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Clinical Biomechanics 19 (2004) 71–77

CLINICAL BIOMECHANICS

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The role of selected extrinsic foot muscles during running

Kristian M. O'Connor *, Joseph Hamill

Biomechanics Laboratory, University of Massachusetts, Amherst, MA 01003, USA

Received 20 November 2002; accepted 4 September 2003

Abstract

Objective. To determine the kinematic, kinetic and EMG responses to perturbations of the foot by running in varus, neutral, and valgus-wedged shoes.

Design. Within-subjects study comparing kinematics, kinetics and EMG while running in three different shoe conditions.

Background. Excessive pronation has been cited as a key contributor to many types of running injuries. However, the roles of the extrinsic foot muscles (those that control motion of the foot) during the stance phase of running have not been adequately identified, which is critical to determining the relationship between pronation and injury.

Methods. Ten males ran in varus, valgus, and neutral-wedged shoes while three-dimensional kinematic and kinetic data and EMG data were collected. Surface EMG data were collected from the tibialis anterior, peroneus longus, medial and lateral gastrocnemius, and soleus. Indwelling EMG was obtained from the tibialis posterior. The net joint moment, power, and total positive and negative work was calculated in the frontal plane. EMG onset, offset, and integrated values were reported.

Results. The maximum eversion angle, maximum inversion moment and total negative work done in the frontal plane were greatest while running in the valgus shoe and least in the varus shoe. The greater joint moment was not accompanied by changes in muscle activation patterns, although the tibialis posterior data were inconclusive in this respect.

Conclusions. Greater pronation leads to greater energy absorption in the foot invertor muscles and tendons. While not conclusive, the EMG data suggest that for these muscles there was not a neuromuscular adaptation to the perturbation.

Relevance

This study reinforces the hypothesized link between excessive pronation and injury and provides valuable insight into the muscular responses (or lack thereof) when foot motion is altered. This information is critical in understanding the effects of shoe design and orthotic interventions.

Keywords: Pronation; Gait; Injuries; Ankle

1. Introduction

Many overuse injuries in running are associated with excessive pronation of the foot during stance (Hintermann and Nigg, 1998; James et al., 1978), although the term excessive has not been clinically defined (Nigg and Morlock, 1987). The etiology of leg injuries is uncertain and certainly multi-factorial but excessive pronation may impose stress on the extrinsic muscles of the foot that leads to injury. The extrinsic foot musculature includes all muscles that insert on the foot but originate proximal to the foot. These include the gastrocnemius, soleus, tibialis posterior, tibialis anterior, peroneus muscles, and the flexor/extensor digitorum/hallucis muscles. Understanding the functional role of these muscles is critical to understanding the etiology of many running injuries.

The tibialis posterior is generally regarded as the primary invertor of the foot based upon its moment arm about the subtalar joint (Perry, 1983). Hence, it is considered the muscle that primarily acts to control the amount of pronation (eversion) that occurs during the stance phase of running. In support of this, McClay and Manal (1999) reported an inversion moment during running stance that they postulated represented activity in the tibialis posterior. Many of the other extrinsic foot muscles, however, also exert an inversion moment about

* Corresponding author. Address: Department of Human Movement Sciences, University of Wisconsin—Milwaukee, P.O. Box 413, Milwaukee, WI 53201, USA.

E-mail address: krisocon@uwm.edu (K.M. O'Connor).

0268-0033/$ - see front matter © 2003 Elsevier Ltd. All rights reserved.
doi:10.1016/j.clinbiomech.2003.09.001

the subtalar joint. The gastrocnemius and soleus have a small inversion moment arm, but these muscles also exert large forces during running to provide the required plantar flexion moment (Perry, 1983). Reber et al. (1993) reported that peak activity of the gastrocnemius, soleus, and tibialis posterior occurred during early to mid-stance which would support their role in resisting foot pronation. It has also been demonstrated that the tibialis anterior is active during early stance, which is an invertor of the foot. The peroneus muscles act as evertors of the foot, but they may be active during early stance in order to increase overall joint stiffness. The toe flexors and extensors also are capable of providing frontal plane moments based on their lines of action, but based on muscle size and moment arms they likely provide a small contribution to the overall moment at this joint.

Perturbation studies can help to clarify the exact role these muscles play. A perturbation that either accentuates or inhibits pronation may alter joint moments and muscle activation patterns illustrating how these muscles act to control pronation of the foot. Mundermann et al. (2003) reported that wearing a posted orthotic decreased the peak inversion moment during stance. While providing important information about the effects of orthoses on joint dynamics, the subject-specific nature of the orthotic intervention makes it difficult to determine a more general response to frontal plane perturbations that could be useful for shoe midsole design. Also, these results indicated that the kinetics of the foot could be altered, but they do not establish what structures led to the change in joint moments.

Systematic perturbations at the foot have been induced by requiring subjects to run in shoes with varus- and valgus-wedged midsoles (Milani et al., 1995; Perry and Lafortune, 1993; van Woensel and Cavanagh, 1992). The varus-wedged shoes were designed with a thicker midsole along the medial portion of the shoe, and the valgus shoes were designed with a thicker midsole along the lateral portion of the shoe. These studies reported systematic increases in pronation when running in the valgus shoe as compared to the varus shoe. These authors did not, however, investigate joint kinetic parameters or muscle activation patterns. An understanding of how these parameters change with perturbations to the foot is critical to understanding possible mechanisms of injuries.

Therefore, the purpose of this study was to investigate the role of selected extrinsic foot muscles during running. This was accomplished by determining the kinematic, kinetic, and EMG responses to running in varus, neutral, and valgus-wedged shoes. It was hypothesized that the varus shoe should decrease foot pronation and the inversion moment while the valgus shoe should increase pronation and the inversion moment. It was further hypothesized that the EMG activation levels of muscles involved in controlling pronation (tibialis anterior, tibialis posterior, soleus, medial and lateral gastrocnemius) would be greatest in the valgus shoes and least in the varus shoes.

2. Methods

2.1. Subjects and materials

Ten males, classified as rearfoot strikers, foot sizes 9–10, were recruited for this study. Their average height, mass and age were 1.72 (SD 0.07) m, 72.6 (SD 5.3) kg, 27 (SD 5) years, respectively. All subjects were active recreationally and injury-free at the time of the study, and none wore orthotics. All subjects signed an informed consent in accordance with university regulations.

Three pairs of shoes (size 9.5) were custom built for this study. All shoes utilized a midsole constructed of ethylvinyl acetate (EVA) with a durometer of 45 (Shore A). The neutral shoes were constructed with a heel height of 2.5 cm. In order to attain an 8°-varus configuration, the medial aspect of the midsole at the heel was 3-cm thick, and the lateral aspect was 2-cm thick. The valgus shoes were built according to the same dimensions but with the heights reversed. The midsoles tapered to a 1-cm thickness at approximately the metatarsals. Similar midsole designs have been used in other studies (Milani et al., 1995; Perry and Lafortune, 1993; van Woensel and Cavanagh, 1992). The upper portion of the three shoes was modified such that there was no heel counter. This design was employed in order to directly track movement of the calcaneus. The shoe was fastened to the foot through the use of a cuff around the ankle to which straps from the shoe were attached (Fig. 1). In a pilot study, it was determined that subjects were able to run comfortably in these shoes.

2.2. Experimental setup

Kinematic, kinetic and EMG data were acquired from the right lower limb of all subjects. Three-dimen-

Fig. 1. Modified upper portion of the shoe designed to directly measure calcaneus motion. Markers triad placement also demonstrated.

sional kinematic data were collected using a seven-camera Qualisys Pro-Reflex motion capture system (East Windsor, CT, USA). Ground reaction force data were collected with a force platform (model BP6001200, AMTI, Inc., Watertown, MA, USA) mounted flush with the floor. Ag–AgCl dual electrodes (2.0-cm inter-electrode distance) were placed at each recording site (model 272, Noraxon, Inc., Scottsdale, AZ, USA). The in-dwelling electrode consisted of two 44-ga paired hook wires within a 25-ga cannula for insertion (Nicolet Instrument Corporation, Madison, WI, USA). All EMG data were collected from a Telemyo-8 telemetered EMG system (Noraxon, Inc., Scottsdale, AZ, USA). Speed was monitored by recording the time between two photoelectric sensors placed at each end of the testing zone. Kinematic data were collected at 240 Hz while ground reaction force and EMG data were collected synchronously at 1920 Hz.

2.3. Protocol

Four reflective markers on a rigid plate were attached to the leg and a marker triad, also on a rigid plate, was attached to the posterior aspect of the foot. Prior to placement of the surface electrodes, the sites were shaved, abraded, and cleaned with alcohol. Surface EMG data were collected from the tibialis anterior (TA), peroneus longus (Per), lateral gastrocnemius (LG), medial gastrocnemius (MG), and the soleus (Sol). Electromyographic data for the tibialis posterior (TP) were collected using a fine-wire electrode. The needle was inserted approximately one centimeter from the medial edge of the tibia, one hand-width below the tibial tuberosity. Prior to the data collection, a standing calibration trial was collected with the subject standing barefoot. For the standing calibration, additional reflective markers were placed on the subject in order to define segment geometries and the segment coordinate systems. These markers were placed on the skin over the medial and lateral femoral epicondyles, the medial and lateral malleoli and the heads of the first and fifth metatarsals and were subsequently removed prior to the collection of the running trials.

Before collecting the overground running trials, subjects ran on a treadmill at 3.6 m/s for approximately 2 min. Subjects then performed five acceptable running trials in each shoe at 3.6 m/s $\pm$ 5% along a 30-m walkway across the force platform. Subjects were required to land on the force plate with their right foot while kinematic, kinetic and EMG data were recorded.

2.4. Data analysis

The three-dimensional coordinate data were filtered with a low-pass, fourth order, zero lag, Butterworth filter with a 12-Hz cutoff frequency (Hamill et al., 1992). Anatomical coordinate systems for the leg and foot were derived from the marker locations collected in the standing calibration. The leg reference system was right-handed and anatomically based, with the X-axis pointing laterally, the Y-axis point anterior, and the Z-axis pointing superiorly along the long axis of the segment. The foot coordinate system was aligned with the room coordinate system during the standing calibration. Three-dimensional angular data were calculated using an XYZ Cardan rotation sequence (Cole et al., 1993).

Kinematic joint parameters at the ankle were extracted for the frontal plane. While the ankle joint complex contains two separate joints, the ankle and subtalar joints, these were treated as a single universal joint (Cole et al., 1993). The touchdown angle, maximum eversion angle, time to maximum eversion, range of motion and peak eversion velocity were reported.

Kinematic and force data were combined to calculate joint kinetic data using an inverse dynamics Newton–Euler procedure (Bresler and Frankel, 1950). Each segment was modeled as a frustra of a cone. Inertial parameters were derived from Dempster (1955). The ankle joint center was defined by the midpoint between the standing calibration markers placed on the medial and lateral malleoli. Moments and powers were calculated about the ankle joint and reported in the frontal plane of the foot coordinate system. Maximum and minimum joint moments and powers and the times to the moments were reported. Work was computed from the time integral of the power time series. Negative and positive work were reported, as well as the total work defined as the sum of the absolute values of the positive and negative work (Eng and Winter, 1995).

EMG data were high-pass filtered at 20 Hz and full-wave rectified prior to the calculation of variables. The high-pass filter was utilized in order to remove movement artifact and the DC bias (Winter, 1990). Since full stride data were unavailable due to data collection limitations, the period before foot contact equal in duration to the stance time was chosen as the beginning of the time series, and toe off was chosen as the end. Therefore, the data were reported from −100% to 100% of stance with 0% representing foot contact. The integrated EMG and the mean of each period were calculated as well as the onset and offset of activity for each muscle. In order to determine the onset and offset of muscle activity, the data were low-pass filtered at 24 Hz and the magnitudes were normalized to the highest peak from the five neutral condition trials of each subject. The threshold for onset and offset was set at 10% of the peak. Linear envelopes were also created by filtering the rectified signal with a 12 Hz low-pass filter for illustration purposes.

All data were time normalized to 100% of stance, with the EMG data reported as described above. Kinematic, kinetic, and EMG parameters were extracted from each trial and mean curves were calculated for each subject.

2.5. Statistical analysis

A one-way repeated measures ANOVA ($P < 0.05$) was performed on each kinematic, kinetic, and EMG parameter to detect differences between shoe conditions. A Tukey's post-hoc test was conducted where appropriate.

3. Results

There were significant differences in the frontal plane kinematic variables among conditions (Table 1 and Fig. 2(a and b)). There were no differences in touchdown angles, but the varus shoe significantly decreased the range of motion. Maximum eversion was significantly greater in the valgus shoe than the neutral and varus shoes, but the times to maximum eversion were not significantly different.

There was an inversion moment during the first portion of stance, followed by an eversion moment during the second portion of stance in all conditions (Fig. 2c). There was a significantly greater inversion moment while running in the valgus shoe, and a significantly smaller inversion moment while running in the varus shoe as compared to the neutral condition (Table 1). The frontal plane power was characteristically triphasic with energy absorption occurring during early and late stance with energy generation only occurring during a short period during the middle of stance (Fig. 2d). There was significantly greater energy absorption while running in the valgus shoe as compared to the other conditions.

Table 1
Group mean (SD) frontal plane kinematic and kinetic parameters

Variable	Condition		
	Varus	Neutral	Valgus
Touchdown angle (°)	4.1 (10.5)	5.0 (9.9)	4.2 (8.5)
Range of motion (°)*	9.1 (2.9)[a]	11.1 (4.1)[b]	12.8 (5.5)[b]
Peak eversion angle (°)*	−5.0 (8.2)[a]	−6.1 (6.7)[a]	−8.5 (7.0)[b]
Time to peak eversion (% stance)	40.2 (11.9)	40.0 (10.7)	36.4 (10.3)
Peak eversion velocity (°/s)*	−253.8 (42.2)[a]	−294.8 (122.9)[a,b]	−335.4 (127.8)[b]
Peak inversion moment (N m)*	12.5 (14.3)[a]	14.7 (14.5)[a]	21.0 (15.7)[b]
Time to inversion moment (% stance)	43.0 (19.6)	37.4 (23.8)	33.2 (13.5)
Peak eversion moment (N m)	−19.9 (15.8)	−16.3 (10.4)	−17.4 (14.4)
Time to eversion moment (% stance)	62.9 (12.2)	68.3 (15.6)	57.2 (13.0)
Minimum power (W)*	−64.2 (28.2)[a]	−63.9 (25.5)[a]	−106.7 (51.0)[b]
Maximum power (W)	67.6 (47.4)	54.3 (28.5)	66.6 (30.5)
Negative work (J)	−2.8 (1.6)	−2.6 (1.3)	−4.1 (2.2)
Positive work (J)	2.4 (1.6)	1.9 (0.8)	2.4 (1.0)
Total work (J)	5.2 (2.7)	4.5 (1.6)	6.5 (2.6)

* Significant difference ($P < 0.05$). Like letters are not significantly different.

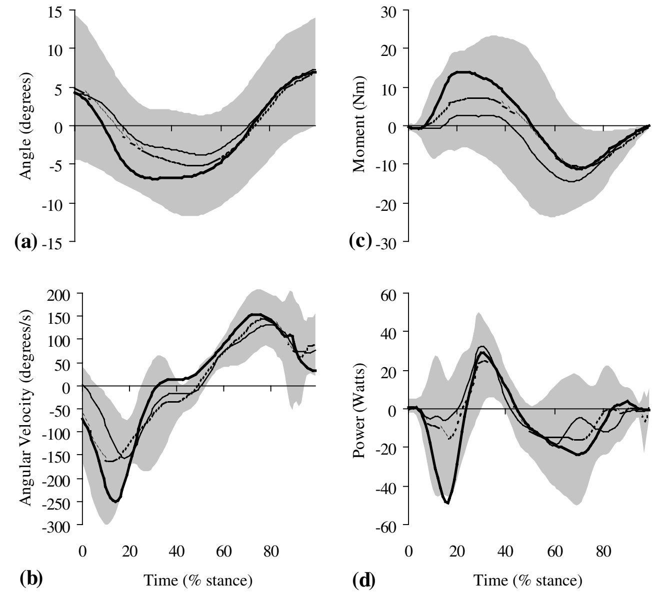

Fig. 2. Mean (a) frontal plane angle, (b) angular velocity, (c) net joint moment, and (d) joint power during the stance phase of running for all subjects. The thin line represents the varus condition, the dashed line represents the neutral condition, and the thick line represents the valgus condition Positive angle, velocity, and moment values represent inversion. Positive power represents energy generation. The shaded area represents ±1 SD (between subjects) from the neutral condition. The large standard deviations are indicative of the differing individual movement patterns that were observed, although subjects responded consistently to the three conditions.

Ensemble activation profiles were compiled for the EMG of each muscle with the number of subjects shown for which acceptable data were collected (Fig. 3). Due to a variety of difficulties in collecting EMG data, data for certain muscles were not obtained on given subjects. In particular, the tibialis posterior proved to be a difficult signal to record from indwelling electrodes for an entire data collection session. The signal degraded in several subjects across the data collection period, leaving only four subjects with full sets of data for this muscle. With only four subjects, statistical tests on this muscle were not performed. There were no significant differences in the integrated EMG, mean EMG, onset, or offset times between conditions for any of the other muscles recorded (Fig. 4).

4. Discussion

The purpose of this study was to examine the role of the extrinsic foot muscles during running by investigat-

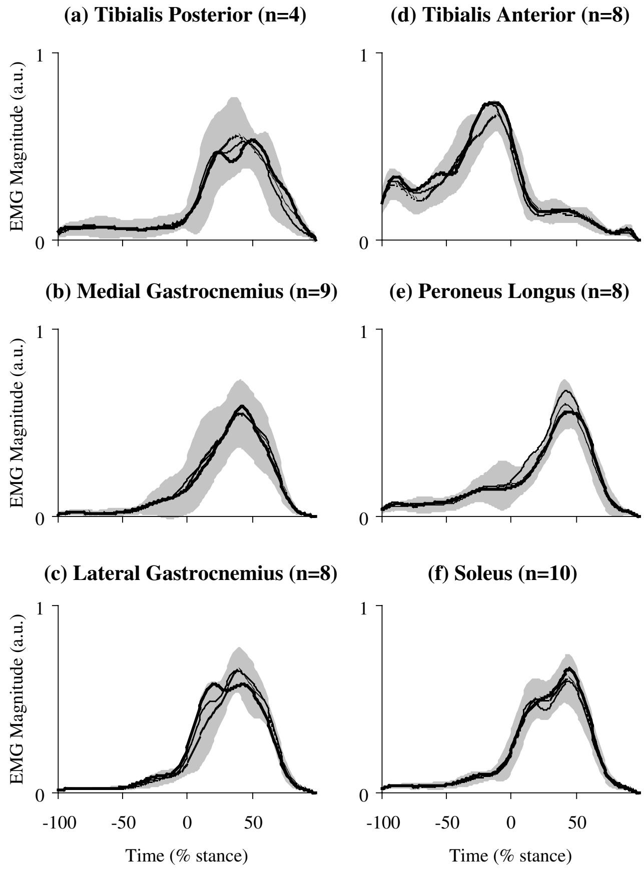

Fig. 3. Group ensemble EMG activity profiles. The thin line represents the varus condition, the dashed line represents the neutral condition, and the thick line represents the valgus condition. The magnitudes are in arbitrary units. Time zero represents foot contact and −100% represents the instance in time before foot contact equal to stance time. The shaded area represents ±1 SD (between subjects) from the neutral condition. The number of subjects included in the ensemble average are shown in the parentheses.

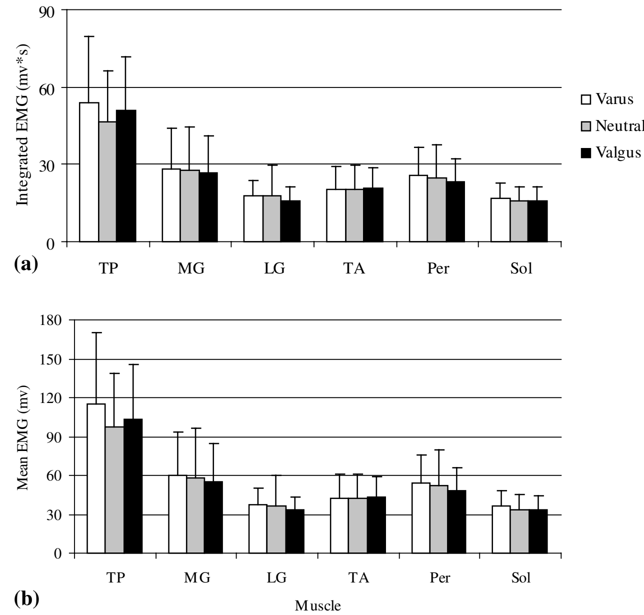

Fig. 4. (a) Integrated and (b) mean EMG values for all subjects. The mean and standard deviation across subjects are represented. There were no significant differences between conditions.

ing the kinematic, kinetic and muscle activation changes that occurred while running in varus, neutral and valgus-wedged shoes. It was hypothesized that the varus shoe would decrease pronation during stance, reduce the net inversion joint moment and reduce activation levels in the invertor muscles (tibialis posterior, gastrocnemius, soleus). The valgus shoe was hypothesized to have the opposite effect relative to the neutral condition. Anticipated joint kinematic and kinetic changes were observed, but there were no significant differences in muscle activation profiles between shoes.

Given the design of the shoe upper chosen for this study, the intent was not to directly compare kinematic results of this study to traditional running shoe studies, nor to infer the motion of the foot within a regular running shoe. Rather, the purpose was to understand the relationship between kinematic, kinetic and EMG variables when motion of the foot is perturbed in some way during running. Although the markers were placed directly on the calcaneus, the kinematic results of the neutral shoe compare favorably to Reinschmidt et al. (1997) who reported three-dimensional ankle kinematics derived from bone pins inserted into the tibia and calcaneus while running in shoes with a heel counter. The joint displacement profiles in their study demonstrated similar patterns in all three planes and the joint excursions were quite similar to the present study. The maximum eversion angle of the bone pins was 8.6° while this study reported 6.1°. In comparison, Reinschmidt et al. (1997) also reported kinematics based on shoe-based markers and reported a maximum eversion angle of 13°. While not central to the purpose of this study, running in the shoes without a heel counter appears to yield kinematic information quite similar to running in a shoe with a heel counter.

With regard to the experimental manipulation, running in shoes with a wedged midsole elicited the predicted kinematic response with the valgus-wedged shoe accentuating calcaneal eversion. Other studies (Milani et al., 1995; Perry and Lafortune, 1993; van Woensel and Cavanagh, 1992) that have employed these shoes have also reported significant differences in rearfoot motion, although these studies based their results upon shoe-based markers. van Woensel and Cavanagh (1992) and Perry and Lafortune (1993) both utilized shoes with 10° wedges while Milani et al. (1995) utilized shoes with 8° wedges. In each case, the differences in rearfoot angle between shoes were approximately equal to the intervention. This could be predicted since they measured the motion of the heel counter, which is rigidly attached to the midsole. The direct measurement of the foot in this study, not using a heel counter, indicated that the 8°-wedge only altered the skeletal motion about 2°–3°. This result seems consistent with other findings where substantial shoe alterations were made that result in little change in foot kinematics (Stacoff et al., 2001). Even so, this magnitude of perturbation was sufficient to alter joint moment patterns.

The frontal plane moments were highly variable among subjects. McClay and Manal (1999), the only study to report normative kinetic data for the secondary planes of motion during running, also reported high variability in the joint moment patterns in the frontal plane. There was also high variability in the joint powers among subjects. Despite this variability between subjects, the peak inversion moment and peak negative power and the peak abduction moment in these secondary planes were significantly different among conditions.

The peak inversion moment was greatest for the valgus shoe which was designed to accentuate pronation. There was 58% more energy absorbed in the frontal plane for the valgus shoe as compared to the neutral shoe (valgus = −4.1 J; neutral = −2.8 J). This increased energy absorption occurred during early stance. While this amount of work is small compared to the sagittal plane (−30 J), it may be sufficient to cause injury if the additional load is primarily experienced in a single structure such as the tibialis posterior muscle. Noyes (1977), for example, reported that the anterior cruciate ligament of rhesus monkeys reached failure when the ligament absorbed 3.5 J of energy. While human tendons are likely thicker than the monkey ACL, the physical properties are similar. Therefore, an additional 1.6 J could be sufficient to damage the tissue when repetitively stressed. While the mechanism of tendon-related injuries is unclear, experimental evidence indicates that increased energy absorption may contribute to injuries (Fisher, 2000). Therefore, increased energy absorption by the muscles caused by hyper-pronation may indeed be a mechanism for injury.

The muscle activity profiles in this study were not reported for a full stride. However, the stance and pre-stance phases are similar to those in the literature for normal running (Reber et al., 1993). All of the muscles other than the tibialis anterior were generally quiet throughout swing. The pre-impact phase for the tibialis anterior does appear to capture the prominent burst of activity in preparation for landing. Given this, the EMG results as represented in this study likely still contain the critical information about their activity during running for most muscles.

There were no significant differences in the integrated and mean EMG values or in the onset and offset times of the five muscles for which statistical tests were performed. The tibialis posterior results, while only the product of four subjects, did not indicate trends that might support the hypothesis that this muscle would systematically alter its activation in response to the experimental manipulation. While these results should certainly be viewed with caution, no clear differences between shoe conditions emerged for this muscle. The EMG data for all muscles generally did not indicate any systematic responses, which could suggest that acute changes in foot frontal plane motion may not require an active response by the neuromuscular system.

The results of this study indicate that musculotendenous injuries may not be directly related to increased activity in muscles controlling pronation of the foot. This study cannot conclude, however, that perturbation of the foot does not alter the forces in the muscles and impose excessive stress at the attachment sites. It has been shown that the passive properties of muscle, such as the force–length and force–velocity characteristics, modulate muscle force (van Soest and Bobbert, 1993) and that the passive properties can regulate external forces (Herzog et al., 2000; Wright et al., 1998). At impact, increased pronation and pronation velocity likely increase the rate of stretch in the invertor muscles, thereby increasing force in the muscle by increasing the eccentric velocity at a given activation level. Also, the hard and soft tissue constraints within the joint may contribute to a resistive moment if the subtalar joint approaches the end of its range of motion (Chen et al., 1988) while wearing the valgus shoe. Therefore, while this study indicates that perturbation of the foot may not elicit adaptations in activation patterns, it is possible that passive properties may lead to greater tissue loads about the ankle. Musculoskeletal modeling may lend insight into the load sharing and the role of passive properties of these muscles and ligaments.

A possible limitation in this study is whether an appropriate amount of time was given for subjects to adapt to each experimental condition. Little data exists on the exact time course of neuromuscular adaptations although there is evidence that the nervous system can quickly adapt to changes in the environment (Belanger and Patla, 1984; Duysens et al., 1992; Ferris et al., 1999; White et al., 2002). While it is possible that activation patterns may alter over time in response to the varus and valgus shoes, the literature supports the assumption that gait adaptations take place relatively quickly.

Another possible limitation may be the shoe design utilized in this study. Although subjects reported being able to run normally in these shoes and the kinematics were similar to Reinschmidt et al. (1997), there may have been changes to muscle activity attributed to the heel-less shoe. Based on the subjective responses and kinematics, the differences in muscular activity as compared to running in a typical shoe were assumed to be minimal.

5. Summary

Running in the varus and valgus shoes altered the joint kinematics and kinetics during running. Greater negative work was performed in the frontal plane while running in the valgus shoes indicating that greater energy was absorbed in the structures that would contribute to an inversion moment. The tibialis posterior, soleus, and gastrocnemius are likely to primarily absorb this energy. There was not, however, a significant change in the muscle activation levels of any of the muscles recorded. These results suggest that passive properties may primarily account for the increased energy absorption when there is greater eversion of the foot. This study was not, however, able to identify which muscle(s) accounted for the loading changes and therefore would be more likely to be injured based upon EMG activity.

Acknowledgements

The authors would like to acknowledge New Balance, Inc for their contribution of the shoes used in this study, and to the International Society of Biomechanics (Matching Dissertation Grant) and Life Fitness Academy (Michael L. Pollock Memorial Grant) for their financial support of this project.

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