NORAXON EMG MEETING 2006

Lecture Abstracts

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Dear Reader,

We are pleased to present you with the abstract proceedings of the NORAXON EMG MEETING 2006 held in Muenster/Germany. In 20 lectures, 18 workgroups from Germany, Austria and Belgium presented aspects of their current EMG projects. The main idea of this meeting was to bring active EMG users in contact and facilitate communication between workgroups.This meeting booklet is educational freeware and can be downloaded and distributed between colleagues at any time.

The variety of topics represents the wide use of kinesiologic EMG. The discussed topics range from basic considerations about processing techniques to applied use of EMG biofeedback in therapy regimes.

We would like to thank all the speakers again for their efforts in creating additional abstracts in English and granting public access to their presentations. We invite you to contact the authors directly by Email if you have any questions, comments or need more information about their current EMG research.

We are in the planning stages for the next meeting in May 2007. Please check the NORAXON (www.noraxon.com) or Velamed (www.velamed.com) web page for actual dates or send us an Email. (On a final note, we would like to thank the Velamed Germany team for their outstanding efforts in organizing the meeting.

We hope to see you at the next NORAXON EMG MEETING!

Best regards

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PERONEAL REACTION TIME MEASUREMENTS IN THE DIAGNOSIS OF ANKLE JOINT INSTABILITY

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INTRODUCTION

The ankle joint is a location that is prone to injuries in sports activities as well as in daily life. Even though the initial injury – an excessive inversion event causing an ankle sprain or ligament injury – can be treated effectively with conservative measures or with surgical procedures, a certain percentage of patients may develop recurrent problems which are termed chronic ankle instability. The problem is quite common especially in young persons participating in high-risk sports demanding a high level of jumping and directional changes.

In the literature, a distinction has been proposed between mechanical and functional instability. The term mechanical instability is more or less clearly defined in terms of increased talar tilt and/or anterior talar drawer due to a mechanical insufficiency of the lateral ligaments. The functional instability, however, is more difficult to diagnose as it has been described as a feeling of giving way, recurrent inversion events and a tendency for swelling after joint loading. No clear-cut criteria or parameters have been proposed and are generally accepted for routine clinical use. The problem is complicated by the fact that these two entities may appear separate but also in a combination in patients.

The cause for the functional instability problem is thought to be related to a neuromuscular insufficiency –some kind of strength deficit – and/or a proprioceptive problem caused by impairments of sensory structures and their feed-back.

The lateral ankle ligaments are the passive stabilizing structure that are not strong enough to withstand unprotected loading of the joint in inversion. Therefore, the peroneal muscles are the active stabilizers that should prevent excessive joint motion and prevent damage to the ligament complex. If the activity of the peroneal muscles (peroneus longus and brevis) appears in a timely fashion, their eversion moment can help to counteract the inversion movement and prevent an impending injury.

METHODS

This fact has lead to research interest that concentrates on the assessment of the respective muscle activity, especially peroneal reaction times (PRT). Measurements are usually performed with surface electromyography that can be recorded from both muscles, the peroneus longus and brevis. An established injury simulation model is a trap-door or tilting platform that induces a sudden inversion, usually limited to a range of 30° (Fig. 1). This extent is not harmful for the subjects but is demanding enough to elicit a reactive response from the muscles of the lower leg.

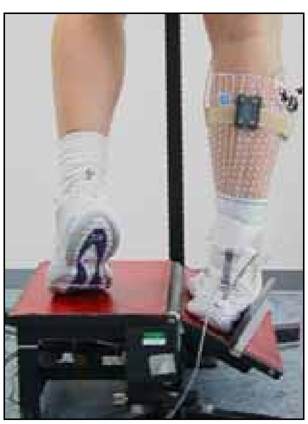

Fig. 1: Set-up for peroneal reaction time measurements on a tilting platform

Electrodes are placed in a bipolar arrangement with an inter-electrode distance of 20 mm on the muscle belly of the peroneus longus (approx. 3 cm below the head of the fibula) and the peroneus brevis (approx. 10 cm above the lateral malleolus). These locations were prepared with abrasive paste and cleaned with alcohol to ensure skin impedance below 6 MΩ. The electrodes were connected to the amplifier unit and A/D-converted at 1000 Hz per channel for raw signal storage (Noraxon MyoSystem & MyoResearch).

In the first stages, the signals were semi-automatically analyzed with self-developed software that detected the onset of muscle activity when the average baseline activity ± 2 standard deviations were exceeded (Fig. 2). More recently, a similar approach was incorporated in the MyoResearch software allowing convenient data processing.

RESULTS

These procedures were applied in healthy subjects as well as in patients that were referred to the hospital for treatment of chronic ankle instability. We used the approach to distinguish patients with and without proprioceptive deficits who might benefit more from neuromuscular exercises than from surgical reconstruction of the ligaments (Rosenbaum et al. 2000). The knowledge about the potential neuromuscular cause of ankle instability lead to the development and application of exercise programs for injury prevention which were evaluated with reflex latency measurements (Eils et al 2001).

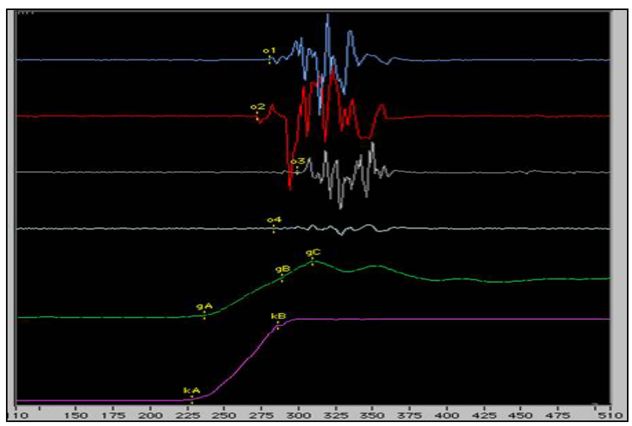

Fig. 2: Example of the EMG signals (PB=blue, PL=red, TA=gray, MG=white) and rearfoot angle (=green) from a single measurement.

DISCUSSION

Even though the is some evidence that peroneal reaction times may benefit from proprioceptive exercises, i.e. lead to a shorter reflex latency it has to be realized that even normal reaction patterns are too slow to effectively prevent lateral ligament overloading. Therefore, only pre-activated muscle activities anticipating the potentially dangerous situation appear to be able to prevent ankle injuries (Konradsen et al., 1997). These effects might be supported by passive stabilizers that provide external support to the ankle joint.

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DATA PROCESSING ALGORITHM FOR REFLEXES OF THE KNEE SENSORIMOTOR SYSTEM

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INTRODUCTION

After injuries to the anterior cruciate ligament (ACL), a functional instability is frequently observed which has been attributed to a disturbed sensorimotor function. In light of the clinical importance of ACL injuries and the resulting functional instability, it is of enormous interest to elucidate the role of sensorimotor pathways that involve the ACL. In animals and humans a direct reflex pathway between the ACL and the hamstrings has been shown. Previous studies only showed a monophasic reflex response after ventral tibia translation after 40-55 ms (1, 2) which was regarded as medium latency response (MLR). However, ventral tibia translation should also induce a stretch of the hamstring muscles and evoke a short latency response (SLR). Before any muscle response after ventral tibia translation can be ascribed to anatomical structures, it is crucial to analyze the obtained muscle reflexes carefully. The aim of the present study was the development of an algorithm to differentiate SLR and MLR responses after ventral tibia translation.

METHODS

In ten healthy subjects reflex responses of the hamstrings after ventral tibia translation and after tendon taps on the biceps femoris were evaluated. To investigate the influence of skin afferents, control experiments were performed after lidocaine injection of the dorsal calf. The subjects were standing upright with 30° knee flexion and 5° external rotation. The thigh was blocked in ventral direction above the patella, which was pressed against a counter pressure plate. An accelerated piston generated a force of 300 N in a posterior-anterior direction 10 cm below the knee joint gap parallel to the tibia plateau in an angle position of 30°. Tibia translation was detected by a potentiometer position transducer. Hamstring reflex activity was recorded with pairs of disk electrodes placed above muscle bulge of the medial and lateral hamstring in the middle between knee joint gap and buttocks gap. EMG signals were amplified and recorded at a sampling rate of 5000 Hz. EMG signals were rectified, averaged and band-pass filtered (10-700 Hz).

Given that the reflex pathway of the SLR corresponds to the tendon jerk reflex first the hamstring tendon jerk reflex was elicited. This monosynaptic response was divided into three parts: from the first onset to the first peak (A), peak to peak (B), from the second peak to the end (C). The duration of these three sections was determined. It is important to note that the onset latencies and the first peak latencies of the tendon jerk reflex are very similar to the corresponding latencies of the first part of the SLR component after tibia translation. Assuming corresponding reflex pathways of the tendon jerk reflex and the SLR component, this would allow predicting the end of the SLR within the complex muscle response after the tibia translation. In the tendon jerk reflexes a constant relation can be shown between the first part and the remaining parts. This constant relation allows for the assessment of the duration of the tendon jerk reflex by just measuring the first part of the EMG signal.

RESULTS

The onset of the tendon jerk reflex showed a mean latency of 21.9 ms. No difference were found between the medial and lateral hamstring. The overall duration of this reflex was 15.7 ms. For section A we found a duration of 4.8 ms, for B 5.8 ms and for C 5 ms

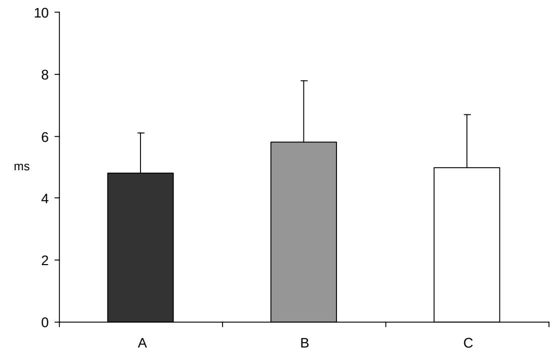

Fig. 1: Latency of the three sections of the tendon jerk reflex. A. onset to first peak; B. peak to peak; C. second peak to end

The duration B and C was expressed as a ratio of section A. The duration of interval B was significantly longer compared with A and C. The mean ratio B/A was 1.24, the mean ratio C/A was 1.04. Thus, the mean factor to calculate the end of the tendon jerk reflex using the duration of section A is 3.28.

Due to the complex hamstring muscle response after tibia translation the onset of the MLR is superimposed by the end of the SLR. With the calculated factor 3.28, the end of the SLR could be reliably assessed, which defines the onset of the MLR response independent from the different signal configuration of the superimposed hamstring reflex response. The onset of the SLR after tibia translation was 20.3 ms. Compared with the mean onset latency of the tendon jerk reflex of 21.9 ms there was no statistically significant difference.

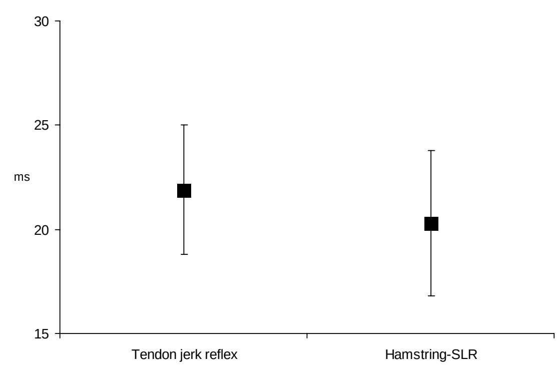

Fig. 2: Comparison of SLR latency between tendon jerk reflex and hamstring SLR. No significant difference was found.

Using the algorithm the mean onset of the hamstring MLR was 38.9 ms. Additionally, in the five subjects investigated after lidocaine injection of the dorsal calf, no significant differences of the SLR and the MLR were found compared with the untreated trials. Therefore, a possible contribution of cutaneous afferent fibers to the hamstring reflex response can be excluded.

DISCUSSION

The main finding of the present study was that an anterior tibia translation evokes a reflex response that could consist of two main parts: the first occurs at a latency of around 20 ms, which corresponds to the latency of the SLR; the second component occurs at 39 ms, which corresponds to the latency of the MLR. Furthermore, an algorithm was developed which allows a safe differentiation between the SLR and MLR components.

Faist et al. (3) reported for the tendon jerk reflex of the biceps femoris latencies of 20 ms, which is comparable to the present results. Taking into account the low range of tendon jerk reflex latencies, a latency of more than 30 ms after the onset of the tibial movement cannot be regarded as a SLR and was therefore regarded as a MLR and a missing SLR. Previous studies (1,2) reported only a monophasic reflex response of the hamstrings after tibia translation which was found after 40 ms and 55 ms, respectively. This may be due to a different way to detect the onset of tibia translation. However, this reflex response was not characterized as MLR.

CONCLUSION

It was demonstrated that by measuring the first part of the SLR from the onset to the first peak the end of the SLR could be predicted and the onset of the MLR component can be assessed reliably. The fact that both SLR and MLR components can be observed after anterior tibia translation underlines the necessity to differentiate the responses before they be ascribed to any anatomical structures. As a basis for future work, the presented algorithm may become a useful tool to differentiate which afferent pathways play a role in initiating hamstring activity.

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FINE WIRE EMG OF THE TIBIALIS POSTERIOR AND FLEXOR HALLUCIS LONGUS MUSCLES IN GAIT AND RUNNING

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INTRODUCTION
Overload running injuries of the lower extremity, particularly the knee, are associated with excessive pronation of the foot resulting in tibial rotation (Nigg et al., 1995). M. tibialis posterior (TP) and M. flexor hallucis longus (FHL) muscles are shown to have an active influence on pronation and the medial longitudinal arch (Gray & Basmajian, 1968; Kaye & Jahss, 1991). Their functional role during running and interaction with footwear is still not clearly understood (Reber et al., 1993; O'Connor & Hamill, 2004). Therefore, the purpose of this study is to investigate the influence of different footwear on the muscles' EMG pattern.

METHODS
11 volunteers (6 for FHL/11 for TP) were tested while walking (1.6 m/s) and rearfoot running (3.0 m/s) on a treadmill (video controlled; 250 Hz). Intramuscular EMG of the TP, the FHL and surface EMG of Mm. peroneus longus, tibialis anterior, soleus, gastrocnemius medialis was recorded (Noraxon, 1500 Hz) under five different conditions: barefoot (BARE), minimal shoe (FREE), conventional running shoe (CS), motion control shoe (MCS) and unstable shoe (MBT). For data analysis the EMG-signals were filtered (20Hz high pass), rectified, smoothed, and amplitude normalized to barefoot maximum. To quantify the myoelectric activity of the muscles the integral (IEMG) of the processed signal was calculated for stance phase and a pre-innervation time of 150 ms.

RESULTS
For the first 7 analyzed subjects in all shoe/barefoot conditions EMG-activity of TP and FHL showed greatest mean and IEMG values between 50-70% (walking) and 35-50% (running) of stance. In the MBT condition the IEMG of TP was statistically higher than in the BARE condition. The linear envelope of TP and FHL EMG-signals over pre-innervation and stance phase are similar to that of the other detected plantar flexors.

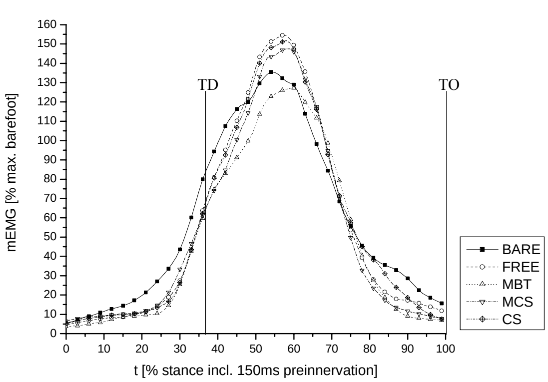

Fig. 1: Average (6 subjects; 10 trials) mEMG curves of FHL during walking running (2) with the events TD, TO and pre-innervation (150 ms) under different conditions.

The maximum of the amplitudes appeared earlier in running than in walking while the pre-innervation was higher in running.

DISCUSSION
TP and FHL EMG-activity peaks in the same phase of running where maximum eversion of the rearfoot is reported (Nigg, 1986). This confirms the theory that TP and FHL tend to counteract the pronation. It seems that the time-history of EMG-activity of the FHL is highly dependent on the velocity of locomotion. The elevated midsole of the unstable shoe attended with a greater lever arm for the horizontal forces acting upon the subtalar joint may result in higher external ankle moments in the frontal plane. It could be hypothesized that in this unstable situation higher TP EMG-activity may be needed to compensate higher ankle moments.

CONCLUSION
This study for the first time showed the EMG-activity of the FHL in running. It could also be shown that footwear design has an influence on TP EMG-activity during treadmill running. Ongoing analysis of IEMG in different phases of stance and its relation to foot kinematics in the frontal plane should lead to a better understanding of the role of TP and FHL and their interaction with their antagonists in walking and running.

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COMPLEX BIOMECHANICAL ANALYSIS FOR PROSTHETIC AND ORTHOTIC TREATMENTS BY USING THE VICONPEAK-KISTLER-NORAXON SETUP

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INTRODUCTION

Biomechanical analysis methods play an important role in prosthetics and orthotics for:

a) Fundamental research;
b) Individual optimization of a prosthesis or orthosis on the patient;
c) Realization of studies regarding functional testing of prosthetic and orthotic components, and
d) New development of prosthetic and orthotic components.

The presentation shows how biomechanical measuring methods can be used to optimize the prosthetic alignment (adaptation of the prosthesis to the locomotor system) on biomechanical basis, so that largely physiological standing and walking becomes possible for the amputee. The method is demonstrated on the basis of prosthetic alignment instructions for transtibial prostheses.

METHODS

In a first experiment, 17 transtibial amputees were examined in order to record long-term adaptations to the situation after the fitting with a transtibial prosthesis (minimal time period after prosthetic fitting: 2 years). The gait was recorded with an opto-electronic camera system (Vicon460) coupled with two Kistler force measuring plates, and at the same time the surface EMG of the vastus lateralis muscle and biceps femoris muscle (NORAXON MyoSystem 2000) was registered. The prosthetic alignment was measured using the LASAR POSTURE static measuring device (/1/). In a second experiment, 5 characteristic alignment situations of 8 transtibial amputees were measured systematically using the same measuring methods.

RESULTS

The results of the first part of the experiment showed that in consequence of the long-term use of the prosthesis the following biomechanical characteristics can be proven:

1. About 60% of the amputees show physiological knee flexion on the prosthetic side during the stance phase; the other amputees have their knee largely extended during the stance phase.
2. The external joint moments on the knee joint are characterized by the fact that the flexion moments, which normally occur during the first half of the stance phase, are considerably reduced on the prosthetic side's joint.
3. The EMG measurements show a considerably reduced muscle activity of the knee extensors on the prosthetic side in comparison with the contralateral side. The ischiocrural muscles, on the other hand, show increased intensity of activity as well as an extended phase of activity (Fig. 1, /2/).

The experiments on prosthetic alignment variations showed high correlations between static and dynamic measuring values. There is an optimal alignment situation, for which biomechanical reasons can be given (/3/) and which can easily be found by the prosthetist for each patient using the corresponding adjustment possibilities of the prosthesis, provided that the static measuring device is used. This alignment situation is achieved, if the load line in the sagittal plane runs approximately 15 mm ventrally to the compromise rotation axis of the knee joint (/4/) and in the frontal plane through the mid-foot and lateral patella edge, and if the physiological stance phase flexion is used while walking (/3/).

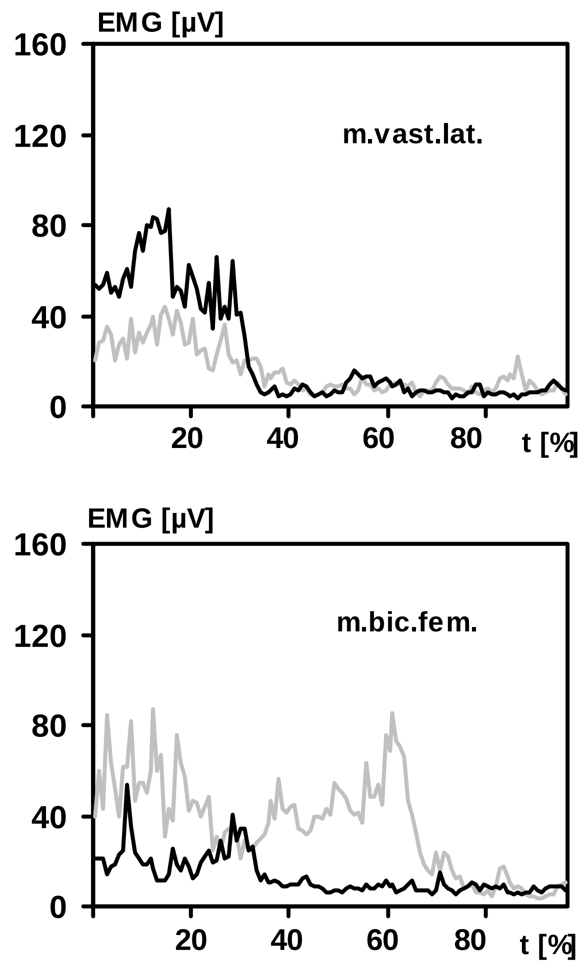

Fig. 1: Typical stance phase surface EMG of a transtibial amputee (grey: amputated limb, black: non-amputated limb)

DISCUSSION

Using the biomechanically founded alignment situation, the transtibial amputee will be enabled to weight the knee joint of the prosthetic side in approximately physiological manner during standing and walking. The EMG measurements show that there is a high correlation between mechanical parameters and muscle activity.

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WAVELET BASED ANALYIS OF MUSCLE ACTIVATION WHILE RUNNING ON DIFFERENT SURFACES

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INTRODUCTION

Several studies revealed no systematic changes of ground reaction forces depending on stiffness of midsole [3; 4]. As a possible explanation, footwear dependent muscle activation is considered. This is viewed as an area for future research because of the potential interaction between boundary conditions in running and muscular activation [5]. Therefore the purpose of this study was to investigate muscle activity depending on running surface condition in order to a) quantify surface related changes in EMG signals and b) investigate inter-individual/gender specific differences in muscular activation.

METHODS

20 female (26±5yrs, 62±5kg, 170±5cm) and 27 male (26±5yrs, 75±8kg, 180±5cm) subjects participated in the study. Surface EMG was recorded (Telemyo 2400T Noraxon®, 3000 Hz) from mm. tibialis anterior, peroneus longus, gastrocnemius lateralis, semitendinosus, vastus medialis and tensor fasciae latae while running (v=3.75±0.25 m/s) barefoot on grass, barefoot on tartan and shod (Straprunner V, Nike) on tartan.

Data were analyzed using wavelet based software, which allows for simultaneously performing frequency, time and intensity analyses by transforming the signals into activity patterns (fig.1) [6]. Activity patterns were computed firstly for each subject (10 steps per surface condition). After a qualitative analysis of the computed activity patterns subjects were organized into different groups, in which subjects with similar activity patterns were collected. Then mean activity patterns were computed for the different groups.

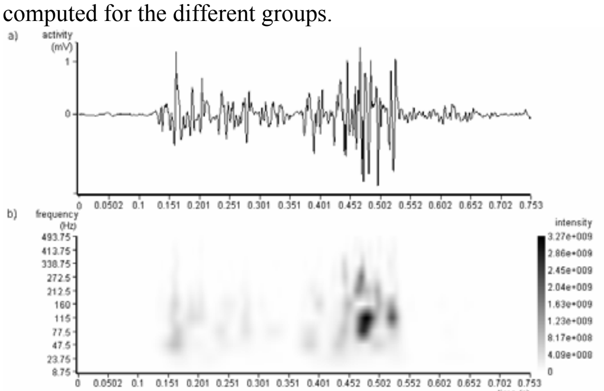

Fig. 1: a) Recorded EMG signal of m. tibialis anterior (one step of running barefoot on tartan) and b) corresponding activity pattern. In activity pattern abscissa represents time, ordinate represents frequency and grey scale represents intensity of the recorded EMG signal.

RESULTS / DISCUSSION

The evaluation of activity patterns of mm. tibialis anterior, semitendinosus and tensor fasciae latae revealed differences in muscle activation between the subjects. The electromyographic behaviour of these muscles was used to group the subjects into different groups. There were systematic differences in activity patterns of several muscles between the groups. Especially activity patterns of m. tibialis anterior revealed conspicuous differences in muscle activation between the subjects (example for grouping in figure 2).

Systematic differences in muscle activation of several muscles between the groups refer to distinct movement behaviour, which gives reason to further studies with simultaneous kinematic/kinetic measurements.

Referring to all of the tested muscles there were differences in muscle activation between men and women. These may be related to differences in body weight, connective tissue, anthropometry or dynamic segment alignment and refer to gender specific movement behaviour. Gender specific differences in movement are described e.g. by [1]. Ferber [2] considers gender specific injuries in connection with gender specific running mechanics. It seems to be advisable to examine the muscular activation in interconnection with kinetic/kinematic analyses to gather information about gender specific movement behavior and thereby aroused loadings.

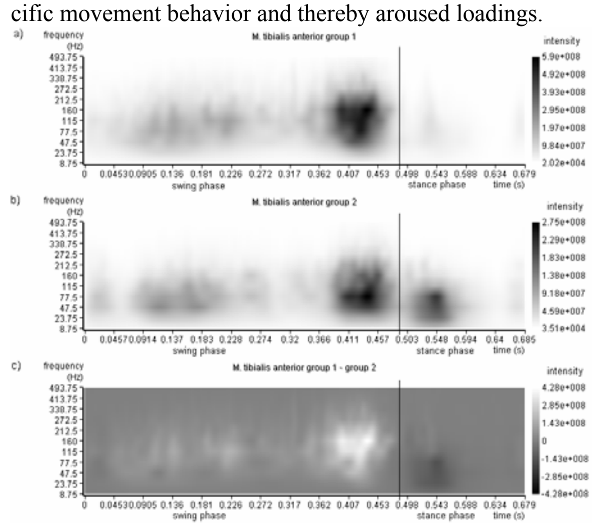

Fig. 2: Activity pattern of m. tibialis anterior while running barefoot on tartan for different groups: a) group 1 (n = 21), b) group 2 (n = 18), c) difference of activity patterns of group 1 and group 2: light zones represent higher intensity in activity pattern of group 1, dark zones represent higher intensity in activity pattern of group 2.

Systematic increase or reduction of muscle activation of the tested muscles depending on running surface could not be detected. However, changes in muscle activity with changing surface could be identified. These changes were highly individual and muscle specific. In mean, the muscular response to surface changes was gender specific. Gender specific muscular response to surface changes is to consider with respect to differences between muscle activation patterns of men and women. It may be also related to gender specific discrepancies in factors like e.g. body weight, connective tissue, anthropometry or dynamic segment alignment.

CONCLUSION

Muscle activity pattern as well as muscular response to changing surfaces showed to be gender specific. Surface related changes in muscular activation were muscle specific. Further studies may treat a possible interaction between muscle activity and anthropometric/kinematic/kinetic data, which could lead to a deeper understanding of gender specific movement and may contribute to new findings concerning footwear design or injury genesis.

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EMG SIGNAL-PROCESSING

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INTRODUCTION

Electromyography (EMG) is a powerful method to analyze non visible activity of muscles. The processing of the signal amplitude and the analysis of onset and offset of EMG provides information about the moment, duration, intensity, and course of electrical activity of muscle. Frequency analysis provides information about fatigue, spectrum characteristics, recruiting, and selection of muscle and fibre types. EMG is also a superposition of action potentials representing all fibre types and is of stochastic nature. The demodulation of EMG and the translation in an analogue output that takes place at the motor units and leads to muscle force. Therefore, in performing essential and required methods of signal processing described in the following, we must always consider the stochastic and modulated content of the EMG signal. The overall guiding principle should be to change of the original EMG as little as possible and only as much as necessary.

SIGNAL AMPLITUDE

The first step in signal processing is the zero-line-correction if necessary. Subtracting the mean value of the whole EMG from every sample is the easiest, non-signal affecting procedure. Using a high pass filter of 3-10 Hz is commonly used, but, is more or less, a slight signal affecting procedure. The advantage of high pass filtering is the removing of movement artifacts, particularly in dynamic measurements.

The second step in signal processing is the rectification of EMG. In addition to the visual information about more or less activity of muscle, the maximum of the rectified EMG results in the peak EMG, a value with low meaning. The area under the rectified EMG, the so called integrated EMG (IEMG), as well as the equivalent mean EMG related to a time sector, results in the first detailed information about muscular activity.

The third step in signal processing is filtering and smoothing of rectified EMG to obtain a more precise envelope and activity pattern of EMG. The usually used method is the low-pass filtering with about 30 Hz. For technical reasons older studies used Butterworth filters of 2nd order to eliminate a phase shift. Digital sampling now allows one to also use digital filters of 1st order without phase shift and is less influential of the EMG-signal.

A rather simple filtering method is the moving average. For example, using a time area of 30 ms, but its limitation is reduced quality against low-pass. The best envelope provides the hardly known method of signal power from moving FFT e.g. with 32 base points (Jöllenbeck, 1999). A rather rough envelope results when using the method in step 2, described IEMG even found in the literature, as a filtering method. The results of filtering respectively smoothing, allows a qualitative, and within limits, a quantitative analysis of EMG concerning coordination and activity of involved muscles.

To provide information about of inter- and intramuscular activity as well as interpersonal comparisons, the next step may be amplitude normalization of EMG by using the isometric MVC in a defined position. It is recommended to use the maximum of rectified and filtered EMG of an MVC for normalization and to express the resulting EMG level as a percent of MVC. To do so, a limited amount of EMG is lost but is usually reproducible. In case of cyclic movements, time normalization may be necessary as well and EMG will be expressed in relation to the percentage of cycle or phase. To do so, time information of the original EMG is lost and is not reproducible.

ONSET AND OFFSET

EMG provides information about non-visible activities and forces. Between EMG and force output there is a time shift called electromechanical delay (EMD). Therefore, to get information from EMG about the mechanical effectiveness of muscle the exact knowledge of EMD is necessary. Unfortunately, the literature demonstrates that variability of the results is large as it relates to EMD without a standardized method and threshold values.

It has to be distinguished between a) the onset at the beginning of activity as the time shift between the first activity of EMG and the concerning force, the so called, electromechanical activation delay (EMAD) and b) the offset at the end of activity as the time shift between the last activity of EMG and the end of the concerning force, the so called, electromechanical relaxation delay (EMRD) (Jöllenbeck, 2001). Onset and offset are determined using the rectified EMG.

The first method to determine the onset is the visual method using a zoomed signal area and is done manually. The method of fixed threshold values is easy but doesn't consider different signal qualities, amplification and noise or dynamic movement which suggests higher thresholds. The method of percentage values may solve the problems of fixed threshold values but noise and dynamic problems remain. The method of using the multiple (e.g. 3-times) standard deviation of EMG signal before activity considers noise and dynamics but is highly dependent on both. The last method to be found in literature uses extreme low pass filtering with 2.5 or 3 Hz resulting in a distortion of EMG content and therefore has to be rejected.

method / muscles		pretreatment of EMG and force	EMG threshold	force threshold	EMD [ms] (*stdev*) from / to
VIS		zoomed Window of 300 ms width and 25% height (related to the maximum value of EMG and force), 'threshold' is determined by the first continuous onset			18,6 (*6,6*) / 22,0 (*8,5*)
FIX_i	KE, KF EE		40 µV, 60 µV, 80 µV 10 µV, 25µV, 50 µV	2 N, 5 N, 10 N 1 N, 5 N, 10 N	14,5 (*9,1*) / 40,4 (*12,5*)
PER_i			3%, 5%, 5%, 10%, 10%, 20%	1%, 2%, 5%, 5%, 10%, 20%	24,4 (*8,4*) / 43,9 (*12,7*)
STD_i			SD 3x, 4x, 5x above mean value		17,9 (*9,7*) / 29,8 (*14,9*)
ELF_i		3 Hz low-pass	3%, 5%, 10%, 25%, 50%, 75%		70,9 (*11,4*) / 128,3 (44,6)
FMA_i	KE, KF EE	EMG: MA 16ms, cos² weighted; force: LF 30Hz, hamming weighted	40 µV, 60 µV 10 µV, 25µV	2 N, 5 N 1 N, 5 N	12,5 (*5,8*) / 30,9 (*12,4*)
PMA_i			3%, 5%	1%, 2%	22,6 (*7,0*) / 37,4 (*11,1*)
SMA_i			SD 3x, 4x above mean value		26,6 (*11,2*) / 45,3 (*16,1*)
FLF_i	KE, KF EE	EMG and force: LF 30Hz, hamming weighted	40 µV, 60 µV 10 µV, 25µV	2 N, 5 N 1 N, 5 N	12,3 (*6,6*) / 32,4 (*12,1*)
PLF_i			3%, 5%	1%, 2%	24,6 (*6,6*) / 42,4 (*12,0*)
SLF_i			SD 3x, 4x above mean value		27,3 (*12,7*) / 44,3 (*16,2*)

Tab. 1: Methods, pre-treatments, threshold values and EMD, index in ascending order related to threshold values.

Comparing all existing methods and different threshold values to determine onset and offset with the same input data results shows wide variability of and absolutely no reliability between methods and thresholds (Jöllenbeck, 2001). That means results found in the literature using different methods or thresholds to determine EMD or onset or offset are not comparable. The best method to determine onset and offset is the visual method, but it is very time-consuming. The best automatic method seems to be the method with percentage values of the moderately filtered (e.g. 30 Hz low pass) EMG signal, but a loss of accuracy of about ±20-30 ms has to be considered. An extensive analysis could show that a) the value of EMAD is about 20 ms in an optimal muscle length, up to 10 ms longer in other muscle lengths and b) EMRD is about 100 ms, but ranging from 60 ms up to 120 ms. When concluding mechanical activity from EMG, both EMAD and EMRD must be considered

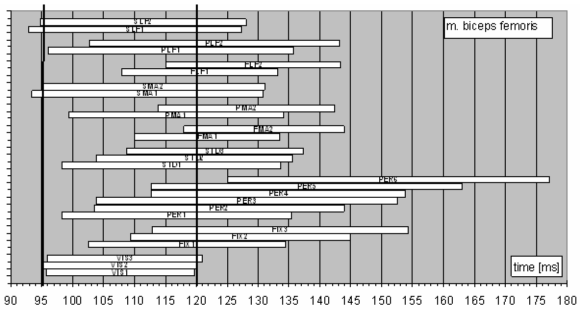

Fig. 1: Time shift of EMD caused by different methods and threshold values exemplary using m. biceps femoris; left end of the beams represents the EMG onset the right end the force onset.

FREQUENCY ANALYSIS

Frequency analysis usually is performed by FFT or, more recently, by Wavelet transformation. Analyzing EMG signals with FFT is a very helpfully tool to ensure that there is no electrical interference of e.g. 50 Hz (Europe) or 60 Hz (USA). Mostly frequently used parameters from frequency analysis by FFT are the median frequency (median value of spectrum) and the mean frequency (mean value of spectrum). Usually the median frequency is a little lower than mean frequency. Both parameters provide information about fatigue e.g. by a decrease during activity, called "frequency shift". The amount of base points for FFT as a potential of 2 depends on the time window to be analyzed. Looking for fatigue, usually 512 or 1024 base points are used following window by window. Looking for frequencing or recruitment order, the use of fewer base points (e.g. 64 or 128) is recommended.

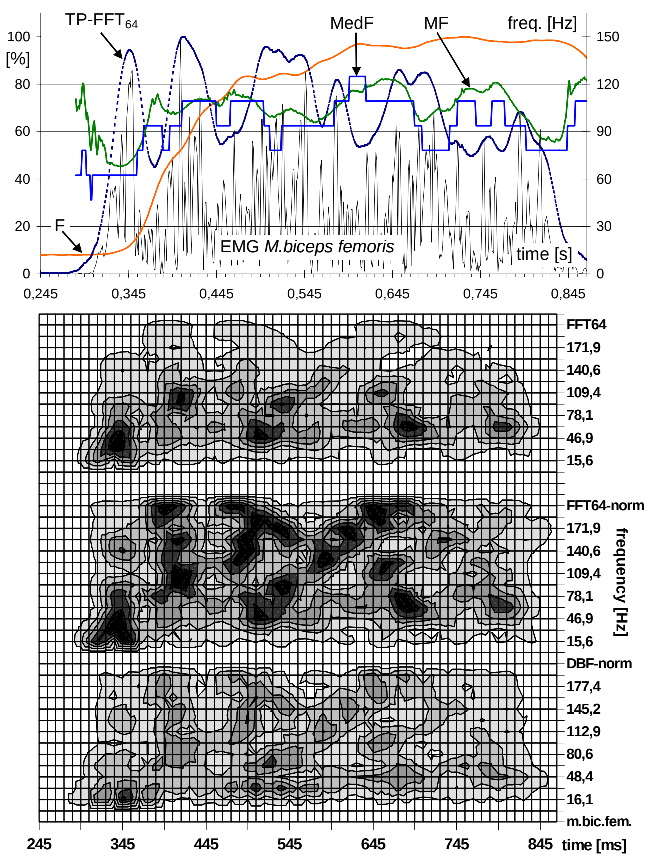

Fig. 5: Top: rectified and normalized ETC and FTC (F) of an EBC with concerning MFT-parameters total power (TP-FFT$_{64}$), median frequency (MedF) and mean frequency (MF). Bottom: Surface diagrams (time / frequency / amplitude) of spectrum analysis by means of MFT (64 ms) and DBF; normal (FFT64) and normalized (FFT64-norm / DBF-norm).

Additionally a moving FFT procedure (similar to moving average) with windowing (e.g. hamming) is recommended (Jöllenbeck, 1999) maintaining time sensitivity while providing a good representation of the desired parameters. The next required step is to transmit every resulting frequency band into a frequency-time-course. The results could allow some realistic speculations about frequencing selections and recruitment order. Unfortunately, for the moment, the proof seems to be elusive. In a similar way using simple digital band-pass filtering with the same bandwidth and transmission in a frequency-time course as FFT, shows similar results, but, for technical reasons is much sharper than the FFT result. Also using the current preferred Wavelet method, the same results as in band-pass filtering are to be seen. But finally, it should be stated that without proof of the results it is not possible to decide which method is the best or the right one.
Not presented here were the rather sparsely used adaptive spectral analysis normally used for EEG as well as the autoregressive method.

CONCLUSION

In processing of the EMG signal two important statements should be considered. We must take care not to fit the EMG-signals to the existing methods as often done in past. Rather, we should use the high potential of the modern computational equipment to develop methods that analyze the real content of the EMG. To do so may assist in avoiding misinterpretations of the past in future work. The presented methods intend to encourage the discussion to create a new quality in EMG analysis and to open the door to a better understanding of the "language of muscle".

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POSSIBILITIES & PITFALLS IN THE APPLICATION OF SEMG: CLUES FROM SCIENTIFIC RESEARCH FOR PRACTICAL APPLICATIONS

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INTRODUCTION

Surface EMG (SEMG) is a reliable tool for functional investigations of muscles. It is easy to apply and non invasive, qualifying it for a wide range of applications. However, only few questions can be addressed by routine SEMG: identification of coordination patterns, identification of muscle fatigue, and identification of muscular strain levels. The two latter issues will be highlighted in the present paper.

MUSULAR FATIGUE

Using both, frequency and amplitude information muscle fatigue is easy to identify: during sub-maximal contraction levels amplitude level will increase and frequency decreases.

To quantify the extent of a certain task both parameters have to be considered, but results remain descriptive. Furthermore, it is not clear if dynamic and static work can be treated the same way. We developed an easy algorithm to judge the amount of both types of fatigue more precisely.

Another problem for fatigue quantification arises from detailed multi-channel investigations which could prove a dependency of both parameters from electrode localization: in the region of the innervation zone (IZ) seriously false calculations occur, more pronounced for bi-polar than for mono-polar recordings.

Therefore, to get reliable results of muscle fatigue quantification the region between IZ and the tendon should be used.

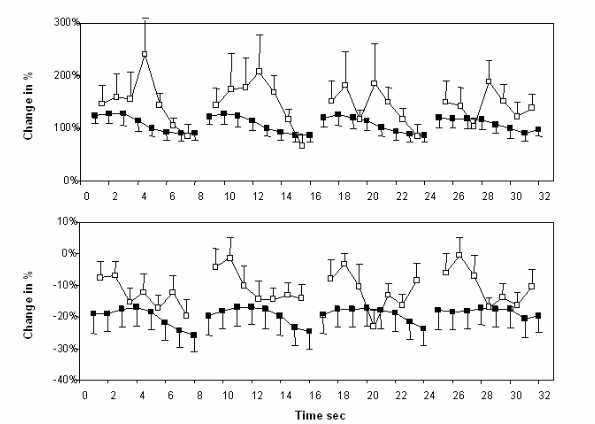

Fig. 1: Relative change of rms (upper panel) and median frequency (lower panel) of fatiguing static contractions (20% MVC) of the biceps muscle. Open squares: bi-polar recording, filled squares: mono-polar recording. Electrodes were positioned in four rows per eight electrodes over the biceps muscle.

STRAIN LEVEL: MUSCLE LENGTH DEPENDENCY

SEMG data have to be normalized to predict a certain strain level. To do this the golden standard is the use of maximum voluntary contraction levels (MVC). With these data sub-maximal force levels can be rated, or, the other way around, measured SEMG levels can be assigned to certain strain levels.

We investigated shoulder muscles at MVC level and 50% of these previously determined force levels. This was done in three planes (sagittal, horizontal, frontal) and both possible force directions, respectively. We found a systematic relationship between muscle length and relative SEMG amplitude with higher amplitude levels at shorter muscle lengths. In other words, shorter muscle lengths required higher SEMG amplitude levels. A possible explanation for this phenomenon could be the presence of large proteins (Titin. Nebulin) in the muscle which have elastic properties: the shorter a muscle is the less these proteins are able to support force production.Therefore, to predict a certain muscular strain level, reference data have to be measured in the same position.

STRAIN LEVEL: EMG-FORCE-RELATIONSHIP

The EMG-force-relationship has been identified as both, linear and polynomial. Solomonow and colleagues could prove that both types are generated by different recruitment strategies: the linear relationship was correlated with increasing firing rate of the respective motoneurons (MN), whereas for the polynomial behaviour, additionally more motor units were recruited.

We investigated trunk muscles at gradually increasing force levels, enabling the identification of the EMG-force-relationship of five different muscles in vivo. We found a highly linear relationship for the both investigated back muscles (lumbar multifidus and erector spinae), but a clear nonlinear behaviour for the investigated abdominal muscles (rectus abdominis, external and internal oblique muscles). Comparisons between male and female subjects revealed equal curve characteristics for the back muscles but considerable differences for the abdominal muscles. Women were characterised by less nonlinear curve characteristics.

These results are in nice correlation with different ratios of Type I and Type II fibers which can be assumed to exist between abdominal and back muscles on one hand and between male and female subjects on the other hand. Therefore, the EMG-force relationship seems to be correlated with fiber composition. This might be the underlying morphologic basis for the different recruitment strategies. Knowledge about the EMG-force-relationship of muscles which have to be investigated seems to be necessary to avoid false assessment of strain levels, even if the muscle length has been kept constant and electrode position was chosen adequately.

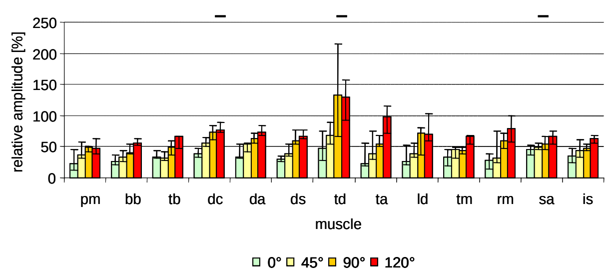

Fig. 2: Relative amplitudes of shoulder muscles during isometric abduction contractions at 50% of the previously determined MVC level. MVC levels were determined for every abduction angle separately. The bars indicate significant differences between the four abduction angles (nonparametric Friedman-test for dependent samples). pm: pectoralis major, bb: biceps brachii, tb: triceps brachii, dc: deltiodeus clavicularis, da: deltiodeus acromialis, ds: deltiodeus spinalis, td: trapezius descendens, ts: trapezius ascendens, ld: latissimus dorsi, tm: teres major, rm: rhomboideus, sa: serratus anterior, is: infraspinatus.

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INTERPRETATION STRATEGY FOR CLINICAL EMG DATA

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INTRODUCTION

Kinesiologic EMG (KinEMG) is a well established quantitative method to measure and analyze the muscle activation during various tasks, exercises and treatment regimes. Today's technology allows an easy access and operation of KinEMG devices. However, the meaningful setup, analysis and interpretation of EMG data are still under the responsibility of the user. Due to the lack of established and standardized clinical protocols, normative data and interpretation techniques, more systematic schemes have to be developed to facilitate the potential use of EMG analysis.

HYPOTHESIS AND ANALYSIS QUESTIONS

The most reasonable way to start and decide for an EMG analysis is the observation of a clinical problem or the need to document objective data on the neuromuscular status of a patient. In a next step, a hypothesis or "expectation" has to be formulated. This again can be translated to an analytical question. Within the range of available kinesiological biomechanical methods, EMG is the outstanding choice for detecting muscle activation. The following categories of analytical questions can be addressed by KinEMG:

- Is the muscle active? Nominal type (on/off)
- Are muscles more or less active? Ordinal type (more/less)
- When is the muscle on/off? Metric (Time scale)
- How much is the muscle active? Metric (MVC normalized)
- Does the muscle fatigue? Metric (Regression/slope)

A variety of sub-categories, using more "clinical" terminology can be derived, e.g. "Does the muscle fire when it should", or "Is the interplay of synergists correct in timing?"

OBSERVATION CRITERIA

To interpret a clinical KinEMG finding it is useful to overview and systematically apply a system of observation categories. Based on the question type (nominal, ordinal, metric) this is done by qualitative terms, calculated values or data tendencies. Due to its relative nature (micro volts vary from subject to subject) it is helpful to describe clinical EMG with a mixture of qualitative terms and signal ratios.

Group A focuses on the single selected muscles and describes the EMG signal in terms of amplitude, frequency and timing characteristics:

- Magnitude of amplitude — increased, decreased, absent
- Timing of activation — too early, late or asynchronous
- Time domain changes — amplitude increase/decrease; frequency increase/decrease; regularity and constancy

Group B describes activation characteristics between muscles:

- Side symmetries — percentage differences
- Activity of antagonists — lack/increase of co-activation; dysfunctional timing
- Activity of synergists — lack/increase of co-contraction (stabilization activity); dysfunctional timing

COMPARISON ANALYSIS

When planning KinEMG investigations it is very important to target reasonable comparison conditions. Because typically patient data cannot be MVC normalized, the strategy is to create ratios and quantity differences between two findings. The following comparison classes can be considered:

- Pre-test : Post-test — to show tendencies
- Left to right side — differences between affected/ unaffected side
- Activity A vs. B — muscle innervation in different test positions
- Signal portion A vs. B — time domain changes of amplitude and frequency
- Muscle A vs. B — qualitative comparison of synergists and antagonists
- Patient vs. Norm-curve — dysfunctional EMG patterns

INTEGRATION OF EMG DATA

An EMG measure monitors the muscle activation, but does not directly reveal information about its cause. To answer the question "***Why***" is probably the most challenging step within interpretation schemes. It strictly requires a set of preliminary information, based on medical diagnosis, physiotherapeutic investigation and anamnesis data. Several biological systems interact and influence any kinesiological finding:

- Psychological and behavioral aspects (e.g. pain expectation, stress)
- Central nervous structural damage/disability (e.g. stroke)
- Lack of motor ability (untrained)
- Wrong or inappropriate motor control programs
- Damage or disability of sub cortical structures and pathways
- Insufficiencies of local reflex systems and sensory receptors
- Biomechanical and structural problems in and around joints
- Physiological status (muscle atrophy, fatigue)

The clinical value of KinEMG depends on the appropriate integration of EMG data and its link to pending therapy decisions.

Clinical Integration of EMG Findings

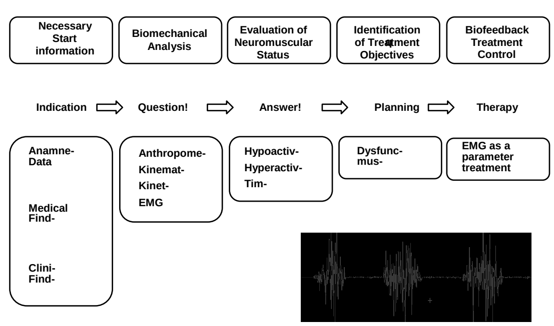

Fig. 1: Process scheme explaining the integration of KinEMG

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MONITORING OF NEUROMODULAR INTERVENTIONS WITH POLY-EMG

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INTRODUCTION

Pain associated with hypertonus due to spasticity is a common symptom of paresis of central nervous origin. The hypertonus induced pain exists in the modification of phasic and tonic spasticity. Besides other methods of neuro-modulation, the epidural stimulation of the spinal cord (SCS) is an efficient method to suppress hypertonus efficiently.

In 1979 Richardson and McLone described the positive modulating anti-spastic effect of SCS for 6 patients with traumatic thoracic spinal cord injury (Richardson and McLone 1979). In later studies, the anti-spastic effect was confirmed by placement of the electrodes below the lesion with the restriction that mild to medium ranged spasticity, but high level spasticity could not be controlled (Dimitrijevic et al. 1986, Barolat et al. 1995).

In early motor control studies done with traumatic paraplegic patients the complex neuronal mechanism of the lumbar spinal cord and its dependency to residual supra-spinal influences below the lesion was described (Dimitrijevic 1988, 1998). Independent of this, it was possible to prove the evidence of a "central pattern generator" (CPG) for locomotion of man evoked by electrical stimulation of the proximal spinal cord (Dimitrijevic et al. 1998).

The design of the recent study is based on the hypothesis that high-grade spasticity of lower extremities can be suppressed by direct stimulation of the lumbar neuronal networks more efficiently than in former studies where electrodes were placed below the lesion, but not specifically corresponding to the defined spinal cord segment (Pinter et al. 2000). Furthermore we allege that the "unspecific" electrode placement was the reason for the efficient modification of mild to middle-grade spasticity, but not high-ranked spasticity

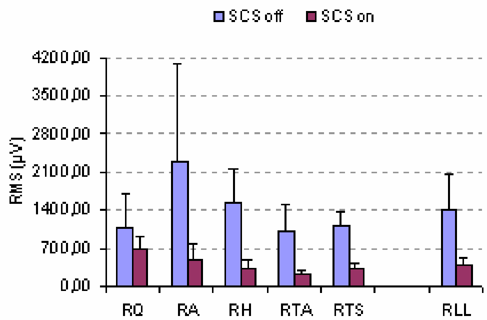
PATIENTS AND METHODS

We evaluated this hypothesis in a study with 8 high-grade spastic patients (mean age – 28.1 years) resulting from a traumatic spinal cord injury. By using a neurophysiologic evaluation the electrodes were placed epidural and directly above the dorsal section of the proximal lumbar segments. The study included patients with exclusive chronic spinal cord injury (1 year post trauma) and high-grade spasticity Ashworth Score > 2.0). These patients did not respond sufficiently to high dosed medication of antispastica, but, segmental reflexes below the lesion were present.
The modification of spasticity by using efficient electrical stimulation was evaluated by semi-quantitative clinical scales and surface electromyography.

RESULTS

For efficient SCS, a dramatic reduction of spasticity was found and objectified by the amplitude of the EMG recordings within the passive hip-knee stretch reflex (Fig. 1).

A



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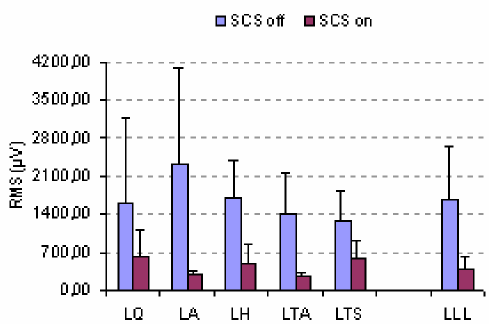

Fig. 1: RMS (Root Mean Square) of the right (A) and the left leg (B) within SCS on compared to SCS off.

There was a significant reduction of the RMS (Root Mean Square) EMG from SCS off and SCS on for both legs. Accordingly a significant reduction in the Ashworth Score could be demonstrated for:

- The left leg from 3.15 (2.3 – 3.8) to 1.15 (1.0-1.5)
- The right leg from 3.2 (2.5-4.1) to 1.3 (1.0-1.6

The efficiency of SCS is directly linked with the placement of the stimulating cathode above the proximal spinal cord segment.

An illustrative example shows that the change of polarity alone can induce a difference in the anti-spastic effect (Fig. 2). In this particular case, we did not only test different locations of the cathode (0-/c+ vs. 3-/c+), but also stimulated in different frequencies (50, 80, 100 Hz) and amplitudes (0, 2, 4, 6, 8, 10 Volts). Interestingly, the change of stimulation parameters caused only a sufficient suppression of high-grade spasticity if the cathode was placed over the proximal lumbar spinal cord segment. Independent of stimulation parameters, the stimulation of the proximal contact of the SCS electrode (contact 0) did not change spasticity as demonstrated by the RMS EMG amplitude measure. But, the stimulation of the distal contact (contact 3), which is located 30 mm below contact 0, showed a significant reduction of RMS amplitude in ratio to the frequency and voltage intensity (Fig. 2).

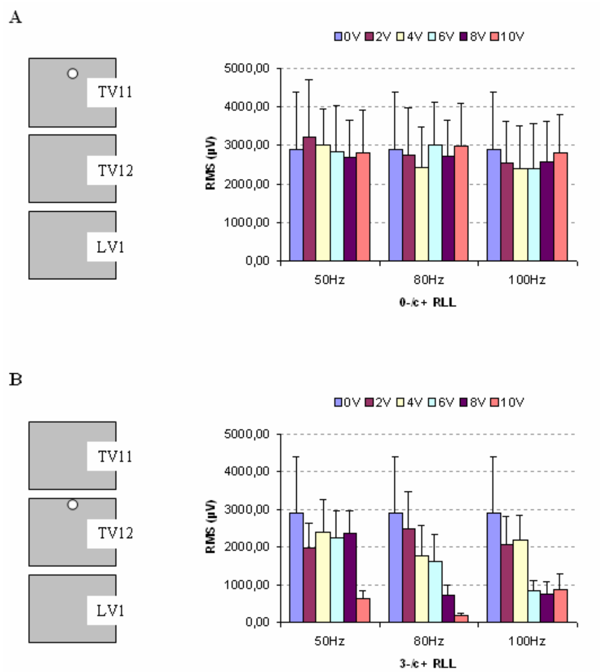

Fig. 2: Antispastic effect of SCS primary being dependent on the placement of cathode, secondary dependent on stimulation parameters

With regards to the variability of lumbar spinal cord segments in relation to the vertebrae, an intra and post surgery neurophysiologic evaluation using "muscle twitches" is a conditio sine qua non (Murg et al. 2000).
For the induction of an efficient SCS, the optimal placement of the electrode is found when in the first step, low voltage (1-4 volts) and low frequency (2.1-5 Hz), muscle twitches can be evoked for the Mm quadriceps and adductors (= key muscles for the proximal spinal cord segment). And in a second step, only an increase of voltage also evokes the Mm tibialis anterior and triceps surae. The variability of localization of lumbar spinal cord segments in ratio to vertebrae ranges from the lower margin of thoracic vertebra (TV 11) up to the upper margin of lumbar vertebra (LV 1).

DISCUSSION

Both clinical and neurophysiologic parameters demonstrated that if applied correctly, the epidural stimulation is a very efficient therapy concept to modulate high graded spasticity of traumatic paraplegia of lower extremities. The efficiency depends on four basic factors:

1. The epidural electrode must be placed directly over the proximal lumbar spinal cord segment.
2. Within constant impulse duration of 210 µsec, a stimulation frequency of 50-100 Hz and a stimulation amplitude of 2-7 volts must be applied.
3. The stimulation parameters must be optimized by systematic testing of the most effective polarity of the quadripolar electrode.
4. Depending on the body position, the amplitude must be adjusted individually.

In consideration of the results, especially the essential meaning of placement, we concluded that within unspecific electrode placement the low and middle-grade spasticity is not modulated by specific inhibitory mechanisms of the dorsal spinal pathways leading to brainstem, whereas high-grade spasticity is exclusively modified by activation of specific neuronal mechanisms within the lumbar spinal cord. Based on the fact that the significant anti-spastic effect was only present if the active cathode was directly placed over the proximal lumbar spinal cord segment, we assume that during electrical stimulation a neuronal network is activated within the spinal cord. Remarkable is the fact that the same specific dependency of placement for the stimulation electrode is given for the induction of the CPG for locomotion (Dimitrijevic et al. 1998).

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EVENT RELATED SURFACE EMG POTENTIALS WITH DELAYED VISUAL FEEDBACK

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INTRODUCTION

The investigation of pointing devices in the field of applied ergonomics showed that using only the PC mouse may have an influence on the development of chronic pain in the upper limbs (Fogleman & Brogmus 1995, Lassen, 2004, Gerr et. al 2002). To understand the underlying process it is necessary to examine experimentally induced strain on the muscular load. The aim of this study was to examine the differences between the muscular load from a single click (340 ms duration time frame) of the PC mouse and the graphic pen (Kotani & Horii, 2003) in two different delay conditions.

METHODS

16 healthy women, with 10 years of PC experience and trained for one day in using the pen, participated in this study. The task was to click on randomly appearing objects (1800) on a computer screen while their forearms were hidden under a plate. A mouse group and a pen group were formed which performed the task under two different strain conditions: non-visual delay (system delay 50 ms) and 120 ms visual delay. Half of the mouse and the pen group began within the 120 ms delay condition, the other half with the non-delay condition. Reaction time (RT) and event related surface EMG (ER sEMG) of 340 ms duration were recorded. ER sEMG from the left and right extensor and flexor were stored on a PC using Ambu Blue Sensor NS-00-S electrodes, a biovision preamplifier and a National Instruments NI-DAQ E-6025 PCI card. The software for recording was based on VB.NET with the NI-Measurement Studio API. MS Excel 9.0 and SPSS 12 were used for analyzing the data.

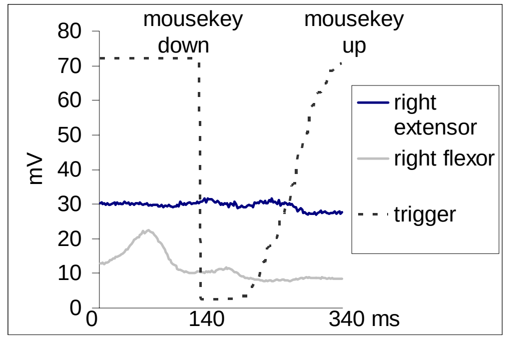
RESULTS

Multivariate analysis of variance with repeated measures showed two significant main effects. The RT of the mouse were significantly lower (Tab.1) than those of the pen ($F = 4{,}98$, $p = 0.42$). The RT in the non-delay conditions were significantly lower than those in the delay condition ($F= 22.36$, $p= .000$). The impact from the delay condition on RT weighted heavier than the impact from the pointing device. There was no interaction between delay and pointing device.

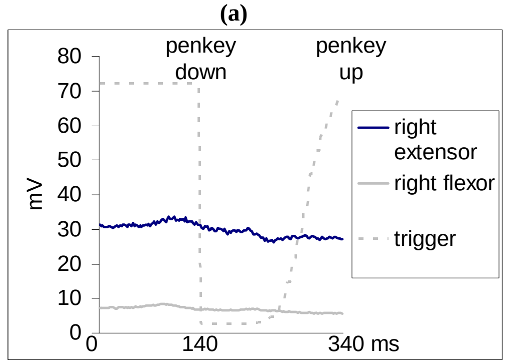
Tab. 1: Reaction Time (ms)

	condition	mean	std. deviation	N
RT mouse	delay	1230,6	36,5	8
	non-delay	1079,4	88,3	8
RT pen	delay	1307,1	149,9	8
	non-delay	1196,3	103,5	8

The ER sEMG showed two different pattern of muscular activity in the right arm for mouse and pen use (Fig. 1 a, b).



(a)



(b)

Fig. 1 a, b: Grand means of the ER sEMG of right forearm of extensor and flexor (N=8) in the delay condition using mouse (a) and pen (b).

The detailed analyzing of the ER sEMG is still in progress and is not finished yet.

DISCUSSION

All women worked on the PC with a mouse for 10 years and had never worked with a pen. Looking for other training scenarios as described in Kotani & Horii (2003) our RT results may bee seen as a lack of training effect. Looking at the experimentally induced strain (delay) it seems that no pointing device could “cope” with it. According to Nijhof (2003) the increase of sensory information (design of the tablet pen) may did not occur. The RT was always smaller in the absence of delay. The muscular pattern of a mouse-click showed high co-contraction in the pre-trigger area (time from 0 to 140 ms when the mouse key is not pressed), while the pen did not. Those co-contractions may lead to lesions in the muscle and therefore they may be described as a risk factor for pain as described by Lund (1991). The absence of muscular co-contraction while clicking with the pen may be seen in the activation of other muscles.

CONCLUSION

In this investigation we examined the impact of visual delay on event related sEMG during single clicks (duration 340 ms time frame) with a PC mouse or a graphic pen. The results showed that the graphic pen may be an alternative in general for preventing mouse injuries. In specific strain situations, like working with old and non-optimized PC systems, we were able to find a muscular benefit although training for a better performance is to advise.

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EFFECT OF SURGICAL TREATMENT ON THE EMG PATTERNS OF CHILDREN WITH CEREBRAL PALSY (CP)

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INTRODUCTION

Electromyography (EMG) has been shown to be a reliable means to evaluate patients with CP during gait [1]. These patients show abnormal EMG patterns and timing during gait [2]. Earlier reports have shown that muscular activation patterns do not change after a surgical intervention [2-6]. In these studies it is supported that no alteration in the muscle activation pattern has to be expected since the surgical intervention is targeted to the periphery (bones, muscles and tendons). Some authors argue that the EMG is a kind of 'fingerprint' in patients with CP and hence the patient cannot adapt to the new biomechanical status after the surgery [3]. Keeping this in mind, EMG measurement after the operation is thought to be redundant. However, these studies have been either assessed in a small number of subjects or conclusions were drawn after visual observation of the raw EMG signals, which is subjective.

More recent studies showed that the activation timing parameters of the semitendinosus and vastus lateralis muscles do change after hamstring lengthening procedures [7]. These changes were attributed not only to biomechanical changes but also to changes in the recruitment of muscles that were not surgically treated. Furthermore, activity of the tibialis anterior muscle is reduced when patients with an excessive plantarflexion at initial contact wear ankle-foot orthoses [8]. Even short-term, intense balance training can improve muscle coordination, but the level of adaptation depends on the severity of involvement of the child [9]. The above recent studies suggest that muscle recruitment in patients with CP may change after an intervention. This might be of interest to the orthopaedic surgeon helping him/her to define more precisely the treatment planning and the amount of surgery and the postoperative rehabilitation treatment. Therefore, the purpose of this study was to evaluate the EMG patterns while walking before and after a surgical treatment in a group of children with CP. Moreover, the potential changes after surgery may be differentiated by the site of the disorder, and therefore we also tested for differences between hemiplegic and diplegic children with CP.

METHODS

Eighteen diplegic and 16 hemiplegic patients with CP aging from 6 to 16 years old (mean±standard deviation [SD] 10.1±3.0 years) participated in the present study. They were able to walk without external supporting devices and did not have any previous surgery, casting, or medical treatment (e.g. botulinum toxin or baclofen). All children were examined 1-3 days before and 1-5 years (2.53±1.23) after multilevel surgery. The study was approved by the local ethical committee. Furthermore, data 20 healthy children of the same age range were taken as a reference group. All subjects underwent full 3D-gait analysis including kinematic, kinetic and EMG data recordings.

The Gillette Gait Index (GGI) [10] for each side of the patients was calculated. According to this index each side was assigned as more or less involved. To preserve a more homogenous group concerning severity we evaluated only the more involved side of each patient.

Seven superficial muscles responsible mainly for the hip, knee and ankle extension/flexion: the vastus lateralis (VLA), the rectus femoris (RFE), the biceps femoris (BFE), the semimembranosus (SEM), the tibialis anterior (TAN), the lateral gastrocnemius (LGA) and the soleus (SOL) muscle. Bipolar surface adhesive electrodes (Blue Sensor, Ambu Inc., Glen Burnie, MD, USA) with inter-electrode distance of 2 cm were applied over each of the examined muscles according to the guidelines given from the SENIAM project [11]. The EMG signal was pre-amplified (×5,000) using the Biovision EMG apparatus (Biovision Inc., Wehrheim, Germany).

The analog data from the force plates and the electromyographer were digitized using a 12 bit A/D card, with a sampling frequency of 1,080 Hz. For the EMG the signal was off-line fully rectified and the linear envelope was calculated with 9 Hz cut-off frequency [12, 13]. The EMG amplitude was normalized to the average value of each stride for the respective muscle and subject. Hence, the EMG signals were expressed as percentage of the mean. Each gait cycle was divided in 7 sub-phases: loading response, mid stance, terminal stance, pre-swing, initial swing, mid-swing and terminal swing. Each of these sub-phases was interpolated to 30 data points and the average of 3-8 gait cycles for each dependent variable and condition (subject, examination, side) was calculated.

To determine a magnitude to describe the similarity of the EMG pattern of a patient to the mean EMG pattern of the norm group, the norm-distance (ND) was calculated, defined as the absolute difference between the EMG of the patient (p) and the mean value of the norm-reference group (N) at a specific time (t), normalized to the standard deviation (s) of the norm group at the same time instance t [14].

$$ND_{pt} = \frac{\left| x_{pt} - x_{Nt} \right|}{s_{Nt}}$$

To quantify the deviation from the normal for each examined parameter, we calculated the mean ND for the whole gait cycle.

A two-way ANOVA was used to determine the effect of the factor DIAGNOSIS (levels: hemiplegic and diplegic patients), the factor TIME for repeated measurements (levels: examination before and after the surgery), and the interaction between these factors for the ND for the EMG of each examined muscle. In case of a significant effect in the above ANOVA a secondary two-way ANOVA (factors TIME and DIAGNOSIS) was applied on each of the 7 sub-phases of the gait cycle. The level of significance was set at $p<0.05$. The statistical analysis was assessed using the SPSS v.13 software (The Apache Software Foundation, IL, USA).

RESULTS

The multivariate test for the norm-distance of all 7 muscles showed a significant effect of the factor TIME ($F_{7,26}=3.58$, $p<0.01$) and non-significant for the factor DIAGNOSIS ($F_{7,26}=1.32$, $p>0.05$). The TIME×DIAGNOSIS interaction was also not significant ($F_{7,26}=1.10$, $p>0.05$). According to the univariate tests for the norm-distance, the soleus, lateral gastrocnemius and tibialis anterior muscle showed significant decrease for the factor TIME. The 95% confidence interval (95%CI) for this decrease was 0.134 to 0.367 for the soleus, 0.032 to 0.294 for the lateral gastrocnemius and 0.014 to 0.275 for the tibialis anterior muscle (see Tab. 1). For the factor GROUP, the norm-distance of the biceps femoris showed marginally higher values for the diplegic group compared to the hemiplegic ($p=0.048$). The 95%CI of this difference was 0.001 to 0.282. All other muscles did not show any other significant differences ($p>0.05$).

	Before	After
SOL	**1.11±0.36**	**0.87±0.26**
LGA	**1.28±0.37**	**1.12±0.29**
TAN	**1.28±0.32**	**1.13±0.35**
VLA	1.16±0.27	1.23±0.30
RFE	1.16±0.28	1.15±0.28
SEM	0.81±0.25	0.78±0.22
BFE	0.85±0.31	0.82±0.17

Tab. 1: Norm-distance (mean±SD) for all examined muscles before and after the surgery. Significant differences between the examinations are designated in boldface ($p<0.05$).

Concerning the secondary statistical analysis, to detect in which phase of the gait cycle systematic changes in the EMG pattern after the operation occur, the soleus and lateral gastrocnemius muscles activation level was decreased during the terminal swing (soleus 95%CI -47.4 to -8.9% and lateral gastrocnemius 95%CI -30.9 to -5.6%) and increased during the terminal stance phase (soleus 95%CI 2.0 to 20.4% and lateral gastrocnemius 95%CI 1.1 to 19.3%). The lateral gastrocnemius muscle activity increased during the pre swing phase (95%CI 0.1 to 11.3%) and a decrease during the loading response (95%CI 23.6 to 2.2%). Furthermore, the tibialis anterior showed a significant decrease during the terminal stance phase (95%CI -13.7 to -2.4%). The above mentioned changes were not group specific (no significant TIME×DIAGNOSIS interaction) and responded to an EMG pattern closer to the one of the group of healthy children.

The spatiotemporal parameters showed that the diplegic patients walked significantly slower (95%CI -0.24 to -0.05 m/s), with prolonged stance phase (95%CI 1.9 to 5.3% of gait cycle) and shorter stride length (95%CI -0.26 to -0.08 m) than the hemiplegic patients. No TIME×DIAGNOSIS interaction was present, revealing that these differences were present before and after the operation. Significant changes between pre- and post-operative data concerning the stance phase duration (95%CI -2.8 to -0.2% of gait cycle), the stride length (95%CI 0.03 to 0.14 m) and the GGI (95%CI -338 to -137) was shown for both groups.

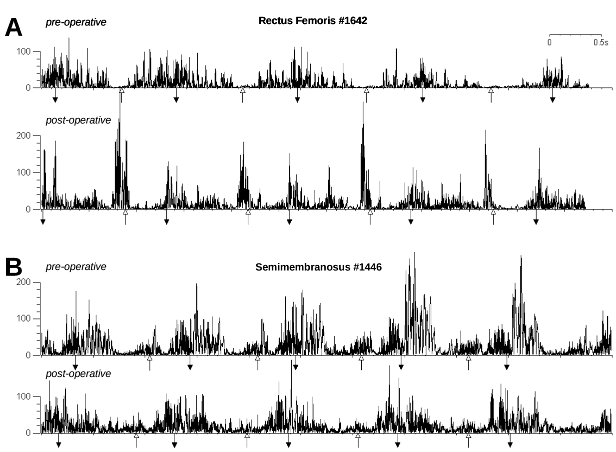

Fig. 1: Fully rectified EMG signals (in µVolts) of 2 patients before and one year after the operation. A. Rectus femoris signal from patient #1642. B. Semimembranosus signal from patient #1446. Upward filled arrows and downward empty arrows represent the initial contact and the toe off, respectively.

DISCUSSION

This study confirms the presence of systematic changes in the EMG patterns after a surgical multilevel treatment in ambulatory children with CP. Taking into consideration the changes in ND for the kinematic and kinetic parameters that occur after surgery, the observed changes in ND for the EMG were relatively small and restricted to the examined shank muscles (lateral gastrocnemius, soleus and tibialis anterior).

The majority (30 out of 34) of the examined children had equinus feet deformity for the more involved side, which was corrected surgically by soleus and gastrocnemius lengthening procedures. Enhanced EMG activity on the lateral gastrocnemius and soleus muscle at the initial contact pre-operatively may result a stretch reflex response. This response was less present post-operatively, resulting in a more normal EMG pattern. The cause for this decrease might be the equinus reduction which allowed the patient to place the foot on the ground in a more dorsiflexed position during the initial contact.

The absence of changes in the EMG patterns for all other muscles indicates that the activation patterns were retained after the surgery. Observing every patient separately and comparing visually the pre- with the post-operative EMG patterns, we often experienced the 'fingerprint' character of the EMG that has been previously reported [3]: pre- and post operative EMG patterns were very similar. However, this was not the case for all examined muscles and patients. Especially muscles which were not surgically treated had more obvious changes in EMG patterns after the operation; nonetheless, the difference was not statistically significant. In general, after visual inspection 15-25% of the examined EMG patterns showed obvious changes after the surgical treatment. This reveals that certain muscles may have the potential to modify their activity. Although these changes were not systematic for all patients or for a subgroup of the examined patients, such cases indicate that muscle activation patterns before the operation that deviate from normal might have been acting (or not acting) compensatory. Variations in the firing patterns in healthy subjects allow the nervous system to develop new ways to organize [15], and this cannot be *a priori* ruled out for children with CP. Measuring EMG after the operation may allow us to understand these compensation strategies of the neuromuscular system, provided that the examined muscles have the ability to adapt to changes. Understanding these strategies will improve our prediction of the surgical intervention result.

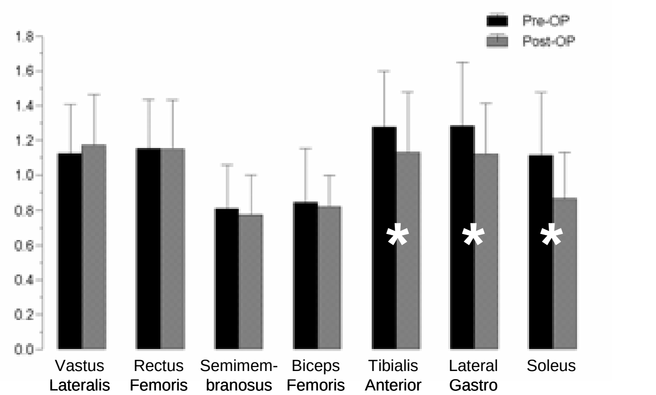

Fig. 2: Norm-distance for the examined muscles before the after the surgery. Asterisks indicate significant differences between the examinations.

CONCLUSION

We observed no differences between patients with diplegia and hemiplegia in the improvement of the EMG gait patterns, and significant changes in the soleus, lateral gastrocnemius and tibialis anterior EMG patterns after the surgery. The question remains whether the observed changes have clinical significance. However, the differences presented here represent systematic changes that occurred for a group of patients; more profound differences within subjects have been observed. Therefore, this study supports that EMG measurement after the operation can describe the clinical picture of the patient better and may assist for planning the surgery and the post operative treatment. Secondly, after analyzing the pre-operative EMG data for a clinical decision, it should not be taken for granted that the activation patterns will remain the same no matter what biomechanical changes will occur after the orthopaedic surgery. Further controlled studies with more homogenous samples and surgical treatment may enlighten compensatory mechanisms that are acting before the operation.

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EMG AND ACCELERATION MEASUREMENTS FOR THE DIFFERENTIAL DIAGNOSIS OF TREMOR – CONCEPTUAL EVALUATION OF A QUANTITATIVE CLINICAL MEASUREMENT PROTOCOL

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INTRODUCTION

Frequency analysis of accelerometer signals has successfully been used to document and classify tremor types in clinical practice (1). The parallel detection muscle innervation of involved muscle groups adds beneficial information to the differential diagnosis of tremor types. Surface EMG recordings of e.g. forearm flexors and extensors allow an easy investigation of agonist-antagonist patterns within varying test positions.

METHOD

An eight channel EMG amplifier (TELEMYO 2400 - NORAXON INC.) was used to detect the surface EMG signals from the flexor carpi radialis and extensor carpi radialis group. The raw EMG signals were recorded with dual silver-silver chloride electrodes (inter electrode distance 2 cm) at 1500 Hz sampling rate. A two dimensional 1G accelerometer (NORAXON INC) was mounted on the back of the hand by using a special fixation velcro strap (Y axis = gravity line, Z axis = horizontal line) and recorded at the same sampling speed. A digital video was recorded in time synchronization to allow a full documentation of each test activity. The raw EMG signal is documented within an analysis report to allow visual inspection of the recording quality. For the co-activation analysis the raw EMG was full-wave rectified and smoothed with a Root Mean Square (RMS) of 100 ms.

TEST PROCEDURES

To establish a clinical meaningful but easy to use test protocol in a pilot study design (N=3), we programmed a test sequence consisting of 3 major activities:

Rest– patient arm is in rest position on his thighs.
Hold– patient holds both arms in horizontal extension position
Reach – patient moves his finger from thigh to the nose

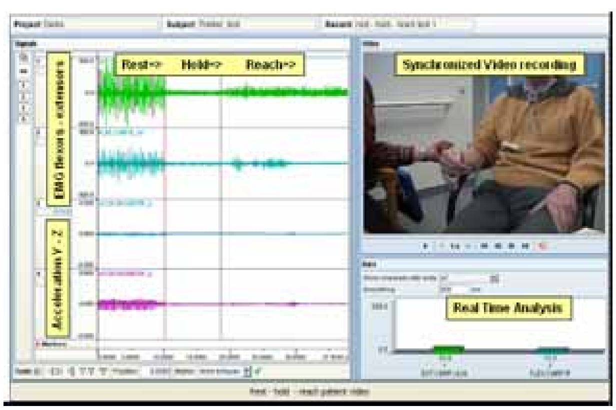

Fig. 1: Combined recording of 2D accelerometer (lower two traces and 2 EMG traces (upper traces) with synchronized DV-video

(EMG processing and analysis software MyoResearch XP). An automatic analysis report calculates the key parameters as explained in more detail below. It may be suitable to add more activities to the protocol routine, which can be operated by a "free to configure" activity protocol. The standardized sequence is fully guided by an electronic protocol assistant and allows an easy and quick operation.

ANALYSIS REPORT

The analysis concept is based on a PC based replay and review function which allows a post hoc inspection both of the video and raw signal recordings. This approach allows a qualitative inspection of motion specific aspects with testing. The analysis report calculates the frequency contents of both accelerometer signals by means of Fast Fourier Transformation (FFT) and documents the power spectrum around the center frequency. The peak value of the power spectrum is considered as the main tremor analysis parameter, typically ranging from 1 to 15 Hz.

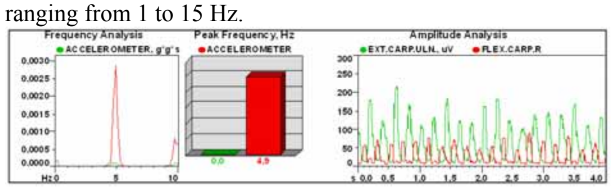

Fig. 2: FFT based frequency analysis of accelerometer signals and RMS innervation patterns for a "Rest"- activity – identifying a rest tremor with center frequency of 4.9 Hz

The RMS amplitude analysis is based on a curve overlay of both muscle traces. This graphical presentation allows for an easy detection of agonist or antagonist tremor activation and illustrates the co-activation patterns of involved muscles.

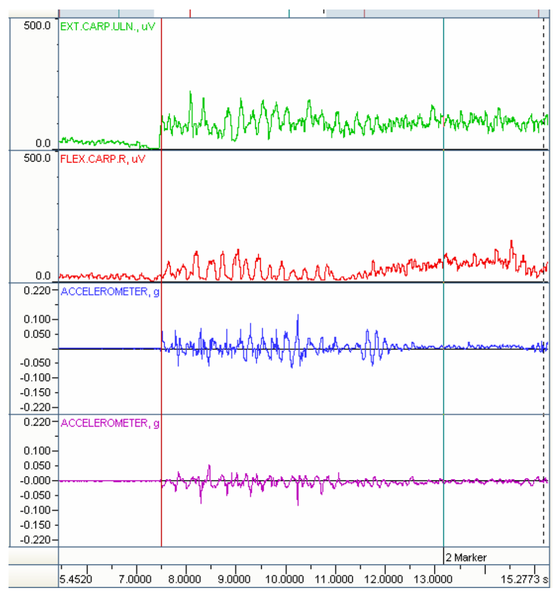

Fig. 3: Free protocol examining the water glass pouring exercise

CONCLUSION

First test runs proved a high clinical evidence by combining surface EMG recording with the accelerometer based motion parameter. The combination of a base test "Rest-Hold-Reach" with additional user configured activities like writing tasks, emptying a glass of water or bilateral activities allows for a meaningful use of the PC assisted automatic protocols. The quantitative analysis of tremor types adds important information to the classical clinical diagnosis of tremor. An algorithm to quantify the amount of co-activation still needs to be implemented in the concept.

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SURFACE-EMG-BIOFEEDBACK IN NEUROMOTOR REHABILITATION

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INTRODUCTION

Although there is evidence that sEMG-guided motor rehabilitation is beneficial for example in post-stroke rehabilitation [4], or in cerebral palsy [5], surface-EMG-Biofeedback (sEMG-BFB) has not yet been sufficiently incorporated into routine neuromotor rehabilitation concepts especially in Europe.

METHODS

Review. sEMG-BFB treatment strategies are discussed in the light of motor-learning concepts.

DISCUSSION

sEMG-Feedback is effective in producing relaxation, active muscle control, and awareness, but the acquisition mechanisms are not clear. Investigation of short-term performance during and after feedback training may not relate to long-term learning [2].

Voluntary movements are goal directed and improve with practice as a result of feedback and feed-forward mechanisms: Motor programs are continuously refined by learning. There are changes and shifts of anatomical location of representations of motor programs as motor behavior progresses, through learning, from being novel to being automatic [3]. Early mapping experiments stimulating the cortical surface electrically initially led to the simplistic idea that the primary motor cortex […] controls individual muscles or small groups of adjacent muscles. More detailed studies […] demonstrate that neurons in several cortical sites project axons to the same target. In addition, most stimuli activate several muscles, with muscles rarely being activated individually […] An implication of this redundancy in muscle representation is that inputs to motor cortex from other cortical areas can combine proximal and distal muscles in different ways in different tasks [3].

The somatotopic organization of the motor cortex is plastic. It can be altered during motor learning and following injury. The idea that the organization of at least some mature motor circuits can change depending on sensory or motor activity holds important promise for the rehabilitation of patients [3].

Motor processing begins with an internal representation, namely the desired result of a movement. The brain represents the outcome of motor actions independently of the specific effector: A purposeful movement is represented in the brain in some abstract form rather than a series of joint motions or muscle contractions (motor equivalence) [3].

In providing sEMG-Feedback about the activity of only one or a few muscles a non-physiologic information is given. This seems especially useful in motor-learning contexts with a high fraction of voluntary-online control: New tasks or in exercises with emphasis on precision instead of speed.

Tapering down the extent and promptness of sEMG-Feedbacks in the course of growing skill acquisition seems to be essential for skill retention and transfer. But optimal sEMG-BFB training schedules still have to be evaluated. Practice schedules that promote the rapid acquisition of performance ability are not necessarily those that optimize skill retention and transfer. For example frequent or immediate knowledge of results (KR) or knowledge of performance (KP) may promote a cognitive dependency on the extrinsic feedback, impeding the formation of an intrinsic reference. In the acquisition phase of motor learning a blocked schedule might be preferred, whereas later on, randomized practice might be superior [2].

A thorough evaluation of the patient aims considering activities of daily living warrants motivation. The use of goal directed, functional meaningful and variable tasks is advisable in the light of the motor-learning-concepts presented. A combination of therapeutic functional knowledge and creativity is necessary to figure out an individual training program. The need for combining knowledge about the nature of the neurological lesion, anatomic and kinesiologic facts, strongly limits the applicability of the sEMG-BFB training in a psychological setting.

Kasman et al [2] summarize sEMG-Training techniques: 1. isolation of target muscle activity. 2. relaxation-based down training. 3. threshold-based uptraining/downtraining. 4. tension recognition threshold training. 5. tension discrimination training. 6. deactivation training. 7. generalization to progressively dynamic movement. 8. sEMG-triggered neuromuscular stimulation. 9. left/right equilibrium training. 10. motor copy training. 11. promotion of correct muscle synergies and related coordination pattern. 12. postural training. 13. body mechanics instruction. 14. therapeutic exercises. 15. functional activity performance. Some of the above mentioned training techniques are discussed in detail.

Finally, studies incorporating sEMG-BFB treatment strategies with a sophisticated task oriented approach are presented [6-9]. They serve as an example for future treatment protocols.

CONCLUSION

s-EMG BFB should be integrated more often in neuromotor rehabilitation plans. But considerable efforts are still necessary to standardize the methodology applied (f.e.electrode placement, short- and long-term treatment protocols). There is a need for further clinical studies.

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EMG ACTIVITY DURING STEP EXERCISIS

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INTRODUCTION

During the rehabilitation process of the lower extremity stair walking is an important factor. Sometimes, instead of using stairs, stepper machines are used for training purposes. A minimum of force and coordination is required to be able to walk over stairs. The question is, whether the muscle activation is comparable and intensity is similar.

METHODS

The EMG activity of walking over stairs (up and down) and the activity of two different stepper machines were recorded. One stepper was a machine where both pedals are joined together (called Ministepper). If one goes up the other one goes down. The second stepper machine (called Powerstepper) has no connection between the pedals. The range of motion of the pedal was the same for both machines. Also the height of the stairs was comparable to the stepper. Electrogoniometers were used to measure the knee joint movement and the movement of the stepper excursion.
The activity of 8 muscles was recorded during a period of 30 seconds on the stepper. We used a wired EMG unit (Neurodata) and the Software MyoResearch98 to capture the signals. A moving average (100 ms) was used for smoothing and a minimum of 10 cycles was used to build an average. The cycle was defined from the highest point of the standing surface of the right side until the same position is reached. The order of the measurements was randomized. For further analysis Matlab was used.

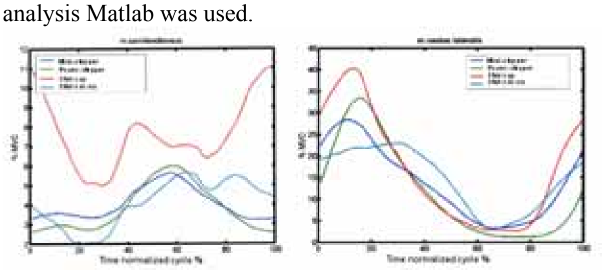

Fig. 1: Left: M. semitendinosus mean pattern of all four conditions over one cycle. Right: EMG pattern of M. vastus lateralis over one cycle

RESULTS

For both stepper machines the knee joint kinematics were quite similar. Also the movement pattern of the knee joint during the extension phase of stair climbing was comparable with the stepper movement. The knee joint movement of walking downstairs showed a different pattern.
During the change from the extension to the flexion movement the M. gastrocnemius medialis showed a 50% higher activity during stair walking in comparisons to the stepper movement. The M. vastus lateralis showed a similar pattern for all exercises. In the second half of the cycle the M. rectus femoris had a higher activity for stair walking. The hamstring had a comparable pattern for both stepper machines but the activity for stair walking was higher in the beginning and at the end. The same differences showed the M. gluteus maximus.
The M. erector spinae lumbal had a low activity just before the maximum knee extension is reached during stair climbing. A second phase of low activity exists at about 80% of the cycle.
For the steppers this second period of low activity exists also and the duration of this low activity phase was longer.

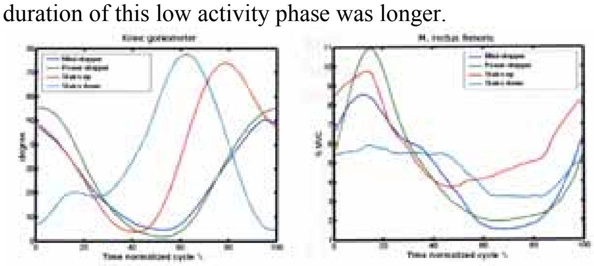

Fig. 2: Left: knee joint angle of four conditions. Right: EMG pattern over one cycle of the M. rectus femoris.

DISCUSSION

The differences of the EMG pattern during stepping and stair climbing of M. gastrocnemius medialis are the result of an active push off during stair walking up. The active flexion of the hip joint, which is needed for walking stairs up, introduces a higher EMG amplitude for the M. rectus femoris. It seems that the higher EMG activity of the hip extensors is the result of the upward and forward movement during stair walking. For the stepper only an upward movement is needed.

CONCLUSION

We were able to show that the stepper machines had lower EMG activities compared to walk stairs upwards. The EMG pattern of both stepper machines was similar during stepping down. During the upward movement we found small differences in the activation pattern.

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CHANGES IN THE EMG MEDIAN FREQUENCY DURING VOLUNTARY MUSCLE ACTIVITY OF DIFFERENT INTENSITY IN RELATION TO BLOOD CHEMICAL PARAMETERS

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INTRODUCTION

The reduction of muscle AP propagation velocity (CV) is often discussed as the major cause of the alterations in the EMG spectrum during sustained muscle contraction, since both effects have been found to show a close correlation (Lindstrom et al., 1970; Stulen & De Luca, 1981; Kranz et al., 1983; Merletti et al., 1990). As causes for the decrease in CV, an elevation of extracellular potassium (Bigland-Ritchie et al., 1979, Petrofsky 1981; Moxham et al., 1982, Kössler et al., 1990) as well as a decrease in muscle pH (Lindström et al., 1977, Mortimer et al., 1970) are under consideration. We examined the effects of exercise-induced changes in pH and electrolyte concentration on muscle electrical activity during voluntary exercise.

METHODS

9 subjects performed handgrip exercises with a hand-ergometer. The tests consisted of the 15 min warm-up phase und six 1 min static exercise periods separated by a 4 min rest. During the exercise bouts, the subjects had to lift weights and to keep them at a constant level. The weights were varied randomly between 5 and 30 kg, with steps of 5 kg. Blood was taken from a cubital vein. Acid-base-state, plasma electrolytes Na+ and K+ and [Lac-] were determined. For the recording of EMG the differential recording was used. M-wave and EMG was recorded from flexor muscles of the forearm. Muscle power, median frequency (MF) and RMS were calculated.

RESULTS

We observed distinct changes of MF during the exercise bouts (Fig. 1). During the exercise bouts, the shift of MF towards low frequencies correlated with the increase of weight ($r=-0.73$, $P<0.001$) or $[K^+]_v$ ($r=-0.76$, $P<0.001$), but not with the changes in pH. The range of the CV changes was from +4 % at 5 kg to –2 % at 30 kg. CV did not correlate neither with changes in venous pH and [K+] nor in MF.

DISCUSSION

Our data contradict the results of other studies, which have shown a relation between MF and CV (Lindstrom et al., 1970; Stulen & De Luca, 1981; Kranz et al., 1983; Merletti et al., 1990). Also the significance of extracellular potassium elevation or decrease in pH for the changes in MF was not shown. For in vivo studies, the reason for these discrepancies might be found in the different activation mechanisms in voluntary and evoked muscle exercise. Some in vitro experiments were performed under non-physiological conditions. During voluntary exercise Bigland-Ritchie et al. (1981) also did not find a relation between changes in MF and CV.

CONCLUSION

Under the conditions of our experiment: 1) the decrease in the CV was not the cause of the decrease in muscle force, 2) the effects of changes in electrolyte concentration and in pH on the CV may compensate each other or may be counterbalanced by other effects (e.g. increased T°). It can be concluded that changes in CV are not the only cause and, maybe, are not the main cause of MF shifts. We mean that in our study these shifts might be caused by the changes in the recruitment of different muscle fiber types.

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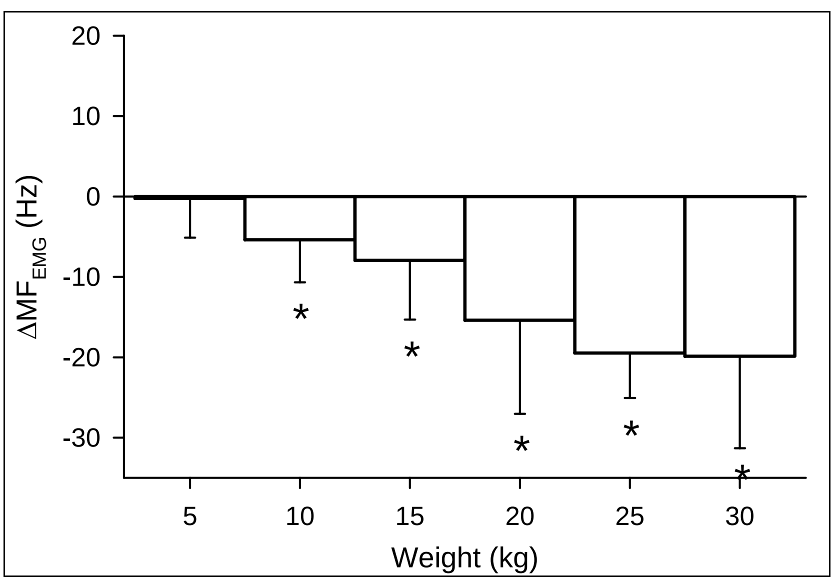

Fig. 1: Changes of MF during the exercise. Means ± SD. *P<0.02.

SURFACE EMG MEDIAN FREQUENCY ANALYSIS OF LOCALIZED BACK AND HIP MUSCLE FATIGUE DURING A MODIFIED BIERING-SØRENSEN TEST

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INTRODUCTION

Epidemiological research has pointed out that chronic low back pain (LBP) is a major problem in the Western world, with lifetime prevalences between 49% and 70%. LBP patients have demonstrated weak and excessively fatigable back muscles compared to healthy persons. In 1984, Biering-Sørensen described a test for measuring back muscle endurance. This test has been shown to be able to discriminate between healthy individuals and LBP patients, and may predict the occurrence of first-time LBP. Originally, the test consisted of measuring the time subjects were able to keep their unsupported upper body part horizontal in a prone position as long as possible. The outcome of this test may however be influenced by factors such as pain, motivation, … Surface electromyography (EMG) has been widely used during the Biering-Sørensen test as an objective tool for evaluating localized muscle fatigue. Fatigue causes a decrease of the EMG power spectrum, mostly described by the EMG median frequency (MF) decline as a function of time. However, several issues remain unsolved in the process of assessing back muscle fatigue with surface EMG.

To gain a better understanding of surface EMG localized muscle fatigue analysis during the Biering-Sørensen test, the aims of the present study were (1) to analyse which statistical model (linear, quadratic, logarithmic or exponential) best described the fatigue-related MF changes over time, (2) to investigate the MF differences between eight bilateral back and hip muscles, (3) to examine the surface EMG signal stationarities, and (4) to investigate the correlation coefficients between traditionally applied short-time Fourier (STFT) and continuous wavelet transforms (CWT).

METHODS

Twenty healthy subjects performed a modified Biering-Sørensen test. For the testing position subjects were placed on an examination couch in a prone position, with their upper body extending beyond the edge and their lower body strapped to the couch. During the test the subjects were instructed to maintain the unsupported body in a horizontal position as long as they could, with their hands touching the forehead, their elbows out to the side and leveled with the trunk and their head in a neutral position looking downward at a visual fixation point. Verbal and tactile feedback throughout the test was given by the examiner. The surface EMG activity of the latissimus dorsi, longissimus thoracis pars thoracis and lumborum, iliocostalis lumborum pars thoracis and lumborum, multifidus, gluteus maximus and biceps femoris were bilaterally measured. In addition, the subjects' lumbar curvature was recorded with a three-dimensional movement analysis system. A software algorithm incorporating STFT and CWT transforms was used to calculate the surface EMG MF parameters. Statistical analyses were performed with SPSS v12. In pursuit of aim (1) a general linear model was used to find out whether significant differences existed between the coefficients of determination of the several models applied. Aim (2) was assessed by calculating simple linear regressions of MF parameters as a function of time, yielding slope and intercept parameters. A general linear model was then used to point out significant differences between median frequency slope values of the various muscles. For aim (3), the EMG signal stationarity was investigated using reverse arrangement tests. Finally, Pearson correlation coefficients were calculated between STFT and CWT based MF parameters (aim 4).

RESULTS

Significant differences could be demonstrated between the four mathematical models used to describe the median frequency decline as a function of time, depending on the muscle investigated. For each muscle, the quadratic model showed statistically significant higher coefficients of determination than the other three models (figure 1). However, very low coefficients of the quadratic term in the regression equation were demonstrated.

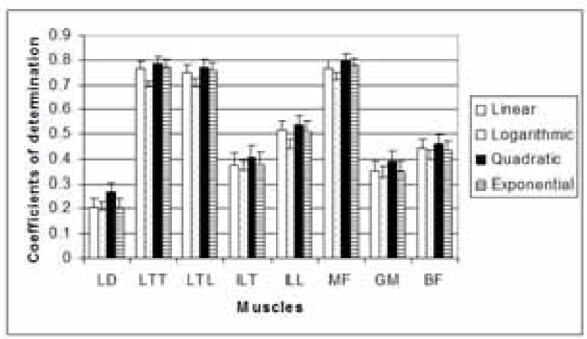

Fig. 1: Coefficients of determination (LD = latissimus dorsi, LTT = longissimus thoracis pars thoracis, LTL = longissimus thoracis pars lumborum, ILT = iliocostalis lumborum pars thoracis, ILL = iliocostalis lumborum pars lumborum, MF = multifidus, GM = gluteus maximus, BF = biceps femoris)

Investigating the MF slope of simple linear regressions of MF as a function of time, significant differences between back and hip muscles could be demonstrated. The latissimus dorsi showed the least rapid and the multifidus the greatest decline in MF slope. The iliocostalis lumborum muscles showed lower MF slope values than multifidus and longissimus thoracis muscles. The lumbar parts of both longissimus thoracis and iliocostalis lumborum muscles showed slightly higher MF slope values than their thoracic counterparts, although the differences were not statistically significant. Concerning the results of the hip extensors in the present study, clear fatigue of the gluteus maximus and biceps femoris was demonstrated (figure 2).

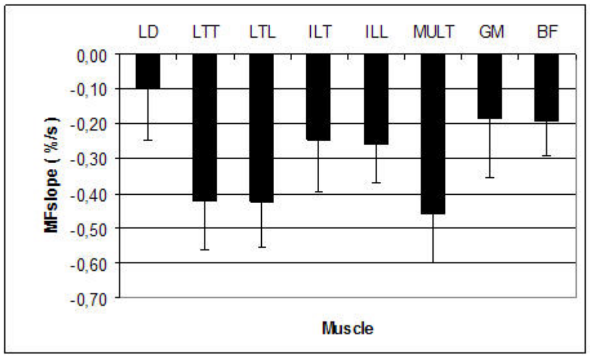

Fig. 2: Median frequency slope (%/s) for each muscle (LD = latissimus dorsi, LTT = longissimus thoracis pars thoracis, LTL = longissimus thoracis pars lumborum, ILT = iliocostalis lumborum pars thoracis, ILL = iliocostalis lumborum pars lumborum, MULT = multifidus, GM = gluteus maximus, BF = biceps femoris)

In the analysis of the EMG signal stationarity, the reverse arrangement tests showed no significant trends in 91.6% of the EMG signals, meaning 91.6% of the analyzed EMG windows could be considered as stationary signals. Finally, generally high correlation coefficients were demonstrated between STFT and CWT based MF parameters.

DISCUSSION

Several studies have relied on simple linear regressions to describe the MF decline over time. Using a simple linear regression has the advantage that single parameters such as slope and intercept can be used for further analyses. Based on the results of the present study, one could suggest using a quadratic model. However, the very low coefficients of the quadratic term in the equation favor the use of simple linear regressions.

Significant differences in MF slope parameters between those back muscles demonstrated in this study are compatible with the notion of a functional subdivision between these muscles, as described in literature. Several hypotheses were mentioned to explain these differences in fatigability (e.g. differences in fiber type characteristics, muscle length, …). As the hip extensor muscles also showed muscle fatigue during the Biering-Sørensen test, this test cannot be seen as a specific evaluation for back muscle fatigue alone.

Pearson correlation coefficients revealed that STFT and CWT in general provide similar information with respect to the surface EMG MF variables during isometric back extensions, and as a consequence either STFT or CWT can be used. However, taking into account previous published results, the latter method is favored in the analysis of localized muscle fatigue.

CONCLUSION

- Significant differences in MF characteristics could be demonstrated between the various back and hip muscles
- The Biering-Sørensen test cannot be seen as specifically evaluating back muscle fatigue alone
- The use of simple linear regression models still seems the most appropriate when investigating the time-dependent MF decline during an isometric back extension task
- STFT and CWT provide similar information regarding localized back and hip muscle fatigue during the modified Biering-Sørensen test

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THE INFLUENCE OF SPECIFIC TRAINING ON TRUNK MUSCLE RECRUITMENT PATTERNS IN HEALTHY SUBJECTS DURING BRIDGING STABILIZATION EXERCISES AND STABILIZATION EXERCISES IN FOUR-POINT KNEELING

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INTRODUCTION

The purpose of the present study was to investigate whether specific lumbar stabilization training has an effect on muscle recruitment patterns in a healthy population

METHODS

Thirty subjects were recruited to perform two types of testing exercises i.e. bridging exercises and exercises in four-point kneeling, both before and after training. Surface electromyographic (EMG) data of different abdominal and back muscles were obtained.

RESULTS

After training, analysis of the relative muscle activity levels (percentage of maximal voluntary isometric contraction) showed a higher activity of the local (segmental-stabilizing) abdominal muscles, but not of the local back muscles; minimal changes in global (torque-producing) muscle activity also occurred. Analysis of the local/global relative muscle activity ratios revealed higher ratios during all exercises after training, although not all differences were significant.

Tab. 1: Mean values, standard deviations (SD) and p-values of the ratio local muscle activity/global muscle activity (mean) during the different exercises (α = 0.05)

Exercise			**1**					**2**		
	Pre		**Post**		**p**	**Pre**		**Post**		**p**
	Mean	**SD**	**Mean**	**SD**		**Mean**	**SD**	**Mean**	**SD**	
IO/RA	3.31	1.76	7.62	6.23	0.001	3.19	2.19	7.38	5.93	0.001
IO/EO	1.19	0.91	2.93	2.18	< 0.001	0.80	0.54	2.34	1.85	< 0.001
MF/ICLT	1.21	0.50	1.81	1.12	0.01	1.27	0.56	1.59	0.97	0.09

Exercise			**3**					**4**		
	Pre		**Post**		**p**	**Pre**		**Post**		**p**
	Mean	**SD**	**Mean**	**SD**		**Mean**	**SD**	**Mean**	**SD**	
Ipsilateral										
IO/RA	8.26	3.40	9.90	6.28	0.20	4.04	3.03	5.88	4.65	0.04
IO/EO	2.70	1.96	3.35	1.95	0.02	0.43	0.32	0.95	0.61	< 0.001
MF/ICLT	1.00	0.36	1.48	0.78	0.15	2.09	1.36	3.21	2.25	0.02
Contralateral										
IO/RA	3.34	1.86	4.37	2.91	0.01	10.02	5.07	10.77	9.66	0.72
IO/EO	0.77	0.38	1.53	1.22	0.01	1.83	1.50	2.43	1.72	0.11
MF/ICLT	1.48	0.79	2.48	1.92	0.31	0.86	0.39	1.25	1.99	0.32

Exercise			**5**					**6**		
	Pre		**Post**		**p**	**Pre**		**Post**		**p**
	Mean	**SD**	**Mean**	**SD**		**Mean**	**SD**	**Mean**	**SD**	
Ipsilateral										
IO/RA	3.42	3.35	5.27	3.49	0.02	3.36	2.84	4.99	3.83	0.05
IO/EO	0.30	0.22	0.67	0.50	< 0.001	0.40	0.25	0.88	0.68	0.001
MF/ICLT	1.63	1.05	2.58	1.50	0.004	1.11	0.37	1.64	1.05	0.01
Contralateral										
IO/RA	10.13	6.47	11.89	9.77	0.44	8.08	5.65	8.23	6.36	0.92
IO/EO	1.65	1.33	2.43	1.75	0.04	1.57	1.09	2.17	1.70	0.06
MF/ICLT	0.89	0.65	0.92	1.38	0.91	1.01	0.28	1.27	1.21	0.31

IO = internal oblique; MF = lumbar multifidus; RA = rectus abdominis; EO = external oblique; ICLT = iliocostalis lumborum pars thoracis.

DISCUSSION

Stabilization training involved isolated local muscle contraction and an integration of the local and global muscle systems during particular movement patterns.[1] It was thought that such training with specific attention paid to the transversus and multifidus[1,2] would significantly increase the relative muscle activity of the local muscles in healthy subjects and consequently change the local/global ratio. Although all ratios increased after training, only the activity of the local internal oblique (representative for the transversus abdominis) increased significantly and not the activity of the local multifidus. Patients often report more difficulties in performing an isolated local back muscle contraction than an isolated local abdominal muscle contraction. The use of surface EMG might also have influenced the results.[3] The relative muscle activity of the RA also increased, but isolated local abdominal muscle contraction was often reported to be associated with co-contraction of other abdominal muscles.[2,4]
Similar stabilization training in symptomatic chronic low back pain patients with clinical evidence of spondylolysis or spondylolisthesis also resulted in a significant increase in the IO/RA ratio during an abdominal drawing in manoeuvre.[5]

CONCLUSION

The results of the present study indicate that muscle recruitment patterns can be changed in healthy subjects by means of a training program that focuses on neuromuscular control. Additional studies are needed to evaluate this type of training as a prevention strategy.

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POLYMYOGRAPHIC STUDY OF THE PELVIC FLOOR INNERVATION IN DAILY AND THERAPY RELATED ACTIVITIES

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INTRODUCTION

The innervation characteristics of the pelvic floor muscles (PFM) in the processes of daily activities and therapy activities remains inconclusive. Current treatment and training strategies tend to concentrate on isometric activation of the target muscles and the efficiency of such activities is unknown. Knowledge about the coordinative involvement of the PFM is very important for planning preventive and therapeutic trainings strategies and concepts. The aim of this study is the evaluation of the involvement of the PFM in daily and therapy related activities. The results can promote training of the target muscle in a variety of ways.

METHODS

The 35 participants of this study were students of the German Sports University at Cologne and women of a midwife's practice. All participants were subjectively free of problems concerning incontinence and also had no specific experience with pelvic floor training. The neuromuscular activity was measured by a NORAXON EMG-System. The following muscles were detected: Mm. pelvic floor, M. obliquus internus, M. obliquus externus, M. rectus abdominis, M. longissimus, M. multifidus, M. gluteus maximus, Mm. adductores. The collection of the EMG data occurred according to the guidelines of ISEK and SENIAM. The pelvic floor muscle was recorded via a vaginal probe (Fa. Medicheck). The raw EMG was sampled with 1500 Hz in a band of 10-500 Hz. All data was analysed by the EMG software "MyoResearch XP". The raw EMG data were full wave rectified and smoothed (RMS 100 ms) and amplitude normalized to the highest activity level of a MCV test sequence. Afterwards all data were statistically analysed via a multifactorized analysis of variance by using the software EASYSTAT.

RESULTS

The results give evidence for the important duty of the PFM involvement in activities of daily living such as therapeutic investigations. In activities of daily living the PFM was seen as the most active muscle in relation the others. This would imply that they are important muscles for trunk stabilization. The study of the therapeutic investigations shows that the target muscle can be trained by many different kinds of exercises. An additionally willful activity of the PFM tends to result in more activation and can therefore be integrated in exercises of higher abdominal pressure.

n= 35		standing	lifting	p-value	standing	stabilization	p-value	standing	Crunch	p-value	standing	Leg-press	p-value	standing	spelling	p-value
Pelvic floor	x	12,92	34,06	0,001	12,92	34,58	0,001	12,92	30,7	0,001	12,92	38,23	0.001	12,92	14,88	n.s.
	s	6,6	17,35		6,6	18,22		6,6	15,04		6,6	19,86		6,6	16,88	
M. obliquus internus	x	7,33	12,31	0,001	7,33	26,1	0,001	7,33	44,17	0,001	7,33	5,17	n.s.	7,33	13,93	0.05
	s	6,23	8,36		6,23	25,03		6,23	15,7		6,23	3,54		6,23	14,59	
M. rectus abdominis	x	3,33	5,17	0.05	3,33	5,3	0,001	3,33	46,01	0,001	3,33	3,29	n.s.	3,33	3,69	n.s.
	s	3,01	5		3,01	3,19		3,01	10,10		3,01	2,55		3,01	2,3	
M. obliquus externus	x	5,38	7,72	0,001	5,38	13,07	0,001	5,38	38,1	0,001	5,38	4,96	n.s.	5,38	7,14	0.001
	s	3,31	4,52		3,31	4,55		3,31	16,52		3,31	2,94		3,31	4,81	
Mm. adductores	x	1,4	11,7	0,001	1,4	5,54	0,001	1,4	7,73	0,001	1,4	10,09	0.001	1,4	1,67	n.s.
	s	1,35	9,25		1,35	6,23		1,35	6,37		1,35	6,85		1,35	2,04	
M. erector spinae	x	3,70	25,05	0,001	3,70	11,69	0,001	3,70	8,16	0,001	3,70	12,60	0,01	3,70	5,6	n.s.
	s	2,58	11,91		2,58	8,11		2,58	4,61		2,58	9,28		2,58	5,6	
M. multifidus	x	4,08	29,99	0,001	4,08	11,76	0,001	4,08	6,98	0,001	4,08	13,78	0,001	4,08	6,64	0,05
	s	3,00	11,75		3,00	7,60		3,00	5,3		3,00	8,93		3,00	7,01	
M. glutaeus maximus	x	3,3	25,68	0,001	3,3	5,21	n.s.	3,3	4,82	0,05	3,3	17,17	0,001	3,3	4,3	n.s.
	s	3,82	15,94		3,82	4,51		3,82	3,57		3,82	8,45		3,82	5,78	

Tab. 1: Mean data (x) and standard deviations (s) and p-value of the activities standing compared to activities of daily living and therapeutic investigations.

Exercises that were combined with training machines such as the "leg-press" or the "hip adductor" yielded the greatest change. Some established exercises like voice-based or "Kegel-exercises" were unable to reach an adequate activation level (≤ 30% MVC) that is necessary for PFM training.

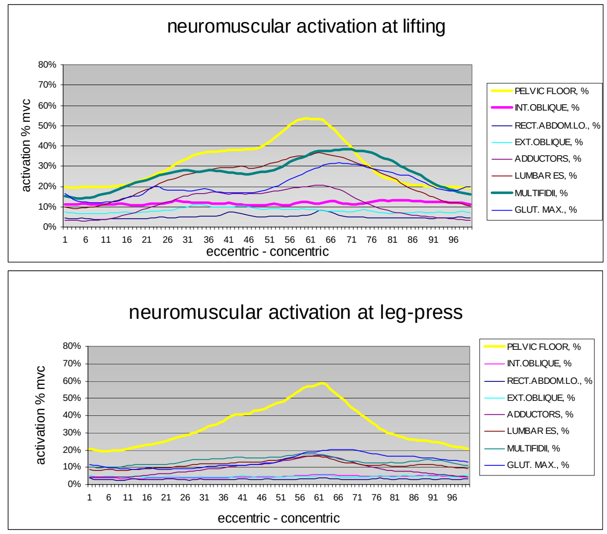

Fig. 1: The average activation profiles (N=35) of the analyzed muscles for the movements "lifting" and "leg-press".

DISCUSSION

The results show in an impressive way the importance of the integration of PFM in activities of daily living and therapeutic investigations. The data revealed that PFM can effectively be trained by dynamic exercises in preventive and therapeutic strategies. Training with both machines demonstrates high activation while the abdominal pressure is low. Furthermore, it seems reasonable to infer that with the use of complex exercises the non-voluntary activation of the PFM can be facilitated. This is important in therapeutic fields, where the effect of PFM training in the treatment of i.e. incontinence is still unclear.

CONCLUSION

The results of this study emphasizes the involvement of the pelvic floor muscle in the movements of daily activities such as therapeutic investigations. In activities of daily living PFM seem to stabilize and protect the trunk and the viscera in healthy women. The results of the therapeutic investigations seem to offer many new and more interesting possibilities of training the pelvic floor muscle. Especially for prevention, but also in therapeutic fields.

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ROLE OF PELVIC FLOOR INTRAVAGINAL SURFACE EMG IN THE DIAGNOSIS AND THERAPY OF FEMALE URINARY INCONTINENCE

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INTRODUCTION
Different methods for diagnosis and evaluation of therapy exist in the therapy of incontinence. The sensitivity and specificity of these methods are very different. Surface EMG (sEMG) is an easy, safe and helpful technique in the therapy and diagnosis of incontinence.

PURPOSE
Aim of this presentation is to discuss the clinical value of the intra-vaginal sEMG for selection of the right therapy.

MATERIAL AND METHODS
We recorded in 120 female patients with stress or urge urinary incontinence standardized vaginal sEMG. The sEMG recording was conducted in a comfortable lying position via periform intra-vaginal probe after a standardized protocol. sEMG was performed during maximal voluntary contractions over 3 times, 10 seconds and once over 60 seconds each interrupted by 10 seconds rest. All data were analysed by the EMG software with a simple moving average calculation (Gymna 420, Fa. Treumedizin). We used in this presentation the mean value of the pelvic floor contraction subtracted by the resting value.

Patients:
The mean age was 53(28-71), parity 2(0-10), already done pelvic floor exercises: 57. 69 patients presented with urge symptoms, 99 patients with stress incontinence symptoms. The mean urination frequency was 9.1 (4-32 SD 3.98), mean incontinence incidence per day 9.6 (0-35 SD 9.6).

Therapy:
60 patients were measured after therapy. 22 patients participated in group training with education, kinaesthetic, pelvic floor and general exercises. 25 patients underwent biofeedback and electro stimulation treatment under supervision. 13 patients received a home biofeedback /electro stimulation device.

RESULTS:
The sEMG values were shown to decline with age, parity and body mass index. Furthermore, we could show that a gynaecological operation significantly reduced the sEMG values. However, there was no correlation to the grade and intensity of incontinence, prior treatment with pelvic floor exercises alone, and the presence of stress or urge urinary incontinence. Furthermore, the number of incontinence, frequency of urination, amount of nocturnal urination were not correlated with significant changes in the sEMG values.

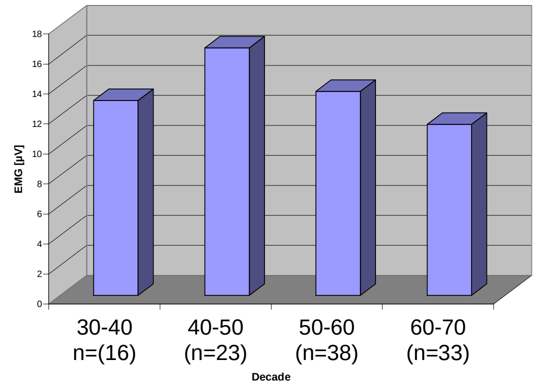

Fig. 1: Group mean EMG separated to in 4 decades

Patients treated by supervised biofeedback and electro stimulation showed significantly greater increases in sEMG values than the other two groups. 10 out of 15 patients learned with biofeedback/electro stimulation to contract the pelvic floor sufficiently. In the group setting, there were only 3 out of 13. However, there were no significant differences in the subjective outcomes between the groups. 54 out of the 60 patients reported a significant improvement of urinary incontinence.

DISCUSSION
There exist significant correlations between the sEMG values and several patient conditions. Older patients, with higher BMI, a history of gynaecological operation or multi-parity should receive a pelvic floor re-education program, preferably with biofeedback, because we expect a lower ability for a sufficient pelvic floor contraction. Mainly patients with a poor pelvic floor activity benefit from EMG-based biofeedback.

CONCLUSION
sEMG can help to select the most suitable conservative therapeutic method and to assess the outcome of therapy. sEMG biofeedback should be integrated more often in incontinence therapy.

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AGE, SYMPTOMS, AND MODE OF DELIVERY AFFECT ANAL SPHINCTER ASYMMETRY

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INTRODUCTION

Information concerning the motor unit (MU) structure of the external anal sphincter may be very useful in a number of clinical situations: for description of asymmetry, for investigation and quantification of muscle activity during stress, relaxation, and fatigue, for the differentiation between myogenic and neurogenic pathologies of the muscle, and for the identification of areas that may be more vulnerable to trauma and/or surgery because of the resulting possible denervation (1-4) or because of increased risk of post-intervention symptoms due to asymmetry of sphincter innervation (5). Two previous papers demonstrated the possibility to extract this information from the external anal sphincter (EAS) by means of S-EMG via an array of electrodes placed along the direction of the muscle fibers. Such methodology is based on an anal probe comprising three arrays, each of 16 equally spaced electrodes placed around the circumference of the probe and able to detect the generation, propagation and extinction of MUAPs produced by the MUs of this muscle (6,7). Assessment of external anal sphincter (EAS) innervation by surface-EMG (S-EMG) multi-electrode arrays (MEA) has recently been shown to be a reliable screening tool for routine procedures (8).

AIM

Surface-EMG (S-EMG) of external anal sphincter (EAS) by multi-electrode arrays (MEA) was used for a large series of gynecologic patients for the first time. We wished to determine the degree of asymmetry of sphincter innervation, its determinants, and the relationship to fecal incontinence.

MATERIAL AND METHODS

A cylindrical probe, 14 mm in diameter and carrying three arrays of 16 silver bars (1 mm diameter, 10 mm length, 2.75 mm apart) were specifically designed and built to record MUAPs circumferentially along the muscle fibers of the EAS during voluntary contractions. SEMG signals were recorded differentially between adjacent pairs of electrodes of the same 16-electrode array during maximal voluntary contractions (MVC) for 10 sec at each of three levels within the anal canal (Figure 1).

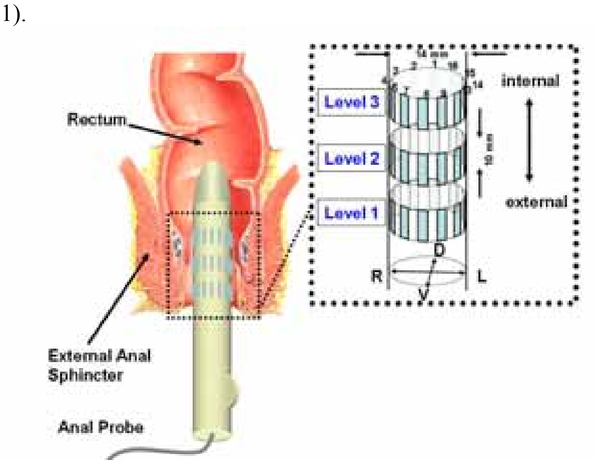

Fig. 1: Position and description of the anal probe.

In 129 continent women (age: 17 – 83) following childbirth and in 40 patients (age: 33 – 82) with fecal incontinence recruited in a gynecological hospital S-EMG of the EAS was performed and generation and propagation of individual motor unit action potentials (MUAP) were recorded in the anal circumference at 3 levels within the anal canal. Innervation zones (IZ) of MUAP, their circular distribution in the anal canal as measure of symmetry, as well as the statistical characteristics of amplitude and frequency of MUAP were compared between and across subgroups (age, continence status, number of deliveries).

RESULTS

IZ and amplitude were overall lower in incontinence, and differences between the three levels were less pronounced in incontinent patients. Age reduced the number of IZ and the amplitude ($p< .01$) (Fig.2). In incontinent subjects, lower amplitudes and higher frequency of EMG were found. Significant asymmetry of innervation was found for dorsal versus ventral, but not for left versus right comparison of amplitudes ($p= .004$) and frequency ($p= .001$) and was more pronounced in the distal anal canal for IZ distribution ($p=.007$) and in the proximal canal for EMG amplitude. Number of vaginal deliveries was associated with significant asymmetry of EMG parameters, especially in incontinent women (amplitude: $p<.001$, frequency: $p<.03$).

DISCUSSION

There are significant differences in sphincter innervation between female patients with fecal incontinence and controls. Older women showed decreased number of IZ and amplitude, but increased frequency. Number of vaginal delivery is associated with dorsal – ventral asymmetry. A considerable amount of information can be extracted from the S-EMG of the EAS. This study deals only with the number of IZs, their circular distribution in the anal canal and describing parameters (amplitude and frequency) at three levels during maximal voluntary contractions. Except for our recently reported work showing that it is possible to identify MUAPs of the EAS and that there is large interindividual variability, no previous experience on this topic has been reported.

CONCLUSION

Asymmetry of innervation of the EAS seems to play a role in incontinence, with contribution of age, number of childbirth, and the mode of delivery. (Supported by grants from the EU and the Fresenius Foundation, Germany)

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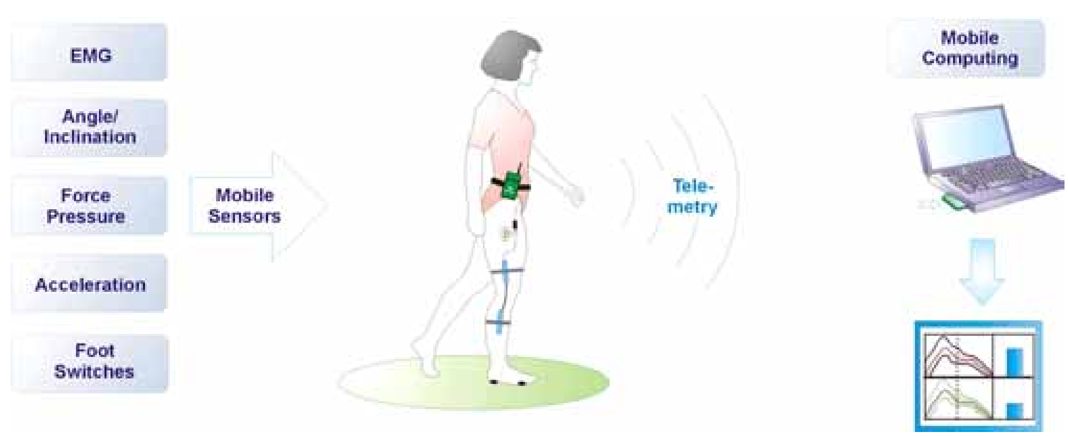

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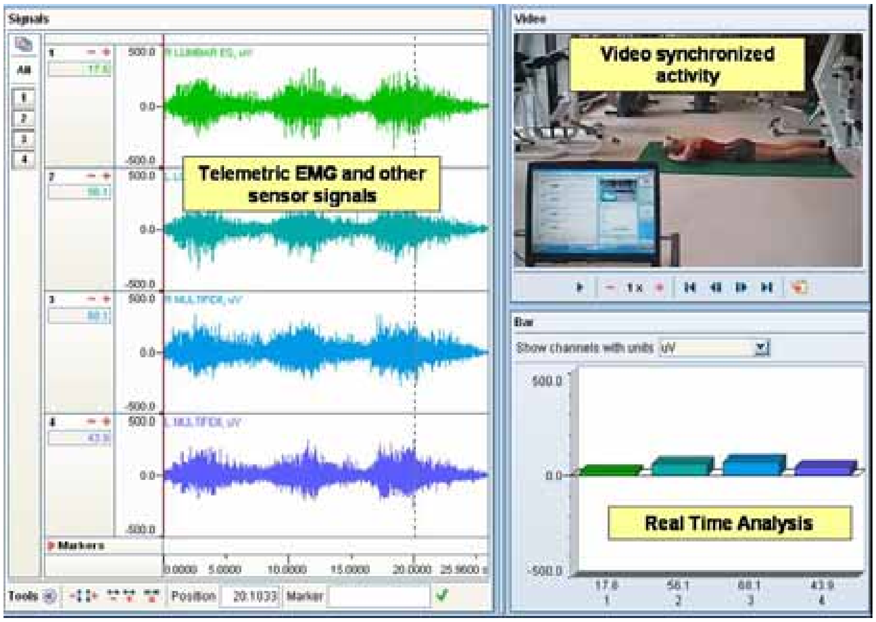

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