

## Foot orthoses affect frequency components of muscle activity in the lower extremity

Anne Mündermann<sup>a,b,\*</sup>, James M. Wakeling<sup>b,c</sup>, Benno M. Nigg<sup>b</sup>,  
R. Neil Humble<sup>d</sup>, Darren J. Stefanyshyn<sup>b</sup>

<sup>a</sup>Division of Biomechanical Engineering, Department of Mechanical Engineering, Stanford University, Stanford, CA, USA

<sup>b</sup>Human Performance Laboratory, Faculty of Kinesiology, University of Calgary, Calgary, Alta., Canada

<sup>c</sup>Department of Basic Veterinary Sciences, The Royal Veterinary College, London, UK

<sup>d</sup>Division of Podiatry, Faculty of Medicine, University of Calgary, Calgary, Alta., Canada

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### Abstract

The purpose of this study was to quantify the effects of selected foot orthoses on muscle activity in the lower extremity during running. Nine male and 12 female recreational runners, clinically and functionally classified as ‘pronators’, volunteered for this study and performed over-ground running trials at 4 m/s in each of four experimental conditions: control, posting, molding, and posting & molding. Electromyographic (EMG) signals were recorded from seven lower extremity muscles. Wavelet analysis was performed to obtain EMG intensities in two frequency bands that were averaged for the pre-heel-strike and post-heel-strike intervals and for 30–100% of stance phase. Posting and custom-molding of foot orthoses increased the global EMG intensity of most muscles of the lower extremity for the stance phase of running ( $P < 0.05$ ). The increases in EMG intensity were greater in the high- than in the low-frequency bands for some lower extremity muscles ( $P < 0.05$ ). The effects on muscle activity of posting and custom-molding of foot orthoses differed between the three phases of running gait. The three tested foot orthoses did affect lower extremity muscle activity differently and these effects were specific to the phases of running gait. Combinations of increased requirements of controlling joint motion and minimizing soft tissue vibrations may have led to greater increases in shank muscle activity for the posted condition. The substantial changes in EMG due to orthotic interventions found in this study documents the importance of the study of muscle activity as a reaction to shoe inserts and foot orthoses.

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### 1. Introduction

Recreational runners use foot orthoses to prevent injuries, to rehabilitate from injuries, to increase comfort and/or to improve performance. Foot orthoses are typically classified as non-posted or posted and as non-molded or custom-molded, and medial or lateral posts are often added to custom-molded foot orthoses [1].

Foot orthoses are generally believed to align the skeleton and to reduce the loading of biological structures in the

lower extremities. However, results of most kinematic studies (e.g. [2,3]) showed that the effects of foot orthoses/inserts on foot eversion were small and non-systematic. Results of a comprehensive recent study showed that foot orthoses do have small but significant effects on maximum foot eversion, maximum foot inversion during the second half of stance phase, vertical ground reaction forces and vertical loading rate [4]. However, the functional reason for a potential improvement of the loading situation of the musculoskeletal system during running due to foot orthoses is not well understood.

Potential candidates for explaining the effects of foot orthoses are changes in the activation of muscles of the lower extremities. Initial studies have documented substantial subject specific changes in lower extremity muscle activity

\* Corresponding author. Present address: Department of Mechanical Engineering, 205 Durand Building, Stanford University, Stanford, CA 94305-4038, USA. Tel.: +1 650 724 9684; fax: +1 650 725 1587.

E-mail address: [amuender@stanford.edu](mailto:amuender@stanford.edu) (A. Mündermann).

in reaction to foot orthoses [5]. To explain these changes, it was proposed that an orthosis supporting the natural joint motion (determined by the geometry of the articular surface and the ligaments) will reduce muscle activation, and that an orthosis counteracting natural joint motion will increase muscle activation to maintain natural joint motion [6].

Changes in muscle activity may occur with respect to intensity and/or frequency. In addition to changes in intensity, which have been quantified traditionally, substantial changes in the frequency content of muscle activation have been demonstrated during prolonged running activities [7] and for running with shoes with different midsole constructions [8]. Thus, it is speculated that foot orthoses may cause a change in intensity and frequency of the EMG signals of lower extremity muscles.

The primary functions of muscle activity vary throughout the stance phase of running. It has been suggested [9], that before and immediately after heel-strike, muscles are activated to stabilize the joints and to minimize possible soft tissue vibrations. After heel strike, muscle may also be activated due to stretch-reflex-related responses [10,11]. Muscle activation during the rest of stance phase is related to muscle forces to support and to accelerate the body and muscle forces that provide stability during locomotion. It is speculated that the effects of foot orthoses on EMG signals may vary for these time intervals. However, to date no evidence has been provided to support this speculation.

The purpose of this study was to quantify the effects of custom molded and posted foot orthoses on muscle activity in the lower extremity during running. It was hypothesized that (a) posting and custom-molding of foot orthoses affect both the intensity and frequency content of the EMG signal in lower extremity muscles similarly for pronating runners and (b) the effects of posting and custom-molding of foot orthoses are specific to different intervals before and within the stance phase of running.

## 2. Methods

### 2.1. Subject population

Nine men ( $27.4 \pm 1.8$  yr;  $174.2 \pm 2.6$  cm;  $65.0 \pm 2.4$  kg) and 12 women ( $23.9 \pm 1.6$  yr;  $167.2 \pm 1.3$  cm;  $63.7 \pm 2.2$  kg) participated in this study. All subjects gave informed written consent according to the guidelines of the University of Calgary Ethics Committee prior to their participation.

All subjects were recreational runners classified as “pronators” with a weekly running distance of 15–40 km. Subjects were only included in the study when their foot eversion (determined from 2-dimensional high-speed video pictures) was greater than  $13^\circ$  during running at 4 m/s on a treadmill [12]. All subjects were clinically assessed by a podiatrist to have a normal range of motion of the joints and normal strength and flexibility of the muscles of the lower extremities [13]. Leg length discrepancy between the left

and right leg was required to be less than 0.5 cm. The detailed results of the clinical assessments have been published earlier [14].

### 2.2. Experimental conditions

Three orthotic and one control condition were used in this study. The top layer of all orthotic conditions was composed of 3 mm Spenco (Spenco Medical Corporation, Waco, TX). The bottom layer of the control condition consisted of 3 mm flat ethylene vinyl acetate (EVA; Solflex [Shore C: 50–55], Phoenix, AZ). The bottom layer of the posted condition consisted of a 6 mm full-length EVA wedge. Plaster casts of both feet in a subtalar neutral position were taken from each subject. Polypropylene shells were fabricated to positive molds obtained from the negative casts. The molded condition consisted of the polypropylene shell with no extrinsic posting, while a 6 mm extrinsic EVA post was added to the medial rearfoot and forefoot areas of the polypropylene shell to obtain the posted & molded condition. The control and all experimental conditions were similar in mass. Running sandals were used for all experiments (Model: Bryce Canyon; The Rockport Company, Canton, MA). The original inserts of the running sandals were removed and replaced by each of the four experimental conditions: (a) control, (b) posted, (c) molded and (d) posted & molded conditions. A more detailed description of the casting and fabrication techniques has been published earlier [14].

### 2.3. Testing procedure

Subjects completed two weeks of their regular running schedule in the control condition (running sandal plus control insert). Following this initial phase, each subject was tested three times per week for three weeks (nine sessions per subject). In each of the nine sessions, subjects ran 200 m on an indoor running track with each of the four insert conditions. Subjects were then set up for biomechanical testing at the Human Performance Laboratory at the University of Calgary. The four experimental conditions were tested in randomized order in each session. Before testing each of the three orthotic conditions, subjects ran 50 m in the control condition. Electromyographic and ground reaction force data were collected for 12 over-ground running trials at  $4.0 \pm 0.2$  m/s per experimental condition (heel-toe running; 48 trials per subject per session). Running trials were only accepted if the subject's speed was within  $4.0 \pm 0.2$  m/s. Subjects required 12–14 trials to obtain 12 trials within this range. This experimental design allowed for a comparison of values for each variable for the orthotic conditions to values for the control condition within each session and, thus, possible variability between sessions due to, for instance, electrode placement was eliminated. Although fatigue is unlikely to occur within a 30-min sub-maximal running protocol, this study design using

randomization eliminated bias due to potential fatigue effects. Since our previous study [15] using the same subjects did not show any trends over the nine experimental sessions that would indicate accommodation to the experimental conditions, differences between the orthotic conditions and the control condition were averaged across the nine sessions for each subject.

#### 2.4. Electromyographic data

The collection and processing methods of the electromyographic (EMG) signals used in this study have previously been described in detail [15]. Briefly, EMG signals were recorded from seven lower extremity muscles of the right leg. Bipolar surface electrodes (Ag–AgCl) were placed on the vastus lateralis and medialis, rectus femoris, biceps femoris (long head), tibialis anterior, peroneus longus and gastrocnemius medialis muscles after removing the hair and cleaning using isopropyl wipes, and then secured using Cover-Roll stretch tape (Beiersdorf AG, Hamburg, Germany). Each electrode was 10 mm in diameter with an intra-electrode distance of 22 mm. A ground electrode was placed on the tibial tuberosity. The EMG signals were pre-amplified at source and recorded using a BioVision system (BioVision, Wehrheim, Germany) at 2400 Hz. Timing of heel-strike and toe-off for one step per trial was taken from the ground reaction force data. Using these two events, the EMG data could be related to different phases of ground contact during running. EMG data for each trial was checked for crosstalk by cross-correlating the raw EMG signals between muscles. The correlation coefficients for all muscle combinations of accepted trials were smaller than 0.500.

Wavelet analysis was used to resolve the EMG signals simultaneously into their intensity in time and frequency space [16]. The intensity obtained using this wavelet analysis represents a close approximation of the power of the EMG signal. A filterbank of 11 wavelets was used [16] and a wavelet domain was defined as the EMG intensity over time corresponding to each of the 11 wavelets. Based on results of a previous study [7], a low-frequency band was defined as frequencies between 25 and 82 Hz and a high-frequency

band as frequencies between 142 and 300 Hz. The choice of wavelet domains 2 and 3 as a representation of the low-frequency band and wavelet domains 6–8 as a representation of the high-frequency allowed for a clear distinction between the low- and high-frequency bands. The global EMG intensity was defined as the sum of EMG intensities for wavelet domains 1–8.

The intensities at the global, low- and high-frequency bands ( $I$ ,  $I_{low}$  and  $I_{high}$ , respectively) were normalized for each subject and session so that the maximum of the total intensity for the control condition had a value of one. The global, low- and high-intensities were averaged over the pre-heel-strike interval (50 ms before heel-strike), the post-heel-strike (50 ms after heel-strike) and phase 1 (30–100% of stance phase) resulting in nine EMG variables per muscle (Table 1). Due to technical problems with the ground reaction force measurements, subject 1 was eliminated from this study. Therefore, the results and discussion sections are based on data for 20 subjects.

#### 2.5. Statistical analysis

Significant differences between the three orthotic conditions for all EMG intensity variables were determined using single-factor repeated measures analyses of variance (ANOVAs). The dependent variables used for these tests were  $I$ ,  $I_{low}$  and  $I_{high}$ . Experimental conditions were used as factor in the ANOVA. Each ANOVA contained values from all three orthotic conditions and all 20 subjects. Significant differences in correlation coefficients  $R$  describing correlations of EMG intensities for frequency bands and muscles between the pre-heel-strike and post-heel-strike intervals and phase 1 and orthotic conditions were detected using repeated measures ANOVAs. The dependent variables for these tests were  $R_{pre}$ ,  $R_{post}$  and  $R_{phase3}$  for each muscle. Orthotic conditions were used as factor in the ANOVA. Each ANOVA contained values from the three orthotic conditions and 20 subjects. Significance was tested at a confidence level of  $\alpha = 0.05$ . Repeated measures Student's  $t$ -tests were performed where a significant factor effect was indicated by the ANOVA.

Table 1  
Definition of the EMG variables included in the analysis

Symbol	Definition
$I_{m,pre}$	Global EMG intensity of each of the seven muscles during the pre-heel-strike interval <sup>a</sup>
$I_{m,high,pre}$	EMG intensity in the high-frequency band of each of the seven muscles during the pre-heel-strike interval <sup>a</sup>
$I_{m,low,pre}$	EMG intensity in the low-frequency band of each of the seven muscles during the pre-heel-strike interval <sup>a</sup>
$I_{m,post}$	Global EMG intensity of each of the seven muscles during the post-heel-strike interval <sup>a</sup>
$I_{m,high,post}$	EMG intensity in the high-frequency band of each of the seven muscles during the post-heel-strike interval <sup>a</sup>
$I_{m,low,post}$	EMG intensity in the low-frequency band of each of the seven muscles during the post-heel-strike interval <sup>a</sup>
$I_{m,phase1}$	Global EMG intensity of each of the seven muscles during 30–100% of stance phase <sup>a</sup>
$I_{m,high,phase1}$	EMG intensity in the high-frequency band of each of the seven muscles during 30–100% of stance phase <sup>a</sup>
$I_{m,low,phase1}$	EMG intensity in the low-frequency band of each of the seven muscles during 30–100% of stance phase <sup>a</sup>

EMG intensity ( $I$ ) was calculated for seven lower extremity muscles, three frequency bands and three time intervals.

<sup>a</sup> The seven lower extremity muscles tested were: ta, tibialis anterior muscle; pl, peroneus longus muscle; gm, gastrocnemius longus muscle; bf, biceps femoris muscle; vl, vastus lateralis muscle; rf, rectus femoris muscle; vm, vastus medialis muscle.

Post-hoc power analyses showed that the statistical power of all ANOVA tests was greater than 90%.

### 3. Results

The inter-subject variability (S.E.M.) in EMG intensity was small (between 2.9 and 10.0%) for both the high- and low-frequency bands (Fig. 1). The ratio of the EMG intensity for the high- and low-frequency band,  $I_{\text{high}}/I_{\text{low}}$ , was between 0.5 and 1.3 for the shank muscles and approximately 0.2 for the thigh muscles. The intensity of the high-frequency EMG

signals was, in comparison to the low-frequency signals, strong for the shank muscles and weak for the thigh muscles.

The tested orthotic interventions produced in general an increase in global EMG intensity in comparison to the control condition (Fig. 2; Table 2). The relative increases were highest for the tibialis anterior, the peroneus longus and the biceps femoris muscles. A significant orthotic effect was observed in 10 of the 21 muscle-time combinations (Table 2). There was a general trend that relative changes in EMG intensity due to orthotic interventions were greater in the high- than in the low-frequency band (Fig. 2). The increases in EMG activity between high- and low-frequency

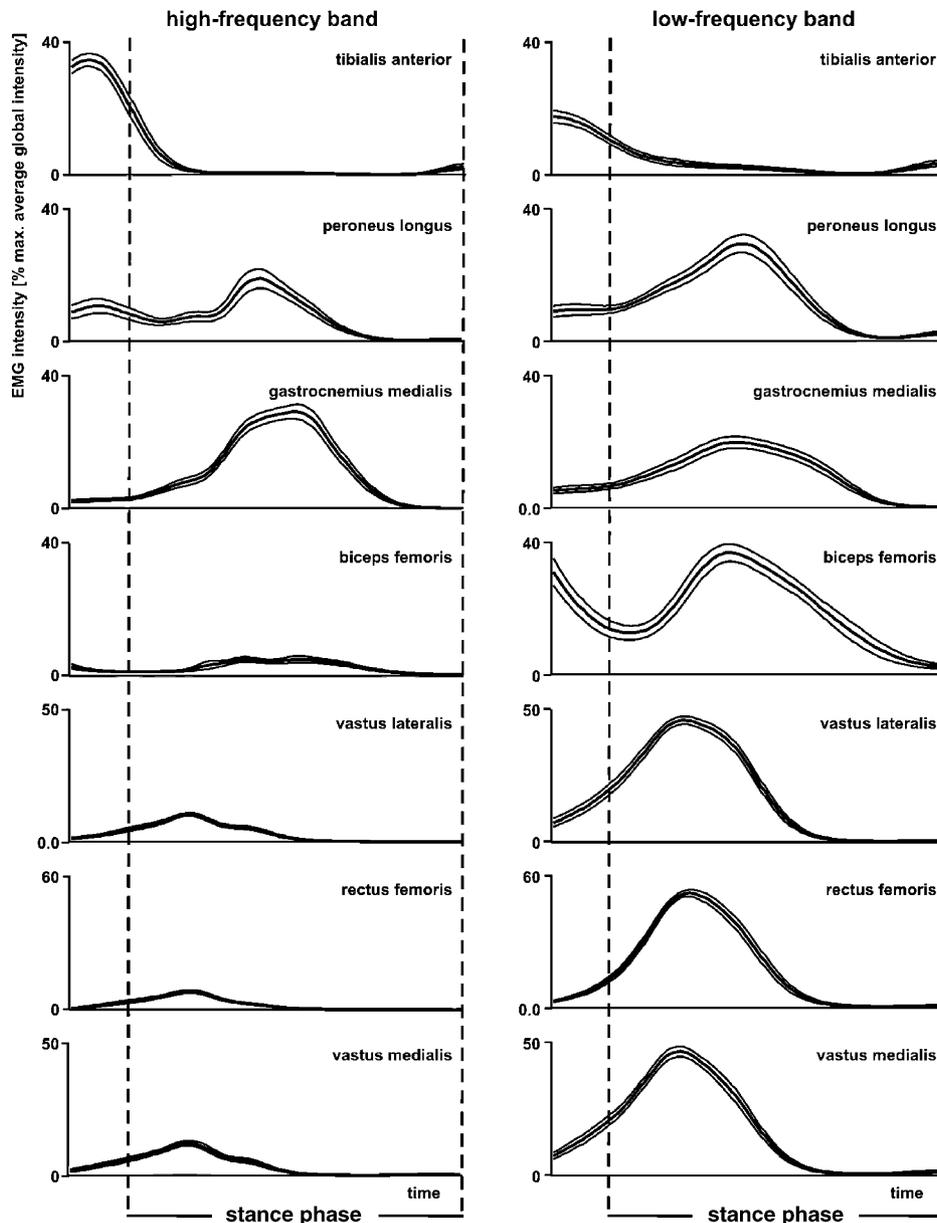


Fig. 1. Average curves (SEM) of EMG intensity for high- and low-frequency bands for seven lower extremity muscles. Data are shown for all subjects for running in the control condition ( $n = 20$ ). The time of the EMG intensity traces were normalized to stance phase. Note that EMG intensity before heel-strike is shown. The magnitude of the EMG intensity traces for the high- and low-frequency bands of each session were normalized to the maximum of the average global EMG intensity trace for the control condition.

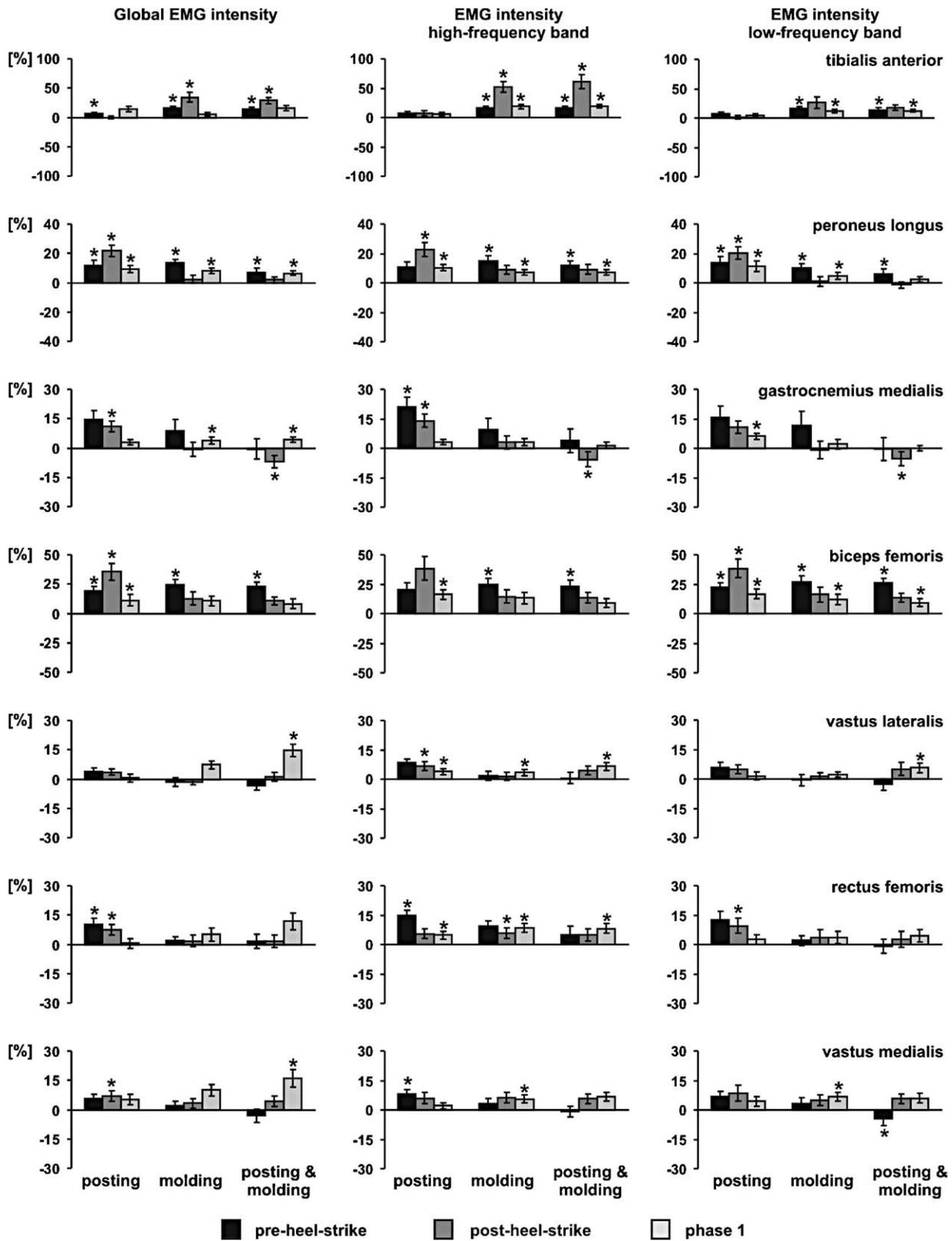


Fig. 2. Average difference (SEM) in EMG intensity of seven lower extremity muscles between orthotic conditions and the control condition. \*Significant difference compared to control ( $P < 0.05$ );  $n = 20$  subjects.

Table 2  
*P*-values values of differences in muscle response between orthotic conditions and between frequency bands (compared to control)

	Pre-heel-strike	Post-heel-strike	Phase 1
<b>Tibialis anterior</b>			
Condition	0.004*	<0.001*	0.002*
Frequency	0.662	0.001*	0.023*
Condition × frequency	0.696	0.007*	0.261
<b>Peroneus longus</b>			
Condition	0.401	0.001*	0.023*
Frequency	0.443	0.016*	0.276
Condition × frequency	0.294	0.615	0.430
<b>Gastrocnemius medialis</b>			
Condition	0.064	0.003*	0.073
Frequency	0.348	0.397	0.863
Condition × frequency	0.064	0.544	0.306
<b>Biceps femoris</b>			
Condition	0.567	0.001*	0.091
Frequency	0.579	0.887	0.983
Condition × frequency	0.976	0.932	0.904
<b>Vastus lateralis</b>			
Condition	0.010*	0.065	0.186
Frequency	0.214	0.865	0.401
Condition × frequency	0.973	0.813	0.870
<b>Rectus femoris</b>			
Condition	0.041*	0.660	0.606
Frequency	0.077	0.976	0.056
Condition × frequency	0.363	0.309	0.637
<b>Vastus medialis</b>			
Condition	0.006*	0.834	0.339
Frequency	0.440	0.853	0.612
Condition × frequency	0.327	0.638	0.434

Values (ANOVA results,  $n = 20$  for each test) are shown for the probability that no interaction occurred between orthotic conditions and frequency bands and EMG intensity (compared to the control condition) for pre-heel-strike and post-heel-strike intervals and for phase 1. For all muscles and time intervals, the degrees of freedom (df) for the sources of variance were  $df_{\text{condition}} = 2$ ,  $df_{\text{frequency}} = 1$ , and  $df_{\text{condition} \times \text{frequency}} = 2$ .

\* Significant interaction,  $P < 0.05$ .

bands were statistically significant for the tibialis anterior and peroneus longus muscles for the post-heel-strike interval and for phase 1 (Table 2;  $P < 0.05$ ).

*Posting* produced the highest increases in global EMG intensity pre- and post-heel-strike for the peroneus longus and biceps femoris muscles. The increases due to posting were smallest during phase 1 for all muscles except the tibialis anterior muscle. Posting produced only small changes in EMG activities for the tibialis anterior muscle. Posting produced substantial and often significant increases in high (five out of nine)- and low-frequency (seven out of nine) EMG intensities for the peroneus longus, the gastrocnemius medialis and the biceps femoris muscles. The increases in peroneus longus activity with the posted condition were observed for all 20 subjects (Electronic Addendum 1).

*Molding* produced the highest increases in global EMG intensity pre- or post-heel-strike for the peroneus longus, gastrocnemius medialis and the biceps femoris muscles. The

increases were substantial and often significant in high (six out of nine)- and low-frequency (six out of nine) EMG intensities for the tibialis anterior, the peroneus longus and the biceps femoris muscles. The increases were small during phase 1 for all shank muscles and for the biceps femoris muscle. The changes in global EMG intensity were highest during phase 1 for the other three thigh muscles. The increases in tibialis anterior activity with the molded condition were observed for all 20 subjects (Electronic Addendum 1).

*Posting & molding* produced the smallest increases in EMG intensity for the peroneus longus, gastrocnemius medialis and biceps femoris muscles. The increases were still substantial and also significant in high (six out of nine)- and low-frequency (six out of nine) EMG intensities for the tibialis anterior, the peroneus longus and the biceps femoris muscles. The changes in global EMG intensity were highest during phase 1 for the other three thigh muscles. The increases in tibialis anterior activity with the posted & molded condition were observed for all 20 subjects (Electronic Addendum 1).

Average correlation coefficients between changes in EMG intensity of the high- and low-frequency bands due to orthoses for 63 muscle-orthotic-time combinations are shown in Electronic Addendum 2. The average correlation coefficients for all orthotic conditions were similar. For all muscles, the average correlation coefficients were highest for phase 1. For the shank muscles, the sequence of the average correlation coefficients was always  $R_{\text{pre}} < R_{\text{post}} < R_1$ . For the thigh muscles, the sequence of the average correlation coefficients was always  $R_{\text{post}} < R_{\text{pre}} < R_1$ . There was a significant time effect for all muscle-orthotic combinations ( $P < 0.05$ ).

#### 4. Discussion

The results of our study showed systematic changes in muscle activity, which were specific to the orthotic conditions and to the selected muscles (Electronic Addendum 1). Thus, different structural components of foot orthoses do affect muscle activity in the lower extremities in a systematic way during running. The systematic and significant changes of EMG activity are assumed to be the result of the selection of the subjects and the specific definition of the variables. The subjects included in this study were a well-defined group of pronators with no history of lower extremity pain or injuries while the subjects in a previous study [5] were not well defined. It is not surprising that a homogeneous group shows a more homogeneous reaction than a heterogeneous group of subjects. The variables investigated in a previous study [5] showing subject specific results were defined as absolute variables and were affected by changes in boundary conditions such as daily fluctuations due to fatigue and/or positioning of the EMG electrodes. The variables in our study were defined

with respect to a control situation. Thus, daily changes in muscle activity due to fatigue and/or electrode placement were minimized since changes were quantified for fixed boundary conditions.

EMG intensities of the peroneus longus and gastrocnemius medialis muscles for the posted condition were significantly greater than for the molded and posted & molded conditions during the post-heel-strike interval and phase 1 (Fig. 2). Results of a previous study [4] using the same subjects and orthotic conditions showed that maximum foot eversion was significantly reduced with the posted condition and was not affected by the molded and posted & molded conditions. Increased EMG intensity in the peroneus longus and gastrocnemius medialis muscles suggest that the body attempted to stabilize the subtalar joint when running with the posted condition, a condition that the body may perceive as unstable.

The speculation that muscle activity of some muscles will be changed using foot orthoses was based on the concept of the *preferred movement path* [6,17,18]. It was speculated that if an orthotic intervention supports the preferred movement path of a joint, muscle activation will be reduced. On the other hand, if an orthotic intervention counteracts the preferred movement path, muscle activation will be increased to maintain the preferred movement path. In the current study, EMG intensity was smallest for the control condition. Thus, according to the proposed concept, the control condition would be considered the optimal condition. The results of our study may have been influenced by the fact that subjects were exposed to the orthotic conditions only during the experimental sessions and that the orthoses may have acted as disturbances to the musculoskeletal system. Future studies should investigate the effects of foot orthoses on EMG intensity following longer wear-periods but in a similarly controlled setting. To date, no conclusive statement can be made whether this concept is in fact correct, as an objective measure for the 'optimal' condition does not exist.

The results of this study showed that the increases in EMG intensity due to foot orthoses were greater in the high-frequency band than in the low-frequency band (Fig. 2). Furthermore, the correlation between increases in EMG intensity in the high- and low-frequency bands was small for most orthotic-time combinations (Electronic Addendum 2), which is in agreement with the observation that peaks in EMG intensity occur often independently in the high- and low-frequency bands [19]. There is strong evidence that the frequency content of EMG signals is related to the relative activation of fast and slow twitch fibers [7,19–23]. Thus, orthotic interventions tested in this study did affect the recruitment of motor units. Increased activity of less fatigue-resistant fast muscle fibers may lead to an earlier on-set of fatigue [20]. It has been suggested that fatigue of the peroneus longus muscle is the dominant cause of lack of foot stability [24] and that muscular fatigue in the tibialis anterior muscle is the origin of running injuries such as

tibial stress fractures [25]. In addition, increases in muscle activity will result in greater metabolic expenditure [26]. Thus, differences in muscle activity as observed in this study may result in earlier on-set of fatigue and/or lower performance.

Increases in EMG intensities immediately after heel-strike may have been influenced by muscle activation required for muscle tuning [9]. It has been reported previously [4] for the same subject population and orthotic conditions that the vertical ground reaction forces and loading rates were higher for the posted than for the molded and posted & molded conditions. Thus, the increased EMG activity of the shank muscles in the posted condition may have been required to dampen soft tissue vibrations as the greater loading rate may have moved the frequency of the ground reaction force closer to the natural frequency of the muscle packages of the lower extremity [27]. However, in the current study we did not quantify the characteristics of soft tissue vibrations nor did we determine the natural frequency of the muscle packages.

The correlations between changes in EMG intensity of the high- and low-frequency bands varied between muscles and time intervals. The shank muscles (tibialis anterior, peroneus longus and gastrocnemius medialis) showed low-correlations between the high- and the low-frequency bands during the pre-heel-strike interval (Electronic Addendum 2;  $R < 0.500$ ). The thigh muscles showed higher correlations between changes in EMG intensity of the high- and low-frequency bands. A possible explanation for this result may be found in the shock wave travelling through the body after heel-strike. The magnitude of this shock decreases as it travels proximally [28]. Thus, the more proximally a muscle is located the less muscle activity is required to dampen possible soft tissue vibrations. Results of a recent study suggested that a burst of fast-twitch muscle activity occurring immediately before impact is predominantly responsible for the dampening of soft tissue vibrations [29]. If this was the case one would expect changes in EMG intensity of the high-frequency band to occur independently of changes in EMG intensity of the low-frequency band and, thus, the correlation between both frequency bands to be low as it was observed in this study. Most muscles showed high-correlations ( $R > 0.500$ ) between the two frequency bands for phase 1 (Electronic Addendum 2), which coincides with the time of largest force requirements for take-off.

In conclusion, the results of this study showed that structural components of foot orthoses affect lower extremity muscle activity. Changes in EMG intensity were consistent and significant across subjects for some muscle-orthoses combinations and subject specific and not significant for others. Changes in EMG intensity as a response to foot orthoses were greater in the high- than in the low-frequency bands. Reported kinematic changes due to foot orthoses are typically small. Thus, one should conclude that one of the main effects produced by foot orthoses is a change in the activation of muscles in the lower extremities.

These changes may occur in the intensity, time or frequency of the EMG signal. Analysis and evaluation of shoe inserts and foot orthoses should, therefore, always include a frequency-time related EMG analysis.

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