Contributions to Leg Stiffness in High-Compared with Low-Arched Athletes

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ABSTRACT

POWELL, D. W., M. R. PAQUETTE, and D. S. B. WILLIAMS III. Contributions to Leg Stiffness in High-Compared with Low-Arched Athletes. Med. Sci. Sports Exerc., Vol. 49, No. 8, pp. 1662–1667, 2017. Purpose: High-arched (HA) athletes exhibit greater lower extremity stiffness during functional tasks than low-arched (LA) athletes. The contributions of skeletal and muscular structures to stiffness may underlie the distinct injury patterns observed in these athletes. The purpose of this study was to compare skeletal and muscular contributions to leg stiffness in HA and LA athletes during running and landing tasks. Methods: Ten HA and 10 LA female athletes performed five overground running trials at a self-selected pace and five step off bilateral landing trials from a height of 30 cm. Three-dimensional kinematics and kinetics were collected using a motion capture system and a force platform. Leg stiffness and its skeletal and muscular contributions were calculated. Independent t-tests were used to compare variable means between arch type groups and Cohen's d were computed to assess effect sizes of mean differences. Results: In running, HA athletes had greater leg stiffness (P = 0.010, d = 1.03) and skeletal stiffness (P = 0.016, d = 0.81), although there are no differences in muscular stiffness (P = 0.134). During landing, HA had greater leg stiffness (P = 0.015, d = 1.06) and skeletal stiffness (P < 0.001, d = 1.84), whereas LA athletes had greater muscular stiffness (P = 0.025, d = 0.96). Conclusions: These findings demonstrate that HA athletes place a greater reliance on skeletal structures for load attenuation during running and landing, whereas LA athletes rely more greatly on muscle contributions during landing only. These findings may provide insight into the distinct injury patterns observed in HA and LA athletes. Key Words: RUNNING, BAREFOOT, KINETICS, INJURY, ANKLE, ARCH, FOOT

A berrant foot structure is associated with greater incidence and prevalence of lower extremity injury in athletes (8,19). Individuals with high-arched (HA) feet experience greater rates of bony injury to the lower extremity such as tibial stress syndrome, tibial stress fractures, and lateral metatarsal stress fractures (8,19). Conversely, low-arched (LA) athletes have a propensity to experience soft tissue injuries of the lower extremity, including Achilles tendinopathy and patellofemoral pain (8,19).

These distinct injury patterns have been attributed to altered force transmission through the lower extremity as a result of the structural and morphological characteristics of the foot (10–13,16,19). Zifchock et al. (21) demonstrated that HA feet are generally rigid, whereas LA feet tend to be flexible. The functional differences in foot characteristics between HA and LA feet and the translation of foot function to lower extremity biomechanics have been further demonstrated during dynamic tasks such as running and landing. Smaller ankle joint work is found in HA compared with LA runners during the stance phase of running and this ankle work difference between arch types is greater during propulsion (i.e., positive work) (13). These findings may be attributed to the fact that HA runners have greater ground reaction force (GRF) lever arms to the ankle joint compared with LA athletes (13). HA athletes also exhibit smaller negative knee and hip joint work during a bilateral landing task, indicating a greater relative contribution of the ankle joint to load attenuation during landing (12). LA athletes have shorter GRF lever arms to the ankle joint (13), and greater muscle forces may be required to generate similar plantarflexor moments in LA compared with HA athletes. Thus, these greater plantarflexor muscle forces may contribute to the greater prevalence of soft tissue injuries in LA athletes (13,16). A potential contributing factor to tissue-specific injury in HA and LA athletes may be lower limb stiffness.

Stiffness is a composite measure of a system’s response to an applied load. Stiffness has been described as the relationship between the deformation of a body in response to an applied force and has been investigated from the level of a single muscle fiber to the modeling of the entire body as a spring (2). High stiffness values indicate that a given load results in small deformations, whereas low stiffness values
indicate that a given load results in large deformations. HA athletes have been shown to have greater leg stiffness and ankle and knee joint stiffness than LA athletes during a running task (13,16). These findings are similar to assessments of vertical stiffness during a landing task (12). It has been previously suggested that greater lower limb stiffness observed in HA athletes results in greater loading to the skeletal system during load attenuation and may underlie the distinct injury patterns experienced by HA athletes. Two studies have sought to differentiate muscular and skeletal contributions to leg stiffness during a downward stepping task and level walking (4,15). DeVita and Hortobagyi (4) modeled the age-related differences in skeletal versus muscular contributions to leg stiffness during a downward stepping trial, whereas Wang et al. (15) quantified skeletal and muscular contributions to gait in children with cerebral palsy. However, no previous investigation has attempted to quantify skeletal and muscular contributions to load attenuation or stiffness in HA compared with LA athletes. Identifying differences in skeletal and muscular contributions to leg stiffness may provide a foundation for clinical interventions to reduce tissue-specific injury rates in these athletes. Therefore, the purpose of this study was to quantify the muscular and skeletal contributions to leg stiffness during running and step off landing tasks. We hypothesized that HA athletes would exhibit greater leg stiffness values than LA athletes during running and landing. Further, it was hypothesized that HA athletes would exhibit greater skeletal contributions and smaller muscular contributions to leg stiffness than LA athletes during running and landing tasks.

MATERIALS AND METHODS

Participants. One hundred and fifty-seven female athletes participating in university club sports were screened for inclusion in this study. Twenty female recreational athletes (10 HA and 10 LA) were identified and recruited to participate in this study (Table 1). All potential participants were active in university club sports at the time of testing and participated in training or competition at a moderate or high level of intensity at least 3 d wk⁻¹ for a duration of at least 90 min per session. All participants were currently free of injury and for the 6 months preceding testing. Only participants who used a non–rearfoot strike (non-RFS) during running were included in the study. We included only a single foot strike pattern because RFS and non-RFS patterns yield different lower limb joint stiffness during running (6,9). Shod running strike pattern was confirmed using foot contact angle (1) before experimental testing. To emphasize the role of the ankle plantarflexors (17), participants performed experimental testing barefoot.

In addition, participants were included and assigned to HA or LA groups based on arch height index (AHI) measurements. AHI was measured as the vertical height of the dorsum at half of the total foot length divided by the truncated foot length (18). All foot measurements were recorded using the arch height index measurement system: a valid and reliable measurement system for characterizing the structure of the foot (3). The HA group was characterized by AHI values greater than 0.377, whereas the LA group was defined as having AHI values less than 0.290. These AHI values correspond to 1.5 SD above and below a mean of 604 feet previously reported in the literature (20) and represent the top and bottom 10% of the population. These values are also similar to previously reported AHI in the literature (3,18). The experimental protocol was approved by the University Institutional Review Board, and all subjects signed informed consent before study participation.

Experimental protocol. Three-dimensional kinematics and GRF were recorded simultaneously using an eight-camera motion analysis system (240 Hz; Vicon Motion Systems Ltd., Oxford, UK) and force platform (1200 Hz, OR-7; AMTI, Watertown, MA), respectively. Participants wore spandex shorts and a T-shirt during testing to limit marker occlusion. The skeleton was modeled using 9.5-mm retroreflective markers placed on the superior and inferior calcaneus, sustentaculum tali, and peroneal tubercle. After a standing calibration, all anatomical markers were removed leaving only tracking markers for the pelvis, thigh, shank, and foot. Participants performed blocks of five overground running trials and five step off landing trials presented in a randomized order. These experimental movements were selected as they represent common load attenuation tasks occurring during sport participation and may lead to injury. Running trials were performed at a self-selected velocity described as the pace at which the participant would run for approximately 30 min. Self-selected running velocity was determined during a series of three practice trials and was maintained (±5%) during testing using an infrared timing system set 5 m

<table>
<thead>
<tr>
<th>Group</th>
<th>Age (yr)</th>
<th>Height (m)</th>
<th>Mass (kg)</th>
<th>AHI (unitless)</th>
<th>Running (m s⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>HA</td>
<td>20.8 ± 2.5</td>
<td>1.62 ± 0.07</td>
<td>58.3 ± 5.4</td>
<td>0.386 ± 0.010*</td>
<td>2.79 ± 0.28</td>
</tr>
<tr>
<td>LA</td>
<td>21.1 ± 2.3</td>
<td>1.63 ± 0.07</td>
<td>58.9 ± 10.9</td>
<td>0.259 ± 0.04</td>
<td>2.73 ± 0.29</td>
</tr>
<tr>
<td>P</td>
<td>0.393</td>
<td>0.390</td>
<td>0.440</td>
<td>0.010</td>
<td>0.356</td>
</tr>
</tbody>
</table>

Participants had similar anthropometric measures except HA athletes had a significantly higher AHI compared with LA athletes.

**Significant difference (P < 0.05) between HA and LA athletes.**
Kinematic and GRF data were filtered using a fourth-order, zero-lag, low-pass Butterworth filter with cutoff frequencies of 10 and 50 Hz, respectively. Running and landing data were analyzed during the load attenuation phase of each movement from initial contact to time of peak vertical GRF. Initial contact was identified with a vertical GRF threshold of 20 N.

The lower extremity was modeled as a linear spring, which included the pelvis, thigh, shank, and foot segments as previously described (4,15). Leg stiffness ($k_{Leg}$) was calculated as the ratio of the peak vertical GRF ($F_{Max}$) and the axial displacement of the lower extremity ($ΔL$) (4,5). $ΔL$ was calculated as the difference in the length of the distance between the metatarsal head and the hip joint between initial contact and $F_{Max}$. The muscular and skeletal contributions to lower extremity stiffness were partitioned based on a previously developed model (4). This model differentiates muscular and skeletal contributions on the basis of the position of lower extremity segments and the orientation of $F_{Max}$ (Fig. 1).

Specifically, skeletal and muscular contributions to lower extremity stiffness were calculated as a function of the angle between the $F_{Max}$ and the tibia as described in equation 1.

$$k_{Leg} = k_{Leg} \cos^2 ϕ + k_{Leg} \sin^2 ϕ$$  \hspace{1cm} \text{[1]}$$

where $k_{Leg}$ is the observed lower extremity stiffness, $ϕ$ is the angle between the orientation of the GRF vector at $F_{Max}$ and the longitudinal axis of the tibia, $k_{Leg} \cos^2 ϕ$ is the skeletal contribution ($k_{Skel}$) to lower extremity stiffness, and $k_{Leg} \sin^2 ϕ$ is the muscular contribution ($k_{Musc}$) to stiffness (4,15). Using this model, muscular contributions to lower extremity stiffness increase with increased knee flexion and ankle plantarflexion. For instance, an individual performing a step off landing with the lower extremity fully extended would land with a vertically oriented GRF vector resulting in a $ϕ$ approaching 0 and all stiffness due to skeletal contributions. Participant means were calculated as the average of the five trials in each condition. Participant means were included in the statistical analyses.

**Statistical analysis.** Independent-sample $t$-tests were used to compare anthropometric measurements and running velocities between HA and LA athletes. Three independent-sample $t$-tests were used to compare dependent variables for running and landing, respectively, including $k_{Leg}$, $k_{Skel}$, and $k_{Musc}$. Additional independent $t$-tests were used to compare variables underlying stiffness calculations, including $ΔL$ and $ϕ$. To adjust for multiple comparisons, a Holm–Bonferroni adjustment was used to alter the level of significance based on the number of comparisons for each movement condition. The Holm–Bonferroni adjustment was conducted for stiffness variables from both experimental movements independently. Cohen’s $d$ effect sizes were also reported to further assess mean differences in $k_{Leg}$, $k_{Skel}$, and $k_{Musc}$ in HA compared with LA athletes using the interpretation of Hopkins (i.e., small, $d < 0.6$; moderate, $0.6 \leq d \leq 1.2$; large, $d > 1.2$) (7). All statistical comparisons were conducted using the Statistical Package for the Social Sciences version 21.0 ($α < 0.05$).

**RESULTS**

No differences were observed between HA and LA athletes in age, height, body mass, or self-selected running velocity (Table 1). To confirm the HA and LA groups, AHI values were significantly different between groups (Table 1). Running. Table 2 presents outcome variables in HA compared with LA athletes during running. During running, HA athletes had significantly greater $k_{Leg}$ ($P = 0.010$, $d = 1.03$) and $k_{Skel}$ ($P = 0.016$, $d = 0.81$) than LA athletes, whereas no differences were observed between HA and LA athletes.
athletes for kMusc ($P = 0.134, d = 0.52$; Fig. 2). $\Delta L$ was smaller in HA athletes compared with LA athletes ($P = 0.020, d = 1.00$). Moreover, $\varphi$ was smaller in HA athletes than LA athletes ($P = 0.038, d = 1.24$).

**Landing.** Table 2 presents outcome variables in HA compared with LA athletes during landing. During the landing task, HA athletes had significantly greater kLeg ($P = 0.015, d = 1.06$) and kSkel values ($P < 0.001, d = 1.84$; Fig. 3). Conversely, kMusc was smaller in HA compared with LA athletes ($P = 0.025, d = 0.96$; Fig. 3). $\Delta L$ was smaller in HA compared with LA athletes ($P = 0.024, d = 0.90$). Further, $\varphi$ was smaller in HA compared with LA athletes ($P = 0.038, d = 0.84$).

## DISCUSSION

The purpose of this study was to quantify the skeletal and muscular contributions to leg stiffness in HA compared with LA athletes during running and step off landing. Although previous studies have identified differences in vertical, leg, and joint stiffness values between HA and LA athletes, this is the first study to quantify skeletal and muscular contributions to stiffness during dynamic loading tasks.

In agreement with our first hypothesis, HA athletes exhibited greater leg stiffness during running and landing compared with LA athletes. These data are consistent with previous findings of greater stiffness values in HA compared with LA athletes during running and landing movements (12,13,16). In a comparison of stiffness in HA and LA athletes during running, Williams et al. (16) reported greater leg and knee joint stiffness in HA athletes compared with LA athletes during overground running. These differences in stiffness were explained by smaller center of mass excursions and knee flexion excursions in HA compared with LA athletes. An investigation of ankle joint stiffness and work in HA compared with LA athletes during running revealed that HA athletes also exhibit greater ankle joint stiffness during the braking portion of the stance phase of running (13). Finally, a comparison of landing biomechanics in HA and LA athletes revealed that HA athletes exhibited greater vertical stiffness than LA athletes during a step off landing (12). Although no differences in peak vertical GRF were observed between HA and LA athletes, similar to the study by Williams et al. (16), HA athletes had significantly smaller COM excursions to explain the greater vertical stiffness compared with LA athletes. The evidence that HA athletes exhibit greater lower limb stiffness compared with LA athletes in running and landing is compelling. However, these findings are impractical to clinicians and sport scientists without fully understanding how stiffness is specifically partitioned within the musculoskeletal system.

<table>
<thead>
<tr>
<th>Task</th>
<th>Group</th>
<th>kLeg (kN m)</th>
<th>kSkel (kN m)</th>
<th>kMusc (kN m)</th>
<th>$\Delta$ (m)</th>
<th>$d$ (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Running</td>
<td>HA</td>
<td>29.3 ± 11.3</td>
<td>15.8 ± 10.2</td>
<td>13.5 ± 4.3</td>
<td>0.055 ± 0.013</td>
<td>27.7 ± 3.2</td>
</tr>
<tr>
<td></td>
<td>LA</td>
<td>20.6 ± 3.7</td>
<td>9.6 ± 3.6</td>
<td>11.0 ± 5.3</td>
<td>0.067 ± 0.012</td>
<td>31.1 ± 2.2</td>
</tr>
<tr>
<td></td>
<td>$P$</td>
<td>0.017</td>
<td>0.044</td>
<td>0.134</td>
<td>0.020</td>
<td>0.038</td>
</tr>
<tr>
<td>Landing</td>
<td>HA</td>
<td>31.7 ± 7.7</td>
<td>22.1 ± 6.0</td>
<td>9.6 ± 3.3</td>
<td>0.044 ± 0.010</td>
<td>47.3 ± 16.5</td>
</tr>
<tr>
<td></td>
<td>LA</td>
<td>24.4 ± 6.0</td>
<td>12.1 ± 4.8</td>
<td>12.5 ± 2.7</td>
<td>0.058 ± 0.016</td>
<td>59.9 ± 13.2</td>
</tr>
<tr>
<td></td>
<td>$P$</td>
<td>0.015</td>
<td>&lt;0.001</td>
<td>0.025</td>
<td>0.024</td>
<td>0.038</td>
</tr>
<tr>
<td></td>
<td>$d$</td>
<td>1.06</td>
<td>1.84</td>
<td>0.96</td>
<td>0.90</td>
<td>0.94</td>
</tr>
</tbody>
</table>

Also presented are changes in limb length ($\Delta L$) and the angle of the GRF vector relative to the orientation of the longitudinal axis of the tibia ($\varphi$).
Our second hypothesis was partially supported as greater skeletal and smaller muscular contributions to leg stiffness were observed in HA compared with LA athletes during the step off landing task. However, greater skeletal contributions to leg stiffness were found in HA athletes with no differences in muscular contributions to leg stiffness between groups during running. In HA athletes, the skeletal component (kSkel) contributed 58% and 70% of leg stiffness (kLeg) during running and landing, respectively, while contributing only 40% and 48% of leg stiffness in LA athletes. The implications of skeletal contributions to leg stiffness may be more important during landing as a greater effect magnitude was observed in landing (d = 1.84) compared with running (d = 0.81) when comparing HA and LA athletes. The model used to quantify kSkel and kMusc in the current study (4) defined skeletal and muscular contributions to leg stiffness as a function of the change in limb length (ΔL) and the angle of the tibia (φ) with smaller ΔL and φ values associated with greater kSkel and smaller kMusc values. These variables (ΔL and φ) could be considered analogs of center of mass and knee flexion excursions. Previous research has demonstrated that HA athletes exhibit smaller center of mass excursions (12,16) and knee flexion excursions (16) during running and landing tasks. These data support previous research findings and suggest that HA athletes performed the running and landing tasks in a more erect lower extremity posture characterized by smaller center of mass excursions and knee flexion angles than their LA counterparts. The running and landing biomechanics observed in HA athletes, including greater relative contributions of the skeletal component, may underlie the greater prevalence of skeletal injuries reported in HA athletes (8,19). Conversely, LA athletes were shown to have greater muscular contributions to leg stiffness as well as a greater reliance on muscular structures for load attenuation. Greater muscular contributions to load attenuation may provide mechanistic evidence of the greater rate of soft tissue injuries in LA athletes (8,19). The findings of the current study describe a mechanism by which lower extremity loading may be preferentially applied to skeletal or muscular structures resulting in divergent musculoskeletal injury patterns. These findings may also provide a foundation by which athletes may alter running and landing biomechanics to reduce skeletal or muscular loading in response to injury. However, the role of loading volume in lower extremity overuse injury must be considered as a central factor in future studies. Although the current study presents novel findings pertaining to skeletal and muscular contributions to leg stiffness during running and landing, some limitations do exist. Ten HA and 10 LA female athletes were included in the current study, which could be considered a small sample size with potentially limited statistical power. However, moderate to large Cohen’s d effect sizes were observed between the HA and the LA athletes. In addition, only female athletes were included in this study, and therefore the results may not be generalizable to male athletes because sex differences have been observed during running and landing tasks (14). Finally, a set drop height of 30 cm was used during the step off landing trials which may have provided a variable mechanical demand between athletes. However, because the average body height of athletes between groups was not different, the mechanical demand during landing between groups was likely similar. In addition, the height of 30 cm was selected to reduce discomfort and the risk of injury as athletes performed the experimental movements barefoot.

CONCLUSION
The findings of this study demonstrate that HA athletes exhibit greater leg stiffness in running and landing than LA athletes, and these differences are due to reduced knee flexion and center of mass excursion. Importantly, the current study provides novel findings that HA athletes have greater skeletal contributions to leg stiffness and load attenuation during both running and landing but smaller muscular contributions to leg stiffness during landing only compared with LA athletes. These unique running and landing contributions to lower limb stiffness may underlie differences in injury patterns experienced by HA and LA athletes during different movement tasks.

No funding was received in support of this research. The results of the study are presented clearly, honestly, and without fabrication, falsification, or inappropriate data manipulation. None of the authors or anyone involved with this research had a conflict of interest, and the results of this study do not constitute endorsement by the American College of Sports Medicine.

REFERENCES


