Forefoot Midsole Stiffness Affects Forefoot and Rearfoot Kinematics During the Stance Phase of Gait

Renan A. Resende, MSc*
Sérgio T. Fonseca, ScD*
Paula L. Silva, PhD*
Antônio E. Pertence, PhD†
Renata N. Kirkwood, PhD*

Background: The forefoot midsole stiffness of the shoe may affect the kinematics of the foot segments. We evaluated the effects of two different levels of forefoot midsole stiffness on the angular displacement of the forefoot and rearfoot in the three planes of motion during the stance phase of gait.

Methods: Thirty-six participants walked on a 10-m walkway at their self-selected speed wearing shoes having either low or high forefoot midsole stiffness. Three-dimensional kinematic data of the foot segments were obtained during the stance phase of gait using an eight-camera motion analysis system synchronized with a force platform. The dependent variables were forefoot and rearfoot total range of motion and maximum and minimum angle values in the sagittal, frontal, and transverse planes of motion.

Results: Reduced forefoot midsole stiffness produced significantly greater forefoot total range of motion in the sagittal plane (1.59°). The low-stiffness condition also increased the magnitude of the forefoot dorsiflexion angles (4.14°). Furthermore, the low-stiffness condition increased the magnitude of the rearfoot inversion (1.21°) and adduction (11.38°) angles and reduced the rearfoot abduction angle (12.1°).

Conclusions: It is likely that reduced stiffness of the forefoot midsole stretched the plantar fascia, increasing rearfoot stability during the stance phase of gait. Increased muscular contraction may also explain increases in rearfoot stability. Therefore, the integrity of the plantar fascia and ankle muscles' force and resistance should be considered when choosing a shoe with reduced or increased forefoot midsole stiffness for walking. (J Am Podiatr Med Assoc 104(2): 183-190, 2014)

Walking is an activity commonly chosen by individuals as a form of physical exercise. In this context, considerable efforts have been directed toward developing footwear that provides adequate stability to the foot during running, but little is known about walking. However, although the loads imposed on the musculoskeletal tissues during walking are smaller than those produced during running, the occurrence of overuse injuries typically involves the incapability of individuals to deal with small amounts of stresses applied repeated and over long periods to the musculoskeletal system. Thus, the influence of footwear characteristics on foot motion, specifically, the influence of midsole stiffness, may contribute to the development of overuse injuries during walking.

A forefoot midsole possessing low stiffness is expected to be less effective as a lever arm system for the medial arch of the foot during the stance phase of gait. As a result, increases in foot motion may occur, which have been associated with the development of musculoskeletal injuries. On the other hand, a forefoot midsole with increased forefoot midsole stiffness could hamper the windlass mechanism by reducing the magnitude of dorsiflexion of the metatarsophalangeal joints, which would, consequently, hamper foot stability and lower-limb progression during late stance. Alternatively, a rigid lever system for the medial
arch of the foot could reduce subtalar motion and, consequently, decrease rotational stresses at the hip and knee. Therefore, a forefoot midsole with adequate stiffness could minimize the flow of stresses in the lower extremity associated with excessive motion in the foot complex and, speculatively, help in preventing injuries not only at the foot but also at other joints in the kinematic chain.

The rearfoot midsoles of most shoes are designed to absorb impact at the calcaneus and reduce reaction forces on the lower limb during heel strike. However, it has been speculated that the high midsole compliance required for appropriate cushioning at the rearfoot during heel strike may not provide adequate support for the forefoot during late stance. This understanding is reflected in current shoes available in the market, which are molded with anisotropic (multiple-density) midsoles. Surprisingly, most studies that investigated the influence of midsole stiffness on gait kinematics were conducted with shoes having a single-density midsole designed as hard (dense) or soft (less dense). Hence, the strategy of improving shoe design by selectively increasing the forefoot stiffness has been mainly empirical. Arguably, more systematic attempts to use this approach would require a description of the effects of a selective manipulation of forefoot midsole compression stiffness on the kinematics of the foot segments during gait. The present study was designed to provide such a description. In particular, we evaluated the effects of two different levels of forefoot midsole stiffness on angular displacement of the forefoot and rearfoot during the stance phase of gait.

Methods

Participants

Thirty-six healthy individuals (mean ± SD age, 24.3 ± 2.9 years; mass, 65.5 ± 8.3 kg; and height, 169 ± 5.2 cm) participated in the study. The inclusion criteria were the following: age between 18 and 35 years old and no history of surgery or injuries in the lower limbs or lumbar-pelvic complex in the past 6 months. The exclusion criterion was the report of any discomfort during data collection. All of the participants signed an informed consent form. This study was approved by the Universidade Federal de Minas Gerais (Belo Horizonte, Brazil) Ethics Research Committee.

Footwear

Two identical pairs of commercial shoes specially designed for walking, with a midsole composed of ethylene vinyl acetate (EVA), size 7.5, were used in the study. The tongues of the shoes were removed to allow placement of a cluster with three passive markers on the forefoot. The shoes possessed a very sturdy externally mounted heel counter, which allowed simple application of tools to cut an opening (1.9–2.0 cm) at the back part to allow the placement of a rearfoot cluster with three markers on the calcaneus. The opening was surrounded by Velcro brand hook and loop fasteners (Velcro USA Inc, Manchester, New Hampshire) to close the back of the shoe and ensure firmness during walking. This method was used to obtain a more accurate assessment of forefoot and rearfoot motion during stance because it has previously been reported that markers placed on the shoe overestimate rearfoot motion. The compression stiffness of one of the pairs of shoes was selectively manipulated. Four 6-mm-diameter holes were drilled in the corners of each square centimeter of the forefoot midsole (length, 12 cm; width, 10 cm) (low-stiffness shoes). The midsole stiffness of the other pair of shoes remained the same (high-stiffness shoes).

After the experiment was finalized, the compression stiffness of the forefoot midsole of the high- and low-stiffness shoes was determined by removing the upper part of the shoes and applying, in the circular central area of 706.5 mm², a force of 98.2 N. The force device consisted of a metal rod fixed on a structure with negligible mass. A metric dial indicator was placed above the metal rod, with an analog scale of hundredths of a millimeter fixed on a magnetic base. The deformation of the midsole after application of the force was measured by the dial indicator. Three trials were performed on each midsole, and the mean ± SD values of the trials were obtained. The compression stiffness of the forefoot midsole was determined as the ratio of the applied force by the deformation of the midsole, and it was measured in Newtons per millimeter. The low- and high-stiffness shoes had forefoot midsole mean ± SD stiffness values of 29.8 ± 0.54 and 49.1 ± 0.58 N/mm, respectively.

Instrumentation and Experimental Design

Three-dimensional kinematic data of the foot segments were obtained during the stance phase of gait with an eight-camera motion analysis system (ProReflex; Qualisys AB, Gothenburg, Sweden)
synchronized with one force platform (OR6-6 model; AMTI – Advanced Mechanical Technology Inc, Watertown, Massachusetts) placed in the center of a 10-m walkway. This system has an accuracy of 0.6 mm as specified by the manufacturer. Anatomical markers and clusters of tracking markers were used to determine the coordinate system of the segments and motion at the shank, rearfoot, and forefoot according to recommendations for minimizing soft-tissue artifacts.

A reference position, considered as the neutral position for the foot segments and tibia, was obtained during a barefoot bipedal relaxed standing trial. For this static trial performed with participants standing barefoot, anatomical markers were placed on the following locations: medial and lateral epicondyles of the femur, lateral and medial malleoli, sustentaculum tali, fibular trochlea, and base and heads of the first and fifth metatarsals (Fig. 1). The anatomical markers at the foot were used to create a laterally directed X-axis, an anteriorly directed Y-axis, and an upwardly directed Z-axis for the rearfoot and forefoot. Clusters containing tracking markers were attached to the participant’s shank, rearfoot, and forefoot. The cluster on the shank was rigid, consisting of rigid plates attached to the neoprene girdles, with three tracking markers. Forefoot and rearfoot motions were measured with a rigid cluster firmly attached to the top of the second metatarsal and another cluster attached exclusively to the calcaneus.

After the static trial, the anatomical markers were removed, the shoes were put on without interfering with the positions of the tracking clusters, and kinematic data were collected while participants walked on a 10-m walkway at their self-selected speed (Fig. 2). The force platform, located in the center of the 10-m walkway, was used to collect ground reaction forces during single stance of each trial. Participants performed six trials with the high-stiffness shoe and six trials with the low-stiffness shoe. Only trials in which force platform contact was made with the correct foot were accepted. The order of data collection using the high- or low-stiffness shoe was randomized.

A pilot test was conducted with ten participants to define the test-retest reliability of the examiner on the measurements taken during the study. The results demonstrated test-retest reliability from moderate to excellent, with a mean ± SD intraclass correlation coefficient of 0.77 ± 0.10, varying from 0.65 to 0.91.

Data Reduction

The mean values of the six trials for each participant and condition were considered for analysis. The stance phase was determined by the contact and lack of contact of the foot on the force platform. Data were collected at a frequency of 120 Hz and were filtered with a Butterworth low-pass filter with a cutoff frequency of 6 Hz. The software Visual3D (C-Motion Inc, Rockville, Maryland) was used to calculate the following kinematic waveforms: 1) rearfoot dorsiflexion-plantarflexion (X-axis), inversion-eversion (Y-axis), and adduction-abduction (Z-axis), represented by the motion of the rearfoot relative to the shank (positive values for dorsiflexion, inversion, and adduction) and 2) forefoot dorsiflexion-plantarflexion (X-axis), inversion-eversion (Y-axis), and adduction-abduction (Z-axis), represented by the motion of the forefoot relative to the rearfoot (positive values for dorsiflexion, inversion, and adduction).

All of the kinematic waveforms were time normalized and were sampled at each 1% from 0%
to 100%, yielding a total of 101 samples of each gait measure for each of the 72 waveforms.

**Data Analyses**

All of the statistical analyses were performed with SPSS for Windows (version 15.0; IBM Corp, Chicago, Illinois). Descriptive statistics, Shapiro-Wilk tests for normality, and Levene tests for homogeneity of variance were performed for all of the outcomes. To test significance, the paired Student t test or the Wilcoxon signed rank test (when the data were not normally distributed) was used to assess the effects of the different conditions for the following measures: 1) rearfoot and forefoot total range of motion in the sagittal, frontal, and transverse planes of motion and 2) maximum rearfoot and forefoot dorsiflexion, plantarflexion, inversion, eversion, adduction, and abduction during the stance phase of the gait cycle. Bonferroni correction was applied (α = 0.0167) considering that three comparisons were made for each stance phase waveform. The mean differences between the conditions are graphically presented, but not statistically analyzed, to demonstrate differences associated with the conditions (Fig. 3).

**Results**

**Gait Speed**

The low- and high-stiffness conditions had mean ± SD gait speeds of 1.31 ± 0.15 and 1.32 ± 0.15 m/sec, respectively, and this difference was not significant (P = .35).

**Range of Motion**

The total range of motion of the forefoot in the sagittal plane was significantly greater for the low-stiffness condition (Table 1), with a mean ± SD difference of 1.50° ± 4.71°. The standard error of measure (SEM) of this variable was 0.76°. This value was smaller than the mean change generated by the low-stiffness condition. There were no other statistically significant differences between shoe stiffness conditions (Table 1).

**Maximum Angle Values**

The low-stiffness condition had the following effects compared with the high-stiffness condition: 1) increased rearfoot plantarflexion, a mean ± SD of 1.4° ± 3.09° (the SEM of this variable was 2.11° and, thus, larger than the mean change generated by the low-stiffness condition); 2) increased rearfoot inversion, a mean ± SD of 1.21° ± 1.8° (the SEM of this variable was 0.99° and, thus, smaller than the mean change generated by the low-stiffness condition); 3) increased rearfoot adduction, a mean ± SD of 11.38° ± 10.8° (the SEM of this variable was 2.08° and, thus, smaller than the mean change generated by the low-stiffness condition); 4) reduced rearfoot abduction, a mean ± SD of 12.1° ± 11.37° (the SEM of this variable was 2.07° and, thus, smaller than the mean change generated by the low-stiffness condition); 5) increased forefoot dorsiflexion, a mean ± SD of 4.14° ± 14.36° (the SEM of this variable was 3.35° and, thus, smaller than the mean change generated by the low-stiffness condition); and 6) increased forefoot inversion, a mean ± SD of 0.83° ± 1.54° (the SEM of this variable was 0.99° and, thus, greater than the mean change generated by the low-stiffness condition) (Table 2).

**Discussion**

The results of this study demonstrate that reducing forefoot midsole stiffness increased total range of motion in the sagittal plane and the magnitude of the dorsiflexion angle of the forefoot during the stance phase of gait. The low-stiffness condition also increased the magnitude of the rearfoot inversion and adduction angles and reduced the magnitude of the rearfoot abduction angle during the stance phase. Such effects were greater than the SEMs calculated for each variable and were, thus, not produced by possible measurement errors related to the study procedures.

The increase in the total range of motion of dorsiflexion/plantarflexion in the sagittal plane of the forefoot may be attributed to the shoe’s reduced forefoot midsole stiffness. As the forefoot contacts the ground and the foot reaches full support during midstance, the midsole influences the amount of movement of the foot segments because it is located between the foot and the ground. In the present study, the reduced forefoot midsole stiffness offered less mechanical support to the forefoot dorsiflexion/plantarflexion movements, which contributed to the increased total range of motion. The influence of the forefoot midsole stiffness on the total range of motion is reinforced by the finding of increased forefoot dorsiflexion also demonstrated during the low-stiffness condition.

The magnitudes of the inversion and adduction angles at the rearfoot were increased and the
Figure 3. Joint angle profiles across the stance phase of gait for the rearfoot and forefoot in the sagittal (A and B), frontal (C and D), and transverse (E and F) planes of motion. Solid lines indicate the low-stiffness condition; broken lines, the high-stiffness condition.
The magnitude of the abduction angle was reduced by the low-stiffness condition. Although the present study did not measure muscle activation, it is possible that during the low-stiffness condition, the participants actively controlled the magnitude of inversion angles in response to less mechanical support offered by the forefoot midsole. Our interpretation of the present results is supported by the findings of Romkes et al., who reported that individuals walking with a less stable shoe presented increased activity of the tibialis anterior and gastrocnemius muscles in consequence of the soft midsole. According to the authors, the increase in the muscle activity occurred to enhance stability during walking with footwear that provides insufficient control of foot motion. Hence, future studies should measure muscle activation under manipulations of forefoot midsole stiffness to investigate whether the increased inversion and adduction angles are, in fact, a result of individuals actively compensating for the suboptimal shoe structure.

Alternatively, the increased magnitudes of rearfoot inversion may be explained by the influence of different levels of forefoot midsole stiffness on the windlass mechanism of the foot. The plantar fascia simulates a cable attached to the calcaneus and the metatarsophalangeal joints. In this context, dorsiflexion of the metatarsophalangeal joints dur-

### Table 1. Total Range of Motion Values and Results of Statistical Tests Designed to Identify Differences Between the Low-Stiffness (LS) and High-Stiffness (HS) Conditions

<table>
<thead>
<tr>
<th>Region and Plane</th>
<th>Total Range of Motion (Mean ± SD [°])</th>
<th>Effect Size</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>LS Condition</td>
<td>HS Condition</td>
<td></td>
</tr>
<tr>
<td>Rear foot</td>
<td></td>
<td></td>
<td></td>
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<tr>
<td>Sagittal</td>
<td>26.5 ± 3.9</td>
<td>26.1 ± 3.99</td>
<td>0.14</td>
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<tr>
<td>Frontal</td>
<td>10.92 ± 3.78</td>
<td>10.27 ± 3.30</td>
<td>0.31</td>
</tr>
<tr>
<td>Transverse</td>
<td>13.00 ± 4.63</td>
<td>11.99 ± 3.98</td>
<td>0.33</td>
</tr>
<tr>
<td>Forefoot</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sagittal</td>
<td>15.23 ± 5.98</td>
<td>13.64 ± 3.11</td>
<td>0.46</td>
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<tr>
<td>Frontal</td>
<td>7.92 ± 4.1</td>
<td>7.23 ± 4.18</td>
<td>0.31</td>
</tr>
<tr>
<td>Transverse</td>
<td>6.88 ± 7.54</td>
<td>5.91 ± 4.17</td>
<td>0.15</td>
</tr>
</tbody>
</table>

*Statistically significant difference between conditions is noted for α = 0.0167.

### Table 2. Maximum Angle Values and Results of Statistical Tests Designed to Identify Differences Between the Low-Stiffness (LS) and High-Stiffness (HS) Conditions

<table>
<thead>
<tr>
<th>Region and Outcome</th>
<th>Maximum Angle Values (Mean ± SD [°])</th>
<th>Effect Size</th>
<th>P Value</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>LS Condition</td>
<td>HS Condition</td>
<td></td>
</tr>
<tr>
<td>Rear foot</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Maximum dorsiflexion</td>
<td>26.53 ± 14.1</td>
<td>27.51 ± 13.68</td>
<td>0.18</td>
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<tr>
<td>Maximum inversion</td>
<td>8.13 ± 6.05</td>
<td>6.92 ± 5.85</td>
<td>0.56</td>
</tr>
<tr>
<td>Maximum adduction</td>
<td>−3.01 ± 15.64</td>
<td>−14.39 ± 9.37</td>
<td>0.73</td>
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<tr>
<td>Maximum plantarflexion</td>
<td>0.02 ± 13.05</td>
<td>1.42 ± 12.31</td>
<td>0.29</td>
</tr>
<tr>
<td>Maximum eversion</td>
<td>−2.79 ± 4.9</td>
<td>−3.35 ± 4.59</td>
<td>0.28</td>
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<tr>
<td>Maximum abduction</td>
<td>−14.29 ± 15.67</td>
<td>−26.39 ± 9.2</td>
<td>0.73</td>
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<tr>
<td>Forefoot</td>
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<tr>
<td>Maximum dorsiflexion</td>
<td>−8.12 ± 22.56</td>
<td>−12.26 ± 17.66</td>
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<tr>
<td>Maximum inversion</td>
<td>13.0 ± 6.01</td>
<td>12.2 ± 5.96</td>
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<tr>
<td>Maximum adduction</td>
<td>7.73 ± 9.81</td>
<td>9.33 ± 10.75</td>
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<tr>
<td>Maximum plantarflexion</td>
<td>−25.16 ± 18.8</td>
<td>−25.9 ± 16.99</td>
<td>0.16</td>
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<tr>
<td>Maximum eversion</td>
<td>−4.17 ± 7.51</td>
<td>−4.97 ± 7.96</td>
<td>0.36</td>
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<tr>
<td>Maximum abduction</td>
<td>0.85 ± 11.47</td>
<td>3.42 ± 9.48</td>
<td>0.35</td>
</tr>
</tbody>
</table>

*Statistically significant differences between conditions are noted for α = 0.0167.
ing late stance winds the plantar fascia around the head of the first metatarsal, which shortens the distance between the calcaneus and the metatarsals, elevating the medial longitudinal arch and possibly increasing supination of the segments of the foot. Thus, it is possible that the reduced forefoot midsole stiffness has increased metatarsophalangeal joint dorsiflexion, stretching the plantar fascia anteriorly and, consequently, increasing the rearfoot inversion angles. Thus, although reduced forefoot midsole stiffness seems to offer less mechanical support to the forefoot, it seems to enhance rearfoot supination during late stance. Nonetheless, the potential link between reduced forefoot midsole stiffness and the increased stress to the plantar fascia and rearfoot stability remains speculative at this point and in need of further scientific scrutiny.

The observed kinematic effects of forefoot midsole stiffness must be interpreted with caution in light of the possible inaccuracy of quantitative measures of the lower extremity obtained with external markers attributable to soft-tissue artifacts. In addition, the kinematic waveforms of the forefoot and rearfoot may not represent the exact angular values because the dynamic trials were analyzed based on the static trial with the participants barefoot. However, the differences found between conditions were not influenced by that procedure because both conditions were analyzed using the same static trial, and cluster configuration and placements were performed according to recent recommendations.

The results of the present study demonstrate that different levels of forefoot midsole stiffness affect the angular displacement of the forefoot in the sagittal plane and of the rearfoot in the frontal and transverse planes during the stance phase. One may argue that the loads imposed on the musculoskeletal tissues during walking are smaller than those produced during running and that the isolated kinematic effects observed in the present study are not likely to have clinical significance. However, the occurrence of overuse injuries typically involves the inability of individuals to deal with small amounts of stresses applied repeatedly or over long periods to the musculoskeletal system. Therefore, the significance of the kinematic changes observed in the present study will most likely depend on the frequency and duration of physical activities performed with the shoes and the individual’s resources to respond to the demands of such activities.

Conclusions

The present results demonstrate that the reduced forefoot midsole stiffness increased the total range of motion in the sagittal plane and the magnitude of dorsiflexion at the forefoot and increased the magnitude of inversion and adduction and reduced the magnitude of abduction at the rearfoot during the stance phase of gait. The reduced stiffness of the forefoot midsole seems to offer less mechanical resistance to the movements of the forefoot in the sagittal plane, which might explain the increased total range of motion. On the other hand, the lower forefoot midsole stiffness seems to enhance rearfoot supination during stance, which can, consequently, enhance foot stability. Therefore, the present study provides initial evidence that different levels of forefoot midsole stiffness change forefoot and rearfoot kinematics, and the choice of the appropriate shoe for walking will depend on the interaction of the findings provided by the present study and the specific individual’s factors, such as the integrity of the plantar fascia, forefoot and rearfoot structural alignment, and arch height.

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Conflict of Interest: None reported.

References


