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Increased Medial Longitudinal Arch Mobility, Lower Extremity Kinematics, and Ground Reaction Forces in High-Arched Runners

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Context: Runners with high medial longitudinal arch structure demonstrate unique kinematics and kinetics that may lead to running injuries. The mobility of the midfoot as measured by the change in arch height is also suspected to play a role in lower extremity function during running. The effect of arch mobility in high-arched runners is an important factor in prescribing footwear, training, and rehabilitating the running athlete after injury.

Objective: To examine the effect of medial longitudinal arch mobility on running kinematics, ground reaction forces, and loading rates in high-arched runners.

Design: Cross-sectional study.

Setting: Human movement research laboratory.

Patients or Other Participants: A total of 104 runners were screened for arch height. Runners were then identified as having high arches if the arch height index was greater than 0.5 SD above the mean. Of the runners with high arches, 11 rigid runners with the lowest arch mobility (R) were compared with 8 mobile runners with the highest arch mobility (M). Arch mobility was determined by calculating the left arch height index in all runners.

Intervention(s): Three-dimensional motion analysis of running over ground.

Main Outcome Measure(s): Rearfoot and tibial angular excursions, eversion-to-tibial internal-rotation ratio, vertical ground reaction forces, and the associated loading rates.

Results: Runners with mobile arches exhibited decreased tibial internal-rotation excursion (mobile: 5.6° ± 2.3° versus rigid: 8.0° ± 3.0°), greater eversion-to-tibial internal-rotation ratio (mobile: 2.1 ± 0.8 versus rigid: 1.5 ± 0.5), decreased second peak vertical ground reaction force values (mobile: 2.3 ± 0.2 × body weight versus rigid: 2.4 ± 0.1 × body weight), and decreased vertical loading rate values (mobile: 55.7 ± 14.1 × body weight/s versus rigid: 65.9 ± 11.4 × body weight/s).

Conclusions: Based on the results of this study, it appears that runners with high arch structure but differing arch mobility exhibited differences in select lower extremity movement patterns and forces. Future authors should investigate the impact of arch mobility on running-related injuries.

Key Words: foot, running injuries, joint coupling

Key Points

- Foot arch mobility played an important role in lower extremity biomechanics during running.
- High-arched rigid runners demonstrated increased vertical loading rates, which are often associated with lower extremity injuries in runners.
- Joint coupling or coordination is influenced by arch mobility. Changes in intersegment coordination can result in the need for compensation and potential overuse at nearby joints.

Differences in lower extremity kinematics and kinetics between runners with high arches (HAs) and low arches (LAs) have received much attention because of the relationship between arch structure and elevated injury risk.[1–4] The prevalence of individuals with HAs in the population has been reported to be as large as 20%.[5] Runners with HA structure report more heel pain, stress fractures, and other structure-specific injuries.[6] These injuries may be the result of runners with HAs moving with a stiffer lower extremity and higher loading rates during running.[3,4,7] Although the arch structure has received much attention, there has been little focus on the mobility of the medial longitudinal arch. The relationship between arch structure and mobility is a critical component in understanding how structure, biomechanics, and injury are related in runners with the HA foot type.

One specific kinematic factor often associated with foot type is eversion. Eversion is the primary motion at the foot that is responsible for shock attenuation during running.[8] Eversion is coupled with internal rotation of the tibia and pronation of the foot.[2] These coupled motions have been defined as the eversion-to-tibial internal-rotation (EV:TIR) ratio.[2] This relationship is based solely on the structure of the arch and the resting position of the subtalar joint; mobility has not been considered in this relationship.[2] The EV:TIR relationship has previously been compared between runners with HAs and LAs, demonstrating that HA runners have decreased total eversion excursions and a

Table 1. Participant Demographics

	Rigid	Mobile	*P* value
No. of participants	11	8	
Sex	8 men, 3 women	6 men, 2 women	
Height, m	1.77 (0.07)	1.72 (0.07)	.16
Body mass, kg	77.2 (15.1)	73.4 (10.3)	.55
Age, y	49.7 (8.5)	53.1 (11.8)	.47
Distance run, km/wk	30.9 (12.5)	33.8 (23.2)	.72
Training pace, min/mi	9:13 (1:20)	10:05 (2:37)	.36
Arch ratio	0.362 (0.020)	0.361 (0.010)	.86
Arch mobility	0.023 (0.003)	0.042 (0.008)	.00[a]

[a] Significant value.

concurrent increase in tibial internal rotation.[2,9] More specifically, HA individuals demonstrate a lower EV:TIR compared with LA runners.[9,10] Further, a lower EV:TIR has been reported to increase the stress on the knee in HA runners secondary to the relative increase in transverse-plane motion of the tibia.[9] These findings suggest that the orientation of the arch has an effect on the coupling of the lower extremity further up the kinematic chain and may be a factor associated with knee injury in HA runners.

Although arch structure has a clear relationship to lower extremity movement, its effect on lower extremity forces is less conclusive.[2,3,9,11–13] Differences in the magnitude of the vertical impact peak have not been associated with arch height.[3] This may be because at initial contact, the forefoot is not in contact with the ground, which means the shape or mobility of the arch cannot affect the absorption or transfer of forces. Loading occurs at initial contact through the calcaneus and the rear one-third of the foot. At this point, the ground reaction force is transmitted through the heel pad, calcaneus, and talus before moving into the lower leg.[10] The second vertical ground reaction force (VGRF) peak may better represent the effect of arch height and mobility on shock attenuation because the arch is likely to deform under weight. Specifically, a more mobile foot may result in decreased magnitude of the second VGRF peak.

High arches are often considered to be rigid because of the relatively vertical rearfoot-to-midfoot axis relationship.[1,14] However, arch mobility has been reported to be independent of the static height of the medial longitudinal arch.[3,11] Decreased mobility of the arch in HA runners has been suggested to relate to an increased need for compliance at other lower extremity joints, such as the knee.[4] High-arched individuals demonstrate a more supinated position of the foot, which results in decreased pronation throughout the stance phase.[2,9,14] However, a mobile arch or midfoot that compresses under force may result in both midfoot and rearfoot pronation, even in an HA foot. Because of this change in rearfoot pronation and the coupling present in the rearfoot and lower leg, it is likely that an individual with a mobile HA will demonstrate different lower extremity kinematics and kinetics.

The purpose of our study was to compare running kinematics and VGRF characteristics between 2 groups of HA runners: those with rigid arches and those with mobile arches. We hypothesized that the EV:TIR ratio would be different between groups, as we suggested arch mobility has a potential effect on rearfoot motion and subsequent coupling. We also hypothesized that the initial loading rate between the groups would be similar because the force at this point is through the rear of the foot, but the second loading rate and maximum VGRF would be decreased in the mobile group.

METHODS

Participants were recruited from the university, surrounding communities, and local running clubs. The study included a total of 104 runners, ranging in age from 29 to 81 years at the time of data collection. Each participant had a lower extremity musculoskeletal examination by a physical therapist with 12 years' experience specific to the running population. Each volunteer gave written informed consent for participation in the study, which was approved by the University and Medical Center Institutional Review Board. Participants ran a minimum of 6 miles (9.7 km) per week for at least 6 months before the study. Participants were excluded from this study if they had any cardiovascular or neurologic compromise, current lower extremity musculoskeletal injury, joint replacement, or joint fusion. Of the 104 participants in the study, 30 were classified as having HAs (Table 1). This determination was made by calculating the left *arch ratio*, which was defined by the height of the dorsum of the foot in weight bearing at 50% of the foot length divided by the individual's truncated foot length.[15] *Truncated foot length* was defined as the length of the foot from the most posterior portion of the calcaneus to the center of the medial joint space of the first metatarsal-phalangeal joint.[15] An arch ratio of at least 0.342 was needed for inclusion as an HA participant. This value was $\geq$0.5 SD of the mean dorsum height-to-truncated foot length ratio measurement of 0.328 based on the sample of 104 feet. These measurements were made directly on the foot in a barefoot condition in both static standing and static non–weight bearing.

For those included in the HA group, we calculated left arch mobility using dorsum height in bilateral weight bearing subtracted from dorsum height in non–weight bearing divided by truncated foot length. The *rigid group* was defined as participants who were $\geq$0.5 SD below the mean mobility of the HA participants, and the *mobile group* had an arch mobility $\geq$0.5 SD above the mean mobility.[15] Eleven participants were placed in the rigid HA group and 8 participants with the most mobility in their left arch were placed in the mobile HA group. All runners included in the current study ran with a rearfoot strike pattern naturally.

We conducted 3-dimensional running analysis on eligible runners. Participants wore the same brand and model of neutral running shoes during the running analysis (New Balance 825; New Balance, Brighton, MA). A standing calibration trial was collected during which retroreflective markers were placed bilaterally on the segments of the rearfoot on the shoe, shank, thigh, and pelvis (Figure 1). Placement of markers on clustered shells at the distal one-third of the shank and thigh segments was used.[16] Static markers (placed at bilateral greater trochanters, medial and lateral knees, medial and lateral malleoli, and medial and lateral forefeet) were positioned during the standing calibration and removed before the dynamic portion of the data collection. The z-axis was oriented from the distal segment end to the proximal segment end (inferior to superior). The y-axis was oriented in the segment from posterior to anterior. Finally, the x-axis orientation was determined using the right-hand rule and was oriented from

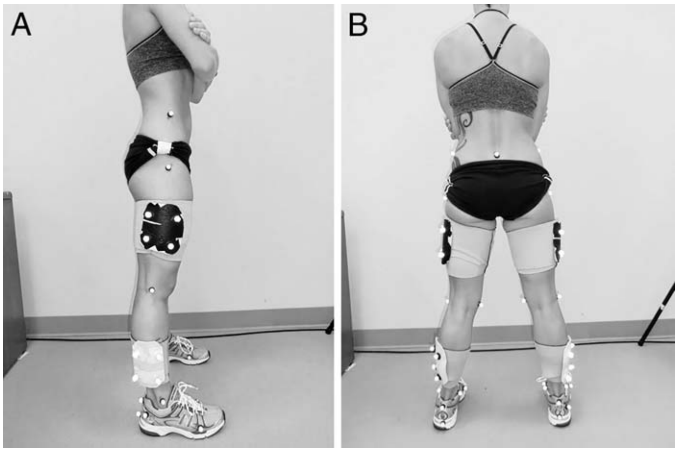

Figure 1. Retroreflective marker placement. A, Sagittal view. B, Frontal view. Markers used for the variables evaluated in the current study were markers located at the knee and distal to the knee.

medial to lateral. Participants were allowed to run along the runway as many times as necessary to feel comfortable with the markers and the laboratory environment. Once the participant was comfortable with his or her running gait and running at a consistent velocity (3.35 m/s), data collection was started. Kinematic data were collected at 240 Hz with an 8-camera motion-analysis system (Qualisys AB, Gothenburg, Sweden). Qualisys software was used to reconstruct the coordinates for each marker. The participants were asked to run along a 20-m runway at a speed of 3.35 m/s (±5%). Running speed was measured using photocells 6 m apart. Two force platforms (Advanced Mechanical Technology, Inc, Watertown, MA) were mounted in the floor of the runway to record VGRF at a sampling frequency of 1200 Hz. Participants were instructed to run with their normal running gait. A total of 10 successful trials were collected for each participant. Most participants were able to achieve 10 successful trials within a total of 20 passes over the force platforms. A trial was considered acceptable if the participant ran with a natural gait over the force plates within the given velocity range and his or her entire left foot hit one of the force plates.

Pelvis, thigh, shank, and foot segments were created using Visual 3-D software (C-motion, Inc, Germantown, MD). The 3-dimensional marker trajectory data were filtered using a second-order recursive Butterworth filter with a 12-Hz cutoff frequency. Ground reaction force data were filtered at 50 Hz with a second-order recursive Butterworth filter. Data were then normalized to the stance phase of the gait cycle. Kinematic data were resolved about a joint coordinate system,[17] with flexion–extension about the x-axis occurring in the proximal segment, internal–external rotation occurring about the z-axis in the distal segment, and abduction–adduction occurring about the intermediate vector, which was the common perpendicular of the other 2 axes.

The independent variables of this study were the individual groups of rigid and mobile HA runners. The dependent variables were the eversion excursion, tibial-rotation excursion, EV:TIR ratio, first peak VGRF, second peak VGRF, and the initial and second loading rates. We calculated the EV:TIR ratio by dividing the eversion excursion by the tibial internal-rotation excursion. Initial and second loading rates were calculated from the VGRF data. The initial loading rate was calculated by dividing the value of the initial peak VGRF by the time to reach the peak. The second loading rate was calculated by subtracting the maximum of the second peak from the minimum of the second peak and dividing it by the time between peaks (Figure 2). A Student *t* test ($P \leq .05$) was used to compare values for dependent variables between groups. Further, 95% confidence intervals and effect sizes were calculated for dependent variables.

RESULTS

The HA runners' arch ratios fell ≥1.0 SD of the collected mean of 104 runners. No difference in arch height index existed between the mobile and rigid groups (Table 1). The groups exhibited different amounts of arch mobility (Table 1).

Comparisons between variables of interest are presented in Table 2. There was a significant difference in EV:TIR between the rigid and mobile groups with a slight overlap in the confidence intervals and a large effect size. Rigid HA runners had more tibial internal-rotation excursion than mobile HA runners. The comparison between groups on tibial internal rotation showed no overlap of the confidence intervals and a large effect size. Rearfoot eversion excursion did not differ between groups (Figure 2).

Second peak VGRF was greater in the rigid HA group compared with the mobile group (Figure 3). The confidence intervals showed a slight overlap and the effect size was moderate to large. Vertical loading rate to the initial peak was found to be greater in the rigid HA group, demonstrating a slight overlap in the confidence intervals and a large effect size. No difference was found in the vertical loading rate for the second peak (Figure 4).

DISCUSSION

The purpose of our study was to compare select lower extremity kinematic and VGRF characteristics between 2 groups of HA runners: those with rigid arches and those with mobile arches. The results of our study suggest that rigid HA runners exhibited a higher initial loading rate, greater peak VGRF, and decreased EV:TIR compared with mobile HA runners. Based on these results, it appears that assessment of static arch mobility, in conjunction with other lower extremity and foot measures, may be an important factor in understanding overall foot and lower extremity function during running. It is also important to note that all runners in the current study ran with a rearfoot-strike pattern. Therefore, these results should be interpreted only as they relate to this population. Midfoot- and forefoot-strike runners have differing biomechanics and therefore may have a different response to foot structure and foot mobility.

A difference was noted between the EV:TIR of the groups in our study. It has been suggested that the angle of inclination of the axis of the subtalar joint is influenced by static arch structure.[18] Therefore, if static structure alone influenced coupling up the kinematic chain, there should have been no difference in EV:TIR, as both groups had the same static arch structure. The current findings demonstrate

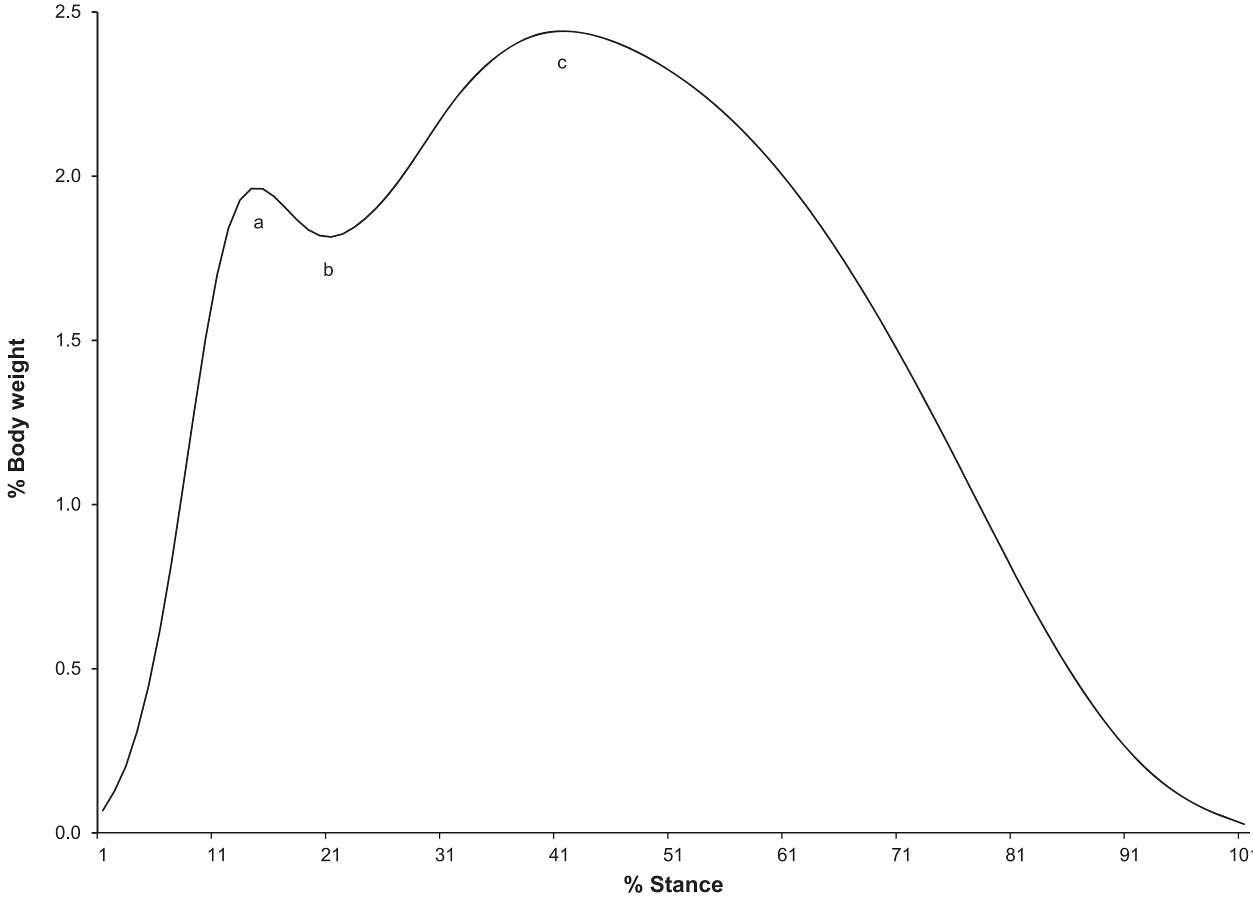

Figure 2. Values used for calculating the primary loading rates: [a] initial peak vertical ground reaction force, [b] minimum of the second peak vertical ground reaction force, and [c] second peak vertical ground reaction force.

an effect of arch mobility on rearfoot coupling independent of static arch structure.

We report a higher value for HA individuals, independent of arch mobility, in total eversion excursion and a lower value for TIR excursion when compared with previous authors.[9,10,17] Furthermore, EV:TIR values in the current study are greater than those previously reported for HA runners.[9,10] However, previous investigators examining the rearfoot and shank motion during running reported that the EV:TIR could vary between 1.3 and 2.2,[19] which is consistent with the values we observed. Although the confidence intervals overlapped slightly, the effect size for this characteristic was large (0.93), suggesting a meaningful clinical effect as well.

Factors that have previously been observed to affect EV:TIR are footwear, arch type, and age. Although

Table 2. Kinematic and Ground Reaction Force Results

Variable	Group, Mean ± SD (95% Confidence Interval) Rigid	Mobile	*P* Value	Effect Size
Eversion excursion, °	11.2 ± 3.4 (9.2, 13.2)	10.1 ± 2.5 (8.4, 11.8)	.22	0.38
Tibial excursion, °	8.0 ± 3.0 (6.2, 9.8)	5.6 ± 2.3 (4.0, 7.2)	.04[a]	0.93
EV:TIR ratio, °/°	1.5 ± 0.5 (1.2, 1.8)	2.1 ± 0.8 (1.5, 2.7)	.05[a]	0.99
First peak, BW	2.0 ± 0.2 (1.88, 2.12)	1.9 ± 0.4 (1.62, 2.18)	.16	0.35
Second peak, BW	2.4 ± 0.1 (2.34, 2.46)	2.3 ± 0.2 (2.16, 2.44)	.05[a]	0.71
Loading rate 1, BW/s	65.9 ± 11.4 (59.2, 72.6)	55.7 ± 14.1 (45.9, 65.5)	.05[a]	0.86
Loading rate 2, BW/s	12.7 ± 3.1 (10.9, 14.5)	11.5 ± 2.9 (9.5, 13.5)	.18	0.42

Abbreviations: BW, body weight; CI, confidence interval; EV:TIR, eversion-to-tibial internal rotation.
[a] Significant value.

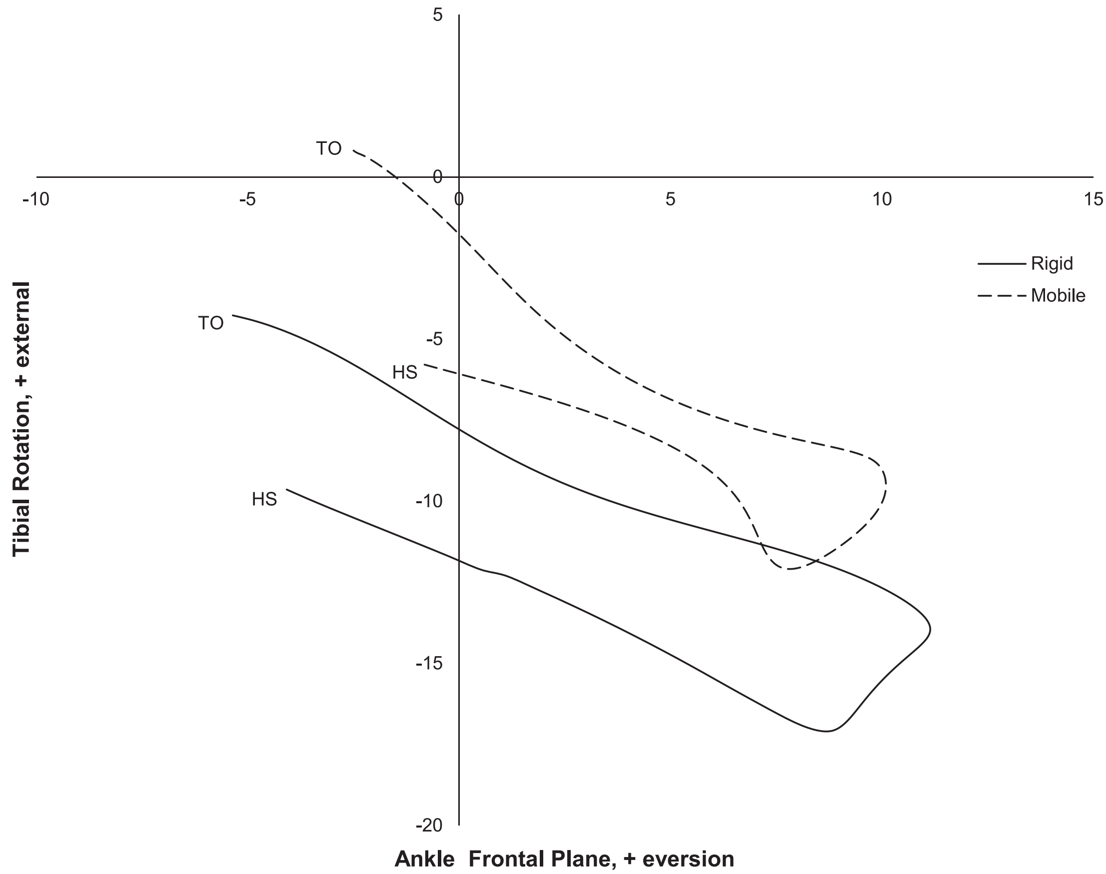

Figure 3. Mean eversion-to-tibial internal-rotation ratios during stance. Abbreviations: HS, heel strike; TO, toe-off.

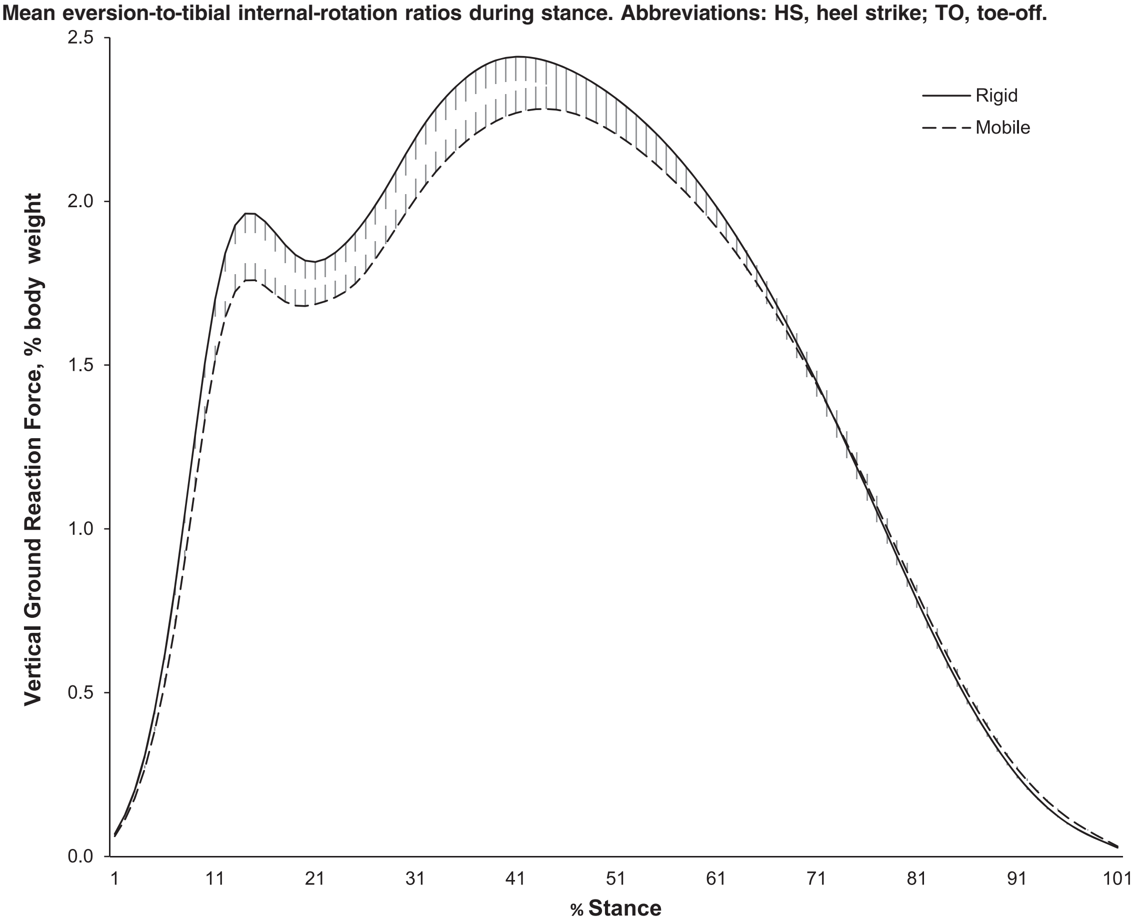

Figure 4. Average vertical ground reaction force (% body weight) during stance.

footwear and arch type were controlled in the current study, the average age of participants was noticeably older than in the previous studies.[2,9,10] The 20-year difference in age may explain the higher EV:TIR in the current study, because as runners age, they typically decrease their range of motion, especially in secondary planes of motion.[20] In a previous study, older runners demonstrated a reduction of 3° of tibial internal–external range of motion during running, which could explain the higher EV:TIR we saw.[21]

During running, arch mobility had a significant effect on tibial-rotation excursion, along with the relative coupling of these values. Specifically, the rigid HA runners exhibited tighter coupling of rearfoot eversion to tibial internal rotation during the first half of the stance phase, as evidenced by the linear slope of the EV:TIR ratio (Figure 3). There was a distinct interruption in the linearity of the slope in the mobile group. This point likely indicates a collapse in the arch consistent with midfoot pronation. This motion causes an increase in the proportion of TIR compared with ankle eversion by transferring rotation through the navicular to the talus and tibia. It is likely that the rigid group did not experience the same amount of collapse in the arch at midstance, which may explain the need for increased eversion excursion to absorb the forces. Arch collapse in the mobile group limited the amount of rearfoot eversion necessary. Further analysis of multisegment foot models in HA individuals will further clarify this relationship.

When we compared VGRF between the groups, the rigid HA group demonstrated higher values than the mobile group at both peaks during the stance phase, although only significantly so at the second peak. Previous authors[2] reported HA runners with a VGRF of 1.97 × body weight, whereas LA runners demonstrated lower values of 1.72 × body weight at the initial peak. Our findings are consistent with previous work showing a mean of 1.95 × body weight for the initial peak. The values reported for the VGRF attained at the second peak in our study were slightly less than those previously found in HA runners.[2] At the speed we used, this difference represents a 4% increase between groups. Although the absolute differences were small, the effect size was large, suggesting that even small effects in VGRF factors could have a clinically meaningful effect. This large effect is likely the result of the small variance typically present in this characteristic.

The initial loading rate was higher for the rigid group than the mobile group. The values appear to be consistent with previously reported data[9] on loading rates for HA runners compared with LA runners. This result is strongly influenced by the differences in the initial peak VGRF between the groups of HA runners. Previous authors[22] showed that runners maintain similar center-of-mass movements on surfaces with different stiffness values by adjusting leg stiffness to accommodate the surface condition. Additionally, when humans hop on various surfaces, they can adjust their leg biomechanics to overcome the effects of surface dampening, allowing them to maintain the same combined leg-surface stiffness.[23] The current findings suggest that runners who are expecting to land on a softer surface (a mobile foot) likely make an adjustment before the foot is in contact with the ground to soften the landing. As the heel contacts the ground, increased mobility of the midfoot may result in compensatory rigidity and potentially decreased shock attenuation through the rearfoot. Subsequently, the loss of attenuation results in necessary adjustments up the chain and therefore a decrease in loading rate. This decrease in loading rate is consistent with the decrease in loading rate seen in LA individuals,[4] further suggesting that arch mobility is an important factor, in conjunction with arch shape, in determining force absorption during gait.

The second loading rate during the stance phase has not been adequately defined or previously studied. Arch flattening has been classified in previous work but not related to active peak or second loading rate.[2] As this loading occurs between the foot-flat and heel-off (when the arch is flattening toward the ground) positions, it is of particular interest because the arch may act to absorb a portion of the VGRF. This value was reduced in the mobile participants, though not significantly. Because the knee is the major shock attenuator at this point in the stance phase, changes at the knee may have a greater influence on second loading rate. Evaluation of arch collapse during running may be a more valuable measurement. However, there are significant methodologic limitations in attempting to collect these data, such as marker placement and the need to run without shoes. Further investigation of second loading rate and midfoot mobility is necessary to further identify its importance in running-related injuries.

Previous authors have identified runners specifically by the height of their arch and not by the amount of mobility in the medial longitudinal arch. We evaluated an important clinical question regarding the effect of arch mobility and biomechanics in HA runners. Although our results are encouraging, it is important to note that the number of participants in the current study was relatively small and they were older than participants in other studies. Replication of the current study with larger numbers of participants in various age groups would further confirm these conclusions. More research into the effects of the mobility of the arch, particularly through multisegment foot modeling, may show that some HA runners need additional stability in their shoes, as opposed to the cushioned trainer that is typically prescribed for these runners. Additional factors that may interact with foot type and mobility include strike pattern and foot orthoses. Finally, understanding the intersegmental foot motion and its relationship to rearfoot and lower leg motion may also aid in shoe prescription and treatment of injuries in the HA population.

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