ARTICLE IN PRESS

Journal of Biomechanics xxx (xxxx) xxx

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech

Effects of footwear cushioning on leg and longitudinal arch stiffness during running

Nicholas B. Holowka [a,b,*,1], Stephen M. Gillinov [a,c,1], Emmanuel Virot [d], Daniel E. Lieberman [a]

[a] Department of Human Evolutionary Biology Harvard University Cambridge, MA, USA
[b] Department of Anthropology University at Buffalo Buffalo, NY, USA
[c] Yale School of Medicine Yale University New Haven, CT, USA
[d] John A. Paulson School of Engineering and Applied Sciences Harvard University Cambridge, MA 02138, USA

ARTICLE INFO

Keywords:
Leg spring
Foot
Running shoe
Midsole
Barefoot running

ABSTRACT

During running, humans increase leg stiffness on more compliant surfaces through an in-series spring relationship to maintain constant support mechanics. Following this notion, the compliant midsole material of standard footwear may cause individuals to increase leg stiffness while running, especially in footwear with very thick midsoles. Recently, researchers have also proposed that footwear stiffness can affect the stiffness of the foot's longitudinal arch (LA) via a similar mechanism. To test these ideas, we used 3D motion capture to record 20 participants running on a forceplate-instrumented treadmill while barefoot, and while wearing three types of sandals composed of materials ranging an order of magnitude in Young's modulus: ethylene vinyl acetate (EVA), and two varieties of polyurethane rubber (R30 and R60). We calculated leg stiffness using standard methods, and measured LA stiffness based on medial midfoot kinematics. While there was an overall significant effect of footwear on leg stiffness ($P = 0.047$), post-hoc tests revealed no significant differences among individual pairs of conditions, and there was no effect of footwear on LA stiffness. However, participants exhibited significantly greater LA compression when barefoot than when running in EVA ($P = 0.004$) or R30 ($P = 0.036$) sandals. These results indicate that standard footwear midsole materials are too stiff to appreciably affect leg stiffness during running, meaning that increasing midsole thickness is unlikely to cause individuals to alter their leg stiffness. However, use of footwear does cause individuals to restrict LA compression when compared to running barefoot, and further research is needed to understand why.

1. Introduction

During running, the human body can be modeled as a spring-mass system, with the leg as a spring that is connected to a point mass and that is capable of storing and releasing elastic energy in muscle–tendon units and other connective tissues (Alexander, 1991; Blickhan, 1989). Leg stiffness in humans is relatively constant at low running speeds (Farley et al., 1993), but increases on more compliant surfaces to maintain similar support mechanics, including foot contact time, stride frequency, and center of mass motion (Ferris et al., 1998; Kerdok et al., 2002). This leg stiffness (k_{leg}) adjustment is believed to follow an in-series spring relationship that maintains constant overall stiffness of the leg-surface system (k_{syst}), such that

$$\frac{1}{k_{syst}} = \frac{1}{k_{leg}} + \frac{1}{k_{surf}} \tag{1}$$

where k_{surf} is the stiffness of the running surface. This relationship has been supported empirically (Ferris et al., 1998; Kerdok et al., 2002), but in theory, running footwear with midsoles made out of elastic material such as ethylene vinyl acetate (EVA) should add one additional component to this system, such that

$$\frac{1}{k_{syst}} = \frac{1}{k_{leg}} + \frac{1}{k_{surf}} + \frac{1}{k_{ftwr}} \tag{2}$$

where k_{ftwr} is the footwear midsole material compressive stiffness (Ferris et al., 1998). A prediction of this model is that more compliant footwear should cause individuals to increase their leg stiffness during running,

* Corresponding author at: Department of Anthropology, University at Buffalo, Buffalo, NY, USA.
E-mail address: nbholowk@buffalo.edu (N.B. Holowka).
[1] Nicholas B Holowka and Stephen M. Gillinov are co-first authors based on their contributions to the work.

https://doi.org/10.1016/j.jbiomech.2021.110869
Accepted 16 November 2021
Available online 19 November 2021

and while this prediction is often discussed in regard to the effects of footwear on running biomechanics (Butler et al., 2003; De Wit et al., 2000; Divert et al., 2005), it has not been tested directly.

Kulmala et al. (2018) recently found that overall leg stiffness increases in individuals running in shoes with thicker EVA midsoles, and since midsole thickness should be inversely proportional to stiffness, this finding supports the expectation that leg stiffness increases in more compliant footwear. In contrast, Bishop et al. (2006) found that during single-leg hopping, which also follows spring-mass mechanics (Ferris and Farley, 1997), participants exhibited no difference in leg stiffness when wearing two different types of running shoes that differed nearly two-fold in midsole stiffness. One possible reason for these different results is that both studies used commercially available shoes that may have had different design features other than just midsole stiffness that could potentially affect limb mechanics (e.g., heel-to-forefoot height offset), thereby confounding results. In addition, single-leg hopping may not involve the same proprioceptive feedback as running.

Another factor to consider is the foot's longitudinal arch (LA), which may also be affected by the compressive stiffness of footwear. The stiffness of the LA is commonly attributed to plantar aponeurotic and ligamentous tissues (Ker et al., 1987), as well as the activity of extrinsic and intrinsic foot muscles (Farris et al., 2020; Kelly et al., 2014, 2015). Kelly et al. (2016) observed that individuals had greater intrinsic foot muscle activation and reduced LA compression when running in shoes than when running barefoot, leading them to hypothesize that footwear compressive stiffness and LA stiffness are related through an in-series spring system like the leg-surface relationship. Contrary to this hypothesis, Birch et al. (2021) found that LA stiffness was unaffected by ground surface stiffness during hopping. However, neither study directly measured the effects of footwear compressive stiffness on LA stiffness during running.

To test the relationships between footwear compressive stiffness, LA stiffness, and overall leg stiffness, we measured participants running barefoot and in custom-designed footwear made from materials that varied by an order of magnitude in Young's modulus. Following the hypothesis that the leg and running surface function as in-series springs, we predicted that individuals would have more compliant legs when running in stiffer footwear. Additionally, following Kelly et al. (2016), we predicted that individuals would exhibit lower LA stiffness in stiffer footwear. Finally, we predicted that individuals would exhibit the lowest leg and LA stiffness when running barefoot on a rigid (steel) treadmill surface.

2. Methods

2.1. Participants

We enrolled 20 participants (12 males and 8 females; age, 21.2 ± 3.3 yrs; height: 174.3 ± 10.3 cm; weight: 63.4 ± 9.3 kg;) who were all experienced runners and members of a university running club. Participants reported no lower limb musculoskeletal injuries in the previous 6 months or pre-existing neuromuscular conditions. All experimental procedures were approved by the Institutional Review Board of Harvard University, and participants provided written informed consent.

2.2. Experimental procedures

We measured participants running in four different footwear conditions: barefoot, and in sandals made from different materials. The sandals included Luna MONO sandals (Luna, Seattle, WA), with a midsole consisting of a flat, 1.2 cm-thick layer of EVA without any features that restrict natural LA motion (Fig. 1a). These 'EVA' sandals were used to produce molds for two other identically-shaped sandals cast from Reoflex polyurethane rubber (Smooth-On, Macungie, PA). We used Reoflex with 30A and 60A Shore hardness to make 'R30' and 'R60' sandals, respectively. To prevent the sandals from bending during swing phase, we added four thin metal wires oriented longitudinally to the sandal midsole to confer greater stiffness against bending, but not compression.

To quantify sandal stiffness, we conducted material properties tests on rectangular pieces cut from the sandals in a custom-made tensile testing machine (ADMET Inc., Norwood, MA), which stretched samples over a 10 mm range at a rate of 0.83 mm/s. We calculated Young's modulus (E) from the slope of the linear portion of the resulting stress–strain curve (Table 1).

We applied spherical reflective markers to 13 anatomical landmarks on the right lower limbs of each participant, following Cappozzo et al. (1995), and placed a marker on the right navicular tuberosity for LA stiffness calculations. We recorded participants running on a split-belt forceplate-instrumented treadmill (Bertec, Columbus, OH) using an eight-camera Oqus motion capture system (Qualisys Corp, Gothenburg, Sweden). Force and marker data were captured at 1000 Hz and 200 Hz, respectively.

Table 1
Abbreviation definitions.

Abbreviation	Variable Definition
A_{foot}	Area of force application of the foot
FSA	Foot strike angle
F_{leg}	Compressive force on the leg
k_{ftwr}	Footwear midsole material compressive stiffness
k_{LA}	Longitudinal arch compressive stiffness
k_{leg}	Leg stiffness
k_{surf}	Surface stiffness
k_{syst}	Leg-surface system stiffness
M_{LA}	External moment on the longitudinal arch
$vGRF$	Vertical ground reaction force
Δl_{leg}	Leg compression
$\Delta\theta_{LA}$	Longitudinal arch compression

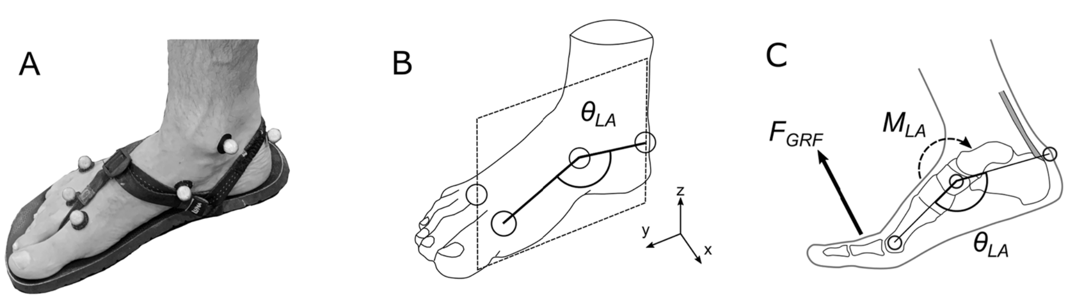

Fig. 1. Foot marker set and longitudinal arch variables used in this study. (A) Foot markers set, with sandals (Luna MONO) used in this study. (B) Markers used to calculate longitudinal arch compression angle (θ_{LA}), as the projected sagittal plane angle between the first metatarsal head, navicular tuberosity, and the posterior calcaneus. (C) Variables used to calculate LA stiffness. The ground reaction force vector (F_{GRF}) creates a plantarflexion moment about the navicular tuberosity (M_{LA}), causing longitudinal arch compression (θ_{LA}).

The order of the footwear conditions (barefoot, EVA, R30, R60) was randomized for each participant. To ensure dynamic similarity regardless of leg length, each participant ran at a 1.0 Froude number (average speed: 2.98 ± 0.08 m/s) (Alexander and Jayes, 1983), calculated as

$$Froude = \frac{v^2}{g^*l} \tag{3}$$

where v is velocity, g is gravitational acceleration, and l is leg length (height of the greater trochanter during standing). For each footwear condition, participants were acclimated by running for at least three minutes. After three minutes we asked participants if they felt comfortable, and visually assessed their gait for consistent stride frequency and joint kinematics in both legs. Once these criteria were satisfied, which took 3–4 min total, we recorded them running for 30 s.

2.3. Data processing

We analyzed ten strides from each recorded trial and used Visual3D v.6 (C-Motion Inc., Germantown, MD, USA) for 3-D kinematics calculations. We filtered marker and ground reaction force data with fourth-order lowpass Butterworth filters with 10 and 50 Hz cutoff frequencies, respectively. We analyzed kinetic and kinematic variables using custom-written MATLAB (The MathWorks, Natick, MA, USA) scripts.

We calculated sandal stiffness (k_{ftwr}) for each condition using the E of the material, along with the area of force application (A_{foot}), based on the equations

$$E = \frac{F/A_{foot}}{\Delta x/x} \tag{4}$$

and

$$k_{ftwr} = \frac{F}{\Delta x} \tag{5}$$

where x is sandal thickness (1.2 cm, all sandals), Δx is change in sandal thickness, and F is force applied to the sandal. We could not measure Δx empirically, so we derived the following equation from Eqs. (4) and (5):

$$k_{ftwr} = \frac{A_{foot}{}^*E}{x} \tag{6}$$

We determined A_{foot} for each participant by recording them running over an Emed q-100 pressure platform (novel GmbH, Munich, Germany) at a self-selected speed. A_{foot} was calculated as the foot contact area at maximum total pressure during the running step.

We calculated leg stiffness (k_{leg}) in the sagittal plane as quasi-stiffness following Liew et al. (2017):

$$k_{leg} = \frac{maxF_{leg}}{\Delta l_{leg}} \tag{7}$$

where max F_{leg} is the peak force on the leg in the sagittal plane, and Δl_{leg} is the change in leg length. We measured leg length as the distance from the greater trochanter to the center of pressure under the foot in the sagittal plane, and calculated F_{leg} as the dot product of the sagittal plane ground reaction force and leg length vectors. We calculated Δl_{leg} as the change in leg length from the beginning of stance phase to the moment of maximum F_{leg}, and deducted the change caused by sandal compression (Δx), which we determined by rearranging Eq. (4) as

$$\Delta x = \frac{\max F_{leg}{}^*x}{E^*A_{foot}} \tag{8}$$

We could not use a 3-D multi-segment foot kinematics model because the sandals restricted marker placement on participants' feet (Fig. 1a). Following Holowka et al. (2021), we calculated LA stiffness as quasi-stiffness during loading by modeling the foot with two segments: the rearfoot, defined by the markers on the posterior calcaneus and navicular tuberosity; and the forefoot, defined by the markers on the navicular tuberosity and the first metatarsal head (Fig. 1b). LA motion, θ_{LA}, was calculated as the angle between these segments, projected onto the sagittal plane of the overall foot segment. We calculated LA compression ($\Delta\theta_{LA}$) as the difference between θ_{LA} at initial foot contact and maximum θ_{LA} during running. To estimate LA loading, we calculated the external moment (M_{LA}) about the navicular tuberosity caused by the ground reaction force vector in the sagittal plane, and assumed M_{LA} was 0 before the center of pressure crossed anterior to the navicular tuberosity (Fig. 1c). We calculated LA quasi-stiffness during loading (k_{LA}) as the slope of the linear least squares regression line fit to the corresponding values of θ_{LA} and M_{LA}, up to the peak M_{LA}.

To assess possible effects of foot strike posture, we calculated the projected sagittal plane angle between the long axis of the foot (the vector between the posterior calcaneus and the fifth metatarsal head markers) and the treadmill surface. We subtracted this angle at the first frame of stance from the angle during standing to calculate foot strike angle (FSA), with more positive values indicating more rearfoot-type strikes (Lieberman et al., 2010). We also calculated $\Delta\theta_{LA}$, peak vertical ground reaction force (vGRF), duty factor, stride frequency, and foot contact time to determine whether the different sandal materials caused participants to use different support mechanics. See Table 1 for definitions of the abbreviations of variables analyzed in this study.

2.4. Statistical analysis

After removing steps with gaps in marker tracking, we analyzed 8–10 steps per condition per participant. For statistical analyses we used average values calculated from all steps within participants. Statistical tests were carried out using R v. 3.6.1 (R Core Team, 2019). We performed Shapiro-Wilkes normality tests on each variable, and visually checked data for similar variance across conditions. We log-transformed k_{LA} prior to analysis to make it normally distributed. We used the 'lme4' package (Bates et al., 2015) to construct linear mixed effects models, with participant identity as a random effect, and used these models to test for the effects of footwear condition on k_{syst}, k_{leg}, k_{LA}, $\Delta\theta_{LA}$, FSA, peak vGRF, duty factor, stride frequency, and foot contact time. We included FSA and stride frequency as covariates in the models for k_{syst}, k_{leg}, k_{LA}, $\Delta\theta_{LA}$, as these variables are associated with leg and LA stiffness (Günther and Blickhan, 2002; Holowka et al., 2021). Because residual and q-q plots confirmed that model residuals were homoscedastic and normal, we performed omnibus Type 3 ANOVAs to test for differences among footwear conditions. When differences were detected, we used the 'lsmeans' package (Lenth, 2016) to conduct post-hoc pairwise contrasts between footwear conditions, with a Holm-Bonferroni P-value correction ($\alpha = 0.05$).

3. Results

Average k_{ftwr} values are presented in Table 2, average values for all other variables are presented in Table 3, and results of statistical tests are presented in Table 4 (omnibus tests) and Table 5 (pairwise tests). Average values of vGRF, Δl_{leg} and $\Delta\theta_{LA}$ over the duration of stance phase

Table 2
Sandal material and mechanical properties.

Sandal Type	Material	Young's modulus (kN/m^2)	Compressive Stiffness (kN/m)*
EVA	Ethylene vinyl acetate	8400	8230 ± 1627
R60	Reoflex 60 polyurethane	2860	2802 ± 554
R30	Reoflex 30 polyurethane	565	554 ± 109

* Average ± SD for all participants. Calculated for each participant following Eq. (6). The average area of force application used in these equations (A_{foot}) was 118 ± 23 cm^2.

Table 3
Average values for spatiotemporal, kinetic, and kinematic variables across all footwear conditions.

Variable	Barefoot	EVA	R60	R30
$vGRF_{max}$ (BW)	2.33 (0.18)	2.4 (0.21)	2.4 (0.19)	2.42 (0.21)
Stride frequency (s^{-1})	1.47 (0.079)	1.45 (0.083)	1.44 (0.077)	1.44 (0.078)
Contact time (s)	0.238 (0.02)	0.242 (0.021)	0.243 (0.019)	0.241 (0.02)
Foot strike angle (°)	3.47 (7.61)	3.48 (8.66)	3.22 (7.42)	2.7 (7.66)
Duty factor	0.35 (0.026)	0.349 (0.028)	0.349 (0.026)	0.346 (0.027)
k_{syst} (kN m^{-1})	25.81 (8.27)	25.56 (8.33)	25.08 (7.11)	25.44 (7.68)
k_{leg} (kN m^{-1})	25.81 (8.27)	25.65 (8.38)	25.33 (7.26)	26.84 (8.43)
k_{LA} (Nm $°^{-1}$)	11.55 (4.43)	11.48 (4.32)	11.06 (4.3)	11.08 (4.38)
$\Delta\theta_{LA}$ (°)	15.94 (4.45)	14.39 (3.57)	15.42 (3.81)	15.14 (4.11)

All variables are presented as mean (SD).

are plotted in Fig. 2.

Compared to running barefoot, when running in sandals participants had 2.7–3.6% greater peak vGRFs ($P < 0.002$) (Fig. 2a), and 1.7–2.3% lower stride frequencies ($P < 0.0002$). There were no significant differences in peak vGRF or stride frequency among different sandal conditions, and no significant differences in k_{syst}, contact time, FSA, or duty factor across all footwear conditions.

For leg stiffness, average load-versus-deformation relationships for the variables F_{leg} and Δl_{leg} are presented in Fig. 3a. Footwear condition had a significant overall effect on k_{leg} ($P = 0.046$), but post-hoc tests revealed no significant differences among individual pairs of conditions ($P > 0.05$) (Fig. 3b). Regarding longitudinal arch stiffness, the average load-versus-deformation relationships for the variables M_{LA} and $\Delta\theta_{LA}$ are presented in Fig. 3c. Footwear condition did not have a significant effect on k_{LA} ($P = 0.81$) (Fig. 3d). However, when running barefoot, participants had 10.3% greater $\Delta\theta_{LA}$ than when running in EVA sandals ($P = 0.004$), and 5.9% greater $\Delta\theta_{LA}$ than when running in R30 sandals ($P = 0.04$) (Fig. 2). There were no significant differences in $\Delta\theta_{LA}$ among the sandal conditions ($P > 0.05$).

4. Discussion

This study investigated how varying footwear stiffness affects leg and longitudinal arch (LA) stiffness during running in four footwear conditions: barefoot, and three sandal types made from materials spanning an order of magnitude in Young's modulus. Footwear stiffness had a significant overall effect on leg stiffness, but post-hoc comparisons revealed no significant differences in leg stiffness among pairs of footwear conditions. Contrary to predictions from the in-series spring model, we found no effect of footwear stiffness on LA stiffness. However, we did find that LA compression was significantly greater when running barefoot than when running in two of the three sandal types, indicating that footwear use generally affects LA function.

The absence of a significant effect of footwear stiffness on leg stiffness in post-hoc tests runs contrary to the findings of Kulmala et al. (2018), possibly because their study compared commercially available shoes that differed in multiple design features, and did not specifically isolate the effect of midsole compressive stiffness, as in our study. For example, the shoes they compared differed in heel-to-forefoot height offset, which could affect foot strike angle and thus leg stiffness (Günther and Blickhan, 2002). In contrast, we found that even a greater than tenfold variation in footwear stiffness failed to effect significant change in leg stiffness. This consistency in leg stiffness across footwear conditions makes sense when considering the in-series spring relationship described in Eq. (2). The surface stiffness (k_{surf}) of the steel plate underneath the treadmill belt was sufficiently high to render its effect on leg stiffness negligible, such that

$$\frac{1}{k_{syst}} = \frac{1}{k_{leg}} + \frac{1}{k_{ftwr}} \tag{9}$$

Thus, when running barefoot, this relationship simplifies to

$$\frac{1}{k_{syst}} = \frac{1}{k_{leg}} \tag{10}$$

The average k_{leg} across participants when barefoot was 25.8 kN/m, and therefore we assume that with footwear, participants would adjust

Table 4
Results of omnibus statistical tests.

Variable	Footwear Condition		Stride Frequency		Foot Strike Angle	
	Chi-square	P	Chi-square	P	Chi-square	P
$vGRF_{max}$	30.8	**<0.0001**	–	–	–	–
Stride frequency	49.8	**<0.0001**	–	–	–	–
Contact time	7.35	0.06	–	–	–	–
Foot strike angle	2.20	0.53	–	–	–	–
Duty factor	2.68	0.44	–	–	–	–
k_{syst}	1.55	0.67	6.88	**0.009**	10.4	**0.001**
k_{leg}	7.95	**0.047**	6.30	**0.012**	10.1	**0.001**
k_{LA}	0.96	0.81	0.92	0.34	27.5	**<0.0001**
$\Delta\theta_{LA}$	14.6	**0.0022**	3.81	0.051	20.1	**<0.0001**

Omnibus tests were Type 3 ANOVAs conducted on model variance from linear mixed models where participant identity was set as a random effect. For k_{syst}, k_{leg}, k_{LA}, and $\Delta\theta_{LA}$, stride frequency and foot strike angle were included as model covariates. Bold: $P < 0.05$.

Table 5
Results of post-hoc pairwise comparisons.

Variable	BF vs. EVA		BF vs. R30		BF vs. R60		EVA vs. R30		EVA vs. R60		R30 vs. R60	
	t	P	t	P	t	P	t	P	t	P	t	P
$vGRF_{max}$	−4.00	**0.0010**	−5.12	**<0.0001**	−4.13	**0.0007**	−1.12	0.68	−0.12	1.00	1.00	0.75
Stride frequency	4.63	**0.0001**	6.01	**<0.0001**	6.16	**<0.0001**	1.37	0.52	1.52	0.43	0.15	1.00
k_{leg}	−0.82	0.85	−2.57	0.059	−0.868	0.82	−2.07	0.18	−0.11	1.00	1.98	0.21
$\Delta\theta_{LA}$	3.55	**0.0037**	2.67	**0.044**	1.91	0.23	−0.77	0.87	−1.64	0.36	−0.87	0.82

Post-hoc tests were carried out on variables for which omnibus tests revealed significant effects ($P < 0.05$) of footwear condition. These tests were conducted using model variance from linear mixed effects models. P-value corrected using a Holm-Bonferroni correction. Bold: $P < 0.05$.

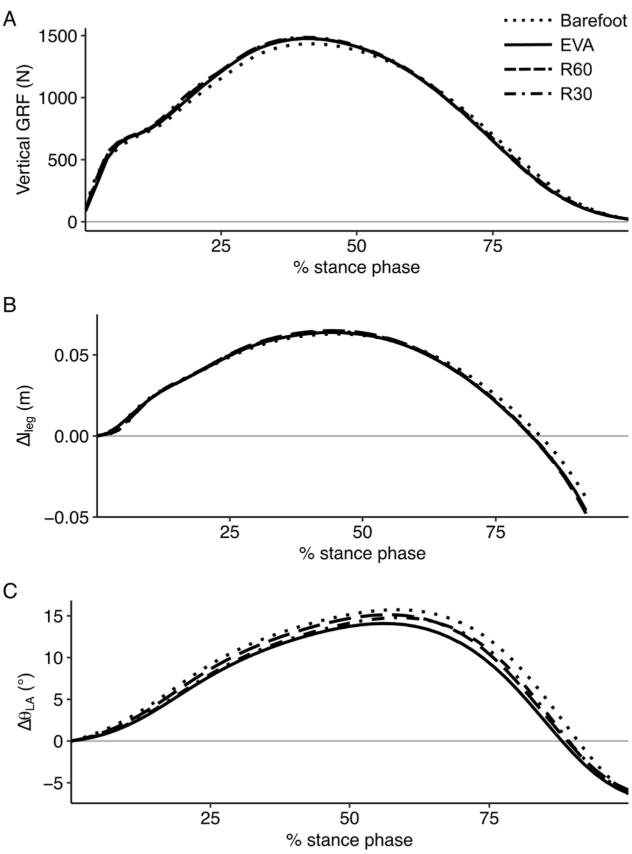

Fig. 2. Average vertical ground reaction force, leg compression, and longitudinal arch compression across stance phase. (A) Vertical ground reaction force (GRF). (B) Leg compression (Δl_{leg}), with '0′ set as the value at initial foot contact. (C) Longitudinal arch compression ($\Delta\theta_{LA}$), with '0′ set as the value at initial foot contact. In all plots, lines represent the average value among all participants (N = 20) at each percentage of stance.

their leg stiffness to maintain a k_{syst} of roughly 25.8 kN/m across all sandal types. The average stiffness calculated for our most compliant sandal (R30) was 554 kN/m. Solving for k_{leg} in Eq. (9) above predicts that individuals should increase their leg stiffness by only 5% in R30 sandals compared to running barefoot to maintain a constant k_{syst}. We found that participants did increase their leg stiffness by about 4% on average when running in R30 sandals versus running barefoot, but that this difference was not enough to achieve statistical significance with a sample of 20 participants ($P = 0.059$). We did not perform a power analysis prior to data collection, and so it is possible that this sample size did not provide enough power to detect a significant difference in post-hoc pairwise comparisons, leaving open the possibility of a small effect of footwear midsole stiffness on leg stiffness. It is worth noting that our results for R30 sandals are consistent with those of Kerdok et al. (2002), who found a < 5% difference in average leg stiffness among participants running on 454 kN/m surfaces compared to 945 kN/m surfaces. However, the effects of standard, commercially available running shoes on leg stiffness are likely to be considerably smaller still than those suggested here for R30 sandals.

The stiffest sandal in this study was composed of the standard EVA polymer used in running shoe midsoles. The major difference in compressive stiffness between our EVA sandals and traditional running shoe midsoles should be due to midsole thickness: all of our sandals were only 1.2 cm-thick, whereas midsole thickness in running shoes tends to range from 2 to 4.5 cm. Had we constructed EVA sandals with thickness near the top of this range by increasing midsole thickness by four times, this design would also decrease the midsole stiffness by a factor of four, following Eq. (6). Even so, these thicker sandals would still be roughly four times stiffer than the most compliant sandal that we tested (R30). Thus, by virtue of this relatively high midsole stiffness, we expect that standard running shoes elicit minimal effects on leg stiffness during running.

Our second main finding, that the compressive stiffness of footwear midsoles does not affect LA stiffness, is consistent with the recent results of Birch et al. (2021), who found that surface stiffness did not affect LA stiffness during hopping. Our results agree with their general conclusion that LA stiffness is not tuned to surface or footwear stiffness. However, we found that LA compression was higher during barefoot than shod

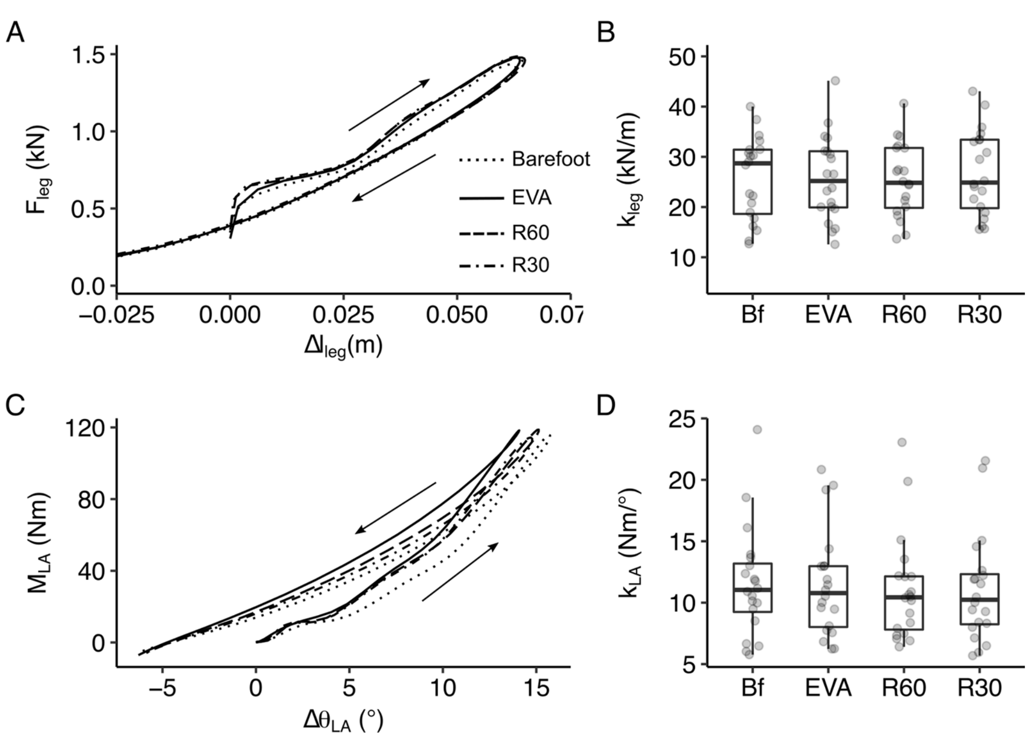

Fig. 3. Variables used to calculate leg and longitudinal arch stiffness. (A) Leg compression (Δl_{leg}) plotted against projected leg force (F_{leg}). (B) Leg stiffness (k_{leg}) boxplots. (C) LA compression ($\Delta\theta_{LA}$) plotted against the longitudinal arch moment (M_{LA}). (D) Longitudinal arch stiffness (k_{LA}) boxplots. Lines (A and C) depict average values calculated from all participants (N = 20), arrows indicate loading (up) and unloading (down). Boxplots (B and D) depict median, upper and lower quartiles and 1.5x interquartile ranges for all participants (N = 20).

running, despite the fact that there were no differences in LA stiffness among footwear conditions. This finding is in concordance Kelly et al. (2016), who also found that individuals increased their intrinsic foot muscle activation when running in shoes compared to when running barefoot. Birch et al. hypothesized that running in conventional shoes may require greater intrinsic foot muscle activation to replace the energy dissipated by viscoelastic materials in the midsole, which could in turn reduce arch compression. Further studies involving electromyography are necessary to evaluate this hypothesis. Regardless, these results support the hypothesis that humans use different LA mechanics when running barefoot than in footwear (Perl et al., 2012).

This study had several limitations. First, to control footwear stiffness, we used sandals made of different types of materials: standard EVA and polyurethane co-polymers (Reoflex). The latter were denser, making the R30 and R60 sandals up to 200 g heavier than the EVA sandals. It is possible that mass differences may have slightly affected overall running mechanics. However, none of the spatiotemporal gait variables or peak ground reaction forces were different among the sandal conditions, indicating that the sandal materials did not cause participants to change their standard support mechanics in any significant way. Additionally, we could not calculate resilience from our material properties tests, and it is possible that the sandal materials had different resilience values. Even if this were the case, it most likely would not have affected our results, since our equations only used leg and LA compression up to peak loads, and therefore the recoil of the sandal material would not affect our stiffness calculations. Because there were no significant differences in spatiotemporal variables among sandal conditions, we also do not think differences in material resilience could have affected overall support mechanics. Finally, our LA stiffness and compression calculations relied on a simple, 2D two-segment foot model, and we assumed that the LA was loaded only when the center of pressure was anterior to the navicular tuberosity (Holowka et al., 2021). Nevertheless, this model produced similar LA compression and stiffness calculations to those of a previous study that used a multi-segment foot model (Farris et al., 2019), as well as a study that used a slightly different two-segment foot model (Kelly et al., 2018). Therefore, we are confident that our results would not have changed with more sophisticated foot models.

Overall, our findings demonstrate that use of footwear in general and degree of footwear compressive stiffness in particular do not appreciably alter leg stiffness during running, because even the most compliant footwear is likely too stiff to elicit the effects previously observed when individuals run on more compliant surfaces. These results have important implications for evaluating the effects of ultra-thick-midsoled running shoes, such as the highly cushioned ‘maximalist’ shoes manufactured by companies like Hoka One One, or high-performance racing shoes like the Vaporfly line produced by Nike. Specifically, our results suggest that the thick midsoles (>3 cm) of these shoes should not cause significant alterations in leg stiffness during running. However, recent studies have demonstrated that footwear *bending* stiffness can affect other aspects of gait, such as push-off mechanics (Cigoja et al., 2020; Day and Hahn, 2021), and thus further research on other possible effects of footwear compressive stiffness during running is warranted. Additionally, while we found no effect of footwear stiffness on LA stiffness, humans appear to restrict LA compression when shifting from barefoot to shod running. This phenomenon deserves further attention, as it may help us better understand the basic function of the human longitudinal arch.

CRediT authorship contribution statement

Nicholas B. Holowka: Conceptualization, Methodology, Software, Formal analysis, Investigation, Writing – original draft, Writing – review & editing, Visualization. **Stephen M. Gillinov:** Conceptualization, Methodology, Investigation, Writing – original draft, Writing – review & editing. **Emmanuel Virot:** Conceptualization, Methodology, Resources, Writing – review & editing. **Daniel E. Lieberman:** Resources, Writing – review & editing, Supervision, Funding acquisition.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Acknowledgements

Thanks to Freddy Sichting and two anonymous reviewers for helpful comments on an earlier draft of the manuscript. Funding for this study came from a grant from the American School of Prehistoric Research (to DEL).

Data Statement

All data are available from the corresponding author upon reasonable request.

References

Alexander, R.M., 1991. Energy-saving mechanisms in walking and running. J. Exp. Biol. 69, 55–69.
Alexander, R.M., Jayes, A.S., 1983. A dynamic similarity hypothesis for the gaits of quadrupedal mammals. J. Zool. 201, 135–152. https://doi.org/10.1111/j.1469-7998.1983.tb04266.x.
Bates, D., Mächler, M., Bolker, B., Walker, S., 2015. Fitting linear mixed-effects models using lme4. J. Stat. Softw. 67, 1–48.
Birch, J.V., Kelly, L.A., Cresswell, A.G., Dixon, S.J., Farris, D.J., 2021. Neuromechanical adaptations of foot function to changes in surface stiffness during hopping. J. Appl. Physiol. 130 (4), 1196–1204.
Bishop, M., Piolkowski, P., Conrad, B., Brunt, D., Horodyski, M., 2006. Athletic footwear, leg stiffness, and running kinematics. J. Athl. Train. 41, 387–392.
Blickhan, R., 1989. The spring-mass model for running and hopping. J. Biomech. 22 (11-12), 1217–1227.
Butler, R.J., Crowell, H.P., Davis, I.M., 2003. Lower extremity stiffness: implications for performance and injury. Clin. Biomech. 18 (6), 511–517. https://doi.org/10.1016/S0268-0033(03)00071-8.
Cappozzo, A., Catani, F., Della Croce, U., Leardini, A., 1995. Position and orientation in space of bones during movement: anatomical frame definition and determination. Clin. Biomech. 10 (4), 171–178.
Cigoja, S., Asmussen, M.J., Firminger, C.R., Fletcher, J.R., Edwards, W.B., Nigg, B.M., 2020. The effects of increased midsole bending stiffness of sport shoes on muscle-tendon unit shortening and shortening velocity: a randomised crossover trial in recreational male runners. Sport. Med. - Open 6 (1). https://doi.org/10.1186/s40798-020-0241-9.
Day, E.M., Hahn, M.E., 2021. Does running speed affect the response of joint level mechanics in non-rearfoot strike runners to footwear of varying longitudinal bending stiffness? Gait Posture 84, 187–191. https://doi.org/10.1016/j.gaitpost.2020.11.029.
De Wit, B., De Clercq, D., Aerts, P., 2000. Biomechanical analysis of the stance phase during barefoot and shod running. J. Biomech. 33 (3), 269–278. https://doi.org/10.1016/S0021-9290(99)00192-X.
Divert, C., Mornieux, G., Baur, H., Mayer, F., Belli, A., 2005. Mechanical comparison of barefoot and shod running. Int. J. Sports Med. 26 (7), 593–598. https://doi.org/10.1055/s-2004-821327.
Farley, C.T., Glasheen, J., McMahon, T.A., 1993. Running springs: speed and animal size. J. Exp. Biol. 185, 71–86.
Farris, D.J., Kelly, L.A., Cresswell, A.G., Lichtwark, G.A., 2019. The functional importance of human foot muscles for bipedal locomotion. Proc. Natl. Acad. Sci. 116 (5), 1645–1650. https://doi.org/10.1073/pnas.1812820116.
Farris, D.J., Birch, J., Kelly, L., 2020. Foot stiffening during the push-off phase of human walking is linked to active muscle contraction, and not the windlass mechanism. J. R. Soc. Interface 17, 20200208. https://doi.org/10.1098/rsif.2020.0208.
Ferris, D.P., Farley, C.T., 1997. Interaction of leg stiffness and surface stiffness in human hopping. J. Appl. Physiol. 82, 15–22.
Ferris, D.P., Louie, M., Farley, C.T., 1998. Running in the real world: adjusting leg stiffness for different surfaces. Proc. R. Soc. B Biol. Sci. 265 (1400), 989–994.
Günther, M., Blickhan, R., 2002. Joint stiffness of the ankle and the knee in running. J. Biomech. 35 (11), 1459–1474.
Holowka, N.B., Richards, A., Sibson, B.E., Lieberman, D.E., 2021. The human foot functions like a spring of adjustable stiffness during running. J. Exp. Biol. 224, 219667 https://doi.org/10.1242/jeb.219667.
Kelly, L.A., Cresswell, A.G., Racinias, S., Whiteley, R., Lichtwark, G., 2014. Intrinsic foot muscles have the capacity to control deformation of the longitudinal arch. J. R. Soc. Interface 11, 20131188. https://doi.org/10.1098/rsif.2013.1188.

Kelly, L.A., Lichtwark, G., Cresswell, A.G., 2015. Active regulation of longitudinal arch compression and recoil during walking and running. J. R. Soc. Interface 20141076. https://doi.org/10.1098/rsif.2014.1076.

Kelly, L.A., Lichtwark, G.A., Farris, D.J., Cresswell, A., 2016. Shoes alter the spring-like function of the human foot during running. J. R. Soc. Interface 13 (119), 20160174. https://doi.org/10.1098/rsif.2016.0174.

Kelly, L.A., Farris, D.J., Lichtwark, G.A., Cresswell, A.G., 2018. The influence of foot-strike technique on the neuromechanical function of the foot. Med. Sci. Sports Exerc. 50, 98–108.

Ker, R.F., Bennett, M.B., Bibby, S.R., Kester, R.C., Alexander, R.M., 1987. The spring in the arch of the human foot. Nature 325 (6100), 147–149. https://doi.org/10.1038/325147a0.

Kerdok, A.E., Biewener, A.A., McMahon, T.A., Weyand, P.G., Herr, H.M., 2002. Energetics and mechanics of human running on surfaces of different stiffnesses. J. Appl. Physiol. 92 (2), 469–478.

Kulmala, J.P., Kosonen, J., Nurminen, J., Avela, J., 2018. Running in highly cushioned shoes increases leg stiffness and amplifies impact loading. Sci. Rep. 8, 17496.

Lenth, R.V., 2016. Least-squares means: the R package lsmeans. J. Stat. Softw. 69, 1–33. https://doi.org/10.18637/jss.v069.i01.

Lieberman, D.E., Venkadesan, M., Werbel, W.A., Daoud, A.I., D'Andrea, S., Davis, I.S., Ojiambo Mang'Eni, R., Pitsiladis, Y., 2010. Foot strike patterns and collision forces in habitually barefoot versus shod runners. Nature 463, 531–535. https://doi.org/10.1038/nature08723.

Liew, B.X.W., Morris, S., Masters, A., Netto, K., 2017. A comparison and update of direct kinematic-kinetic models of leg stiffness in human running. J. Biomech. 64, 253–257. https://doi.org/10.1016/j.jbiomech.2017.09.028.

Perl, D.P., Daoud, A.I., Lieberman, D.E., 2012. Effects of footwear and strike type on running economy. Med. Sci. Sports Exerc. 44, 1335–1343. https://doi.org/10.1249/MSS.0b013e318247989e.

R Core Team, 2019. R: a language and environment for statistical computing. R Foundation for Statistical Computing, Vienna, Austria https://www.R-project.org/.