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Changing the texture of footwear can alter gait patterns

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Abstract

The foot provides an important source of afferent feedback for balance and locomotion. Sensory feedback from the feet can be altered by standing or walking on different surfaces. The purpose was to determine the effects of textured footwear on lower extremity muscle activity, limb kinematics, and joint kinetics while walking. Three-dimensional kinematics and kinetics, as well as muscle EMG, were collected as subjects walked with a smooth and textured shoe insert. Muscle activity was analyzed using a wavelet technique. The textured shoe insert caused a significant reduction in both soleus and tibialis anterior intensity during periods when these muscles are most active. Furthermore, the changes in muscle activity were only seen in the low frequency content of the EMG signal. The foot was significantly more plantar flexed at heel strike with the textured inserts. Small changes were also seen in vertical ground reaction forces and joint moments. It was assumed that the changes in gait patterns were due to a change in sensory feedback caused by the textured shoe insert. The possibilities of altered sensory feedback with footwear are discussed. Sensory feedback from the feet may affect specific motor unit pools during different activities. Changing the texture, without changing the geometry, of a shoe insert can alter muscle activity during walking. This may be useful in the prescription of footwear interventions and suggests that footwear may have sensory as well as mechanical effects.

Keywords: Afferent; Sensory feedback; Biomechanics; Shoe inserts; Muscle; EMG; Wavelet; Kinematics; Footwear

1. Introduction

The human foot is the first point of contact between the body and the external environment, and is ideally positioned to provide sensory information to the central nervous system during static and dynamic tasks. An important source of sensory feedback comes from specialized mechanoreceptors found within both the hairy and glabrous skin of the foot. Afferent feedback from these receptors has been studied in both animal and human models, and in different experimental settings.

Afferent feedback from the feet is important for balance and locomotion. During static postural tasks, cutaneous feedback originating from specialized mechanoreceptors in the foot is thought to have a strong influence on balance stability [16,20] and postural correction strategies [9,17]. While walking and running, both noxious [4] and non-noxious [11] stimulation of cutaneous nerves that innervate the foot can affect α-motoneuron activity in the muscles of the legs, most likely via Aβ reflex pathways [38]. Changes in lower extremity kinematics have also been reported following cutaneous stimulation during the gait cycle [11,41].

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During gait, reflexes are dependent on the task, muscle, phase of the step cycle, and location and intensity of the stimulus [12,28,37,40]. However, electrical stimulation of cutaneous nerves and the study of muscle reflexes does not provide information about the effects of long term changes in cutaneous sensory feedback.

Sensory feedback from the feet may be influenced by changing the characteristics of a shoe sole or surface. Watanabe and Okubo [36] provided evidence that standing on different surfaces can alter the transmission of afferent signals from the plantar surface of the foot. Increased tibial nerve activity was seen when standing on surfaces that were textured with varying densities of semi-circular shot pellets. Wu and Chiang [39] showed differences in the latency of muscle reflexes after perturbations when standing on soft surfaces. The authors implied that sensory feedback from the feet was altered when standing on the different surfaces. Maki and coworkers [21] showed a qualitative improvement in balance recovery in young and elderly populations when sensory feedback was thought to be enhanced with special tubing attached to the subject's feet, although no direct evidence of increased sensory feedback was provided. These studies examined the effects of a prolonged change in sensory feedback during postural tasks. However, these effects have not been examined during cyclical movements, such as walking or running. Specifically, the effects of footwear induced changes in sensory feedback on human gait patterns have not been studied in a normal population.

It has been recently speculated that the ability of shoe inserts or orthoses to alter joint kinematics or kinetics may be influenced by sensory feedback from the feet [23,27]. Subject specific differences in sagittal plane kinematics and joint kinetics have been attributed in part to sensory mechanisms. These speculations have not been justified. By altering on the only the texture, but not the shape of a shoe insert, the effects of altered sensory input can isolated and examined without the confounding effects of mechanical shape changes.

Therefore, the purpose of this study was to determine the effects of textured footwear on lower extremity muscle activity, limb kinematics, and joint kinetics while walking. It was hypothesized that the use of textured inserts would result in: (a) significant differences in muscle activity at different phases of the step cycle; (b) significant differences in sagittal plane limb kinematics; and (c) no differences in joint kinetics or impact forces.

2. Methods

Fifteen subjects, 12 males and three females, volunteered to participate in this study (mean ± SD age: 24.7 ± 2.9 years; height: 177 ± 9 cm; weight: 74 ± 12 kg). All subjects reported that they were free of any known neurological dysfunction or physical impairment that might affect their performance in this test. Ethics approval was obtained from the university's Office of Biomedical Ethics. All subjects were properly informed about the nature of this study and signed written consent forms prior to their participation.

2.1. Experimental protocol

Subjects were required to walk at a speed of 1.5 m s^{-1} along a 30-m indoor pathway in two shoe insert conditions. The speed of walking was controlled with infrared timing lights placed 1.90 m apart. The control insert was made from 3 mm thick EVA foam (shore C 60) that was cut to standard foot insert sizes. The textured insert was a 3 mm thick EVA foam insert that was cut from a commercially available sandal. It was textured with semi-circular mounds with center distances of 8 mm (Fig. 1). The textured insert was worn with the textured side up, in contact with the plantar surface of the feet. It was easily detected by all subjects and strong enough to elicit a distinct preference or dislike for the insert. Shoes were not used. The heel area of the shoe inserts was sprayed with sticky medical adhesive (Hollister Inc. Libertyville, IL) to help hold the insert to the subject's bare feet. A nylon stocking was then placed around the foot to hold the insert in place. Thin elastic wrap (Underwrap, Cramer Products Inc., Gardner, KS) was

Fig. 1. Pictures of the smooth (left) and textured (right) inserts used in this study.

then wrapped around the foot to reduce movement between the foot and the insert, and to increase friction between the nylon and the ground. The increased friction prevented the foot from slipping on the floor surface, especially during the push-off phase in late stance. All subjects reported that they were comfortable and able to produce natural gait patterns with the nylon stockings and elastic wrap. Interventions were applied bilaterally to both feet.

2.2. Kinematics and kinetics

Three spherical reflective markers were attached to each of the right limb segments of interest, the rear-foot, shank, and thigh (see Fig. 2), using medical adhesive spray (Hollister Inc. Libertyville, IL). Reflective markers were also used to define joint centers and were defined as follows: Hip joint – the anterior-superior iliac spine (proximal-distal (p-d) and medio-lateral (m-l) planes) and the greater trochanter of the femur (anterior-posterior (a-p) plane); knee joint – the lateral epicondyle of the femur (p-d and a-p planes), and the center of the patella (m-l plane); ankle joint – the midpoint between the medial and lateral malleoli. Static standing trials were collected to relate the position of each limb's surface markers with the corresponding joint centers. Segment embedded coordinate systems were defined using the methods described by Cole and coworkers [7].

Three-dimensional lower extremity kinematics were collected using seven high-speed infrared video cameras (Motion Analysis Corp, Santa Rosa, USA) collecting at 120 Hz. Ground reaction forces were collected at 2400 Hz using a force plate (Kistler Instrumente AG, Winterthur, Switzerland) embedded in the walkway flush with the ground. Kinematic and kinetic data were filtered using a recursive, fourth-order low-pass Butterworth filter with cut-off frequencies of 6 and 100 Hz, respectively. Analysis of the movement data were performed using KinTrak software (University of Calgary, Canada). Kinematic variables of interest included sagittal knee and ankle angles, and the relative three-dimensional motion between the shank and rearfoot. Kinetic data analyses focused on three-dimensional knee and ankle joint moments, and ground reaction forces. Kinematic and kinetic data for each subject were reported as the mean of 10 footfalls for each condition. A schematic of the experimental set-up is shown in Fig. 2.

2.3. EMG

The skin overlying the soleus (SOL), medial gastrocnemius (MG), tibialis anterior (TA), vastus medialis (VM), rectus femoris (RF), and biceps femoris (BF) muscles of the right leg was shaven and then cleaned with isopropyl wipes. Bi-polar surface electrodes (Ag/AgCl) that were 10 mm in diameter, with inter-electrode spacing of 22 mm were used for data collection. The electrodes were secured to the skin and checked to ensure that skin impedance (Grass Electrode Impedance Meter, Warwick, RI) did not exceed 5 kΩ. The electrodes were then wrapped with Cover-Roll stretch tape (Beiersdorf AG, Hamburg, Germany). EMG signals were collected using a Biovision system (Biovision, Wehrheim, Germany) sampling at a frequency of 2400 Hz. The data were pre-amplified, band-pass filtered (10 Hz–1 kHz) and stored on a computer for further analysis.

EMG data were analyzed using a wavelet technique, which allows the signal to be simultaneously resolved in both time and frequency space [30,34]. This method of analysis is analogous to filtering the EMG signals with a series of band-pass filters. A set of 11 wavelets was used to filter the data with center frequencies that spanned a range from 6.90 to 395.49 Hz, and matched those described by von Tscharner [30]. The center frequencies, $f_c(k)$ for the set of wavelets was defined by the function:

$$f_c(k) = \frac{(k + c_1)^{c_2}}{c_3},$$

where c_1, c_2, and c_3 are scaling factors and k is the wavelet number. The scaling factors used for the analysis

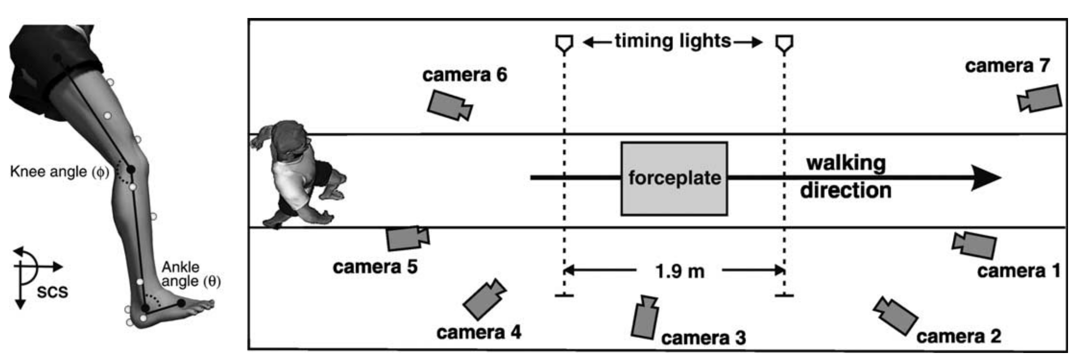

Fig. 2. Schematic diagram of the marker placements on the foot, shank, and thigh for the collection of three-dimensional kinematics. Also included is the segment coordinate system (SCS) reference frame used for inverse dynamics calculations. On the left, a diagram showing the experimental set-up, and high-speed video camera placement during data collection.

were equal to 1.45, 1.959, and 0.3 for c_1, c_2, and c_3, respectively [34].

The intensity of the wavelet transformed signals was calculated as a function of both the square of the amplitude and time-derivative (slope) of the wavelet-transformed signal. A Gaussian filter was then applied to reduce high frequency artifacts that result from finite sampling frequencies. The total signal intensity was defined as the sum of all intensities from all wavelet bands, and is equal to twice the square of the RMS value for the signal when similar time periods are chosen. The wavelet technique has the advantage that it also allows the possibility of examining the signal in different frequency (wavelet) domains. Based on the results of previous studies [32], low and high frequency components of the EMG signal were defined. The low frequency component (I_{low}) was calculated as the sum of the intensities of wavelets two and three [30] with center frequencies of 37.71 and 62.09 Hz. The high frequency component (I_{high}) was defined as the sum of wavelets six to eight, with center frequencies ranging from 170.39 to 271.48 Hz.

The integrated intensity plot for any given wavelet represents the energy of the signal in a given time and frequency domain. For the purpose of this paper, references about the total energy of a given muscle refer to the total energy of the EMG signal of that muscle over a given time frame. The total energy, as well as the low and high frequency components of the EMG signal for each muscle, were calculated for the entire stance phase, defined as the period from heel strike to toe-off as determined with the force plate. The data from each muscle were normalized to each subject's mean energy for the flat insert condition. The stance phase was also broken down in to discrete time intervals and the energy of the wavelet transformed EMG signals was calculated for the first 20% of stance, from 20% to 70% of the stance phase, and the final 30% of stance. These time periods approximate the intervals from heel strike to foot flat, mid-stance, and the propulsion phase from heel lift to toe off, respectively. Mean data were averaged from 10 footfalls for each of the interventions.

2.4. Statistics

SPSS (Chicago, IL) statistical software was used for statistical analysis. Muscle data were analyzed using Wilcoxen signed rank tests. Kinematic and kinetic data for the control and textured condition were analyzed using paired sample *t*-tests. Significance was set a priori at $\alpha = 0.05$. Due to the limited number of females included in the study, no specific gender differences were analyzed.

3. Results

3.1. Muscle activity

The textured insert caused a significant reduction in total SOL energy for the entire stance phase, as calculated from heel strike to take-off. When analyzing the

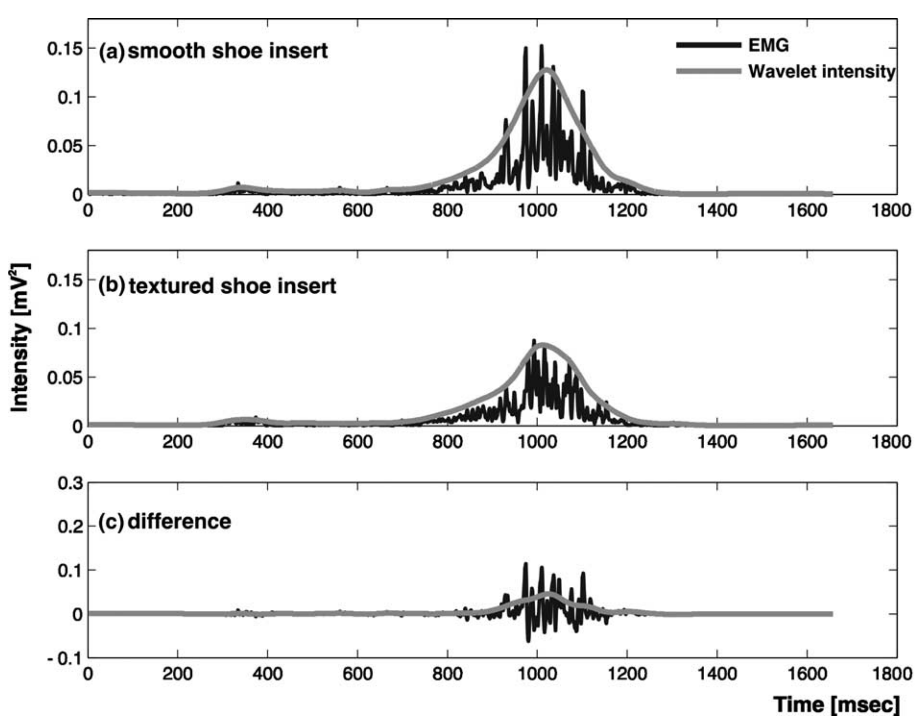

Fig. 3. Mean soleus EMG for one representative subject while walking with (a) smooth and (b) textured shoe inserts. EMG data were rectified and squared. The wavelet total intensities are superimposed on the EMG data. Soleus activity was significantly lower while wearing the textured shoe insert.

different periods of ground contact, no difference was found in SOL energy during the first 20% of stance. A general, but non-significant, 6% decrease in SOL energy was found during mid-stance ($P = 0.06$), and there was a significant 13% decrease in SOL energy in the propulsive phase of stance (Figs. 3 and 4).

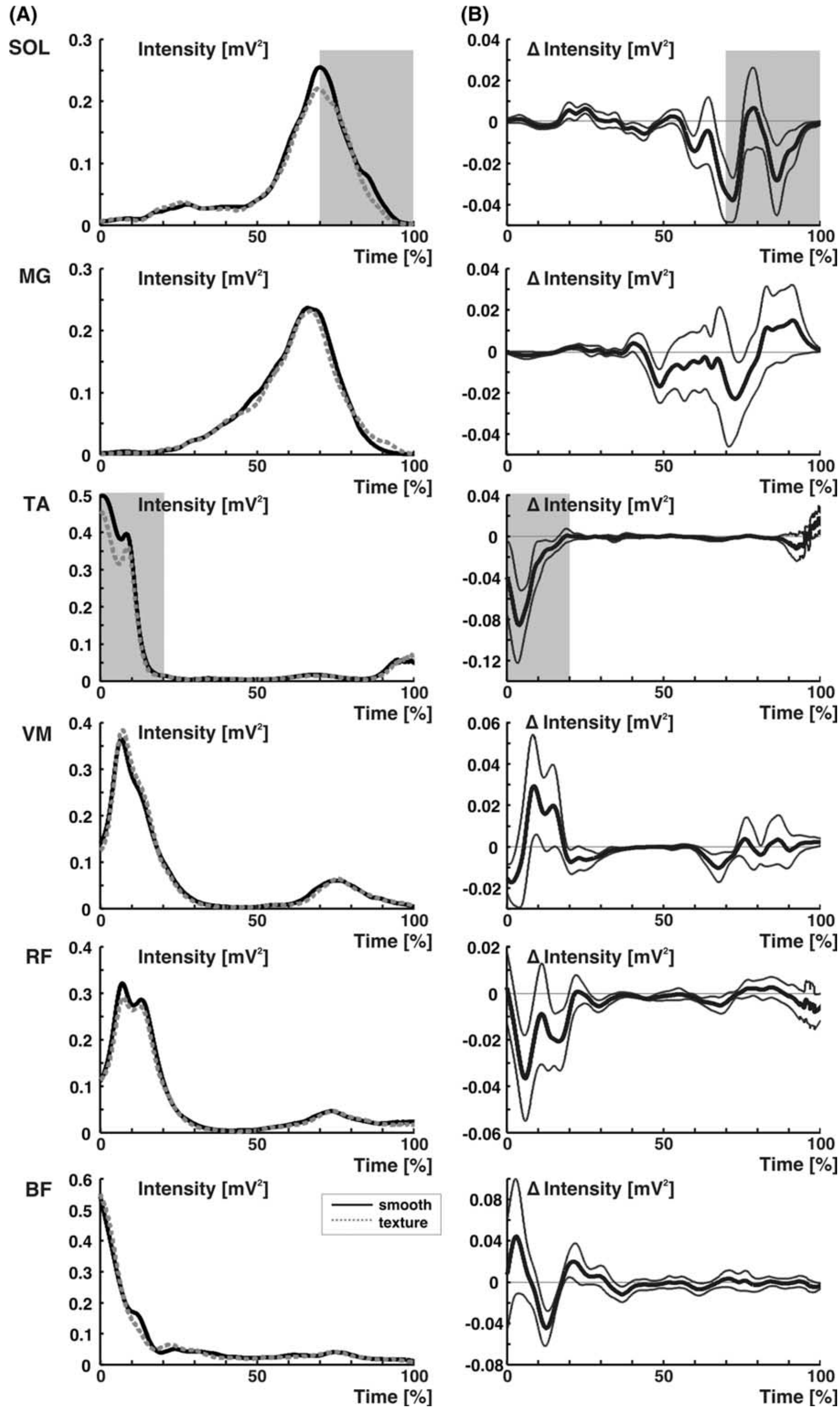

Fig. 4. Intensity and difference plots for soleus (SOL), medial gastrocnemius (MG), tibialis anterior (TA), vastus medialis (VM), rectus femoris (RF), and biceps femoris (BF) muscles. Graphs in the left (A) column represent mean data from all subjects ($n = 15$). Solid black lines show muscle intensity with the flat insert; gray dotted lines are from the textured insert. Standard errors are not shown for clarity. In the right (B) column, mean (and standard errors) differences between the smooth and textured shoe inserts are shown, depicting facilitation or suppression of muscle activity during the stance phase. Significant decreases in intensity were found for SOL and TA (highlighted areas).

A significant decrease in total TA energy was also found for the entire stance phase while wearing the textured insert. In the first part of the stance phase immediately after heel strike, there was a significant 13% decrease in total TA energy (Fig. 4). There were no significant differences found in TA energy for mid-stance, or the propulsion phase of ground contact.

In general, overall total RF energy was decreased during for the entire stance phase ($P = 0.08$). The differences in RF energy occurred primarily in the early part of stance phase. There were no significant differences found for MG, VM, or BF energy while wearing the textured insert during the stance phase of gait.

In analyzing the specific frequency content of the EMG signal, a significant reduction of SOL energy was found in the low, but not the high, frequency domain of the EMG signal when calculated for the entire stance phase. A similar result was found for TA, but the decrease in low frequency energy was not significant ($P = 0.06$). In just the first 20% of the stance phase, a non-significant decrease ($P = 0.06$) was found in the low frequency content of the EMG signal for TA. No differences were found in low frequency SOL energy in this time period. During mid-stance, decreased SOL energy was found in the low frequency domain ($P = 0.08$). No significant differences were observed in the high frequency content of the EMG signal for any muscle or time period analyzed (Fig. 5).

3.2. Kinematics

No significant differences were found in frontal or transverse plane kinematics, such as rearfoot eversion or tibial rotation. There was a small, but significant 0.6° increase in ankle angle (angle between foot and shank) at heel strike with the use of the textured insert,

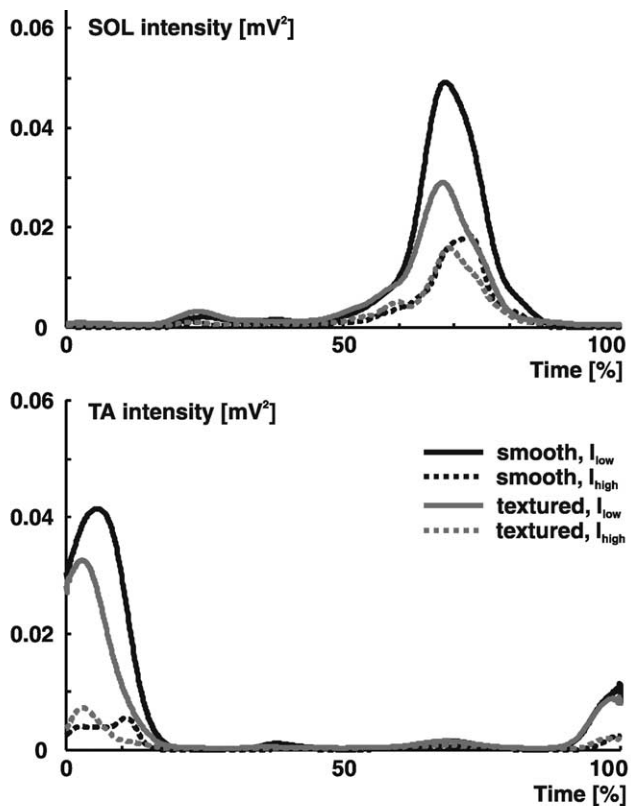

Fig. 5. Low and high frequency intensities of the EMG signal are shown for soleus (soleus) and tibialis anterior (TA) from a representative subject. The smooth insert condition is shown in black; the textured insert in grey. Significant decreases were observed in the low frequency (solid line), but not the high frequency (dotted line), content of the EMG signal.

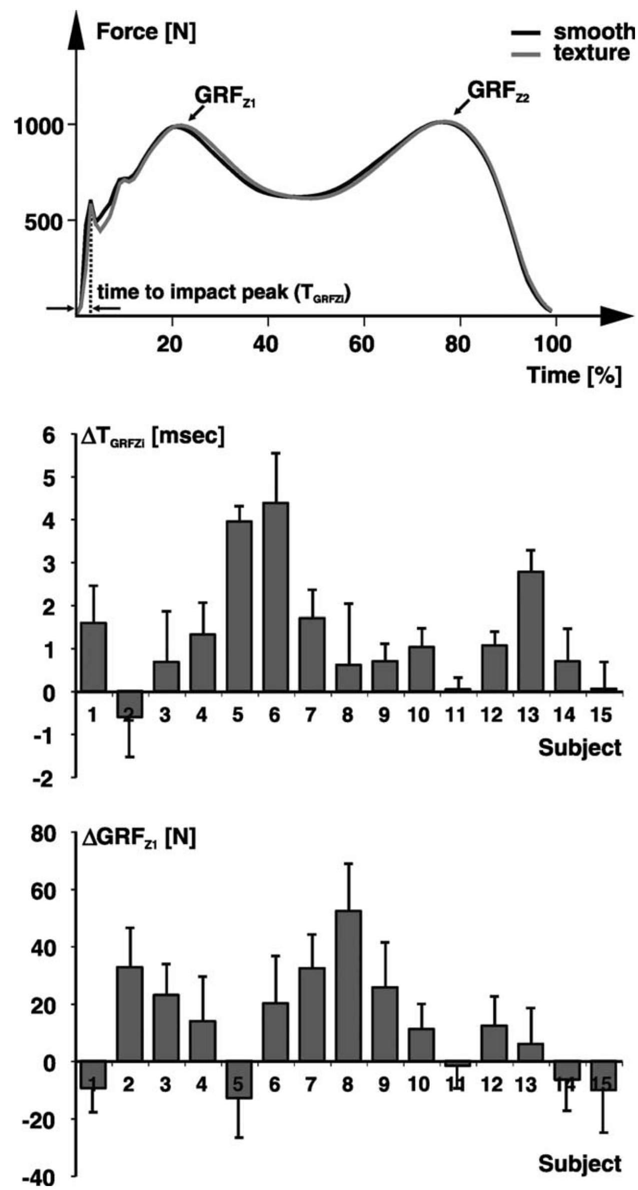

Fig. 6. Differences in vertical ground reaction forces for all subjects. Significant increases in the time to impact peak (T_{imp}), and the first vertical active peak (GRF_{Z1}) were found for the textured insert condition.

resulting in a more plantar flexed position. During the mid-stance phase when the foot is flat on the ground, ankle angles were significantly increased, indicating less rotation of the tibia over the foot. No significant differences were found in ankle angle at takeoff.

There were no overall differences found in knee joint angles at any time period.

3.3. Force and kinetics

There was no significant difference in the magnitude of the peak vertical impact forces between the two insert conditions. There was a significant increase in the time to peak impact force while wearing the textured shoe insert. The time to the impact peak rose 9.6% from 14.5 to 15.9 ms. There was a small, but significant 1.5% increase in the active force peak during the first half of stance phase with the textured insert (Fig. 6). There was no significant difference in the magnitude of the second vertical force peak. No significant differences were seen for the anterior-posterior or medio-lateral components of the ground reaction forces.

There were no significant differences found for ankle joint moments in any plane between the two conditions. At the knee joint, a small (0.25 N) but significant 11.8% decrease in the peak internal rotation moment was found in early stance phase while wearing the textured insert. A difference was not found for peak external rotation moment at the knee during the latter part of stance. A general increase in the first peak abduction moment was found, although this was not significant ($P = 0.08$ and 0.91 N). There were no differences found for the second peak abduction moment during the propulsion phase. No differences were found between the two shoe inserts for sagittal (flexion–extension) knee joint moments.

4. Discussion

The purpose of this study was to use a textured shoe insert as a sensory intervention and examine its effects on human muscle patterns, lower extremity kinematics, and joint kinetics while walking. Changing only the texture of the shoe insert resulted in significant changes in the EMG activity of both ankle flexor and extensor muscles, ankle joint kinematics, and the moments generated at the knee joint.

4.1. Sensory feedback

The basic assumption in this study is that the use of textured shoe inserts in some way altered sensory feedback from the foot. In neurophysiological terms, the concept of altered sensory feedback encompasses a broad spectrum of possibilities, ranging from a change in the discharge rate of specific mechanoreceptor afferents, to alterations in the spatio-temporal firing patterns of a population of sensory afferents. Based on the results of the present study, we cannot make a definitive statement about how afferent feedback was altered, or if feedback was increased or decreased, with the textured inserts.

At present, there is no information from direct recordings from mechano-sensitive plantar afferents to distinguish among these possibilities. Whole nerve neurogram recordings (as described elsewhere in the literature [5,14]) cannot provide definitive answers, since they lack spatial resolution and may not even permit a distinction between sensory and motor neural discharge. To properly address this issue requires direct microneurographic recordings from sizable populations of fully identified mechano-sensitive plantar afferents (i.e. [18,29,31]), while subjects walk with or without textured inserts. This is technically very demanding and probably beyond the scope of current recording techniques.

Nonetheless, it is very likely that textured foot inserts used in this study elicit patterned alterations of sensory discharge from plantar mechano-sensitive afferents while walking. This is supported by the observation that the textured insert provoked an unmistakable subjective response as all subjects reported an immediate change in sensation. The sensory changes induced by the shoe insert were therefore greater than the psychophysical perception threshold, which is commonly used to study the effects of cutaneous feedback on human gait patterns [10,28].

4.2. Muscle activity

Changes in plantar sensory feedback with the textured inserts resulted in decreased soleus and tibialis anterior activity during the period when these muscles are usually active. Similar findings have been presented in the literature, albeit using different stimulation techniques. Electrical stimulation of the tibial nerve, which innervates the plantar foot, has been shown to cause soleus inhibition in static [2] and dynamic (walking) tasks [40]. In addition, Abbruzzese and coworkers [1] suggested that input from the soles of the feet may open oligo-synaptic pathways that have inhibitory effects on soleus motoneurons. These results are consistent with the findings of this study and support the notion that the textured inserts facilitated sensory feedback from the plantar surface of the feet.

Twelve of the fifteen subjects had reduced tibialis anterior activity for the first 20% of stance phase while wearing the textured shoe insert. Again, previous research has shown that a change in cutaneous sensory feedback from the feet can alter tibialis anterior activity in a phase-dependent manner during walking and running. Both Yang and Stein [40] and Duysens and coworkers [10] demonstrated tibialis anterior inhibition in response to electrical stimulation of the tibial nerve during the transition from the swing to stance phase.

In this study, the reduced tibialis anterior energy found at, and immediately after, heel strike may have been a continuance of an inhibitory response caused by increased sensory feedback from the plantar surface of the foot during this transition phase. Although the foot is unloaded immediately prior to touchdown, this transition phase has been suggested as a period when the foot is most sensitive to cutaneous stimuli [13]. The contact of the textured insert on the foot during this critical period may have been sufficient to influence muscle activity. In addition, feedback from the contra-lateral limb during the double support phase of walking may have also provided sensory cues about the nature of the surface underfoot. Tax and coworkers [28] have provided evidence that cutaneous stimuli can cause phase dependent reflex responses in the contra-lateral limb during running. It is reasonable to assume that altering cutaneous feedback from the feet during normal walking with textured inserts may also cause contra-lateral responses.

4.3. Wavelet analysis of EMG signals

The findings of decreased EMG signal energy for soleus and tibialis anterior were based on the total intensity of the wavelet transformed data. This result is the same as would have been found with the frequently used RMS method if the same time periods were compared. The advantage of the wavelet method was that the EMG signals were resolved in both time and frequency space [30]. This allowed an analysis of specific events at different frequency bands within the EMG signal, while maintaining the time resolution.

It has been shown for many different animals that high and low frequency bands within the EMG signal correspond to faster and slower twitch fiber activity, respectively [34,35]. It has been speculated that this is also the case for humans. Motor unit action potentials (MUAPS) from faster motor units have higher conduction velocities and faster rise times which can increase the frequency content of the EMG signal [3,19,26]. The frequency bands used in this study were based on previous work that has already shown them to have different responses in different movement tasks [32,33].

In the current study, decreased soleus and tibialis anterior intensity found while wearing the textured shoe insert was seen in the low frequency spectrum of the EMG signal only. This suggests that the sensory effects of the textured shoe inserts acted primarily on slower motor unit pools. This result is interesting in light of the effects that cutaneous feedback have on muscle reflexes. As previously mentioned, cutaneous feedback from the feet affect lower extremity muscle activity at different phases of the step cycle. If cutaneous feedback from the feet opens inhibitory pathways on lower extremity motoneurons as proposed [1], the effects may be dependent on the specific recruitment and activation patterns during a given task.

During normal level-ground walking, it would be expected that the slower, fatigue resistant muscle fibers were recruited. The textured insert may have affected slower motor units simply due to the fact that they were more predominant during this type of activity. However, afferent feedback may have variable effects on the activity of different threshold motor units in a given task. It would be interesting to examine whether changes in cutaneous feedback can affect high frequency muscle activity that occurs during high intensity activities.

4.4. Kinematics and kinetics

The hypothesis that the textured inserts would affect kinematics only in the sagittal plane was supported. Significant differences were only found for sagittal plane ankle motion, with increased plantar flexion of the foot at touchdown. Subject differences in touchdown flexion angles ranged from $-0.4°$ to $2.9°$, with all but three subjects showing increased plantar flexion. These values are in the same range as differences reported by others following non-noxious electrical stimulation of the tibial nerve during gait [37,41]. The differences in foot kinematics also coincide with the decrease in tibialis anterior activity at touchdown and the early part of the stance phase.

The hypothesis that the textured insert would not affect ground reaction forces or joint kinetics was not supported. The increased time to impact peak with the textured inserts suggest a feed-forward adaptive response to changed sensory feedback, as reflex responses could not occur in this timeframe. No differences were seen in rearfoot eversion, eversion velocity, or sagittal knee joint angles during early stance, all of which have been linked to decreased impact loading rates [6,8]. This difference in time to impact peak may have been due to a more plantar flexed position at touchdown. Assuming that the textured insert facilitated plantar feedback, the changed loading rate may have been an attempt to mitigate large amounts of sensory input at touchdown, especially at the heel. This presumes that loading rates are linked to sensory input and plantar sensation, which may or may not be the case.

4.5. Methodological considerations

The adaptive capabilities of the mammalian neuromuscular system have been demonstrated on many occasions. Movement strategies can be altered in a very short period of time in response to external perturbations [22], or in a phasic manner during the step cycle [40]. Adaptations may also occur over longer periods as new motor strategies are learned.

In this study, the immediate reflexive response to the shoe inserts was not measured. Due to the nature of the experimental design, only data from the fifth or sixth step in a given trial were collected. While it is possible that an adaptive response occurred in the first step, it was not the purpose of this study to determine the step by step adaptations that may occur with a footwear intervention. Rather, it was our goal to determine the steady state changes that resulted from the changes in sensory feedback with textured footwear.

4.6. Functional considerations

Footwear interventions are commonly used to alter gait patterns, improve comfort, and treat a number of lower extremity ailments [15,24,25]. Shoe inserts and orthoses are typically considered to be mechanical interventions. However, it was not possible to determine whether the effects of those interventions were mechanical or sensory in nature. Therefore, we excluded the mechanical component by changing only the texture of the shoe insert.

The results of this study suggest that changing footwear textures may be used as a sensory intervention to alter gait patterns, especially muscle activity, during walking. At this time, the clinical benefits and consequences of such changes are not understood. For example, it is not known whether a decrease in muscle activity, either early or late in the stance phase, represents a more efficient movement pattern, or a potential decrease in task performance.

For some people, increased sensory feedback from the plantar surface of the foot may be desirable, either from a psychological or functional viewpoint [21]. For others, the changes in muscle activity may, or may not affect comfort and/or fatigue during locomotion. At this point in time, it is difficult to make predictions about what subject populations would benefit from increased sensory feedback during dynamic movement tasks. Nonetheless, the results do suggest that footwear may have both mechanical and sensory effects.

From a biomechanics point of view, further research is required to investigate how textured footwear interventions and the changes in muscle activity and ankle kinematics affects performance, comfort, fatigue, and overuse injuries. From a neurophysiology point of view, the effects of textured footwear on afferent plantar feedback, as well as the specific effects on low and high frequency components of muscle activity requires further investigation.

5. Conclusions

Changing the texture, without changing the shape, of a footwear intervention, caused changes in gait patterns, especially lower extremity muscle activation patterns. Changed sensory input may have specific effects on different motoneuron pools. There is a strong indication that the changes in gait patterns were due to a change in sensory feedback from the plantar surface of the foot. This supports the theory that sensory feedback from cutaneous receptors in the plantar surface of the foot is important in determining movement strategies during human locomotion. Furthermore, the results of this study suggest that textures may be used as a sensory intervention to alter gait patterns in subjects who seek footwear interventions. The benefits and consequences of such changes are not yet known.

Acknowledgments

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References

[1] M. Abbruzzese, V. Rubino, M. Schieppati, Task-dependent effects evoked by foot muscle afferents on leg muscle activity in humans, Electroencephalogr. Clin. Neurophysiol. 101 (1996) 339–348.
[2] A.M. Aniss, S.C. Gandevia, D. Burke, Reflex responses in active muscles elicited by stimulation of low-threshold afferents from the human foot, J. Neurophysiol. 67 (1992) 1375–1384.
[3] L. Arendt-Nielsen, K.R. Mills, The relationship between mean power frequency of the EMG spectrum and muscle fibre conduction velocity, Electroencephalogr. Clin. Neurophysiol. 60 (1985) 130–134.
[4] M. Belanger, A.E. Patla, Corrective responses to perturbation applied during walking in humans, Neurosci. Lett. 49 (1984) 291–295.
[5] D. Burke, Unit identification, sampling bias and technical issues in microneurographic recordings from muscle spindle afferents, J. Neurosci. Methods 74 (1997) 137–144.
[6] G.K. Cole, B.M. Nigg, A.J. van den Bogert, K.G.M. Gerritsen, Lower extremity joint loading during impact in running, Clin. Biomech. 11 (1996) 181–193.
[7] G.K. Cole, B.M. Nigg, J.L. Ronsky, M.R. Yeadon, Application of the joint coordinate system to three-dimensional joint attitude and movement representation: a standardization proposal, J. Biomech. Eng. 115 (1993) 344–349.
[8] S.J. Dixon, A.C. Collop, M.E. Batt, Surface effects on ground reaction forces and lower extremity kinematics in running, Med. Sci. Sports. Exerc. 32 (2000) 1919–1926.
[9] M.C. Do, B. Bussel, Y. Breniere, Influence of plantar cutaneous afferents on early compensatory reactions to forward fall, Exp. Brain. Res. 79 (1990) 319–324.
[10] J. Duysens, M. Trippel, G.A. Horstmann, V. Dietz, Gating and reversal of reflexes in ankle muscles during human walking, Exp. Brain. Res. 82 (1990) 351–358.
[11] J. Duysens, A.A.M. Tax, M. Trippel, V. Dietz, Phase-dependent reversal of reflexively induced movements during human gait, Exp. Brain Res. 90 (1992) 404–414.
[12] J. Duysens, A.A.M. Tax, M. Trippel, V. Dietz, Increased amplitude of cutaneous reflexes during human running as compared to standing, Brain Res. 613 (1993) 230–238.

[13] J. Duysens, A.A.M. Tax, S. Nawijn, W. Berger, T. Prokop, E. Altenmüller, Gating of sensation and evoked potentials following foot stimulation during human gait, Exp. Brain Res. 105 (1995) 423–431.
[14] S.C. Gandevia, J.P. Hales, The methodology and scope of human microneurography, J. Neurosci. Methods 74 (1997) 123–136.
[15] M.L. Gross, L.B. Davlin, P.M. Evanski, Effectiveness of orthotic shoe inserts in the long-distance runner, Am. J. Sports Med. 19 (1991) 409–412.
[16] R. Hayashi, A. Miyake, S. Watanabe, The functional role of sensory inputs from the foot: stabilizing human standing posture during voluntary and vibration-induced body sway, Neurosci. Res. 5 (1988) 203–213.
[17] F.B. Horak, L.M. Nashner, H.C. Diener, Postural strategies associated with somatosensory and vestibular loss, Exp. Brain Res. 82 (1990) 167–177.
[18] R.S. Johansson, A.B. Vallbo, Tactile sensibility in the human hand: relative and absolute densities of four types of mechanoreceptive units in glabrous skin, J. Physiol. 286 (1979) 283–300.
[19] L. Lindström, R. Kadefors, I. Petersen, An electromyographic index for localized muscle fatigue, J. Appl. Physiol. 43 (1977) 750–754.
[20] M. Magnusson, H. Enbom, R. Johansson, I. Pyykkö, Significance of pressor input from the human feet in anterior-posterior postural control, Acta. Otolaryngol. (Stockh) 110 (1990) 182–188.
[21] B.E. Maki, S.D. Perry, R.G. Norrie, W.E. McIlroy, Effect of facilitation of sensation from plantar foot-surface boundaries on postural stabilization in young and older adults, J. Gerontol. A 54 (1999) M281–M287.
[22] L.M. Nashner, M. Woollacott, G. Tuma, Organization of rapid responses to postural and locomotor-like perturbations of standing man, Exp. Brain Res. 36 (1979) 463–476.
[23] B.M. Nigg, A. Khan, V. Fisher, D. Stefanyshyn, Effect of shoe insert construction on foot and leg movement, Med. Sci. Sports Exerc. 30 (1998) 550–555.
[24] B.M. Nigg, M.A. Nurse, D.J. Stefanyshyn, Shoe inserts and orthotics for sport and physical activities, Med. Sci. Sports Exerc. 31 (1999) S421–S428.
[25] M. Razeghi, M.E. Batt, Biomechanical analysis of the effect of orthotic shoe inserts. A review of the literature, Sports Med. 29 (2000) 425–438.
[26] M. Solomonow, C. Baten, J. Smit, R. Baratta, H. Hermens, R. D'Ambrosia, H. Shoji, Electromyogram power spectra frequencies associated with motor unit recruitment strategies, J. Appl. Physiol. 68 (1990) 1177–1185.
[27] A. Stacoff, C. Reinschmidt, B.M. Nigg, A.J. van den Bogert, A. Lundberg, J. Denoth, E. Stüssi, Effects of foot orthoses on skeletal motion during running, Clin. Biomech. 15 (2000) 54–64.
[28] A.A.M. Tax, B.M.H. van Wezel, V. Dietz, Bipedal reflex coordination to tactile stimulation of the sural nerve during human running, J. Neurophysiol. 73 (1995) 1947–1964.
[29] M. Trulsson, Mechanoreceptive afferents in the human sural nerve, Exp. Brain Res. 137 (2001) 111–116.
[30] V. von Tscharner, Intensity analysis in time-frequency space of surface myoelectric signals by wavelets of specified resolution, J. Electromyogr. Kinesiol. 10 (2000) 433–445.
[31] J.P. Vedel, J.P. Roll, Response to pressure and vibration of slowly adapting cutaneous mechanoreceptors in the human foot, Neurosci. Lett. 34 (1982) 289–294.
[32] J.M. Wakeling, S.A. Pascual, B.M. Nigg, V. von Tscharner, Surface EMG shows distinct populations of muscle activity when measured during sustained sub-maximal exercise, Eur. J. Appl. Physiol. 86 (2001) 40–47.
[33] J.M. Wakeling, S.A. Pascual, B.M. Nigg, Altering muscle activity in the lower extremities by running with different shoes, Med. Sci. Sports Exerc. 34 (2002) 1529–1532a.
[34] J.M. Wakeling, M. Kaya, G.K. Temple, I.A. Johnston, W. Herzog, Determining patterns of motor recruitment during locomotion, J. Exp. Biol. 205 (2002) 359–369b.
[35] J.M. Wakeling, D.A. Syme, Wave properties of action potentials from fast and slow motor units of rats, Muscle Nerve 26 (2002) 659–668.
[36] I. Watanabe, J. Okubo, The role of the plantar mechanoreceptor in equilibrium control, Ann. N Y Acad. Sci. 374 (1981) 855–864.
[37] B.M.H. van Wezel, F.A.M. Ottenhoff, J. Duysens, Dynamic control of location-specific information in tactile cutaneous reflexes from the foot during human walking, J. Neurosci. 17 (1997) 3804–3814.
[38] B.M.H. van Wezel, B.G.M. van Engelen, F.J.M. Gabreëls, A.A.W.M. Gabreëls-Festen, J. Duysens, Aβ fibers mediate cutaneous reflexes during human walking, J. Neurophysiol. 83 (2000) 2980–2986.
[39] G. Wu, J.-H. Chiang, The significance of somatosensory stimulations to the human foot in the control of postural reflexes, Exp. Brain Res. 114 (1997) 163–169.
[40] J.F. Yang, R.B. Stein, Phase-dependent reflex reversal in human leg muscles during walking, J. Neurophysiol. 63 (1990) 1109–1117.
[41] E.P. Zehr, T. Komiyama, R.B. Stein, Cutaneous reflexes during human gait: electromyographic and kinematic responses to electrical stimulation, J. Neurophysiol. 77 (1997) 3311–3325.

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