High-arched runners exhibit increased leg stiffness compared to low-arched runners

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Abstract

Leg stiffness between high-arched (HA) and low-arched (LA) runners was compared. It was hypothesized that high-arched runners would exhibit increased leg stiffness, increased sagittal plane support moment, greater vertical loading rates, decreased knee flexion excursion and increased activation of the knee extensor musculature. Twenty high-arched and 20 low-arched subjects were included in this study. Leg stiffness, knee stiffness, vertical loading rate and lower extremity support moment were compared between groups. Electromyographic data were collected in an attempt to explain differences in leg stiffness between groups. High-arched subjects were found to have increased leg stiffness and vertical loading rate compared to low-arched runners. Support moment at the impact peak of the vertical ground reaction force was related to leg stiffness across all subjects. High-arched subjects demonstrated decreased knee flexion excursion during stance. Finally, high-arched subjects exhibited a significantly earlier onset of the vastus lateralis (VL) than the low-arched runners. Differences exist in leg stiffness and vertical loading rate between runners with different foot types. Differences in lower extremity kinetics in individuals with different foot types may have implications for new treatment strategies or preventative measures.
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1. Introduction

The relationship between lower extremity structure and mechanics has received attention in the recent literature [1–4]. Since the foot is the interface with the ground during gait, differences in foot structure may result in differences in mechanics of the entire lower extremity. Studies which have addressed the relationship between foot structure (specifically, medial longitudinal arch structure) and lower extremity mechanics have demonstrated mixed results [4–6]. This may be due to the similarity in arch structures compared. Evaluation of individuals with extremely different medial longitudinal arch structure may provide more definitive information regarding the relationship between structure and mechanics in the lower extremity during running.

One mechanical parameter that may be related to arch structure is leg stiffness. Stiffness may be an important factor during running as it represents the ability of the entire lower extremity to attenuate the excessive forces generated during the stance phase. If the system becomes too compliant, there may be an overload of structures associated with force attenuation (i.e. eccentric muscle activity). If there is decreased compliance (increased stiffness), there may be an increase in forces up the kinetic chain. Lower extremity stiffness has been defined in numerous ways [7–9]. One method of representing the stiffness property of the entire lower extremity during gait is to assess the amount of vertical deflection of the leg spring (as a function of the body's center of mass (COM)) for a given ground reaction force [9]. Stiffness has recently been found to differ in individuals with different dynamic foot orientations during running [4].

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However, individuals with pes planus and pes cavus have not previously been evaluated.

The structure of the foot may influence leg stiffness through the mechanical coupling of the foot and the knee. For example, in the HA individual, an increased height of the medial longitudinal arch is often associated with a more supinated position of the foot. Therefore, the foot may exhibit decreased pronation throughout the stance phase. During gait, pronation is associated with tibial internal rotation and knee flexion [10–13]. Therefore, a decrease in pronation may result in decreased knee flexion excursion in these individuals and an associated higher stiffness. These kinematic parameters may also be associated with neuromuscular function during running. It has been shown that lower extremity stiffness increases with increased activation of a single muscle around a joint [14,15] or increased cocontraction around a joint [16].

The purpose of this study was to compare leg stiffness between two groups of runners: those with high arches and those with low arches. Specifically, it was hypothesized that high-arched runners will demonstrate greater stiffness, which will result in a greater total extension moment (support moment) in the sagittal plane. Greater total support moment was hypothesized to be associated with increased knee extensor activation and a subsequent decrease in knee extension motion. Finally, as a result of a stiffer leg, the rate of loading from initial contact to the impact peak was expected to be increased.

2. Methods

The study included 20 high-arched (HA) (10 females; 10 males) and 20 low-arched (LA) (12 females; 8 males) subjects between the ages of 18 and 50 (mean = 27.8 years; S.D. = 8.1) with no lower extremity injury at the time of the experiment. Subjects were excluded if they had a history of ACL deficiency, foot surgery or lower extremity surgery within the previous 12 months. Subjects ran at least 6 miles per week at a minimum 8 min/mile pace. Prior to the experiment, all subjects signed informed consent forms.

Subjects were screened for inclusion in the HA or LA group using an arch ratio [17]. The arch ratio was defined as the height to the dorsum of the foot from the floor at 50% of the foot length (DORS) divided by the individual's truncated foot length (TFL). TFL was 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. An arch ratio of at least 0.356 was needed for inclusion in the HA group and less than or equal to 0.275 for the LA group. These values fell at or outside 1.5 standard deviations of the mean DORS/TFL ratio measurement of 0.316 (σ = 0.027) based on a sample population of 102 feet.

Subjects who met the inclusion criteria returned to the laboratory for a complete gait analysis. Retroreflective markers (anatomical markers and at least three tracking markers per segment) were placed unilaterally (side of greatest previous injury involvement) on the segments of the rearfoot, shank, thigh and pelvis (Fig. 1). The three rearfoot markers were placed directly on the heel and extended through windows cut in the shoes.

An anatomical coordinate system was established for each of the foot, shank, thigh and pelvis in order to determine joint kinematics and kinetics. 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 medial to lateral. Anatomical markers were then removed following a standing calibration. Kinematic data were collected using six camera VICON motion analysis system (Oxford Metrics Limited, UK). All subjects wore the same brand and
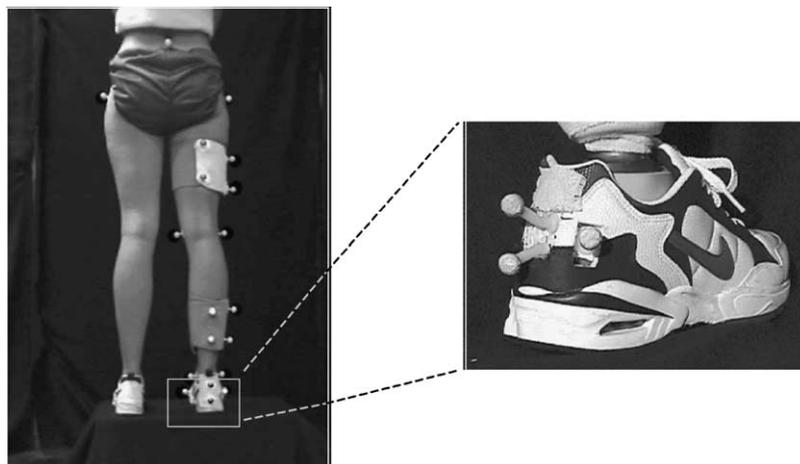

Fig. 1. Retroreflective marker placement with heel marker placement. Markers were placed on the segments of the pelvis, thigh, lower leg and rearfoot. Anatomical markers were placed over greater trochanters, femoral condyles, malleoli and forefoot. The inset shows markers placed directly on the skin, projecting through holes cut in the heel counter of the running shoe.
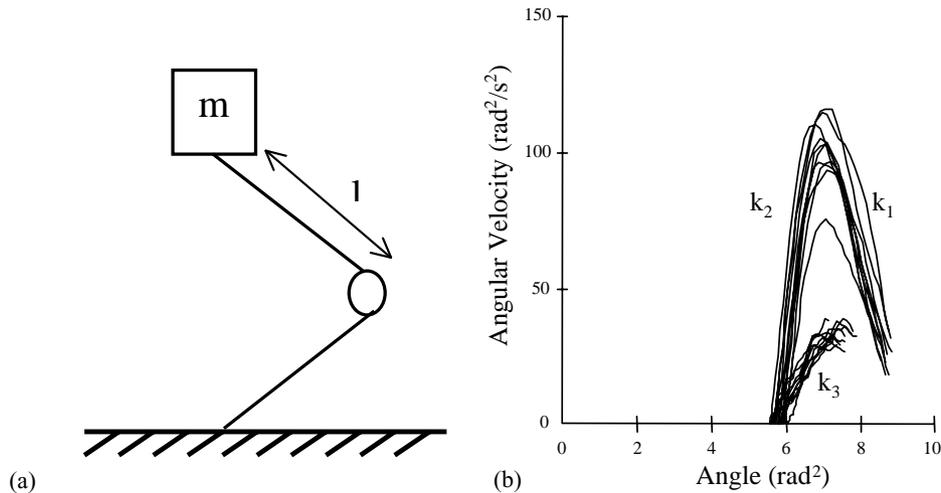
Fig. 2. (a) The block (m) represents the body mass of the subject. The length (l) from the center of the mass to the center of rotation represents the distance between the joint centers of the hip and knee. (b) The slopes of the lines from these sample data indicate the knee stiffnesses (k1–k3) throughout the stance phase.

model of shoes. The subjects then ran along a 25 m runway at a speed of 3.35 m/s (8 min/mile pace). Speed was monitored with photocells, and only trials within ± 5% of the target speed were accepted. A force plate (BERTEC, Worthington, OH) mounted in the center of the runway recorded ground reaction forces. Kinematic data were sampled at 120 Hz and force data at 960 Hz. Ten footstrikes were collected and averaged for each subject.

The three-dimensional coordinates of each marker were reconstructed using the VICON motion analysis software. The three-dimensional coordinates were filtered using a second-order recursive Butterworth filter with 8 Hz low-pass cut-off frequency. Force data were low-pass filtered at 50 Hz. All data were analyzed between heel strike and toe-off and normalized to 100 data points, each representing 1% of the stance phase of gait. MOVE3D software (National Institutes of Health, Bethesda, MD) was used to determine joint kinematic data. Kinematic data were resolved about a joint coordinate system, [18] 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 two axes. Joint moments were resolved about a helical axis into the proximal segment's coordinate system.

Leg stiffness was estimated during running using a mathematical spring-mass model [9,19–21]. Stiffness was further scaled to body mass as body mass has been previously found to be positively correlated to leg stiffness [22]. The vertical deflection of the center of mass was determined by using the second integral of the vertical ground reaction force [23].

Since more motion during the first half of stance in the sagittal plane comes from the knee [24] when compared to the hip and ankle, a post-hoc assessment of knee joint stiffness was employed to help explain differences in leg

stiffness and knee extensor muscle activity. Knee stiffness was modeled as a block and a rotational spring (Fig. 2a)

$$\frac{k}{I} = \frac{\Delta\omega^2}{\Delta\theta^2} \tag{1}$$

where k corresponded to knee stiffness, I is the mass moment of inertia of the thigh, ω the angular velocity of knee flexion and θ the angular displacement of knee flexion [8]. Three linear regions (stiffnesses) of the plot of $\Delta\omega^2/\Delta\theta^2$ can be calculated using this method. $k1$ is calculated from heel strike to peak knee flexion velocity. $k2$ is calculated from peak knee flexion velocity to peak knee flexion angle. $k3$ is calculated from peak knee flexion angle to toe-off (Fig. 2b). Knee stiffness was evaluated during the weight acceptance phase of stance, which corresponds to $k2$.

Support moment [25] was determined throughout the stance phase and was analyzed at the time of the impact peak of the vertical ground reaction force (SMFZ), peak knee flexion angle (SMKF) and peak posterior propulsive ground reaction force (SMAP). These events were chosen as they occur at transition points during the gait cycle. Finally, vertical loading rate was determined by calculating the average rate of rise of the heel strike transient of the vertical ground reaction force over the interval spanning 20–80% of this peak.

Surface electromyography (EMG) was also used to provide a neuromuscular explanation for differences that might be found in stiffness. EMG was recorded at 960 Hz using a 16-channel system (Motion Lab Systems Inc., Baton Rouge, LA) to determine muscle onset times, and the amount of co-activation present in selected lower extremity muscles during running. EMG electrodes were placed unilaterally over the lateral hamstrings (LH), vastus lateralis (VL), lateral head of the gastrocnemius (LG) and tibialis anterior (TA) muscles as described by Winter [26]. EMG data were

collected during a maximum voluntary isometric contraction prior to motion trials in order to normalize muscle activity. EMG data were filtered at the collection site with a four-pole band pass filter with a range of 20–350 Hz. Raw EMG data were processed with custom software (Labview, National Instruments, Austin, TX). A linear envelope was created using full wave rectification and a phase corrected second-order Butterworth filter with a low-pass cut-off frequency of 20 Hz.

All EMG data were time synchronized with the kinematic and kinetic data. Co-activation values were determined for the pairs of VL and LH, VL and LG, and LG and TA and averaged over the 10 trials. Co-activation was determined with the following equation:

$$\text{co-activation} = \left(\frac{\text{LM}}{\text{MM}}\right)(\text{LM} + \text{MM}) \qquad (2)$$

At each 1% of stance, the level of activity of the less active muscle (LM) was divided by the level of activity of the more active muscle (MM). This value was then multiplied by the value of activation of the sum of the muscles [27]. Co-activation was determined from 83 ms before heel strike to 83 ms before toe-off in order to account for electromechanical delay [28]. Muscle onset was determined by finding the time of peak EMG activity, moving backwards and stopping when three consecutive data points within three standard deviations of resting EMG activity were located [29]. Time to peak was determined as the time to absolute peak EMG activity. Time to onset values are reported in reference to heel strike with negative values referring to pre-heel strike and positive values post-heel strike.

Comparison between HA and LA subjects were made using a one-tailed Student's t-test ($P \leq 0.05$) to determine whether differences in stiffness, support moment, peak knee flexion and vertical loading rate existed between these groups. Additionally, all variables were correlated with each other to further describe their inter-relationships across subjects in both groups.

3. Results

Subjects were similar in age and height between the two groups (Table 1). HA and LA runners' arch ratios were found to fall outside +1.9 and −1.7 standard deviations, respectively, of the previously collected sample mean of 102 feet.

HA runners showed significantly greater leg stiffness compared to LA runners. HA subjects also showed significantly higher knee stiffness than LA subjects. The forward velocity of running was not significantly different between the HA and LA runners ($P = 0.36$) (Table 2). These differences represent approximately 10 and 17% higher stiffness in the HA runners for leg extremity stiffness and knee stiffness, respectively. HA runners had shorter support times and smaller COM excursions when compared to the LA runners.

Table 1
Mean ± standard deviation of subject characteristics

	High arch	Low arch	P-value
Number of subjects	20	20	
Sex	12 F, 8 M	10 F, 10 M	
Height (m)	1.72 ± 0.09	1.74 ± 0.10	0.510
Mass (kg)	66.45 ± 9.49	72.73 ± 17.89	0.174
Age (years)	28.0 ± 8.1	27.7 ± 7.5	0.904
Arch ratio	0.37 ± 0.01	0.27 ± 0.02	0.000*
Leg length (m)	0.90 ± 0.05	0.90 ± 0.06	0.639

m: meters and kg: kilograms.
* Statistically significant.

Table 2
Mean ± standard deviation for stiffness characteristics between high-arched and low-arched subjects

	High arch	Low arch	P-value
Leg stiffness (kN/m kg)	7.17 ± 1.16	6.46 ± 1.01	0.03*
Knee stiffness (kN m/kg)	0.14 ± 0.04	0.12 ± 0.02	0.04*
Forward velocity (m/s)	3.53 ± 0.07	3.55 ± 0.07	0.36
Contact time (s)	0.26 ± 0.02	0.27 ± 0.02	0.03*
Vertical displacement (m)	0.07 ± 0.01	0.08 ± 0.01	0.01*

kN: kilo Newtons; m: meters and s: seconds.
* Statistically significant.

No support moment variables were found to be significantly different between the two groups (Table 3). HA subjects demonstrated a weak trend of greater average support moments during impact (SMFZ and SMAP) while LA subjects demonstrated a mild trend of greater mean support moment during peak knee flexion (SMKF).

No differences were observed in co-activation values between groups (Table 4). VL onset was the only timing variable correlated to leg stiffness ($R = -0.31$, $P = 0.029$). It was activated significantly earlier in HA runners when compared to LA runners ($P = 0.002$).

Knee flexion excursion was significantly lower ($P = 0.007$) in the HA group by 4.2° when compared to the LA group (Fig. 3a), although decreases in joint excursions at the hip, knee and ankle could account for an overall decreased COM deflection.

Table 3
Mean ± standard deviation for loading rate and support moment variables

	High arch	Low arch	P-value
Loading rate (N/s)	62.48 ± 13.62	52.05 ± 10.79	0.01*
SMFZ (N m/kg m)	1.47 ± 0.28	1.45 ± 0.33	0.44
SMKF (N m/kg m)	3.51 ± 0.41	3.72 ± 0.60	0.12
SMAP (N m/kg m)	2.21 ± 0.46	2.10 ± 0.69	0.28

N: Newtons; s: seconds; m: meters and kg: kilograms.
* Statistically significant.

Table 4
Mean ± standard deviation for electromyographic muscle co-activation and timing variables

	High arch	Low arch	P-value
QH	5.88 ± 2.34	5.54 ± 2.01	0.31
QG	9.68 ± 3.34	9.05 ± 3.52	0.29
GT	5.76 ± 3.15	5.46 ± 2.19	0.37
VL onset (% stance)	−20.95 ± 4.77	−13.20 ± 9.54	0.00*
LG onset (% stance)	−20.75 ± 7.34	−20.30 ± 8.16	0.43
TA onset (% stance)	−20.30 ± 14.46	−24.05 ± 6.78	0.18
LH onset (% stance)	0.65 ± 16.88	−3.05 ± 15.07	0.27

QH: quadriceps to hamstrings co-activation ratio; QG: quadriceps to gastrocnemius co-activation ratio; GT: gastrocnemius to tibialis anterior co-activation ratio; VL: vastus lateralis; LG: lateral gastrocnemius; TA: tibialis anterior and LH: lateral hamstrings.

* Statistically significant.

Loading rate was significantly greater in the HA runners when compared to the LA runners (Table 3). The lower loading rate is a result of a lower impact peak that occurs over a longer period of time (Fig. 3b).

4. Discussion

The overall purpose of this study was to determine if structural deviations in the lower extremity would be associated with differences in lower extremity kinetics. This may be important, as it has been suggested that variations in lower extremity kinetics may be a factor in injuries to the lower extremities [30,31]. The results of this study suggest that HA subjects have greater overall leg stiffness and a higher vertical loading rate when compared to LA runners. One possible limitation of this study is that the stiffness measure evaluates total lower extremity stiffness over the entire stance phase of gait, which only provides limited information given the biphasic nature of the ground reaction force. In order to gain information regarding the impact peak, ver-

tical loading rate was evaluated and found to be greater in the HA subjects. HA subjects also demonstrated knee extensor activity prior to heel strike that may have assisted with controlling knee extension and contributed, in part, to the increased stiffness. Although a strict protocol was employed for EMG collection and analysis, this information should be interpreted with caution given the inherent variability of EMG data, especially during running.

Leg stiffness values obtained in this study are less than what has been reported in the past [4,19,21,22]. This is consistent with the longer contact times and greater center of mass excursion found in this study. Additionally, the previous studies did not scale stiffness values to individual body mass. Increased body mass has been reported to be associated with increased stiffness [22]. Unscaled values are more similar to previously reported values as they were slightly greater ($\mu = 8.41$, $\sigma = 1.78$ range = 4.7–14.6) than scaled values. This may account for some of the differences observed.

The results of this study supported the hypothesis that HA runners exhibited greater leg stiffness. HA runners showed shorter contact times and less vertical deflection of the center of mass (due largely to decreased flexion at knee), which is consistent with the increased knee stiffness observed. Increased stiffness in the leg may explain, in part, the greater incidence of bony injuries reported in the HA subjects [32]. Although differences appeared to be small between groups (10%), these differences are likely reliable. Stiffness is estimated using data collected from the force plate. These data have been reported to be the most sensitive and reliable [13].

Knee joint stiffness was also evaluated and the results were consistent with the total leg stiffness with HA runners exhibiting higher values. As previously discussed, knee range of motion has the largest contribution to overall sagittal plane excursion of the joint in the lower extremity and therefore, likely contributes the greatest amount to overall lower extremity stiffness. Since the knee goes through approximately 45° of flexion during the loading phase of running gait, it seems reasonable that knee stiffness and leg
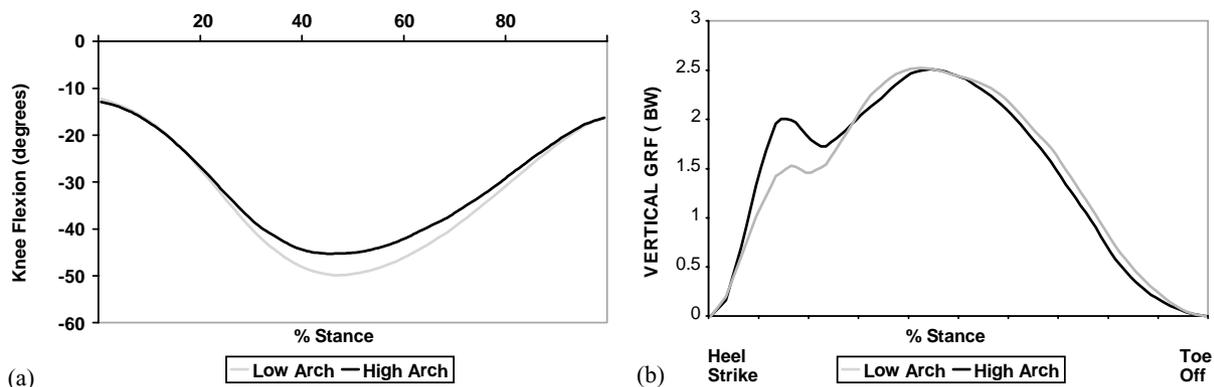

Fig. 3. (a) Knee flexion excursion during stance. Low-arched runners reached a statistically significant ($P = 0.007$) greater peak knee flexion as a result of having greater knee flexion excursion during the first half of stance. (b) Representative ground reaction force curves and loading rates for high-arched and low-arched subjects [38]. Actual vertical loading rates were found to be significantly different between groups.

stiffness would be related. The current results differ from the results of Farley and Morgonroth [33] who state that leg stiffness is driven by ankle stiffness during hopping. Hopping is characterized by a forefoot strike landing, while the runners in the current study were all rearfoot strikers. Future evaluation of joint stiffness in rearfoot strikers and forefoot strikers may provide further insight into the relative contributions of joint stiffness to total lower extremity stiffness during running. These differences may allow for specificity in training or rehabilitation for individuals with different foot types or footstrike patterns.

While the support moment variables were not different between groups, there was a significant positive correlation with leg stiffness across all subjects for SMFZ ($R = 0.56$, $P = 0.001$). SMFZ is the total extensor moment occurring at the impact peak of the vertical ground reaction force. This peak typically occurs in the first 20% of stance. This may suggest that an increased extensor moment during initial loading played a role in increasing leg stiffness. This moment may be a response to spindle activation during heel contact as these reflexes are thought to occur within the initial 25% of stance phase [34,35]. This increase may also be related to the earlier knee extensor activity in the HA group.

The results of the EMG analysis may offer an explanation for the observed differences in stiffness in this study, as suggested by Heise et al. [36] that muscle activity may account for lower extremity stiffness. The VL became active sooner in the HA runners which may explain the decreased knee flexion excursion in this group. Based on these data, it appears that during running the activation of the agonist for knee eccentric control (quadriceps) may contribute more to knee stiffness than co-activation of the muscles surrounding the knee. This is likely due to the large external flexion moment generated at the knee by gravity. This requires an extension moment to be generated by the quadriceps and requires little activity of the hamstrings, thus decreasing co-activation. The center of mass excursion was also reduced. This reduction in the COM excursion was likely a result of the decreased knee flexion range of motion. These results provide rationale for possible training of the knee extensors to increase or decrease knee activation before heel strike in order to regulate knee stiffness.

The larger vertical loading rate in the HA runners is interesting in conjunction with the stiffer gait pattern in this group. Loading rate is calculated during the impact peak of the vertical ground reaction force. Force transmission was decreased passively during this portion of the stance phase. Leg stiffness takes into account active loading. The earlier activation of the knee extensors (VL) may have allowed for a more rigid limb at heel strike and contributed to the increased loading rate. Additionally, this activation may have decreased the knee flexion excursion occurring later in the stance phase, increased the knee stiffness and contributed to the difference in overall stiffness seen in the HA runners.

Vertical loading rate has previously been suggested to increase the risk of knee pathology [37]. Loading rate values

were consistent with those previously reported by Radin et al. [37]. An 18% difference was noted in this parameter between HA and LA runners. Although the magnitude of this difference alone may appear relatively small, it likely becomes more detrimental in these subjects as ground reaction forces and foot strike repetitions are high during running. Consistent with this finding, the HA runners in the present study did have significantly higher rates of bony lower extremity injuries [32]. Because vertical loading rate is considered passive and occurs quickly during initial loading, further study is necessary to determine if loading rate can be augmented during running.

Based on the current results, HA runners exhibit higher leg and knee stiffness. These increased stiffnesses are accompanied by an increased vertical loading rate, decreased knee flexion excursion and earlier activity of the knee extensors. Based on these relationships, training runners to increase lower extremity compliance may result in decreasing their risk of future bony injury. Future work should focus on measuring medial longitudinal arch mobility, as static structure alone cannot completely dictate lower extremity mechanics. Currently, it is difficult to objectively measure the midfoot and forefoot due to marker placement difficulties, skin movement issues and shoe interference. Objectively, quantifying dynamic arch mobility may determine if a stronger relationship can be established between arch structure, arch stiffness and leg stiffness. This relationship may result in better treatment and injury prevention in runners with specific clinical presentations.

Acknowledgements

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