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Fine Wire EMG of the Tibialis Posterior and Flexor Hallucis Longus Muscles in Gait and Running

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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.

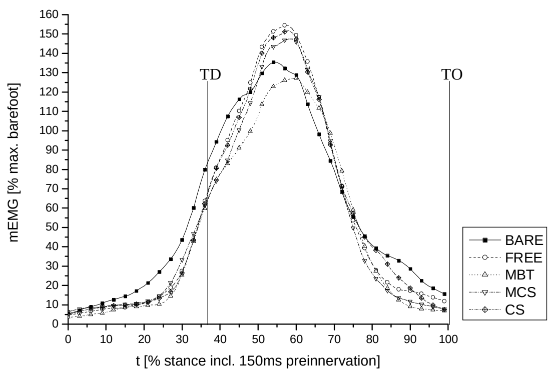

Fig. 1: Average (6 subjects; 10 trials) mEMG curves of FHL during walking running (2) with the events TD, TO and pre-innervation (150 ms) under different conditions.

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/).

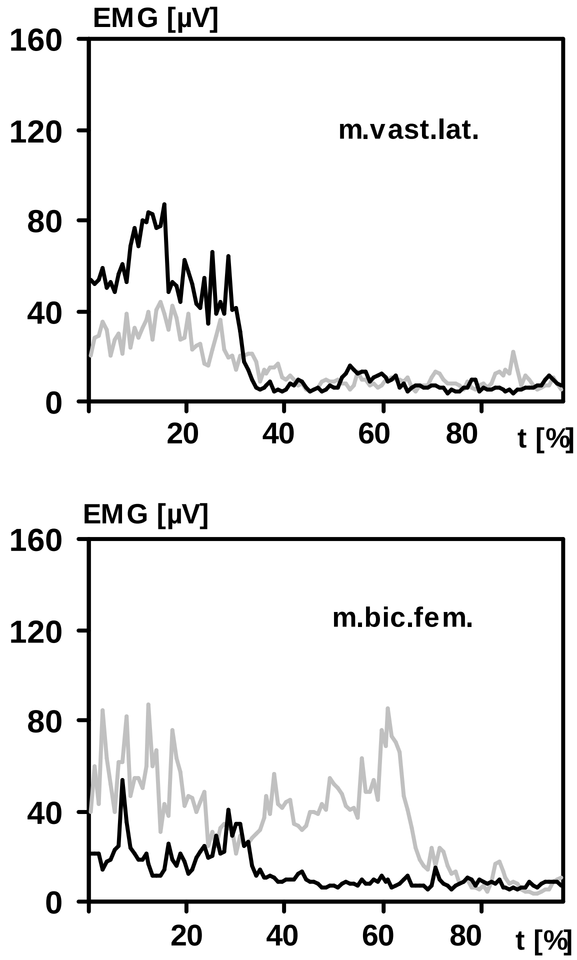

Fig. 1: Typical stance phase surface EMG of a transtibial amputee (grey: amputated limb, black: non-amputated limb)

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.

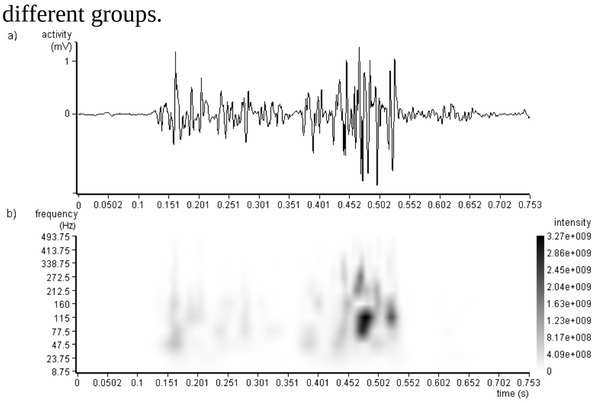

Fig. 1: a) Recorded EMG signal of m. tibialis anterior (one step of running barefoot on tartan) and b) corresponding activity pattern. In activity pattern abscissa represents time, ordinate represents frequency and grey scale represents intensity of the recorded EMG signal.

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.

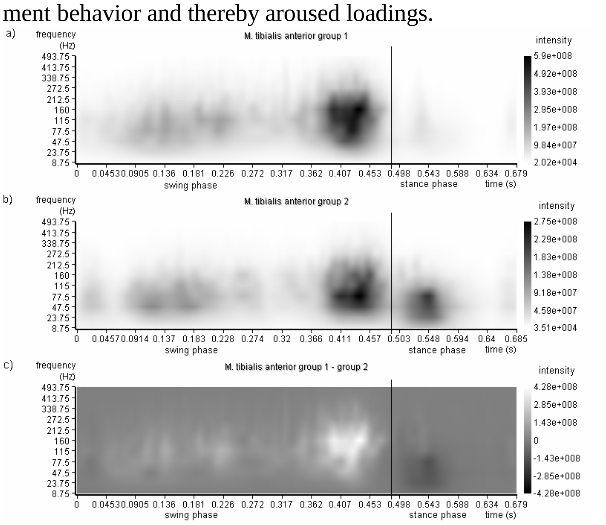

Fig. 2: Activity pattern of m. tibialis anterior while running barefoot on tartan for different groups: a) group 1 (n = 21), b) group 2 (n = 18), c) difference of activity patterns of group 1 and group 2: light zones represent higher intensity in activity pattern of group 1, dark zones represent higher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