NORAXON EMG-Meeting 2007

Erweiterte Kurzfassungen

MONITORING OF MOTOR RECOVERY AND RESTAURATION DUE TO NEUROPHYSIOLOGICAL INTERVENTIONS

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INTRODUCTION

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 not high level spasticity could 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 neurophysiological 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).

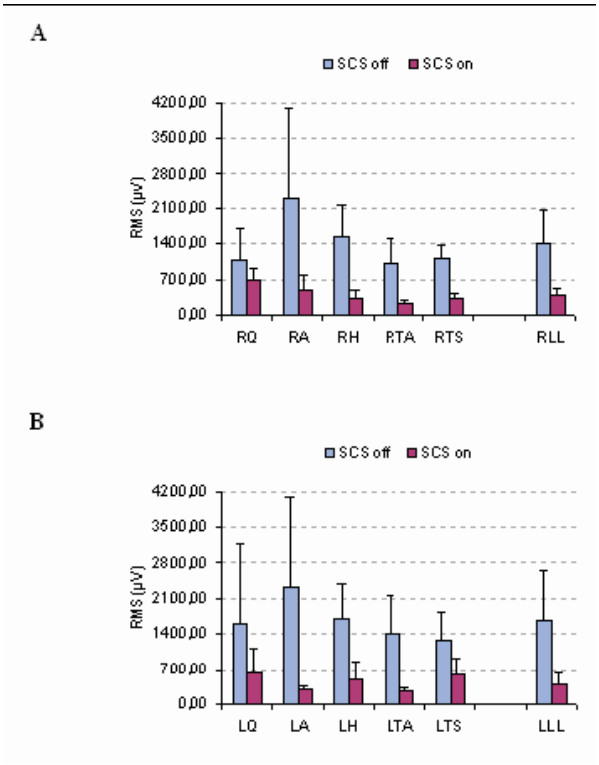

Figure. 1:
RMS (Root Mean Square) of the right (A) and the left leg(B) within SCS on compared to SCS off.

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

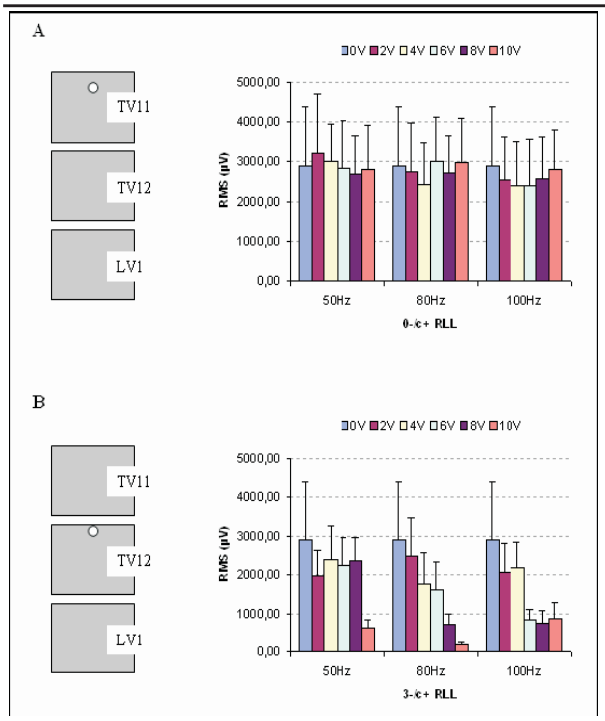

Figure 2
Antispastic effekt of SCS primary being dependent on the placement of cathode, secondary dependent on stimulation parmaters

polarity alone can induce a difference in the anti-spastic effect (Fig. 2).

In this particular case, we did not only test different location 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).

With regards to the variability of lumbar spinal cord segments in relation to the vertebrae, an intra and post surgery neuro. physiological evalution 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 tibilias anterior and triceps surae. The variability of localisation 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 neurophysiological 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 an 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.

CONCLUSION

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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EVALUATING MOTOR RECOVERY IN NEUROLOGICAL REHABILITATION WITH SURFACE-EMG

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INTRODUCTION

CNS lesions often change motor control ability of the CNS with attendant modification or even elimination of voluntary movements as we can find it after stroke. Multi-muscle sEMG in accordance with the brain motor control assessment (BMCA) protocol[1] is a method to determine the status of brain reorganisation for features of motor control. Simplest volitional movements are performed as well as reinforcement and passive manoeuvres. Features of motor control are assessed by sEMG and are visually analysed by a tester. Therefore commonly used sEMG parameters like amplitude, frequency, recruitment, inhibition and coactivation parameters in the ipsilesional and contralesional side are assessed. Unfortunately, no quantitative analysis method exists for patients with central lesions and for the upper extremities. This is due to limitations of sEMG recording and measurement during voluntary movement and to the choice of strategies of motor control employed by the subjects[2]. Approaches for quantitative analysis were only made for persons with incomplete spinal cord injuries and for the lower extremities 2, 3. So the aim of this study was to find appropriate parameters to quantify features of multi-muscle sEMG patterns which prescribe the changes in motor control as we have seen them by visual analysis.

METHODS

We analysed retrospective multi-muscle sEMG recordings from persons with stroke history (5 male, 4 female, 48±14 years) post stroke (6±5 month post stroke, 4 left side lesion, 5 right side lesion) with noticeable alterations in motor control features seen in sEMG between two sessions. We concentrated on two volitional manoeuvres, the wrist extension - flexion manoeuvre (EFM) and a finger tapping manoeuvre (FTM). We analysed the sEMG data from the wrist extensor muscles (WE), wrist flexor muscles (WF), biceps brachialis (BiBr) and the first dorsal interosseus muscle (FDI). A bandpass of 10 to 500 Hz had been used for recording. Event markers had been set to denote the delivery of the cuing tone to initiate voluntary manoeuvres. sEMG data were transferred from our recording software (Pegasus) to Noraxon software and RMS was used for smoothing. For EFM we set markers at each fall of the event to separate extension and flexion. Then we analysed the area, mean and peak values of each phase and calculated the quotients for agonist/antagonist (extension: antagonist coactivation extension = 100 * area(WF) / area(WE) and flexion: antagonist coactivation flexion = 100 * area(WE) / area(WF) and the ipsilateral proximal cocontraction (extension: proximal coactivation extension = 100 * area(BiBr) / area(WE) and Flexion: proximal coactivation flexion = 100 * Area(BiBr) / Area(WF)) of the paretic limb and compared these quotients in between the sessions. For FTR we concentrated on the interburst – inhibition (IbInh). We separated the burst (active period) from the interburst period (paused period) and calculated the IbInh index as ratio of the Mean RMS of theses two periods: IbInh = 100 - 100 * mean (FDI) pause / mean (FDI) active. The IbInh index was compared between two sessions. Because of the low number of patient data and the preliminary character of this investigation, no further statistics was done so far.

RESULTS

EFM: The antagonist coactivation in extension decreased from 109±94 to 63±36, in flexion from 61±37 to 43±12, the proximal coactivation in extension from 184±247 to 48±35 and in flexion from 212±289 to 54±17. In 4 patients we saw a lower antagonist coactivation in wrist extension and also in 4 patients in wrist flexion. But only in two persons we could confirm our findings of a lower antagonist coactivation from visual analysis in both extension and flexion. In proximal coactivation all 6 patients showed less proximal coactivation in the extension and 5 in the flexion manoeuvre.

V FTM: The mean amplitude increased from 28±26 to 87±54 in the post session in FDI muscles. The interburst inhibition index increased from 31±22 to 44±20. The post test of all 6 patients showed higher mean amplitudes. 4 patients showed an increased interburst inhibition at the post test, two did not.

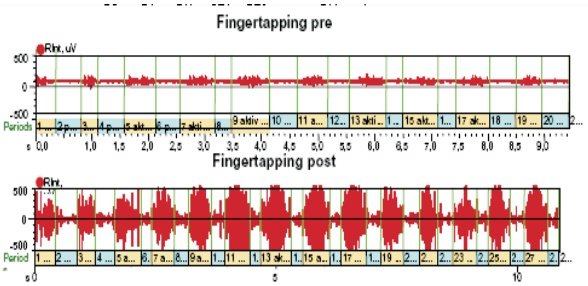

Fig. 1: sEMG raw signal from FTM with markers.

DISCUSSION

The results show us, that agonist/antagonist coactivation and proximal co contraction as well as the interburst-inhibition as calculated in our study could be appropriate parameters for evaluating motor recovery. But we could not confirm the results of the visual analysis in all cases. This confirms the difficultness for quantitative evaluation of motor control measured with sEMG due to high parameter variability across subjects and measurements. Another source of variability refers to different phases of restorations of motor function[6] in stroke patients. This and our findings suggest that the parameters used are not appropriate for a general use in evaluation of features of motor control. Therefore inclusion criteria for a selective use of this parameters are necessary. The best consistency we found in the proximal coactivation of the extension manoeuvre, where all patients showed a lower coactivation in the post test. This result is similar to findings from Sherwood et al. [4,5], who could show a superiority of sEMG data over Ashworth category as an objective quantification of altered motor control (spasticity).

Nevertheless considerations while evaluating these data made us think that the interburst-inhibition could be a valid and reliable parameter which cannot be strongly influenced by tester and subject.

CONCLUSION

The purpose of this study was to find an approach to the quantitative analysis of sEMG data from a multi-muscle sEMG examination assessed with a BMCA protocol. We have been focusing on two voluntary motor tasks which are very simple and do not allow many strategies for performance. We used 3 parameters of which we expected to consistent with our visual analysis. The results show that the used parameters for quantitative documentation of multi-muscle sEMG are useful but show limitations in the evaluation of alterations in CNS motor control. This supports that further investigations are needed to find appropriate inclusion criteria for a selective use of parameters.

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MUSCULAR STRAIN AND COMPLAINTSAT THE UPPER EXTREMITIES DURING OFFICE WORK

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INTRODUCTION

Persons performing office work often complain of pain in the musculoskeletal system in spite of the relatively low muscular forces, which have to be exerted during such work.Complaints may result from the long-term activation of the muscles needed for the stabilisation of the posture at the workplace and from disadvantageous working conditions. In a field study on office workers, working conditions, musculoskeletal symptoms, myoelectrical activities and the work flow were investigated.

METHODS

69 employees of the German tax authority working in the same tax office were involved in the study. The working conditions were studied using a questionnaire regarding the activity of the persons, the used working tools and the local arrangement of the equipment. Musculoskeletal symptoms were investigated applying a modified and enlarged version of the 'Nordic Questionnaire' of Kuorinka et al. [1].

For a subgroup of 13 employees, a detailed workphysiological study was performed. During total working shifts 4 surface electromyograms (EMG) were recorded in the right shoulder/arm region of the subjects from the m. trapezius (pars descendens), m. deltoideus (pars acromialis and pars clavicularis) and from the m. extensor carpi ulnaris.Simultaneously the actual task of the persons was documented using a classification procedure described by Luttmann et al.[2]. Additionally, at four points in time during the working day (begin of the working shift, before and after the noon break, end of the shift) the subjects were asked to indicate the degree of actual musculoskeletal complaints. For this purpose a questionnaire developed by Corlett and Bishop [3] and extended by Schmidt et al. [4] was applied which is based on a schematic diagram representing the contour of the human body. The persons under study were asked to indicate painful body areas according to the pain intensity.

RESULTS

Questionnaire

In the first part of the study regarding work content, working conditions and the musculoskeletal symptoms of the 69 employees, it was found that 70 % of the persons performed computer work during up to 50% and paper work up to 40 % of their working time. Computer work was predominantly performed using a desktop with discrete keyboard and screen; only 13 % of the subjects have used a laptop.

About 50 % of the persons indicated complaints for at least 8 days during the last 12 months for the neck and the lower back, and about 40% and 30% for the right or the left shoulder, respectively.

Work-physiological recordings

In figure 1 typical results of the electromyographical measurements are summarized. The Electrical Activity (EA) of the right trapezius muscle, representing the rectified and time averaged raw EMG, is shown for the four most important activities (four upper diagrams) and for the total activity (lowest diagram). Findings are shown for 12 subjects, since for one of the 13 persons under study the EMGs were excluded from the data analysis due to a poor signal quality. In the right column the mean value ± standard deviation for the 12 persons is presented. It can be concluded that for the trapezius muscle the highest values of the Electrical Activity were observed during paper work, whereas the lowest values were found during mouse application. During the use of the keyboard and during secondary activities intermediate EA values were recorded for the trapezius. Similar results were obtained for the other shoulder muscles. For the forearm, the highest muscular activity was found during keyboard operations.

Temporal changes in the EMGs were analysed with respect to the EMG amplitude and the frequency spectrum. A previously developed method for 'Joint Analysis of the EMG

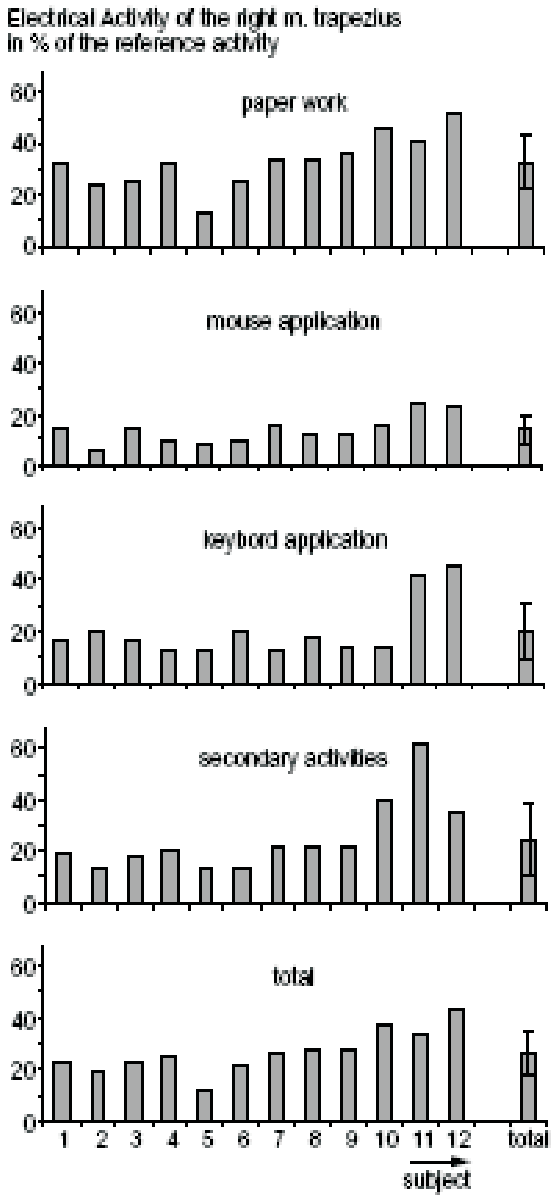

Figure 1:
Mean Electrical Activity of the right m. Trapezius during full working shifts for different activities for 12 persons and mean value ± standard deviation for all persons (right column); adapted from [5]

change into the categories 'fatigue', 'recovery', 'force increase' or 'force decrease'. Application of this method to the EMGs of this study [7] reveals, that for several persons muscular fatigue – as indicated by a time-related increase in the EMG amplitude and a simultaneous left-shift in the EMG spectrum – was found for the hand extensor during keyboard operation and for the right trapezius muscle during paper work.

For the majority of persons an amplitude decrease and a simultaneous spectral left-shift in the course of the working shift was observed. According to the JASA method, for such EMG changes a decrease in the force production of the studied muscles during the day has to be assumed. Further analyses reveal a correlation between such time-related EMG changes and the number of actual complaints. In particular, persons with a steeper decrease in the EMG amplitude of the shoulder muscles mentioned a lower number of shoulder complaints during the working day than persons with a smaller change in the EMG amplitude.

DISCUSSION

The analysis of the temporal changes in the EMG parameters Electrical Activity and Median Frequency reveals that for a part of the subjects muscular fatigue was found, in particular for the forearm during keyboard operations. Furthermore, for the forearm a relatively high number of complaints with the tendency to increase over the working day was indicated. These findings may refer to a connection between the development of muscular fatigue and pain, especially in the forearm, where, on the one hand, static activity is needed for an exact positioning of the hand and, on the other hand, repetitive movements with high accuracy have to be performed, in particular, during typing.

For the majority of the subjects a decrease in the force production during the working day was observed. For the interpretation of this finding it may be assumed, that the working capacity of the persons is lowered in the course of the working day. Accordingly, a constant or increasing muscular activity may result in an overload for a certain part of the working day. Instead, lowering the muscular activity in the course of the day may reflect a self adjustment of the muscular activation to the actual working capacity in order to avoid a possible overload. This finding suggests that lowering of muscular activation during the day may be helpful to prevent muscular complaints.

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THE REGULATION OF MOTION FOLLOWING STRUCTURAL CHANGES OF THE SENSORIMOTOR SYSTEM.

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INTRODUCTION

The sensorimotoric system results from the circular combination of the following structures: sensors – afferent pathways – the central neural structures – efferent pathways – musculature. The muscle-tendon complexes, the joint connective tissues and the skin are the locations of the sensors. Before the movement they provide information about posture and joint positions and during the movment they characterize the course. The movements are the origin of feed back information. Thereby the curcuit of the SMS is closed and the fine regulation of the movements is possible.

METHODS

SurfaceEMG(Noraxon; MyoResearch 1.26.21) of the v.astus medialis, vastus .lateralis and rectus .femoris muscle was measured (1kHz) during a maximal isometric contraction by 24 ACL-patients at the day before reconstruction and weekly up to the 6th week, by 76 ACL-patients between the 7^{th} and 12^{th} week and by 103 total knee arthroplasty (TKA)-patients at the day before, 1 week and 3 weeks after operation. Also the EMG of the gluteus medius muscle during gait by 24 total hip arthroplasty(THA)-patients at the begin and after 3 weeks of a rehabilitation process in a rehabilitation clinic was documented. RMS over a period of 2 or 4 seconds (TKA) or over 10 gait cycles (THA) was calculated. Furthermore correlation analyses were also carried out between the results of the investigation appointments.

RESULTS

The 24-patients ACL-group showed at the day before the reconstruction on the injured site RMS-values of 54 % (median values), after 1 week 14 % and after 6 weeks 45 % of the healthy site. By the 76-patients ACL-group the EMG was reduced 50 – 65 % of the contralateral site and the strength defizite was 49 10 %. The EMG of the 103-TKA-group was diminished by 30 % at the day before, nearly 80 % after 1 week and 50 % after 3 weeks. The results for the correlation between the EMG at the day before and after 1 week are: TKA-site/ non TKA-site rectus fem.: 0,42/0,66; v.astus med.: 0,40/0,67; vastus .lat.: 0,57/0,85 respectivly. The correlation betewen the testand retest-EMG of the gluteus med. muscle is 0,84.

There are some additional results of the EMG from patients following shoulder injuries, femoral fracture and tibia osteotomy with strong diminished activities.

DISCUSSION

Joint injuries and degenerative diseases cause a disturbed function of the sensorimotoric system. The patients show a distinctive “functional paralysis”. This functional paralysis probably concerns the fast muscle fibre population and in the older patients the “fast part” of the slow muscle fiber population too. The EMG data show, that the reduction of IEMG is about the same for all three parts of the quadriceps muscle and the same can be found for the gluteus medius muscle and the muscles of the shoulder.

CONCLUSION

A functional scar will be the final result. In the consequence of all facts the muscles can be activated no longer physiologically and the therapeutic training effects must be clearly decreased and need more time.

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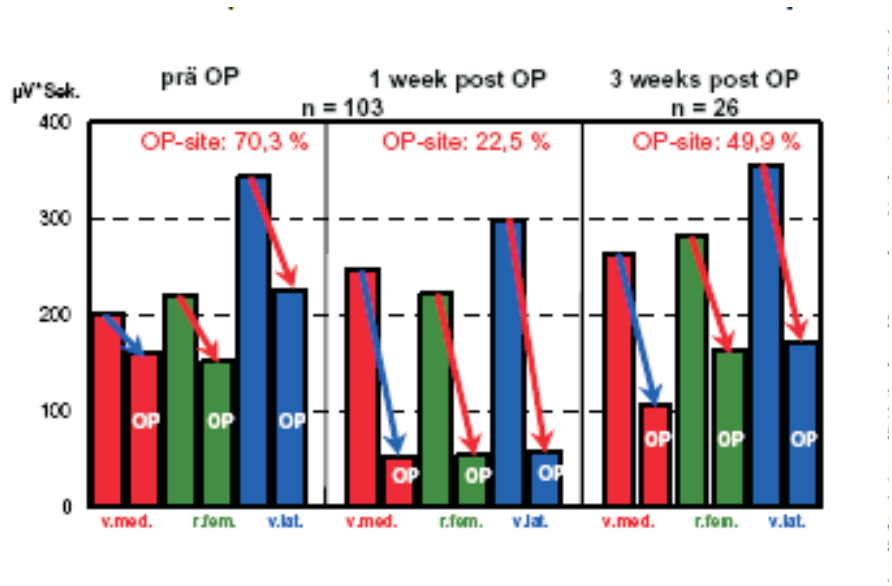

Figure 1
EMG-activities of the 103 patient TKA-group before, 1 week and 3 weeks following operation

NEW GENERATION OF SEMG TELEMETRY DEVICE OPERATING WITH DIRECT SIGNAL TRANSMISSION FROM THE ELECTRODES

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INTRODUCTION

Surface EMG technique is the most common and simple method to analyse the functionality of the muscle during isometric contractions or dynamic movements.

Especially during dynamic activities, traditional sEMG systems involve some practical difficulties, such as movement artefacts, patient conditioning, long time preparation of patients. What we propose is a commercial system fully digital and wireless which makes sEMG measurements very practical, reliable and quick.

METHODS

The system is made by miniaturized SMD (surface mount device) and low power consumption electrical components. Basically the system is composed by two parts: the main unit (receiver) and wearable EMG probes (see fig. 1 below). Each probe is able to collect, A/D convert, condition and amplify the EMG signals and transmit EMG data through a specific WiFi protocol. No connection cable are adopted and existing between the probes and the main unit. All the probes are synchronized more than 150 times per second, in order to assure maximum reliability.

RESULTS

ZeroWire system is able to guarantee transmissions between probes and main unit up to 20 meters. Thanks to a specific WiFi digital transmission protocol, the system is interference free, either versus LAN WiFi systems, Bluetooth devices and mobile telephones. The maximum sampling rate is 2 kHz per single channel, and the bandwidth is 10 – 500 Hz for sEMG and 10 – 1000 Hz for fine-wire EMG. The EMG probes are small and light: 33x23x19mm and 12 grams weight. No ground reference electrode is needed.

Two special probes are provided for gait cycles detection (footswitches); in this case up to 4 FSR sensors can be connected for each probe.

Figure 1
ZeroWire EMG electrodes worn by a 4 years old child.

The main unit provides both digital (USB) and analog outputs, simultaneously available. ZeroWire system assures complete compatibility and connectivity with the main movement analysis systems, in particular with Vicon and Noraxon MyoResearch XP software, thanks to trigger and I/O sync ports and SDK libraries.

Product is certified as Medical Device within class IIa.

DISCUSSION

ZeroWire reveals adequate for several applications, such as rehabilitation, neurology and neurophysiology, sport science and ergonomics. ZeroWire system provides an extremely high quality level of signals and allows the user to drastically reduce the necessary time period for patient preparation. The particular proposed solution allows to remove the ground reference electrode and significantly reduces the artifact due to the movement of wires.

The absence of cables around the subject, the extremely small size and the low probe weight offers high comfort for the patient, free to move without constrains.

In turn, ZeroWire allows to maximize the patient performance (more natural movement) and the reliability of the measurements.

CONCLUSIONS

ZeroWire system results reliable and particularly suitable for clinical and research purposes, as demonstrated by the worldwide success it got.

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- Technion University – Israel
- Technical University of Tokyo - Japan
- University Hospital UZ Leuven – Belgium
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MACHINE-BASED PELVIC FLOOR MUSCLE TRAINING – IS IT POSSIBLE?

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INTRODUCTION

Whether it is possible to strengthen the pelvic floor muscles with training machines is a question that has not been seriously looked at to date. With some 24% of Germans suffering from urinary incontinence (1), it would be quite significant if such a method could be found, particularly since over 6% of Germans are already members of fitness clubs (2). Putting incontinence patients through a machine-based restistance training programme would be much simpler than the known and practiced pelvic floor strengthening regimes.

The aim of this study was to examine the use of pelvic floor muscles (PFM) in machine-based resistance exercises under two conditions: regular training (RT) and active pelvic floor contractions during the concentric phase of each repetition (APFCcon). The results are discussed and reflect pelvic floor activity levels at rest and with APFC during lying, sitting, and standing.

METHODS

The study involved 6 females (height: 167.7 (±8.2) cm, weight: 64.5 (±11) kg, age: 33 (±6.7) years) without pelvic floor problems and no pathological findings using a multi-activity test (3). All had done machine-based weight training before although not all regularly and not necessarily on the machines used in the study.

The following resistance training machines were used to examine the potential involvement of the PFM: hip adduction (HA), knee flexion in prone position (KF), hip extension in lateral position (HE), hip flexion in lateral position (HF), and leg press (LP). The HA and LP were done with backrest in both steep and flat positions. Therefore, the subjects did 7 different exercises in each of the RT and $APFC_{con}$ modes. Training intensity was 55% of 1 repetition maximum (RM). Each repetition lasted 8 seconds (4 concentric / 4 eccentric) and 8 continuous repetitions were recorded. We recorded surface electromyograms (EMGs) of the PFM (Innocept Stimpon vaginal probe), m. obliquus internus, m. rectus abdominis, m. gluaetus maximus and m. adductores using NORAXON® Telemyo 2400T and a bandwidth of 10-500 Hz. Signals were A/D converted with 1500 Hz and stored in MyoResearch XP software followed by ECG reduction, full wave rectification, smoothing with RMS 100 ms and amplitude normalization to the highest activity level during Maximum Voluntary Contraction (MVC) detected from 5 different MVC tests. The mean EMG-curves of subjects were put into MyoResearch normative groups. SPSS® 15.0 was used for the statistics.

RESULTS

RT results showed mean PFM activity of 19%-25% of the highest activity level during MVC for all machine- based exercises but H^{A}steep/flat. ^{A}PFCcon increased mean activity levels during those exercises to 27-29%. Mean EMG-elevations through $APFC_{con}$ were statistically significant for all exercises other than $_{LP}$flat (all p=0.05; Wilcoxon). Activation was much higher during the two HA-exercises (Table 1).

Tab. 1: HA_{flat} and HA_{steep} – PFM activity as % of highest activity level during MVC

Exercise	RT			$APFC_{con}$		
	con	ecc	rpt	con	ecc	rpt
HA_{flat}	59±17	38±17	47±20	69±13	40±17	53±21
HA_{steep}	51±11	30±11	40±15	60±9	32±11	45±17

The positive influence of the flat backrest position was not significant but clearly evident in 3 of the 6 subjects. $APFC_{con}$ led to significantly higher mean and concentric EMG-values than RT during HA (Wilcoxon, p=0.003).

For all exercises, mean concentric values were higher than mean eccentric values; with the smallest difference in ^{LP}flat (RT: 5%, $APMC_{con}$ 9%) and highest in HA_{flat} (RT: 23%, $APMC_{con}$ 30%).

DISCUSSION

Regular machine-based resistance training offers mostly a PFM activation from 16% - 30%. This lies within the range of APFC during lying, sitting, and standing (see Table 2) which can be taken as typical for regular isolated PFMT. Other, more complex PFMT-exercises as segmental stabilization or crunch with rotated pelvis might cause higher PFM activation. However, knowledge about such values is scarce.

Machine-based hip adduction showed very high activation levels compared to assumed values for regular isolated PFMT of up to 82% (group mean peak during RT at $_{HAflat)}$. Due to the lack of reference values for traditional PFMT-exercises the potential advantages of machine-based resistance training are not yet proven but require further attention and research.

Tab. 2: Reference values – activity of the pelvic floor muscle group in % of highest activity level during MVC

Activities	*without APFC*	**with APFC**
Lying	3	25
Sitting	3	30
Standing	8	23

CONCLUSION

We showed that PFMs are clearly involved in tested machine based exercises. Pelvic floor activation during the concentric phase was at least as high as during isolated APFC. We found remarkable PFM activation during machine-based hip adduction. All the machine-based exercises done in the study

allowed an intensification of PFM activity by ^{A}PFCcon. Recommendations for the PFM exercise dosage are based on the exercise physiology principles for normal skeletal muscle (4). In any case, these are mostly used in machine-based weight training. The activation levels observed during machine-based PFMT seem sufficient for PFM training purposes and so there are several reasons why the suitability of machine-based PFMT for preventive PFM strengthening may be possible and should be tested. In addition, the therapeutic use of machine-based PFMT and the role of coordination training during therapy must also be clarified.

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THE INFLUENCE OF DIFFERENT TESTING CONDITIONS IN RUNNING ON GENDER SPECIFIC EMG- ACTIVITY AND KINEMATICS OF THE LOWER EXTREMITY – A WAVELET-BASED ANALYSIS

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INTRODUCTION

Running mechanics are considered to be different between male and female runners (Ferber et al., 2003). To investigate running mechanics, different testing conditions such as treadmill- or over ground running on laboratory tracks with different path lengths can be used. Kinematics are found to differ between over ground and treadmill running, but there is little information about changes in EMG activity depending on testing condition (Nigg et al., 1995, Wank et al., 1998). Therefore the purpose of this study was to investigate a) if kinematics and EMG signals in running vary depending on testing condition, b) if there are gender dependent differences in muscle activity or kinematics and c) if the muscular / kinematical response to changes in testing condition is gender specific.

METHODS

Six female and six male volunteers were tested running (3.0m/s) on a treadmill (TM), on a 10m laboratory track (LT) and during continuous over ground running (CR). Surface EMG was recorded (3000Hz) with unaltered electrode placement from mm. tibialis anterior, peroneus longus, gastrocnemius medialis and lateralis, soleus, vastus medialis and lateralis, semitendinosus and biceps femoris. Kinematical data of knee and ankle joint was obtained by a highspeedcamera (125Hz) in the sagittal plane and a custom designed rear foot goniometer in the frontal plane. For each testing condition three sets of data were recorded to test innercondition variability. Selected EMG - variables were computed based on a wavelet analysis of the signals.

RESULTS

For all subjects there are differences in EMG signals as well as in kinematics while running in different testing conditions (example of changes in rear foot angle depending on running condition in fig. 1). These muscular and kinematical responses to changes in testing condition are highly individual and deviate between the tested subjects.

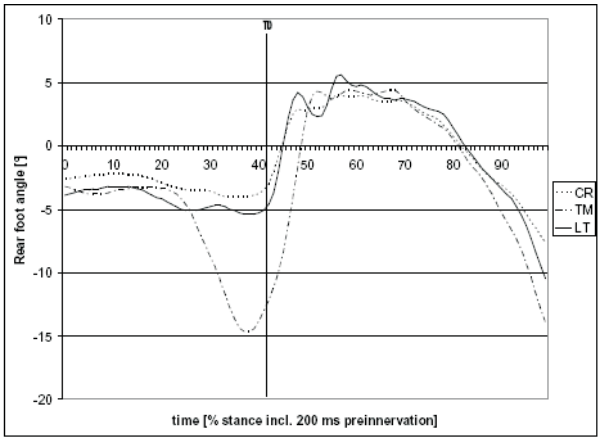

Fig. 1: Mean values of rear foot angle of one male subject while running in the three investigated testing conditions (nine steps per condition; TD = touch down).

Several mean values of male and female runners differ in EMG signals and kinematics ($p<0.05$).

In addition the muscular and kinematical responses to changes in testing condition are gender specific (example of diverging muscular responses of one male and one female subject to the change from LT to TM running in figure 2).

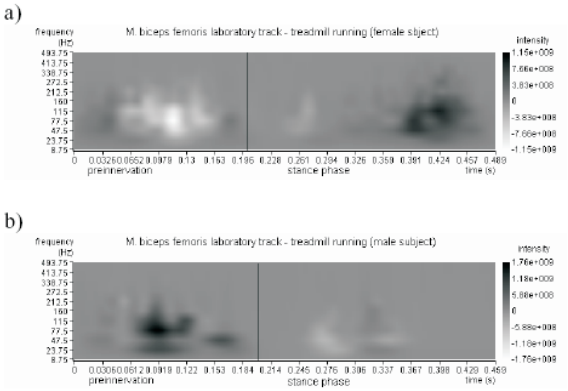

Fig. 2: Gender dependent EMG responses to varying testing conditions (example of a wavelet-based EMG analysis: difference patterns of activity patterns of LT and TM for two subjects: a) female, b) male. Light zones indicate higher intensity in EMG-signals during running on TM, dark zones indicate higher intensity in EMG-signals during running on LT).

DISCUSSION

Measured muscular and kinematical responses to changes in testing condition are highly individual. This corresponds to the results of Nigg et al. (1995), who found kinematical differences between over ground and treadmill running to be individual. Wank et al. (1998) found little differences in EMG signals between over ground and treadmill running, which may be due to investigating mean data and not examining individual data.

Although data are highly individual, statistical significant differences in gender specific EMG responses and kinematics exist. While interpreting these findings, however, it has to be considered that the procedure of averaging data results in smoothing individual responses. These responses have to be taken into account as well.

CONCLUSION

In this study differences in EMG signals and kinematics depending on running condition and gender have been found. Since calculated values are highly individual, care has to be taken by obtaining and interpreting mean values.

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DETERMINATION OF POSITION OF SURFACE ELECTROMYOGRAPHIC ELECTRODES FOR SELECTED EQUINE MUSCLES

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INTRODUCTION

Movement apparatus of horse has been given considerable attention in equine research, which has resulted in development of various imaging techniques for exploring functional details. These techniques provide useful information about the underlying structures but do not provide the desired information about the nerve and muscle function or disorders. In such situations, EMG is a more useful tool (1). Several researchers have used EMG to assess muscle activation patterns when evaluating gait and locomotion (2,3). Of the various techniques in practice in EMG, surface electromyography provides an estimation of muscle activity in large superficial muscle groups (4). It has the advantage of being non-invasive, although it is more difficult to attach the electrodes and they may be sensitive (5). Surface electromyography was used in a study of equine locomotion and was found to be reliable to provide more reproducible and consistent data than electrodes positioned intramuscularly (6, 7).

The recommended position of a superficial electrode is midway between the myotendinous junction and the motor point. As a second choice, it can be the middle of the muscle between the origin and insertion is selected (8).

The absence of approved protocols for SEMG in veterinary sciences adds to the dearth of scientific literature on the topic. This work was done to gain reference material about position of surface electromyogrphic electrodes for selected equine muscles that can give reliable and representative information about the activity of those muscles. Dissections and ultrasonographic imaging was performed to determine the maximum cross-sectional area of selected muscles, where the electromyographic activity will be determined.

MATERIALS AND METHODS

The muscles selected for this study were Longissimus dorsi, Biceps femoris, Gluteus medius, Semitendinosus, and Long digital extensor.

Formalized and fresh specimens from 5 hindlimbs were studied to identify the anatomical midpoints of the muscles. This part of the work was performed in the laboratory of the Institute of Anatomy of the Veterinary University, Vienna. Longissimus dorsi muscle was not dissected as the required information about the maximally active part of the muscle was already available (9).

Ultrasonograhy was done with Siemens transducer type 3C40+. It was performed on 5 clinically sound horses without any history of back or limb pain. Superficial and deep reference points were identified for individual muscles, and their depth was measured between them.

The point thus determined to be representative for individual muscle was used to record the electromyographic activity. The recordings were made with a wireless telemetric system; Telemyo Mini 16 (NORAXON).

RESULTS

The maximum cross-sectional area of the muscles around the midpoint as determined by dissection and ultrasonography was as follows:

(Table 1)

Muscle	Point of maximum cross-sectional area determined by Dissection	Point of maximum cross-sectional area determined by Ultrasonography
Longissimus dorsi	Dissection not performed	Either side of spine of T16
Biceps femoris	Between femur and poverty line, on a line joining lower end of tuber coxae and greater trochanter of the femur	Same as for dissection finings
Gluteus medius	Midway between the lumbosacral joint and greater trochanter	Same as for dissection findings
Semitendinosus	Just distal to the midpoint between the tuber ischii and caudal surface of femorotibial joint	Midway between the tuber ischii and caudal surface of femorotibial joint
Long digital extensor	Mid-way between tibial tuberosity and hock, on the craniolateral surface	Same as for dissection findings

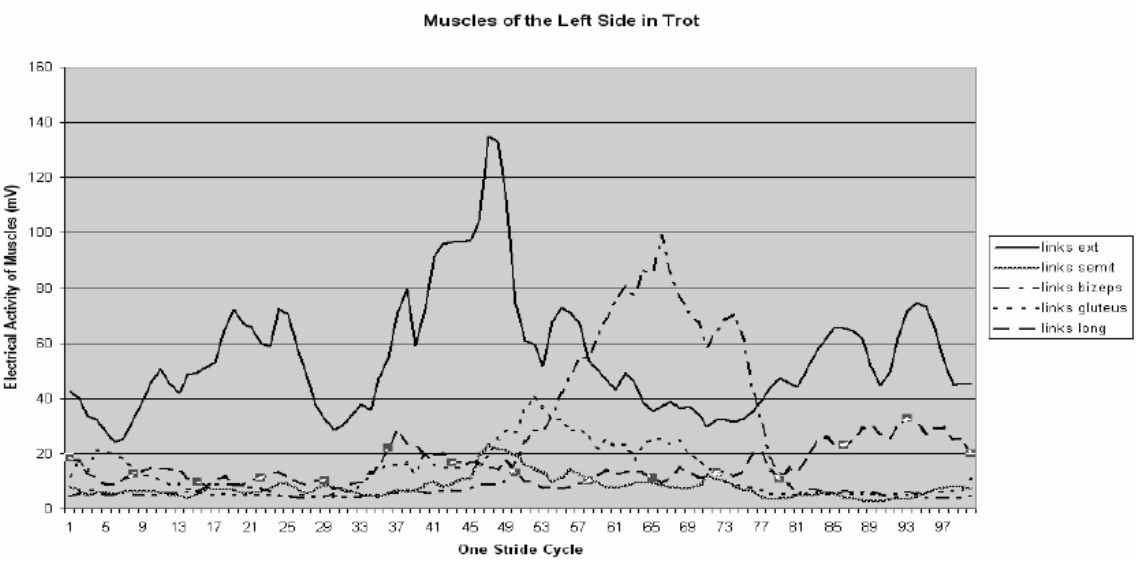

Figure 1: Electrical activity of the muscles of the left side during one stride cycle in trot

Strong and representative signals were obtained from all the muscles at the positions described above. The electrodes for semitendinosus were shifted proximally and laterally to a wider surface area to avoid cross talk without compromising the strength of the signal.

For processing of data, the marker on the Left Fore Hoof was taken as a reference for start of the stride cycle. The activity pattern of these muscles during a stride cycle can be used to understand their function. (Figure 1)

DISCUSSION

The dissection and ultrasonographic findings for Longissimus dorsi, Biceps femoris, Gluteus medius, and Long digital extensor were comparable. For semitendinosus, there was difference in observations.

Looking at the anatomy of the muscle, Semitendinosus is surrounded by soft muscle tissue all around (10) which makes it prone to lose shape and structural relations when formalized for a long time. Also, since it has little bony or compact environment, so living tissue forces have greater chances to affect the re-arrangement of muscle fibers, and thus the slight shift of the point of maximum volume during contraction under tension.

CONCLUSION

The non-invasive character of SEMG makes it more acceptable from the animal handling ethics point of view. But at the same time lack of definite protocols for its use in veterinary sciences make it less reliable as a clinical tool, which indicates the need of the standardization of its protocols. These protocols can include, though not limited to:

1.Documentation of motor points of superficial muscles

2.Use of ultrasonographic images to determine the structure of living muscle tissues

3. Use of fresh cadavers instead of formalized specimens to document the structure of dead muscle tissue

4. Development of consensus over current practices and standards of SEMG among the veterinary clinicians and researchers.

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NEUROMUSCULAR CONTROL OF HAIR COMBING IN HEALTHY SUBJECTS AND STROKE PATIENTS. AN APPROACH FOR FUNCTIONAL BIOFEEDBACK TRAINING.

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INTRODUCTION

EMG-Biofeedback is used since the late 1960s in the rehabilitation of stroke patients. The theoretical objective is that undamaged, yet subliminal pathways can be recruited and assume the function of pathways that were irreversibly damaged (1).

Although EMG studies in rehabilitation are widespread, there is a lack of data with respect to functional neuro-mu scular activation in daily living activities. Thus, no norm data are available to compare with patients data. Therefore, the aim of this study is to examine the neuromuscular control of hair combing in healthy subjects as a basis for biofeedback-training, to implement the results in an approach for functional biofeedback-training.

METHODS

The current study included 13 healthy subjects (height: 173,45±7,55 cm, weight: 64,36±12,35 kg, age: 23,72±3,52 years) without any neurological and orthopaedic disorders. Subjects were asked to carry out 3 times a hair combing movement. The hair combing movement was divided into 7 phases: 1) move the arm cranial, 2) combing movement from anterior to posterior of the middle part of the head, 3) return, 4) combing movement of the left side of the head, 5) return, 6) combing movement of the right side of the head, 7) return to starting position.

We recorded surface electromyogram (sEMG) of the m. trapecius pars ascendens (LT), m. trapecius pars descendens (UT), m. trapecius pars horizontalis (MT), m. serratus anterior (SA), m. biceps brachii (BB), m. deltoideus pars ventralis (AD), m. deltoideus pars middle (MD), and m. deltoideus pars dorsalis (PD) using NORAXON® Telemyo 2400T. Signals were A/D converted with 1500 Hz and stored in Myo Research XP software followed by ECG reduction, full wave rectification, smoothing with RMS 100 ms and amplitude normalization to the highest activity level during Maximum Voluntary Contraction (MVC) detected from 8 different MVC tests. 2D kinematics were obtained synchronously using NORAXON® software. The recordings of the arm in the frontal plane were resampled to 1500 HZ and analysed by SIMI Motion (build 275). To investigate the highest EMG activity in different phases of the hair combing cycle, a Two-Way-ANOVA (phase, muscle, interaction phase x muscle) with a Holm-Sidak correction was performed.

RESULTS

The statistical analysis revealed that there is a significant main effect for the identified phases and the recorded muscles (respectively $_{F6,672=25,0,}$ $^{p<0.001;}$ $_{F7,672=37,0,}$ $p<0.001$). However, the interaction 'phase x muscle' just did not pass the significant level of 0.05 $_{(F42,672=1,4,}$ $p=0.063$). Post hoc analysis showed that with respect to the individual MVC of each muscle, the AD, the LT, and the SA have major contributions with respect to the hair combing movement and the defined phases (see table 1, figure 1).

DISCUSSION AND CONCLUSION

The data of the current study provide insights into the neuromuscular control of hair combing. With respect to the biofeedback training, the current results present a first baseline of functional training in EMG biofeedback for a daily life activity of stroke patients. Possible training parameters like the contribution of the different muscles and the coordination of the muscles, especially AD, LT, and SD, can be dedicated for biofeedback-training. In future, the data will be extended by more (older) subjects. Moreover, the evaluation of the EMG-biofeedback training will be finished.

Tab. 1: Maximum mean sEMG activity of the recorded muscles with respect to the 7 different phases (in % MVC).

Phase	Muscle	Mean MVC	SD MVC
Phase 1	AD	21,7	14,1
Phase 2	LT	28,1	14,7
Phase 3	AD	27,8	10,3
Phase 4	SA	35,2	19,1
Phase 5	LT	21,4	11,5
Phase 6	LT	19,8	11,9
Phase 7	LT	6,7	6,9

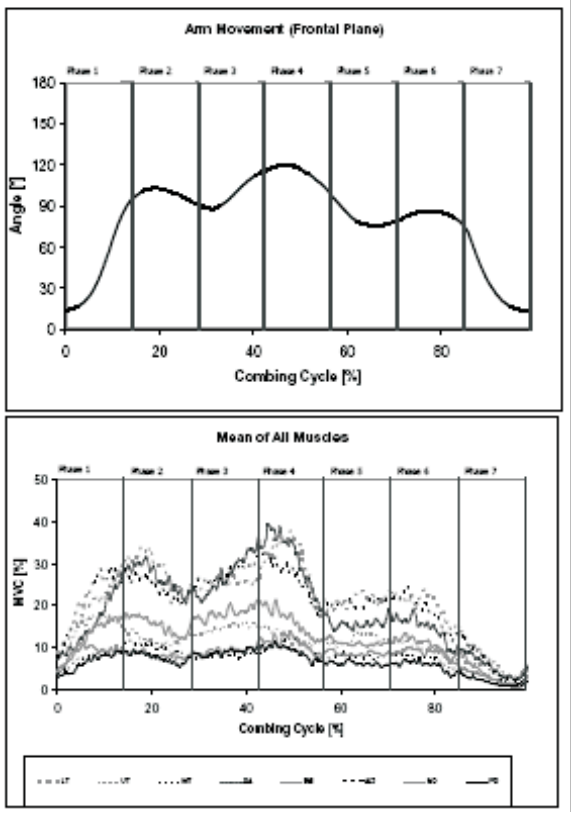

Fig 1a and b: Arm Movement in the frontal plane (mean) and mean sEMG of the recorded muscles during the hair combing movement (N=13, average out of three repetitions)

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FIDELITY OF THE SURFACE EMG SIGNALS AND ELECTRODE PLACEMENT

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INTRODUCTION

Detecting and recording differences of potentials accompanying muscular contraction (electromyography = EMG) between conductive sensors (electrodes) placed on the surface of the skin (sEMG) is extremely easy. Every body, without any anatomical knowledge and with poor equipment, can record a signal more or less contaminated by various type of noise. However, this does not mean that the recorded signal represents the targeted muscle or part of a muscle. In a clinical context, our challenge is to record sEMG with –1- the minimum crosstalk (CT) from adjacent muscles –2- recognise CT from co activation –3- whereas movement artefacts and interference from power hum depend more upon the characteristics of the hardware.

FIDELITY OF THE sEMG SIGNAL

1. Choice of electrodes

It is commonly accepted that selectivity of surface electrodes depends on _1_ size of the sensitive area _2_ inter electrode distance –3- orientation of the electrodes axis and the direction of the muscle fibres. The sensitive area is not limited to the Ag/AgCl component but include all the surface of the conductive gel.

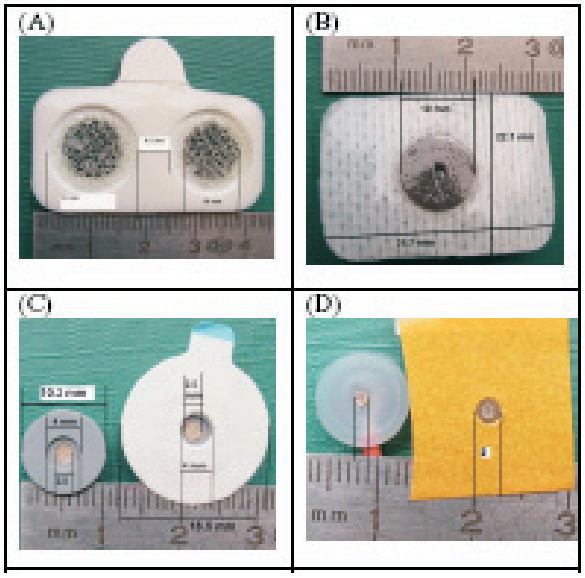

Figure 1: Selection of passive electrodes
Myotronics/Noromed: *Norotrodes 20*
BEAC Biomedical: *Easytrode*
Gereonics: *Miniature skin electrodes*
TMS International: *Miniature electrode with movement artifacts rejection*

Electrode	Ag/AgCl Diameter [mm] & area [mm²]	Overall sensitive diameter [mm] & area [mm²] of the gel	Electrodes gap [mm]
(A)	10→78.54	16 →01.06	Fixed 6.5
(B)	10→78.54	31.7 x 22.3→ 706.9	variable
(C)	2.5→4.90	4→12.56	Variable
(D)	1.5→1.76	2 →3.14	variable

Most labs use passive or active simple differential configuration of electrodes. A consensus based on EMG modelling recommended 2 cm as "best" inter electrode distance (Freriks) without considering size of the segment nor the distance between adjacent muscles. Increasing electrode distance to sense a deepest muscle enlarges the spatial summation and decreases selectivity.

Curtailment of inters electrode distance (IED) increases spatial resolution, which decreases "end of fiber effect" and crosstalk but decreases EMG amplitude. Usually cerebral palsy (diplegics, quadriplegics) exhibits low amplitude EMG. This is not a consequence of filtering by soft tissue but most likely the consequence of change in muscle fibre ratio and/or recruitment of small amplitude motor units.

Figure 2: Example of matrix of electrodes to determine the minimum cross talk area of tibialis anterior, extensor digitorum longus, peroneus longus, peroneus brevis.
Electrodes were also placed on the skin over the tibia bone. Single differential recording, Beckman miniature surface electrodes® interspaced 12mm, sensitive (gel) area 12.5 mm^2. Gain 1000. CMRR > 110dB at 50Hz

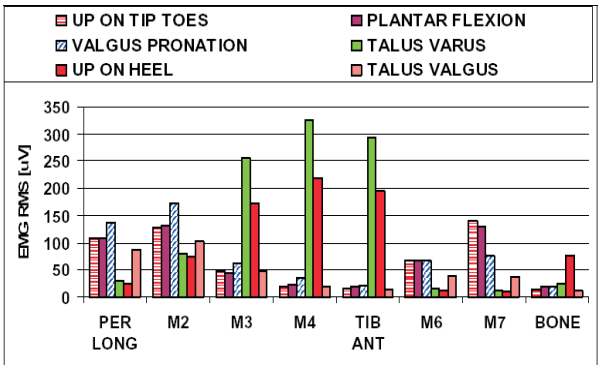

Figure 3:
Example of histogram from which was deduced the area of minimal cross talk. During selective movement of the foot and/or toes, the best site for a given muscle had the highest RMS for its analytical action and the lowest during specific movements of the other muscles. Note that electrodes over the tibia bone records pure cross talk. Other experiments with the reference electrode placed all over the medial side of the tibia, both malleola or head of fibula confirm the diffusion of EMG waves all over the segment. The skin over a bone is not electrically neutral. This observation does not support the recommendation to place the reference electrode on the skin overlying a bone. This recommendation under value the extension of the volume of conduction generated by a muscle contraction. New equipment which does not require ground electrode (ZeroWire EMG ® by AURION) solves this controversy.

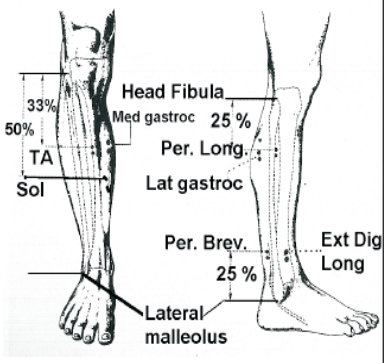

Fig 4: Example of original Basmajian et al's electrode sites except for extensor digitorum longus that was moved on the level of the peroneus brevis, half way between tibia and fibula and tangent to the lateral border of the tendon of Ext Dig Long (Blanc). Congenital club foot and cerebral palsy individuals may have shorter muscle belly and longer tendon of Ext Dig Long. In this case electrodes are moved above the level of reference.

TA = tibialis anterior
Sol = soleus
Med gastroc = Medial gastrocnemius
Lat gastroc = Lateral gastrocnemius
Per. Long. = Peroneus longus
Per. Brev. = Peroneus brevis
Ext Dig Long = Extensor digitorum longus

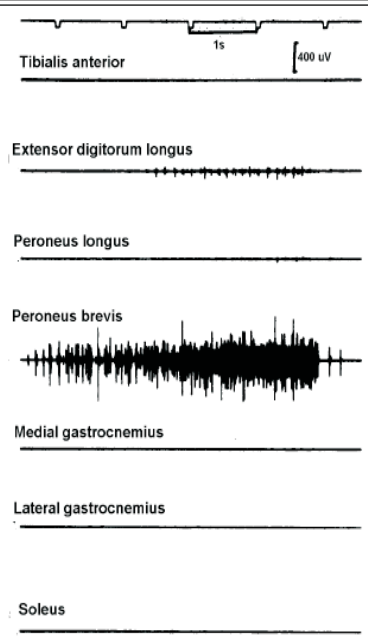

Fig 5: Example of discrimination between leg muscles. Specific movement = active valgus of the hind foot while the foot is supported in 5° of plantar flexion. Trace on extensor digitorum longus is co-contraction. This was confirmed by cross exercises

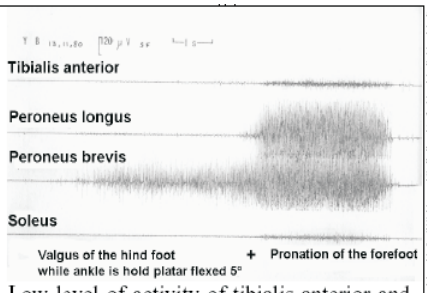

Fig 6: Example of discrimination between Peroneus longus & brevis.

Low level of activity of tibialis anterior and soleus could be CT but this test does indicate its source.

Linear electrode array localises efficiently the projection on the skin of the innervation zone (IZ). This information could improve the efficiency of injection of botulinum toxin.

2. Electrode site for minimal cross talk

In 1979, Basmajian & Blummenstein carried out extensive testing to determine the best sites for EMG biofeedback: most selected sites were expressed in percentage of the distance between bony landmarks. In 1980, Blanc repeated the experiments including children and teenagers; then, in 2004, included some of the intrinsic muscles of the foot (unpublished). Both investigators had validated all sites for electrodes parallel to muscle fibres, interspaced 15 mm max (typically 12 mm), 12.5 mm^2 area of the gel. This approach could look quite empirical but coefficients of correlation and cross correlation between different subjects' muscles, i.e. tibialis anterior, could be as high as 0.2 to 0.4 and do not help to recognize cross talk. Double differential recordings could help. The remaining challenge is to differentiate between co contraction and cross talk. Time shift between channels (bursts of raw EMG) during selective movements is the best indicator to recognize cross talk versus co contraction. Selective contractions check for the selectivity of electrodes and their appropriate placement. If motor command and/or control are deficient, reflexes or synergistic patterns are the alternative solution, therefore it relies on the investigator's skill to trigger the appropriate pattern.

Examples of recording allowed by miniature skin electrodes interspaced 12 to 15 mm

Tensor fascia latae: along a line between Anterior Iliac Spine (AIS) and greater trochanter (GT) at proximal 25% of the distance between these 2 landmarks. Selective movements: active during internal hip rotation without hip flexion. Activity during hip flexion but co contraction of rectus femo- ris. No EMG during external hip rotation. Co contraction with anterior part of gluteus medius in hip abduction. In-toeing gait without bony deformity can results from an abnormal burst of EMG at the transition between stance and swing or in early swing. This could be linked to a second burst of rectus femoris at the same period often seen in "clumsy children".

Gluteus medius: 3 functional portions: posterior, medial (above the GT) and anterior have a different timing of their peak of EMG activity during walking. Electrodes placed across the direction of the muscular fibers above GT IED > 2cm record a summation of the 3 parts. Predominance of posterior part: hip abduction starting in flexion 20°. Predominance of anterior part: hip abduction starting in hip extension. No known mean to activate selectively the medial part.

Sartorius: proximal part of a line between AIS and half width of the medial condyle, on the lateral edge of Scarpa's triangle. Selective movement: hip flexion combined with hip external rotation and knee flexion. A good position: subject seated hip flexion with the heel following the tibial crest.

Semi menbranosus and semitendinosus: upper quarter along a line from ischium to half width of the medial side of the knee (medial collateral ligament if palpable). Selective movement: starting from a knee flexion of 90°, pure internal rotation of the leg without increasing knee flexion. Co

contraction with biceps femoris in knee flexion and/or hip extension.

Biceps femoris: along a line from ischium to head of fibula, tangent to the medial side of the tendon, electrodes below the level of vastus lateralis. Selective movement: starting from 90° of knee flexion, external rotation of the leg without increasing knee flexion, which will trigger a co contraction of medial hamstrings.

Intrinsic muscles of adult foot: abductor hallucis, extensor hallucis brevis, first dorsal interosseus and flexor digitorum brevis. Depending of the foot morphology flexor hallucis brevis can be discriminated from abductor hallucis. However, selective movements are difficult to obtain. Reflex patterns are the alternative solution. Miniature electrode (like TMS International) might extend the in vestigation to children's feet.

CONCLUSION

Reliability of sEMG depends upon the spatial selectivity of the electrodes. During routine evaluation differences of timing, time shift between bursts of Raw EMG differentiate between cross talk and co activation of adjacent muscles.

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