic ethiology) prothesised since January 2002.
- One woman, 36 years old, medical doctor, with rheumatoid arthritis since 1987, presently level II on the Steinbrocker index.
- One man, 68 years old, ex-construction worker, with rheumatoid arthritis since 1998, presently level III on the Steinbroker index.

We excluded any vestibular and visual dysfunctions.

We used a computerized dynamic platafform posturography (CDPP) for the quantitative and functional assessment.

RESULTS / DISCUSSION

The CDPP is used to quantify the force applied by the body to platform equipped with strain gauges; it measures postural sway in several test conditions that are intended to challenge the mechanisms of postural control. So it may be useful in the case of patients with symptoms or signs of chronic balance dysfunction when combined with the clinical history and physical examination. Analysing the multiple graphics, we noted the following items:

- All four patients had normal **weight bearing test**, beside a small difference in more affected side. This test is only made on static position.
- Concerning **limits of stability test,** in both groups, the most important alterations were found in forward and lateral *excursions*. As we know, on one hand, in rheumatoid arthritis maximum ankle plantar flexion and dorsiflexion are diminished in the injured ankle(s); on the other hand, in the amputee, we have the stiffness of the prosthesis ankle (SACH foot). The dynamic characteristics of normal articulation are absent. Also the *movement velocities were* worse in young transfemoral amputee (short rehabilitation time, heavier prosthetic limb), and elderly arthritis (muscular weakness according to his arthritis severity).
- The two amputees demonstrated serious difficulties in **modified clinical test for sensory interaction on balance**, namely the young one in firm platform with open eyes (suggesting abnormality in processing the new limb in the central system – somatosensory and visual feedback).
- The **rhythmic weight test** reflected bad directional control even in slow velocity, specially in the two elderly (probably because of aging).

- In the **walk test** we found diminished step length and speed in the transfemoral amputee (perhaps poor prosthetic fit; greater energy spent) and elderly arthritis (as revealed by gait studies in this population).
- In the **step / quick turn test** all four patients revealed severe difficulties.
- In the **sit to stand test** the young amputee (artificial knee) and the elderly arthritis (pain, weakness and deformities) showed difficulties to stand up. The elderly amputee also demonstrated difficulties in *sway velocity.*
- The two amputees couldn't execute the **step up / over test** with the prosthetic limb side (perhaps due to the lack of ankle moment generator).

CONCLUSION

Features of balance and gait disturbances in rheumatoid arthritis and amputees are not simple. There are numerous factors like age, sex, personality, disease duration / severity / activity (acute, ...), disability, kind of prosthesis, which can cause difficulties for the methodological quality of most of the experimental studies.

In spite of the little number of patients, we really found differences in balance / gait analysis in amputees and arthritis subjects, as mentioned above.

About CDPP, it provides clinical contribution (evaluation / quantification) and may help planning rehabilitation program. The early detection and correction of causative factors of disturbances is of great importance for successful rehabilitation of these patients.

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QUANTITATIVE ANALYSIS OF TENDON TAP REFLEX RESPONSES IN HEMIPARETIC STROKE

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INTRODUCTION

The muscular weakness that follows hemiparetic stroke has been attributed largely to a loss of descending excitation to the spinal segment. While this mechanism presumably plasy a role in causing muscular weakness, there is now increasing evidence that disorganized motor output at the spinal segmental level also contributes. (Tang, et al., 1981, Gemperline, et al., 1995). By disorganization we mean altered recruitment and rate modulation patterns of motoneurons within the pool. In this study, we compare surface EMG characteristics on the involved and contralateral sides of hemiparetic subjects. We chose to investigate muscles of hemiparetic subjects, because the presence of a relatively normal limb provides us with a sound basis for comparison with the paretic limb. We sought to characterize abnormal motor unit recruitment by examining the EMG response in the biceps brachii muscle to a tap of the biceps tendon.

METHODS

All participants were consented via protocol approved by Northwestern University Institutional Review Board. Experiments were performed on both the affected and the contralateral sides of five hemiparetic stroke survivors aged 37 to 62. The forearm was then cast and immobilized in a ring mounted to a six-degrees-of-freedom (ATI, Inc.) load cell designed to record the force exerted at the wrist. Bipolar surface EMGs were recorded from both the biceps and triceps brachii muscles using a Delsys(2.1) surface electrode. The surface electrode was placed on the triceps muscle to identify and prevent cases of co-contraction of the biceps and triceps. Tendon reflexes were elicited in both arms of each subject at approximately five second intervals using a tendon hammer. This device contained a load cell attached at the point of contact with the forearm so as to measure the force with which the biceps tendon was struck. An oscilloscope provided visual feedback of the force the subjects were generating.

The subjects were first asked to remain at rest while a series of stretch reflex trials were performed and then asked to maintain a low level of force so as to evoke a minimal amount of EMG (background) activity while stretch reflex trials were performed. The magnitudes of the surface EMG responses were measured by calculating the root mean square(RMS) of the reflex response. A least squares regression line was fitted to the RMS EMG of the reflex response versus the (force)magnitude of the tendon tap(TT) response, and the slope of this regression line was compared for the involved and contralateral sides of each subject for both conditions.

RESULTS

Four of the five subjects showed a substantially greater slope on the impaired side than on the unimpaired side under resting conditions. Due to technical difficulties data from one subject was not analyzable. These slope values are listed in Table 1. Three of the five subjects showed a greater regression line slope (RMS TT vs. TT magnitude) when the subject was exerting minimal voluntary force as compared to resting conditions in both the impaired and unimpaired limbs. A complete set of data was not obtainable in two of the subjects due to an inability to maintain consistent background activity on the impaired side. The same procedure was performed on two neurologically normal subjects. There was no significant difference in the slope of the regression line of the TT response versus TT force between the two sides.

DISCUSSION AND CONCLUSION

While a significantly increased regression line slope (TT response vs. TT force) would be expected on the impaired side of hemiparetic stroke subjects, the result that the slope on the impaired side was increased during the condition of background activity as compared to rest seems to suggest that the greater slope on the impaired side can be attributed, at least in part to augmented motor unit recruitment. If the increase gain was due to an altered threshold only then a further gain increase would not be expected by a forced threshold change (voluntary activity). Further study is warranted (involving a greater number of subjects) for more conclusive results.

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	Number of successful tendon tap-RMS EMG trials (n)		RMSEMG-Tendon tap magnitude slope (volts/volts)	
Subject	Involved	Uninvolved	Involved	Uninvolved
KR	28	34	.1735	.01157
DM	19	35	.2131	.003537
BT	-	-	-	-
LN	73	20	.1916	.0707
CR	26	44	2.2141	.3551

Table 1. Tendon tap-EMG slope data

CORRELATION BETWEEN HEART RATE VARIABILITY AND SURFACE ELECTROMYOGRAPHY DURING DYNAMIC PHYSICAL EXERCISE.

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INTRODUCTION

Heart rate variability (HRV) has been used as a physiological marker of autonomic control of the heart in different situations such as dynamic physical exercise (DPE). It has been observed that young healthy individuals, when submitted to stress in the form of DPE, have a higher HRV at low effort levels and a lower HRV at higher work powers, with a concomitant slow increase in the heart rate (HR) response(3,5). The execution of DPE with progressive loads not only promotes changes at the cardiovascular system level but also induces an increased recruitment of muscle fibers. Some investigators have reported that surface electromyography (EMGs) is a useful tool for the assessment of the level of muscle recruitment during exercise since the procedure permits the analysis of myoelectric signals which are directly related to the degree of recruitment of motor units (MUs)(2,4). The objective of the present study was to analyze the correlation between HRV and EMGs during a continuous dynamic physical exercise test of the ramp type (TDPEC-R) on a cycle ergometer.

MATERIALS AND METHODS

The study was conducted on 10 healthy men aged 19 to 29 years (man: 23.7 ± 3.02 years) with an active life style. Before the tests, the subjects were submitted to clinical evaluation consisting of anamnesis, clinical examination and laboratory tests, resting electrocardiogram (ECG), physiotherapeutic evaluation, and maximum effort test. The subjects signed a formal term of consent to participate in the study after being informed about the procedures. The study was approved by the Ethics Committee of the Federal University of São Carlos. DPE tests were conducted at the same time of day to reduce the influence of the circadian variations, and in a controlled-temperature environment. The tests were performed using an electromagnetic braking Quinton-Corival-400 cycle ergometer automatically controlled by a microprocessor model Workload Program (Quinton-Groningen, Netherlands). The experimental protocol consisted of a TDPEC-R, with the volunteers resting for 60 s and then starting exercise with a 4 Watts power for 120 s. After this time, 10 W increments per minute were introduced up to a power of 140 W. The volunteers were monitored at the MC5 lead and the ECG signals were captured by a one-channel cardiac monitor (ECAFIX TC500) connected to a microcomputer by means of an analog-digital converter (Lab.PC+ National Instruments, Co.). HR was recorded on a beat-to-beat basis in real time and the R-R (RRI), in ms, were calculated using a signal processing program (6). HR was recorded 60 s before exercise, for 540 s during exercise, and 180 s after exercise. Integrated EMGs was collected from the vastus lateralis muscle using disposable surface electrodes positioned on the muscle belly at a distance of 2 cm from center to center after trichotomy and skin cleaning, coupled to an active bipolar terminal, which in turn was interconnected with a signal conditioner interfaced to a microcomputer. The EMGs signals were captured during the final 20 s of each minute from a computerized 17-channel electromyograph (Lynx Tecnologia Eletrônica Ltda.) using a signal acquisition program (AqDados5) with a sampling frequency of the analog-digital converter of 1000 Hz. For HRV analysis we used the RMSSD index which is the square root of the mean of successive differences between iR-R (ms) . RMSSD indices were calculated at 6 s intervals during the final 20 s of each minute at each power of the TDPEC-R. The 6 s window series were shifted every 3 s. For the analysis of EMGs we calculated the root mean square (RMS) of the last 10 muscle contractions captured during the final 20 s of each minute. Data were analyzed statistically by the nonparametric Spearman correlation test (α=5%).

RESULTS AND DISCUSSION

Figure 1 shows that during TDPEC-R with a power increment there was a decrease in the RMSSD index of the RRI (ms) with a concomitant increase in RMS index for the amplitude of the EMGs signal (μV). The calculation of the correlation between RMS and RMSSD showed that the rs index of - 0.69 was statistically significant ($p<0.05$).

With respect to HRV, the present results agree with literature data reporting that during a dynamic physical effort test at high levels there is a predominance of sympathetic stimulation of the sinus node associated with a decrease of vagal activity in the modulation of the HR response HR[1,5]. The increased MU recruitment observed through the increase in RMS may be attributed to the predominance of fiber type in relation to the specificity of their metabolism. This occurred to satisfy the demand provoked by the progressive increase in power [2,4]. Thus, there was an increase in HR in response to a stimulus mediated by a reflex neural afferent pathway originating from ergoreceptors and metaboreceptors of skeletal muscle in activity towards the cardiovascular center.

CONCLUSION

We suggest that the decrease in HRV with increasing power may be related to the reduction of parasympathetic modulation associated with a predominance of the sympathetic system over the NSA. This occurred in a parallel manner integrated with the degree of muscle fiber recruitment represented by the increase in the amplitude of theEMGs signal during physical exercise.

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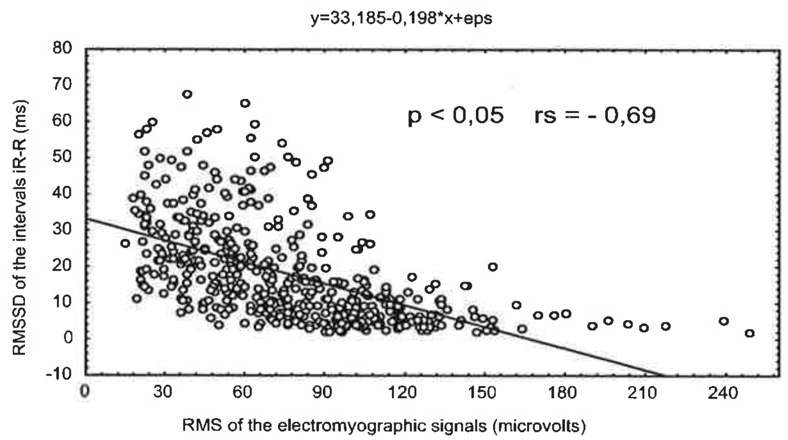

Figure 1: Correlation between the RMSSD index for the R-R intervals (ms) and the RMS of the amplitude of the electromyographic signals of the vastus lateralis muscle of the volunteers studied (n=10).

EFFECT OF MUSCLE MASS OF ISOMETRIC CONTRACTION ON HEART RATE AND TORQUE DEVELOPMENT

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INTRODUCTION

Isometric contractions cause an increase in heart rate (HR) characterized by a rapid initial response attributed to the inhibition of the functioning of the vagus nerve on the sinus node; this occurs up to the initial ten seconds (s) of exercise. Depending on the percentage of the maximum voluntary contraction and the time of contraction, the HR gradually increases, and this slow increase is predominantly due to sympathetic stimulation of the sinus node[1, 2]. With respect to the effects of the quantity of active muscle mass during isometric contraction on the increase of HR there is controversy, since there are studies that show that the magnitude of response of the HR to isometric exercise is greater when there is more muscle mass involved in the contraction[3,4], whereas SILVA *et al.* (1999)[5], studying maximum isometric exercise with different quantities of muscle mass in one or both legs, observed that the magnitude of response to HR was not influenced by the quantity of muscle involved in the contractile process. The objective of this study was to verify the response of the HR and the average torque values during maximum isometric exercise of the flexion and extension of the elbow.

MATERIALS AND METHODS

Eleven healthy men were studied, with ages between 19 and 28 years (23.4 +- 2.6 years) and with an active lifestyle. Prior to the tests they were submitted to a clinical evaluation that consisted of anamnesis, clinical and laboratory examinations, resting electrocardiogram (ECG), and a physiotherapeutic evaluation. The experimental procedures, and agreement to participate, were discussed and agreed upon, and a formal consent form was signed. This work was approved by the Research Ethics Committee of the University of São Carlos.

The tests were conducted at the same time of day to reduce the influences of the variations in circadian rhythms. The experimental protocol consisted of maximum isometric exercise tests of flexion and extension of the elbow of the dominant upper limb. During 10 s the volunteers maintained isometric contraction with resistance provided by the electronic dynamometer *Biodex Multi-Joint System II Inc.* Three contractions were made for each movement imposing variable time intervals between one contraction and the next. The volunteers carried out the exercises in a seated position with the articular angle responsible for the generation of the greatest torque being chosen according to the position of the elbow, that is 90° for flexion and 40° for extensions.

The volunteers were monitored in the MC5 lead, the ECG signals captured by a single channel cardiac monitor (ECAFIX TC500) were transferred by microcomputer by means of a digital analogue converter (LabPC+ *National Instruments, Co.*). The HR was recorded, beat by beat, in real time and the R-R intervals were calculated by means of a signal processing program[6]. HR was recorded as follows: 65 s prior to isometric force, 10 s during the effort and 120 s after the effort, corresponding to the recovery period. The torque measurements were provided by electronic dynamometer readings. To analyze the responses of the HR, the average of the HR deltas of the three contractions in each movement was calculated. The deltas were obtained by determining the average HR during the 65 s prior to the control recording and subtracting the HR peak found during the 10 s of physical effort. The average torque measurements for the three contractions of each movement were also calculated. For statistical analysis the nonparametric *Wilcoxon* test for paired samples ($\alpha = 5\%$) was used.

RESULTS AND DISCUSSION

There was no significant difference between the deltas of the HR in the flexion and extension movements (flexion = 38.0, and extension = 39.6 bpm, $p>0.05$), nevertheless the greatest torque measurement was generated during the isometric contraction for elbow flexion compared to elbow extension. (flexion = 60.6, and extension = 49.3 Nm, $p<0.05$). For elbow flexion four muscles are directly involved, brachial, brachoradialis, round pronator, and brachial biceps, besides the extensor carpi radialis longus that assists in movement, whereas for extension there are only two muscles working, the brachial triceps and the anconeus [(7)]. Murray *et al* (2000)[(8)] analyzed these muscles and concluded that the grouping of flexor muscles of the elbow presents quantitative and functional advantages over the extensor muscle grouping. This fact did not influence our results in terms of the magnitude of the HR response to the isometric physical effort, nevertheless it could have influenced the generation of torque measured during isometric flexion of the elbow. Additionally our results are in agreement with SILVA *et al* (1999)[(5)], who also did not find differences in the magnitude of the HR responses when different quantities of muscle mass involved in isometric contractions were compared.

CONCLUSION

The flexion movement of the elbow caused greater mean torque development than did the extension movements, with no differences in the variation of the HR when these movements were compared. Thus, these results suggest that there is no influence of the quantity of active muscle mass during isometric contractions on the magnitude of the HR response. Nevertheless, different muscle groupings could be responsible for different torque developments.

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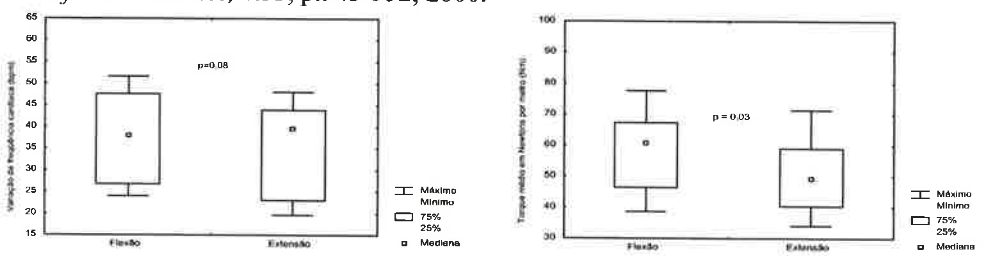

Figure 1: (A) Mean values of the variation in heart rate, in beats per minute (bpm), and (B) mean torque in Newtons per meter during maximum isometric exercise of flexion and extension of the elbow for the volunteers under study.

MOTION ANALYSIS SYSTEM IN SPORTS APPLICATION

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XPress-2000 is a motion analysis system which measures the pressure distribution in a shoe.

Applications may include gait analysis, rehabilitation assessment, shoe research and design, aid in shoe prescription and orthopedic design.

Sports medicine is another field of application for the Xpress-2000, e. g. the system can be used to optimize motions which may lead to lower strain on human joints and ligaments and therefore the risk of injury is minimized.

8 sensors per shoe are used to measure the pressure distribution in a shoe. This sensors can be individually placed on socks of the test subject. The sensors have sticky tapes and so it is no problem to place them on the socks. The offset for the sensors can be done by the computer.

The data is digitized and coded by a microcontroller and then sent via a telemetry route which may reach up to 1km to receiving unit which is fixed to a camcorder.The pressure data is recorded synchronous to the video recording of the motion on the audio track of the video tape.

The video camera produces 25 pictures per second. So we transmit one pressure sample per picture from each sensor. For the transmitting station we use 4 mignon batteries as voltage supply. With 600mAh rechargeable batteries you can work for more than 10 hours. The receiving station gets the power from the camera.

To analyze the measuring the video and audio data are sent to a PC system which uses the Xpress-2000 software to visualize the data.

INTERFACE

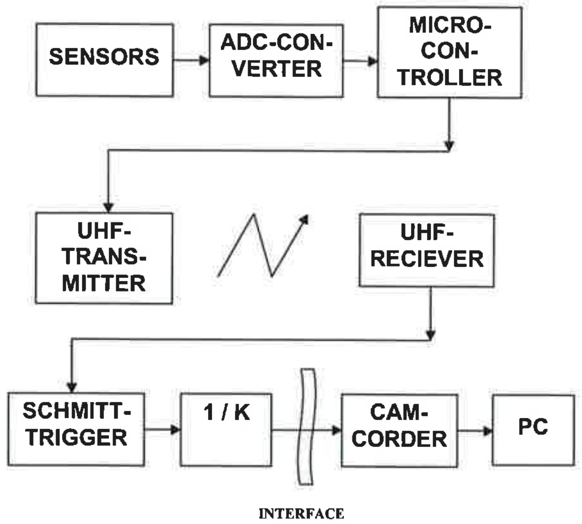
The technical details of the digital pressure dates,
recorded on the audio-track from the video-tape are:

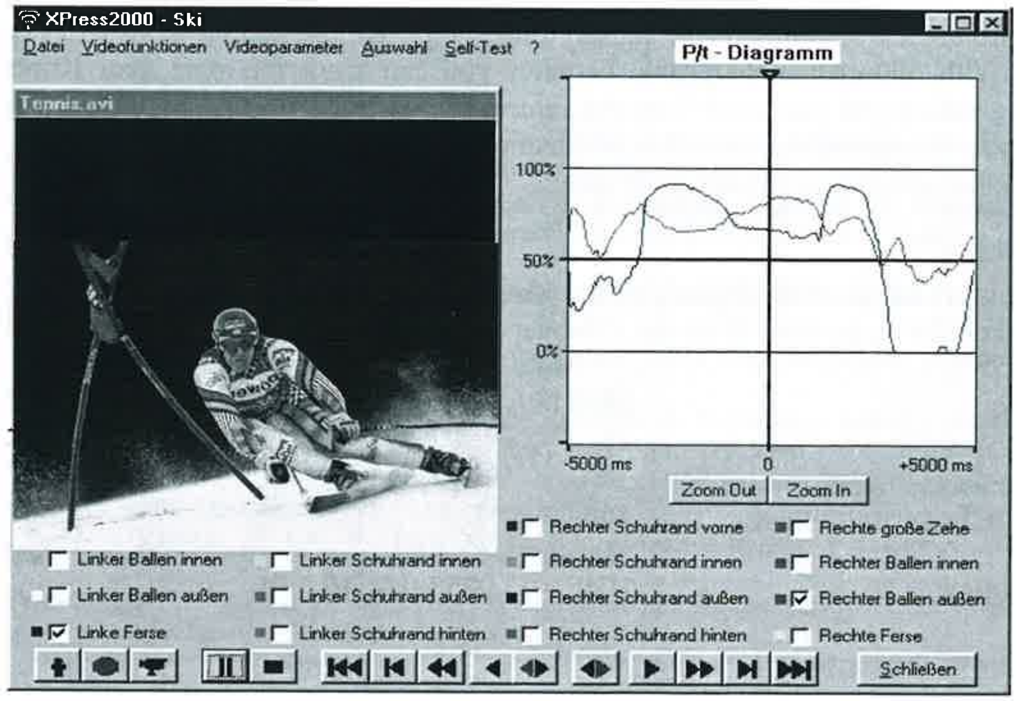
Baudrate:	6800 bit/s
Pictures/second:	25
Pressure-dates/picture:	16
Bitrate of the pressure-rate:	16bit
Code:	Manchester
Transmitting:	UHF, 433,925 MHz, 10 mW

Placement of the sensors:

1.	Big toe
2./3.	Ball inside and outside
4.	Heel
5.-8.	Edge of the shoe

Construction of the System





Visualizing pressure-dates in synchronisation to the video-picture with an assistance of pressure/time-diagramm.

ANALYSIS OF THE TORQUE AND EMG OF THE QUADRICEPS AFTER MUSCLE DAMAGE – A CASE REPORT

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INTRODUCTION

Skeletal muscle injury is common in sports activities and also during several conditions of work. The process of muscle regeneration as well the necessary conditions and limits of this process are well documented (Faulkner et al, 1993). Eccentric exercise has been used as a physiological model to induce muscle injury in humans (Lund et al, 1998). Skeletal muscle injury induced by eccentric exercise has been associated to force reduction (Nosaka & Clarkson, 1996) and changes in the EMG signals (McHugh et al, 2000). Then, the purpose of this study was to evaluated the maximal isometric torque of the quadriceps muscle and surface electromyography of the vastus medialis oblique and vastus lateralis muscles before, immediately after and during the first 7 days after injury induced by maximal eccentric exercise.

METHODS

One sedentary and healthy woman (23 years, 1.68m, 58Kg) was recruited to the study after providing informed consent. This study was approved by Ethics Committee for human studies of Federal University of São Carlos, São Paulo, Brazil. The participant did not have orthopedic pathology and was not previous involved in weight training. Injury was induced in the right quadriceps muscle by 4 bouts of 15 maximal isokinetic eccentric contractions (angular velocity of $5^{o}.s^{-1}$, range of range from 40° to 110° of knee flexion), using an isokinetic dynamometer (Biodex Joint System 2). The maximal isometric torque was evaluated using the same isokinetic dynamometer equipment, with the hip and knee joints positioned to 100 and 90 flexion degrees, respectively. The maximal isometric torque of the quadriceps muscle and the surface electromyography (EMG) of the vastus medialis oblique (VMO) and vastus lateralis (VL) muscles were always acquired simultaneously. The surface EMG was obtained using simple differential active surface electrodes and a digital analogue A/D converter (Lynx Electronics Technologies). The electrodes were fixed to the midline of the muscle belly with the detection surface perpendicular to the muscle fibers (De Luca, 1997). The reference electrode was fixed over the proximal anterior tibia shaft. To be sure that the electrodes were positioned every day in the same local, a plastic mould of the quadriceps was made. The EMG signals were amplified in the active differential electrode with a 10 gain, a rejection index by common modulation of 80 dB, a band width of 20-500 Hz with a roll-off of less than 12 dB/octave; a noise of less than 2 μV MSR (20-500 Hz), input impedance of less than 100 MΩ, low-pass filter of 450 Hz and high-pass filter of 10 Hz. The EMG signals were sampled in a synchronous manner at 1000 Hz frequency. EMG signals were processed (Matlab 5.0 software) to obtain the Root Mean Square (RMS/ μV). The maximal isometric torque and EMG were evaluated before and immediately after four bouts of maximal eccentric contractions, as well once daily during the following 7 days. The data were analyzed descriptively.

RESULTS

Immediately after eccentric exercise the maximal isometric torque of quadriceps decreased 52%, when compared to the pre-exercise level (from 145 to 70 Nm-Figure 1A). At the same period, there was also a decline of RMS values for both VL (decreased 48%, from 119 to 63 μV-Figure 1B), and VMO (decreased 30%, from 459 to 323 μV-Figure 1B).

DISCUSSION

The decrease of isometric torque and RMS immediately after the bouts of eccentric exercise, could be indicated failure in the mechanism of the muscle fibers contraction, probably associated to the muscle fibers fatigue. When the recovery period was analyzed, it was found different comportment among maximal isometric torque and RMS. The maximal isometric torque of the quadriceps gradually recovered pre-injury level until the 5th day (Figure 1A). Although the most reduction of RMS for both VMO and VL muscles occurred at the 2nd day after injury (63% e 66%, respectively), VL muscle recovered the pre-injury values of RMS at the 7th day, while VMO did not recovery it until the 7th (Figure 1B). Previous studies using eccentric exercise to induce injury in the quadriceps muscle also found decrease in both muscle force and RMS (Hortobágyi et al, 1998).

CONCLUSION

The results of this study, within of the experimental conditions used, permit the conclusion following: a) immediately after the bouts of eccentric exercise there was a decline of the maximal isometric torque of the quadriceps muscle, as well of the RMS in both VL and VMO muscles; b) the behavior of the isometric torque of the quadriceps and EMG signals of VMO and VL muscles differed during the regeneration period, especially when VL and VMO were compared; d) maximal isometric torque of the quadriceps muscle is completely recovered a long of one week.

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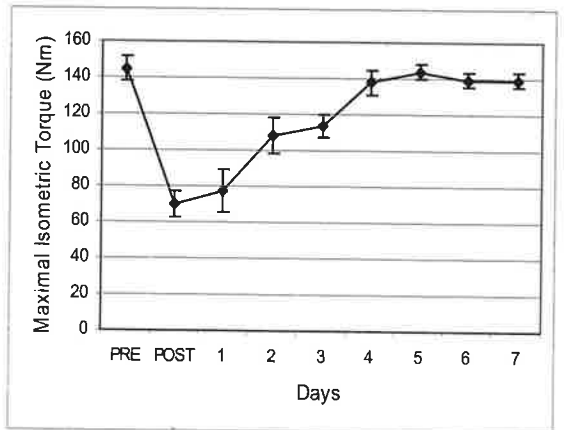

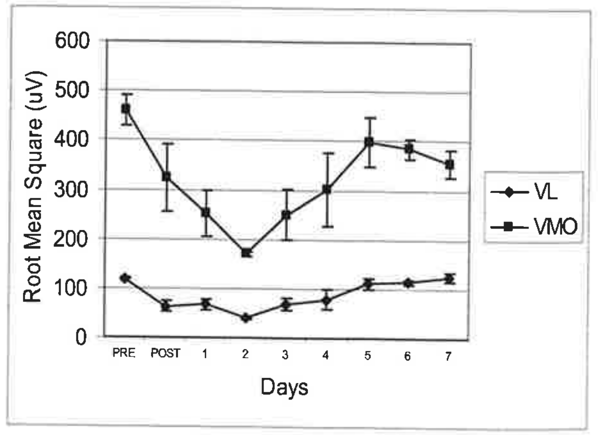

FIGURE 1- Maximal isometric torque (A) and Root mean square (B) evaluated pre, immediately post eccentric exercise, and in the following 7 days after exercise. Results are mean ± standard deviation.

EFFECT OF PROLONGED VIBRATORY STIMULATION ON MOTOR EVOKED POTENTIALS IN LOWER LIMB MUSCLES.

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INTRODUCTION

Previous works in the upper extremity have shown that vibratory stimulation (VS) can enhance corticospinal excitability as measured with transcranial magnetic stimulation (TMS). Motor evoked potentials (MEPs) recorded in the vibrated muscle are larger [1 2], while those recorded in the antagonist muscle are depressed [3]. The facilitatory effect of VS on MEPs has been attributed to increase excitatory drive at the spinal and cortical level arising from activation of primary and secondary spindle afferents in the vibrated muscle [1, 2]. In these studies, the vibration-induced modulation was studied only for brief periods of application (e.g., 1-5 sec). However, in clinical settings, VS is often applied for longer duration (e.g., 1-2 min) to alter motor responses in weak or spastic muscles [4]. In the present experiment, we investigated with TMS the effect of prolonged VS on MEPs in lower extremity muscles.

METHODS

Twelve healthy subjects (age: 18-54 years; 10 M, 2 F) participated in this study. Subjects were seated comfortably with both lower extremities relaxed at 80° of hip flexion and 45° of knee flexion. TMS was delivered using a MagStim 200 (MagStim Co., London, UK) with a 90 mm circular coil. The coil was adjusted over the vertex to evoke optimal responses in the target muscle of the dominant lower extremity. To ensure reproducibility, the site was identified with the aid of a tight fitting bathing cap marked with a co-ordinate system. MEPs were recorded with surface electrodes placed over the *quadriceps femoris* (QF), *biceps femoris* (BF) and *soleus* (SOL) of the dominant leg. Vibration was produced using a commercial therapeutic vibrator (Wahl Massage Master, CA) with a rounded contact surface of 2 cm in diameter. The vibration (2 mm amplitude @60 Hz) was applied continuously for 5 minutes over the patellar tendon. The amount of pressure was adjusted so to avoid induction of a tonic vibration reflex. EMG activity was continuously monitored to ensure that voluntary muscular activity or TVR were not present. The effect of BF tendon vibration was also studied in three males subjects.

In each subject, MEPs were recorded at rest and then at specific time intervals during and after vibration (see Fig.1). Five to 10 MEPs were recorded at each time interval. The amplitude and latency of each MEP was measured off-line and averages were derived for each set of trials. The mean amplitude values were normalized to the mean control values at rest (100%) and then pooled for all subjects. A one-way repeated-measures ANOVA was used to compare changes in amplitude and latency of MEPs across time intervals.

RESULTS

As shown in Fig 1, vibration of the patellar tendon greatly facilitated MEPs in both the QF (A) and BF (B), right at the onset. This facilitation continued to build up in the first minute and then started to decline gradually at 90 sec and 210 sec. A similar pattern of facilitation with the same time course was seen in the remote synergist SOL (mean increase of 276±66% @ 30 sec VS). Immediately after VS was ended, MEPs returned close to their resting values for all three muscles. Interestingly, MEPs in QF were depressed by 48%∀10% and 64%∀8% one and two min post-VS, respectively (Fig1A). No such post-vibration depression was seen in BF and SOL. There was no statistically significant change in MEP latencies for all three muscles during and after MV. In the small group of subjects tested for the effect of BF tendon vibration (n=3), a similar pattern of modulation to that seen with patellar stimulation was observed (i.e., large facilitation in the vibrated muscle BF with somewhat less facilitation in the antagonist muscle QF).

DISCUSSION

Consistent with previous studies, the present study has shown that prolonged VS can greatly enhance corticospinal excitability in lower limb muscles. Our results differed from previous studies, however, in that the effect appeared to be less selective in terms of agonists-antagonists. The facilitation seen in the antagonist BF during patellar vibration is most likely attributable to propagation of VS to the surrounding tissues because of the mode of application chosen (i.e., large amplitude vibration with a large vibrating probe). The fact that comparable effects were seen in QF when vibrating the BF tendon concurs with this interpretation. The time course of the VS-induced facilitation was such that the greatest enhancement occurred within the 1st minute of application and then tended to decline very slowly. This finding provides support to the recommendations of limiting VS to 2 minutes in the context of clinical applications. Clinicians must be aware, however, than propagation of vibratory stimuli may produce spreading facilitatory effects not limited to the vibrated muscle.

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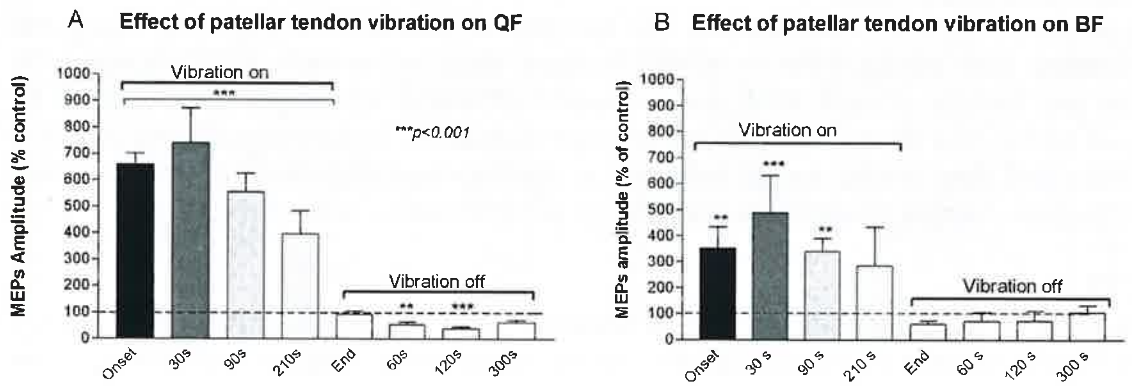

AFFECT OF DIFFERENT INTENSITIES AND TONES OF AUDITORY STIMULATION ON MOTOR CORTEX EXCITABILITY

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INTRODUCTION

Cortical activity relating to voluntary motor control has been widely studied using the techniques of transcranial magnetic stimulation (TMS) of the human motor cortex to elicit motor evoked potentials (MEPs). TMS is one of the most useful non-invasive methods for estimating the motor cortex excitability during voluntary movements in humans. It has been reported that a positive linear relationship is established between amplitude of MEPs and force level (Hess et.al., 1987, Benecke et.al., 1988, Mazzocchio et. al., 1994).

Voluntary contraction during the target movement is affected by the auditory information, visual confirmation, sensibility, actual exerted performance and other feedback. However, few attempts have been made to identify the mechanism of these phenomenon (Furubayashi et. al., 2000).

In this study, we investigated the effect of auditory stimulation on the human motor cortex during 5% maximal voluntary contractions (MVC).

METHODS

Thirteen healthy volunteers aged 22-29 years old were selected. All the subjects provided informed consent for the study. All the subjects were right-handed according to their writing and exercising behavior. None of them had a previous history of neurological illness.

Subjects were seated on a comfortable chair with their eyes open and asked to look forward at an oscilloscope (SS-6122A SYNCHRO SCOPE IWATSU). To maintain a constant voluntary contraction, visual feedback of EMG activities was given to subjects by an oscilloscope in front of them and sustained 5% maximal EMG as accurately as possible.

Transcranial magnetic stimulation - Motor cortex was stimulated using a Magstim Model 200 magnetic stimulator with a round coil of 90 mm mean diameter windings. The round coil was placed at the optimal position for eliciting MEPs from the target muscle. The stimulus intensity was approximately 10% above relaxed threshold.

EMG recordings - EMG recordings were made using surface Ag/AgCl electrodes, 0.9 cm diameter, fixed over the right first dorsal interosseous muscle (FDI). Signals were amplified with filters set at 10 Hz and 3 kHz (NeuropackΔ NIHON KOHDEN) and digitized at the sampling rate of 10 kHz (Maclab, AD Instruments). A reference electrode was placed over the wrist joint. The size of MEP responses to TMS given at different times after an auditory stimulation were compared with responses to TMS alone (control response).

Experimental protocol and Auditory stimulation - Different intensities and tones of sound were used in the three experiments. In the first experiment with loud auditory stimulation, the sound was composed of 35 trains of 135 dB (SPL) pulses of 0.1 ms duration. Between the onset of each pulse was 6 ms. The interval between auditory stimulation and TMS ranged from 10-200 ms in 10 ms increments. Auditory stimulation was presented to both ears through the headphone. In the second experiment with different tones of auditory stimulation, we used two tones, a high tone (composed of 50 trains of pulses of 0.1 ms duration, between the onset of each pulse was 1 ms) and a low tone (composed of 5 trains of pulses of 1 ms duration, between the onset of each pulse was 10 ms) at 60, 70, 80, 90 and 100 dB. In this experiment, the effects of auditory stimulation on the motor cortex were examined at 30 ms. In the third experiment, the auditory stimulation which was composed of 9 trains of 135 dB

(SPL) pulses of 0.1 ms duration was presented to one-sided ear through the headphone. In this experiment, the effects of auditory stimulation on the motor cortex were examined at 30 ms.

RESULTS and DISCUSSION

Loud of auditory stimulation suppressed MEP responses to TMS when it preceded the magnetic stimulation at a peak of 30 ms in all subjects. This result show that inhibition of motor cortex by loud auditory stimulation occure at a peak of 30 ms. However, MEP responses between 10-20 ms and 60-200 ms did not change significantly. While high auditory tone stimulation suppressed MEP responses to TMS at 100 dB, low tone sound stimulation did not give rise to anything. This result indicate that tone of the sound component affect the excitability of motor cortex. Furthermore, Loud of auditory stimulation to one-sided ear also suppressed MEP responses to TMS when it preceded the magnetic stimulation at 30 ms.
This reslult show that the integration of auditory information of one-sided ear in order to inhibit motor cortex occure after auditory cortex.

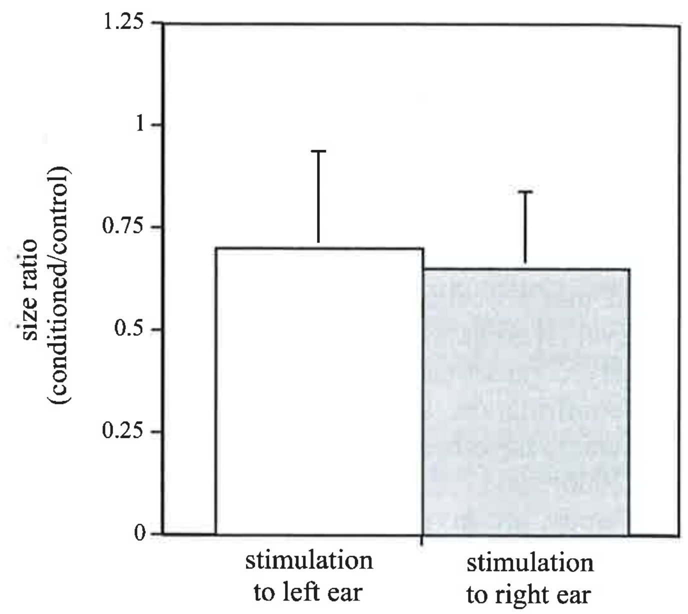

Fig. 1 Effect of left motor cortex excitability relating to auditory stimulation of one-sided ear.

CONCLUSION

These results show that inhibitory action on the motor cortex is highest at 30 ms after loud auditory stimulation and inhibitory action on the motor cortex by auditory stimulation to one-sided ear is in the same way. Additionally, tone of the sound component seems to be related to the magnitude of motor cortex excitability.

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INHIBITORY EFFECTS ON SOLEUS H-REFLEX INDUCED BY DIFFERENT ELECTRICAL STIMULATION PARAMETERS

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INTRODUCTION

Therapeutic Electrical Stimulation (TES) is a popular modality commonly used to decrease the excitability of motoneurons in spastic patients. Several studies have been reported that the repetitive electrical stimulation modulated the soleus H-reflex amplitude in healthy subjects. However, a number of clinical studies have been reported that TES could reduce spasticity in hemiplegic subjects, while others have found either no change or even increase spasticity after TES. The diversity of these findings may be due to variation in patients and in stimulation parameters. Furthermore, the optimal stimulation parameter to reduce spasticity has not been determined. The purpose of this study was to investigate the differences in effects of the preceding conditioning electrical stimulation at different pulse duration, frequencies and intensities for changes of the soleus H-reflex amplitude.

METHODS

Experiments were performed on six healthy volunteers (aged from 24 to 57) after obtaining informed consent. The subjects were comfortably seated and the left leg was fixed with the hip at 90, knee at 120, and ankle at 90 degrees. The control soleus H-reflex was recorded during rest period before applying the conditioning stimulation. The H-reflex was evoked by stimulating the left tibial nerve at the popliteal fossa with a single rectangular pulse of 1 millisecond duration. The intensity of test stimulus was adjusted to obtain a small M-wave with H-reflex. The recording bipolar surface electrodes were placed on the lower part of the soleus. The conditioning electrical stimulation was applied for 5 seconds via surface electrodes to the left common peroneal nerve at the level of the caput fibulae. The frequency of the conditioning stimulus was varied at 20, 50, 100, 200, 333 and 500 Hz. In the conditioning stimulation of 20Hz, the pulse width was set at 0.2 or 1 milliseconds and the intensity was decided at 0.8 or 1.2 times of the threshold for the evoked M-waves (motor threshold: MT) from the left tibialis anterior muscle. In the conditioning stimulation except for 20Hz, 1ms of pulse width and 0.8MT of intensity were used. The time intervals between the test and the conditioning stimulation were varied in steps ranging from 1 to 5 seconds. In each session, the test H-reflex was evoked 1, 2, 3, 4 and 5 seconds after the end of the conditioning stimulation.

RESULTS

Figure 1A shows the changes of the soleus H-reflex amplitude at the stimulation frequency of 20Hz. A significant inhibition of the H-reflex amplitude was observed in 1.2MT-0.2ms and 1.2MT-1ms at the conditioning to test interval of 1 second. On the other hand, soleus H-reflex was hardly inhibited at 0.8MT-1ms. Figure 1B shows the changes of the soleus H-reflex amplitude at the different frequency. At the conditioning to test interval of 1 second, a significant inhibition of the H-reflex amplitude was observed at 200Hz-0.8MT-1ms. As a result, soleus H-reflex was inhibited at the parameter of 20Hz-1.2MT-0.2ms, 20Hz-1.2MT-1ms and 200Hz-0.8MT-1ms at the conditioning to test interval of 1 second. However, the statistical significant differences were not admitted among the three parameters.

DISCUSSION

In the present study, we examined the differences in effects of the electrical stimulation at different parameters for changes of the soleus H-reflex. A disynaptic reciprocal Ia inhibition has been thought to be a mechanism to inhibit antagonist muscle motoneuron in spinal cord. Furthermore, it has been reported that the group Ia fibers are more easily excited by low intensities of stimulus than alpha fibers. When comparing pulse widths and intensities in the conditioning stimulation of 20Hz, the soleus H-reflex was hardly inhibited at 0.8MT-1ms. However, lower intensity without pain is desirable for the conditioning stimulation because the stimulation of 1.2MT intensities could occasionally give subjects some pain. It has been reported that the pulse of 1ms duration stimulate group Ia fibers selectively at threshold. Therefore, we examined the changes of soleus H-reflex at the different frequency of 0.8MT-1ms. The soleus H-reflex was inhibited the most strongly at 200Hz, and was hardly inhibited at 500Hz. Since pulse interval between the first and second stimuli was short (2 ms) at the frequency of 500Hz, the second impulse might be input during the refractory period after the first action potential. As a result, the excitement of the group Ia fibers might not be induced by 500Hz. On the other hand, it is thought that there was no influence of the refractory period in 200Hz. Furthermore, since reciprocal Ia inhibition lasts for only several milliseconds, the inhibition of soleus H-reflex in 200Hz at least 1 second after the end of conditioning stimulation can be based not only on a reciprocal inhibition mechanism but also on other mechanisms. One mechanism we suppose is the posttetanic potentiation induced by the temporal summation of high frequency stimuli (200Hz)

CONCLUSION

We found that, in the conditioning electrical stimulation with 0.8MT-1ms to the common peroneal nerve, the frequency of 200Hz was the most effective to inhibit the soleus H-reflex.

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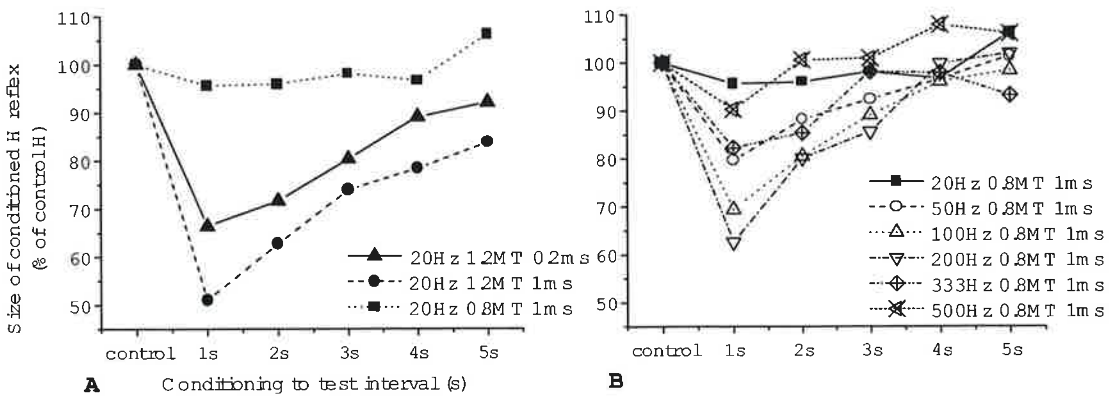

Fig. 1 A B Means of the amount of soleus H-reflex inhibition induced by the different conditioning stimulus at different conditioning to test intervals.

CORTICO-MOTOR THRESHOLD TO TRANSCRANIAL MAGNETIC STIMULATION: VARIABILITY WITH AGE, DOMINANCE AND PARTICIPATION IN SPORTS

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INTRODUCTION

Cortico-motor threshold (CMT) is a widely used parameter in studies using transcranial magnetic stimulation (TMS). Because it is influenced by drugs that affect sodium and calcium channels, CMT is considered a primary index of neuronal membrane excitability[1]. At the representational level, CMT provides information about a central core region of neurones in the muscle representation. In healthy controls, studies that have looked at the effect of factors such as age and dominance on CMT intensities have produced conflicting results with the majority finding no effect with either one or both factors. That dominance have no consistent effect on CMT seems surprising giving the evident asymmetry in skill performance between the two hands. In a recent study, Pearce et al[2] reported that skilled racquet players had lower CMT and an enlarged motor representation in the hemisphere controlling the playing hand. Thus, the level of skills seems to be a factor when studying asymmetries of CMT in hand muscles. In the present study, we used the procedure outlined by Mills & Nithi[3] to estimate CMT at rest in representative muscles of the upper and lower extremity to determine how factors like age, dominance, and participation in racquet sports influence threshold values.

METHODS

Twenty-three healthy adults (age: 18-57 years, 15 M, 8 F) participated in the study. Of 23 participants, 15 (12 M, 3 F) practiced racquet sports (badminton, tennis, squash) on a regular basis (i.e., 3-10 hours weekly). CMT was determined at rest on both sides in the first dorsal interosseous (FDI) and in the quadriceps (QF) using a Magstim 200 (MagStim Co, Wales, UK). Motor evoked potentials (MEPs) in FDI were evoked using a double 70 mm coil, while a 90 mm circular coil was used for QF. The procedure for threshold determination consisted of finding first the maximal intensity that produced no responses out of 10 stimuli. The intensity was then gradually increases at 1 % step to find the minimal intensity that produced 10 MEPs out 10 stimuli. These two values define the lower (LT) and upper threshold (UT), respectively. For the QF, the series of corresponding magnetic stimuli was set to seven to avoid coil heating. The CMT intensity was derived by taking the mean from the LT and UT values. Corticospinal excitability was also assessed by examining inter-hemispheric ratios (Dominant/Non Dominant) of the amplitude of responses evoked during high intensity magnetic stimulation (60% for FDI, 90% for QF) in each muscle. The effects of age, gender, dominance and participation in racquet sports on CMT values and MEPs amplitude were determined with an ANOVA (three fixed factors and one covariate, significance level, *$p<0.05$*).

RESULTS

The ANOVA revealed that CMT measured in FDI were not influenced by age, gender or participation in racquet sports. In addition, threshold values for the two hands were comparable (Fig. 1A). In QF (n=19), age turned out to be a significant factor accounting for ~25% (Fig. 1B) of the variance in CMT measured in both the dominant (r^2=0.25, p=0.03) and non-dominant leg (r^2=0.22, p=0.04). As for FDI, CMT did not vary between the two legs (Fig.1A). Although MEP amplitude tended to be higher in the dominant hand (mean inter-hemispheric ratio 2.17± 1.75), the inter-individual variability was large and none of the factors (age, sports or gender) proved to be significant. In the QF, the mean inter-hemispheric ratio was close to one (1.45±1.2, n=17).

DISCUSSION

In the present study, we found that dominance; age and gender had little influence on CMT measured at rest in FDI. Similar findings have been reported by Mills & Nithi [3]and others[4], Interestingly, our CMT values for FDI were very close to those reported by Mills & Nithi (i.e., 40-45%), an indication of the validity of the method. In addition, our results showed that regular practice of a racquet sports did not seem to translate into difference in CMT between hands. The fact that our sample consisted mainly of recreational players, as opposed to Pearce et al who studied elite players, may explain the absence of asymmetry. In QF, we found that CMT at rest were affected by age (Fig 1B). As pointed out by Rossini et al[5], several mechanisms may account for the elevated CMT found in older subjects: 1) an age-related loss of low-threshold pyramidal track neurones (PTN), 2) desynchronization of descending volleys elicited by TMS due to changes in axonal conduction of PTN, or 3) a depressed excitability at the spinal level. Recent evidence from cat experiments suggests that the 2[nd] mechanism (i.e., change in central conduction) may play a significant role during aging[6]. The fact that age differentially affected CMT measured in the upper and lower extremity may reflect the greater vulnerability of corticospinal projections to lower limb muscles to age-related changes in the CNS.

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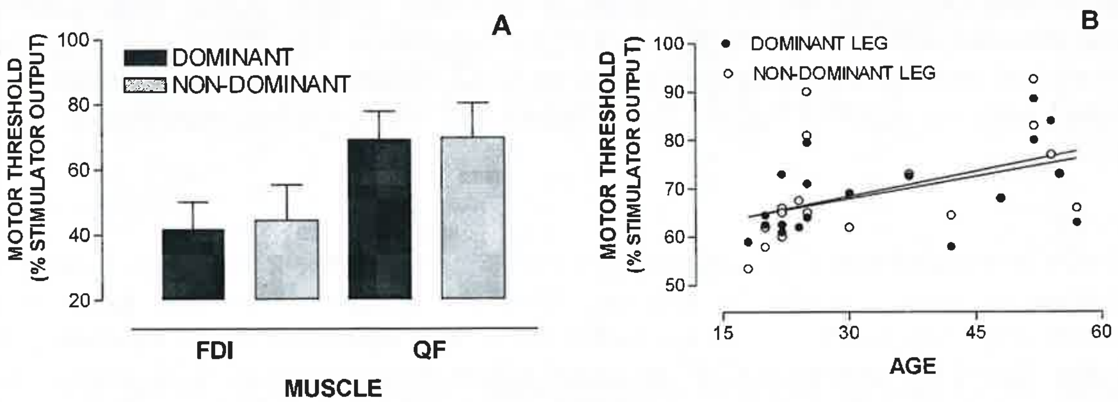

TRANSIENT DEPRESSION AND DELAYED FACILITATION OF MOTOR EVOKED POTENTIALS BY PROLONGED EUROMUSCULAR ELECTRICAL STIMULATION OF THE QUADRICEPS FEMORIS MUSCLE

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INTRODUCTION

NMES is a modality commonly used in the rehabilitation setting to facilitate or retrain weak muscles. NMES is typically used with contraction/relaxation cycles by means of electrodes placed over the target muscle or directly over the mixed nerve. Thus NMES induces a combination of both afferent and efferent information from the muscle and its afferent. Previous studies with transcranial magnetic stimulation (TMS) have shown that peripheral afferents have on important role in modulating cortico-motor plasticity. For instance, Ridding et al., (2000) reported an increase in cortico-motor excitability of the thenar muscles after prolonged electrical stimulation of the median nerve. Similar results have been found in the hypothermal muscles with electrical stimulation of the ulnar nerve, Kealing-Lang et al.,2002). In the present study, TMS and QF H-reflex were used to study the short-term effects of a single session of prolonged NMES on the cortico-motor excitability of knee muscles.

METHODS

Seventeen young healthy (25±5yrs) subjects participated in the study. Subjects were seated comfortably with both lower extremities relaxed. TMS was delivered using a Magstim 200 with 90 mm circular coil. The coil was adjusted over the vertex to evoke optimal responses in the target muscles of the dominant lower limb. To ensure reproducibility, the site was identified with the aid of a tight fitting bathing cap marked with a co-ordinate system. MEPs were recorded at rest with surface electrodes placed over QF and biceps femoris (BF). QF M-response and H-reflex were measured in 5 subjects to assess changes in spinal excitability. Percutaneous cathodal electrical stimulation (Grass-S88) of 1 ms pulse duration was applied to the femoral nerve at the inguinal fossa. The NMES (Respond Select II) was delivered on the femoral triangle (300µsec, 15 Hz) with duty cycle of 10 s. The intensity was set to produce a full knee extension against gravity. Changes in MEPs amplitude were measured at specific time intervals both during (5, 15, 30 min.) and after NMES (30 min. post). The QF M-response and H-reflex were measured pre and 30 min.of the NMES.and 30 min post NMES The peak-to-peak amplitude and latency of each MEP, M-response and H-reflex were measured by averaging 8-10 responses in each subject. Changes in amplitude were expressed in percentage of the resting control values. The non-parametric Friedman test ($P<0.05$) was used to compare the changes in each condition.

RESULTS

In a majority of subjects, MEPs in the QF (13/17) and BF (11/11) were depressed (35% on average during the period of NMES ($p<0.01$). In the period post-NMES (30 min.), MEPs were facilitated (QF by 175±18% and BF by 160±15%, Fig.1B). For the remaining subjects, the pattern of modulation was reversed with facilitation during and returns to control or depression in the period of post-NMES. In five subjects in whom QF H-reflex was measured, no major change in amplitude (±10%) was seen during and after NMES. QF M-response, H-reflex and MEP latencies did not change significantly across experimental conditions.

DISCUSSION

These results indicate that NMES of the QF tends to produce a transient depression in the excitability of the cortical projections to knee muscle during the application (Fig 1A). This effect seems cortical in origin since no noticeable change in QF M-response and H-reflex were noted. The most probable physiologic explanation include: NMES-induced suppression of motor cortical excitability via intracortical inhibition (GABAergic modulation), post-exercise depression (central fatigue) or a gating phenomenon due to movement displacement or to NMES by itself. Other potential mechanisms include antidromic volleys in lower limb motor neurons, hyperpolarisation, recurrent inhibition (Renshaws), activation of autogenic Ib circuits or afferent mediated (group III and IV) depression.

The major finding in the present study is that somatosensory stimulation results in delayed yet substantial increase in the amplitude of QF MEPs following NMES. This results confirms the important role of peripheral afferent inputs from the lower limb muscles on motor cortical plasticity. This delayed facilitation is similar to that reported by Ridding et al. (2000) after 2h stimulation of the median nerve and those of Kaelin-Lang et al. (2002) after 2h stimulation of the ulnar nerve in hand muscles. Our results suggest that the effects are less specific for lower limb muscles, since both the QF and its antagonist (BF) were affected by NMES. The late facilitation seen after NMES (up to 30 min.) is likely cortical in origin since no change in spinal excitability or QF-M-response would have persisted for such duration. The change in corticomotoneuronal excitability elicited by NMES is similar to a post-exercise facilitation of MEPs reported in arm muscles after voluntary contraction (Samii et al., 1996), although this phenomenon occurs only at short intervals (up to 1 s) after contraction (possibly a short term potentiation). It is tempting to suggest that similar but long lasting intra-cortical mechanisms (e,g. long-term potentiation) might be involved in the present experiment.

NMES can be useful in clinical setting for the rehabilitation of brain-injured and stroke patients as well as post- immobilization period following a musculoskeletal injury.

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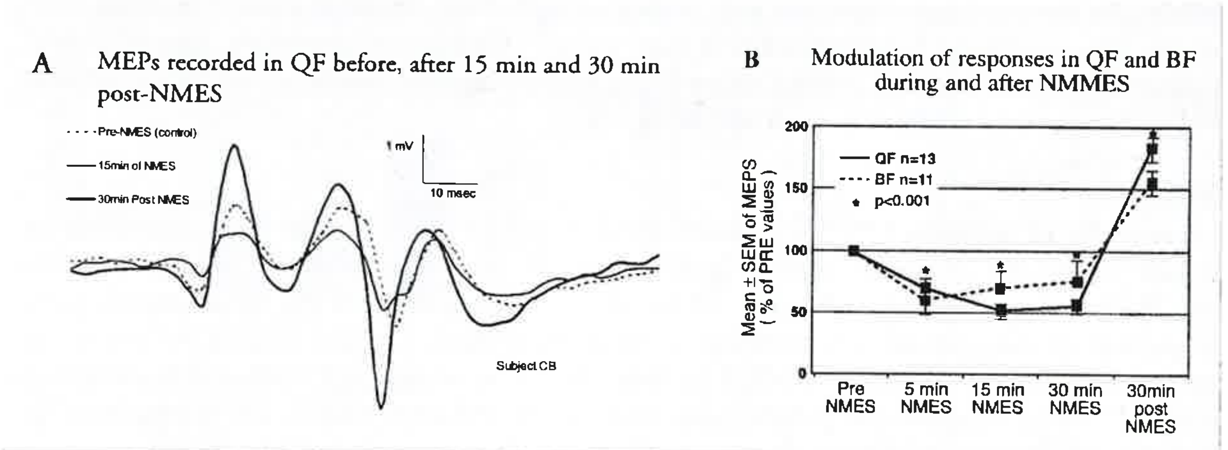

NERVE EXCITATION OF BULLFROG IN AN INHOMOGENEOUS MEDIUM EXPOSED BY MAGNETIC FIELD

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INTRODUCTION

Magnetic stimulation has been used for many clinical studies since Baker's study[1] of transcranial magnetic stimulation on human body. It is, however, difficult to find the local region of nerve excitation, because the current flows in the wide range of human body. But the case of human nerve magnetic stimulation is reported that the maximum excitation tends to come when the induced current is turned on perpendicularly to the nerve fiber[2]. This study focused on the characteristics of nerve excitation for each direction of the current induced by magnetic stimulation.

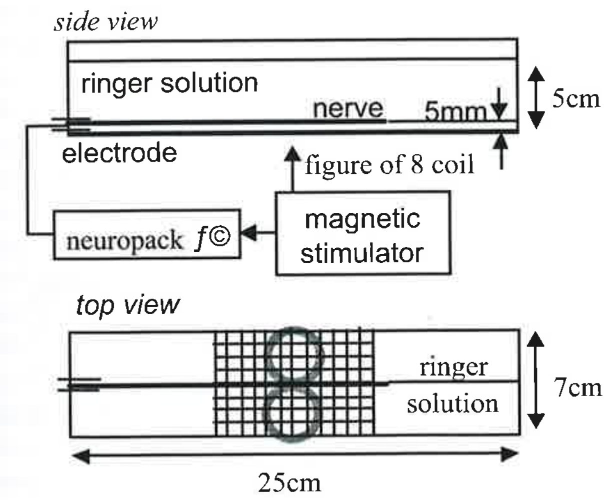

Fig.1 Measuring system with sciatic nerve of bullfrog

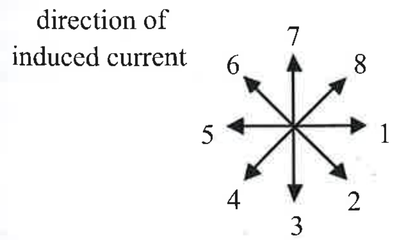

Fig. 2 Direction of stimulation current.

METHODS

10 sciatic nerves taken out from bullfrogs were set straight up in the volume conductor filled with ringer solution as illustrated in Fig. 1. The nerve action potential was recorded by using a suction electrode. We used figure of 8 coil with 2.5cm in diameter, which is attached to the magnetic stimulator (Mode 1200 , Magstim, England). The nerve action potential was recorded by Neuropack Λ(Nihon-Kohden, Tokyo). The nerve is laid above the intersection of coil in ringer solution, and excited by magnetic stimulation for each eight directions as shown in Fig. 1. The intensity of stimulation was 40%(0.8T)

RESULTS

When the changed direction of the induced current by turning the coil stimulated the sciatic nerve, the amplitude of the action potential was measured. The respective data of the nerves were standardized by each maximum amplitude. The action potential was almost none when the induced current was flowed perpendicularly to the nerve fibers in homogeneous volume conductor. Especially, nerve excitation tended to become larger when the induced current was flowed parallel to the nerve fibers. The amplitude of action potential in the direction to the periphery, 4 or 6 in Fig.2, was larger than that in the direction to the central nerve, 2 or 8 in Fig. 2.

DISCUSSION

The distance of virtual cathode can be calculated by both latency difference between action potential in the direction 1 and that in the direction 5 and nerve conducting velocity. The Fig.2 Normalized amplitude of action result in this measurement was 25.6±4.1[mm] potential for each direction of induced (n=10, A.V ± S.D.). Also, the latency of action current in homogeneous volume potential with the induced current flowed in the conductor.
direction of 1 was shorter than that with the induced current flowed in the direction 5. In the case of figure of 8 coil with 25mm in diameter virtual cathode is considered to deviate from the coil intersection to the induced current by 13mm (half of 25.6mm) when the induced current is flowed parallel to the nerve fibers. This experiment is on magnetic stimulation of peripheral nerve under the assumed condition that nerves fibers are straight in the body. A report says that curved nerve increase excitation. There seems more factors which affect the characteristics of nerve excitation by magnetic stimulation.

CONCLUSION

This study confirmed virtual cathode is the key to focality through the *in vitro* experiment using bullfrog's sciatic nerve. Presumptively inhomogeneity of a living body would affect distribution of induced electric field, in which generated virtual cathode would induce nerve excitation. It is necessary to carry out experiment using the model with inhomogeneity of a living body, and to clarify focality of magnetic stimulation in the human peripheral nerve.

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