Contribution of the transverse arch to foot stiffness in humans

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Abstract

Stiffness of the human foot is central to its mechanical function, such as elastic energy storage and propulsion. Its doubly-arched structure, manifested as longitudinal and transverse arches, is thought to underlie the stiff nature. However, previous studies have focused solely on the longitudinal arch, and little is known about whether and how the transverse arch impacts the foot’s stiffness. The common observation that a flexible currency bill significantly stiffens upon curling it transversally underlies our hypothesis that the transverse arch dominates the foot’s stiffness. Through human subject experiments, we show that transversally reinforcing the ball of the foot by means of sports tape can increase the midfoot bending (dorsiflexion) stiffness by 53% on average. Such a result is possible if and only if midfoot bending is mechanically coupled with splaying of the distal metatarsal heads, thus engaging the distal transverse ligament. Using a simplified foot model, we elucidate a mechanism for such coupling to arise because of the transverse arch; namely transverse curvature couples sagittal plane bending with stretching along the transverse direction. We therefore conclude that the previously ignored distal metatarsal transverse ligaments, by way of the transverse arch, likely underlie the midfoot’s elastic response by an amount comparable to or surpassing the plantar fascia, the windlass mechanism and other longitudinal tissues. Our findings impact current practice in clinical, sports and evolutionary biomechanics that focus solely on the longitudinal arch and on two-dimensional sagittal plane mechanics.

1 Introduction

The elastic response of the foot serves multiple roles in the healthy locomotory behavior of humans. These include reducing the energetic cost of locomotion (Ker et al., 1987; Bramble and Lieberman, 2004; Kuo et al., 2005; Zelik and Kuo, 2010), transmitting propulsive forces to the ground (Bojsen-Møller, 1979; Susman, 1983; Bramble and Lieberman, 2004), and supporting the body’s weight without injurious deformations (DeSilva and Gill, 2013; Hastings et al., 2014). The multiple functionalities can however impose competing demands on its response as a mechanical device. Consider for example two contrasting roles that the foot plays in reducing the energetic cost of walking, which impose opposing demands on its stiffness. For the purpose of elastic energy storage the foot undergoes significant deformation from early to mid stance (Holowka et al., 2017), and therefore has to remain sufficiently flexible (Ker et al., 1987; Huang et al., 1993; Zelik and Kuo, 2010). By contrast, it behaves as a stiff propulsive lever during push-off and applies impulsive forces on the ground (Bojsen-Møller, 1979).
Kuo et al., 2005), i.e. the foot undergoes relatively little deformation (Holowka et al., 2017). Both of these beneficial stiffness properties of the human foot are thought to arise because of its arched structure (Morton, 1922; Venkadesan et al., 2017 preprint). However, present understanding of the relationship between the arched structure and mechanical function as an elastic body remains limited to studies that focus solely on the sagittal plane mechanics of the foot, and the longitudinal arch (Morton, 1924b; Elftman H., 1935; Elftman and Manter, 1935; Jones, 1941; Hicks, 1954; Morton, 1964; Bojesen-Møller, 1979; Susman, 1983; Ker et al., 1987; Williams and McClay, 2000; Prang, 2016). Recent work using mathematical models and physical replicas of feet however have hypothesized a more central role for the transverse arch (Venkadesan et al., 2017; Venkadesan et al., 2017 preprint). The mechanism by which the transverse arch influences foot stiffness is the topic of this paper.

We first consider what is known about the longitudinal arch, its role in stiffness of the foot, and the gaps in our understanding of foot elasticity. The medial longitudinal arch (MLA) is regarded as the main contributor to the elastic response within the sagittal plane (Morton, 1924a; Williams and McClay, 2000). When the forefoot is loaded, the ground reaction force bends the foot along its length, and the plantar fascia are stretched (Morton, 1924a; Ker et al., 1987; Huang et al., 1993). This sagittal plane response is analogous to the elasticity of the string that spans across a bow. The stretch in the plantar fascia when the foot is bent is proportional to the height of the arch (Williams and McClay, 2000; Venkadesan et al., 2017). This forms the rationale for the prevalent hypothesis that the height of the MLA predicts the stiffness of the foot when the forefoot is loaded. Partial support for this is found from cadaver studies that transect the plantar fascia and find that the foot’s stiffness is lowered, albeit by less than 25% (Ker et al., 1987; Huang et al., 1993). An additional involvement of the plantar fascia arises from the windlass mechanism (Hicks, 1954). The two ends of the plantar fascia attach to the calcaneus and the phalanges, proximally and distally, respectively. When the toes are extended, such as during push-off, the plantar fascia are stretched and may increase the foot’s stiffness because of both geometric and material effects. The geometric effect is that the increased plantar fascia tension causes the midfoot height to increase (Hicks, 1954; Gelber et al., 2014). The material effect arises from the fascia being comprised of tough elastic fibers that exhibit non-Hookean strain-stiffening elasticity (Kitaoka et al., 1994). However, neither of these hypotheses, the bow-string or the windlass, suffice to account for the remarkable stiffness of the human foot when it sustains substantial forefoot loads. Cadaver estimates show that incorporating the plantar fascia leads to rise in foot stiffness of less than a third (Ker et al., 1987; Huang et al., 1993). By contrast, the arched human feet are 2–3 fold stiffer than flat primate feet (Venkadesan et al., 2017 preprint). A comparison between healthy and flatfooted humans highlights the insufficiency of the mechanical understanding of the foot that is focused solely on the sagittal plane. Kido et al. (2013) compare healthy versus flatfooted subjects and find that under body weight loading, the hallux of flatfooted subjects dorsiflexes (relative to the cuneiforms) twice as much as for healthy people, i.e. the midfoot deforms twice as much. By this measure, healthy arched feet are ≈ 100% stiffer than flatfooted ones, much more than accounted for by the plantar fascia. The windlass mechanism also cannot account for the substantially increased stiffness of healthy versus flatfooted subjects. The engagement of the windlass mechanism relies solely on the diameter of the distal metatarsal region (Hicks, 1954), which is unaffected in the flatfooted subjects. X-ray recordings of the foot during toe-extension show that the windlass is indeed engaged for flatfooted subjects (Gelber et al., 2014). However, in heel-raise tests where the toe is free to extend, flatfooted subjects were unable to support their body weight on their forefoot (Hastings et al., 2014). Furthermore, studies measuring the effect
of longitudinal arch morphology on injury in humans have remained inconclusive (Murphy et al., 2003; Tong and Kong, 2013). These multiple lines of evidence point to gaps in our understanding of the mechanics of the foot, which has so far remained focused solely on the sagittal plane.

Curvature-induced stiffness because of the transverse arch emerges as an alternative hypothesis (Venkadesan et al., 2017 preprint). In particular, the transverse tarsal arch (TTA) (Venkadesan et al., 2017 preprint; Venkadesan et al., 2017) is regarded as the main contributor to the foot’s stiffness. The hypothesized role of the TTA is that it couples bending of the foot within the sagittal plane with stretching of elastic tissues that are orthogonal to the sagittal plane (Venkadesan et al., 2017 preprint). This is analogous to how a thin elastic material (say, a currency bill) significantly stiffens upon curving it in the transverse direction. Such curvature-induced stiffness is known to play a dominant role in other propulsive organs such as rayed fish fins (Nguyen et al., 2017), but has not been previously tested in human feet. Casual observation of the foot under forefoot loading shows that the metatarsals indeed splay apart (Fig. 1a). However, it remains to be shown that the transverse ligaments resist this splay, and it is not simply the soft tissues of the sole being compressed.

Our hypothesized mechanism for transverse curvature-induced stiffness of human feet is illustrated by the model shown in Fig. 1b. Each metatarsal preferentially dorsiflexes along the normal to the transverse arch, and loading the forefoot would splay the distal metatarsal heads apart. This splay is resisted by the ligaments and other soft tissues spanning the transverse direction at the distal end. Therefore, increasing stiffness along the transverse direction at the distal region is predicted to increase the foot’s overall stiffness. In this paper, we experimentally test this prediction by developing techniques to quantify foot stiffness, and measure the effect of externally increasing the transverse stiffness at the distal metatarsal heads.

2 Methods and materials

2.1 Quantification of foot stiffness

Ability of the foot to support loads on the forefoot is quantified by the vertical displacement of the forefoot (relative to the rear) in response to vertical loads at the ball of the foot. This ratio, of the vertical forefoot force to the vertical displacement of the forefoot, is a measure of the foot’s overall stiffness. Three-point bending tests, such as those performed on cadaver feet (Ker et al., 1987; Huang et al., 1993), are used to estimate the foot’s overall stiffness. The estimated stiffness of healthy feet is around 0.7 body weights/mm, i.e. several body weight loads have to be applied to the forefoot in order to obtain a displacement of even a few millimeters. Therefore, safety and comfort issues complicate direct load-displacement measurements with live subjects. We use an alternative estimation method, based on the well-known windlass mechanism of the foot (Hicks, 1954; Gelber et al., 2014). The plantar fascia (also known as the plantar aponeuroses) originate at the calcaneus and pass under the ball of the foot before attaching to the toes. On toe extension, the fascia are stretched, pulling the heel and the forefoot closer together, and raise the midfoot away from the ground (Fig. 2a). The midfoot rises in inverse proportion to the stiffness of the foot. The rise in arch height per degree of toe extension $\Delta h/\Delta \theta$ therefore quantifies overall foot stiffness, and a lower $\Delta h/\Delta \theta$ implies a stiffer foot. In Appendix A we analytically derive the relationship between the midfoot height rise and foot stiffness and show that

$$\frac{\Delta h}{\Delta \theta} \propto \frac{1}{k_f}$$  \hspace{1cm} (1)
Figure 1: Bending-stretching coupling in the foot. (a) Upon loading the forefoot with body weight, the distal metatarsals splay apart in the transverse direction. The overlaid images are before and after loading the forefoot. (b) A simplified mathematical model, with three metatarsals whose bending axes (torsional springs at the proximal end) are misaligned due to the transverse arch, illustrated the hypothesized mechanism for this splay. Distal metatarsal ligaments (linear springs at the distal end) resist the metatarsals splaying, and thus stiffen the foot.
Figure 2: Experimental setup and protocol. (a) Overlaid images of a subject’s taped foot showing retro reflective markers before and after extending the toes by $\Delta \theta$, which led to a rise in the arch height by $\Delta h$. (b) Custom built rig to support subjects’ measurement foot. It comprises of a fixed heel support, a low-friction slider under the forefoot that is free to move in the longitudinal direction, and a toe extension plate attached to the forefoot support with a lock equipped hinge. A thin sliding plastic sheet on the toe extension plate reduced sliding friction and skin shear under the toe. (c) Subjects received live animated feedback on a computer screen in front of them from the force plates about (i) foot load as a fraction of their body weight, and (ii) the center of pressure under the forefoot. Subjects were instructed to keep indicators within their respective targets. (d) Sample traces of the longitudinal center of pressure location and body weight fraction under the test foot. Gray regions indicate when the load parameters were outside the acceptable range, and these samples were not used in the analyses.
where $k_f$ is the stiffness of the foot defined as the ratio of the applied vertical force at the ball of the foot to its vertical displacement.

### 2.2 Experimental protocol

We performed Yale IRB approved human subject experiments with 8 healthy, consenting volunteers (5 females, 3 males, 21–43 years) with no self-reported history of foot injury in the previous 2 years. Kinematics were tracked for 7 bony landmarks (Fig. 2b), which were identified by external palpation: first distal interphalangeal joint, medial and lateral malleoli, first and fifth metatarsophalangeal joints, navicular, and calcaneal tuberosity. Additionally, the mid dorsum was defined as the point directly above the mid point of the foot length, following standard protocols (Saltzman et al., 1995; Williams and McClay, 2000; Butler et al., 2008). With the subject standing, we used calipers to measure the truncated foot length $L$ (defined as the length from the calcaneal tuberosity to the first metatarsophalangeal joint), and the tarsal width $w$ (defined as the width from the medial navicular tuberosity to the lateral facet of the cuboid). The measurement foot was placed on a custom built rig consisting of a fixed heel support, a forefoot support free to move in the longitudinal direction, and a toe extension plate attached to the forefoot support with a lock equipped hinge (Fig. 2a). A sliding plastic sheet on the toe plate reduced sliding friction and skin shear at the toe. The other foot was placed on an adjacent force plate, and its height adjusted so that both the feet were on the same level. Foot landmarks were tracked using retro-reflective markers and near infrared motion capture cameras at 50 Hz (Vicon Motion Systems Ltd.). Foot forces were recorded using a six-axis force plate (AMTI Inc.) at 1000 Hz. As the toes were extended by the experimenter pushing on the toe plate, the toe extension angle ($\Delta \theta$) (defined as the angle of between the Toe–MTP 1 line) and change in the mid-dorsum arch height ($\Delta h$) were recorded (Fig. 2). Subjects received real-time animated feedback on a computer screen that was positioned in front of them (see Fig. 2c). The visual feedback displayed the magnitude of the vertical force (horizontal bar), and the center of pressure (circular dot) relative to the foot. The foot outline in Fig. 2c is for the reader’s benefit, and the subject could only see a box and a dot. The feedback helped to ensure that (i) body weight was distributed equally between the two feet by requiring a visual target (slider within the horizontal bar) to remain within 10% of equal load distribution, (ii) the center of pressure remained under the forefoot within a 140 mm by 70 mm rectangle (dot within a rectangle). This constraining rectangle for the center of pressure was defined for each subject during an initial calibration trial, where the subject was asked to balance on the ball, while we recorded the mean center of pressure location. Subjects were instructed to load the ball of the foot with half their body weight, and maintain the center of pressure within the constraining rectangle for the entire trial. These controls on the vertical load and the center of pressure were imposed in order apply a consistent force and moment on the foot (see appendix A), and additionally may help maintain consistent muscle activity across subjects.

The foot was tested under three conditions (i) unmodified foot with no tape (subscript $\eta_n$, e.g. $\eta_n$), (ii) taped with an elastic tape tightly wrapped around the distal metatarsal heads (subscript $\eta_t$, e.g. $\eta_t$), and (iii) control trials with the elastic tape loosely applied with no tension (subscript $\eta_c$, e.g. $\eta_c$). Length of the tape was customized for each subject to match the circumference of the forefoot, and for condition (ii) it was stretched to the maximum extent possible by hand (same experimenter for all subjects) and wrapped tightly. For (iii), it was wrapped at the forefoot without perceivable stretch. The foot was held unloaded at the time of application of the tape in (ii) and (iii).
2.3 Data Analysis

Kinematic data were smoothed by a moving average filter with a window size of 50 samples (1 s). The average of the heel, lateral malleolus and medial malleolus markers was defined as the origin for all kinematics, thereby adjusting for rigid translations of the foot. When the vertical force magnitude or the center of pressure indicators were outside the specified range (e.g. gray highlighted regions in Fig. 2), those data points were rejected. The initial and final arch heights were used to calculate $\Delta h$, and initial and final toe angles were used to calculate $\Delta \theta$. Data were averaged from two repetitive trials for each foot. For each subject and condition (taped and no-tape), $\Delta h / \Delta \theta$ was normalized by the control trial (loose tape) for each, to yield a relative stiffness measurement $\eta$ as follows: (see Appendices for derivation)

$$\eta(\bullet) = \frac{(\Delta h / \Delta \theta)_c}{(\Delta h / \Delta \theta)(\bullet)} - 1, \quad (\bullet) = t \text{ or } n.$$  \hspace{1cm} (2)

As seen from equations (1) and (2), $\eta$ quantifies the stiffness of the test conditions relative to the control, i.e. $\eta_c = 0$, and $\eta(\bullet) > 0$ implies that the foot was stiffened by the test condition ($\bullet$). Data were analyzed in MATLAB (The MathWorks Inc., Natick, MA) and statistical analyses were performed using R (R Core Team, 2015).

2.4 Statistical methods

Because $\eta$ is a measure of stiffness normalized by control, it suffices to compare it between the taped and untaped conditions, i.e. $\eta_t$ and $\eta_n$ to test for differences between the stiffnesses of these two conditions. To test the statistical significance of this difference, we performed a two-way analysis of variance (ANOVA) with $\eta$ as the dependent variable, and condition (levels: tape, no-tape) and foot (levels: left, right) as the two independent, fixed factors. The ANOVA used Type III sums-of-squares. If the ANOVA was found significant, a single comparison was carried out between the means for each factor, with a significance threshold of 0.05. We also measured the difference between $\eta_t$ and $\eta_n$ as Cohen’s $d$, using the pooled standard deviation $s$ (Cohen, 2013):

$$s = \sqrt{\frac{(n_1-1)\sigma_t^2 + (n_2-1)\sigma_n^2}{n_t + n_n - 2}} \hspace{1cm} (3)$$

$$d = \frac{\bar{\eta}_t - \bar{\eta}_n}{s} \hspace{1cm} (4)$$

The same statistical analyses were performed for the curvature normalized estimate of stiffness $\tilde{\eta}$.

3 Results

3.1 Mathematical model for curvature-induced stiffness

The influence of the transverse tape on midfoot stiffness $k_f$ depends on the strength of the mechanical coupling between midfoot dorsiflexion and metatarsal splaying. The mechanism by which the transverse arch leads to the coupling may be approximated using a mathematical model of the foot shown schematically in Fig. 1b. The mathematical analysis parametrizes the coupling strength, and thereby guides the experimental design and interpretation to test the role of the transverse arch.

The model consists of three rigid bars, representing metatarsals, each of length $L$ and separated from each other by a distance $\ell_0$. The tarsal bones are represented by a rigid base, and the ligaments connecting the metatarsals to the tarsals are represented by torsional
springs of stiffness $k_0$. In this simplified picture, each metatarsal articulates about a single axis of rotation that is predominantly oriented medio-laterally, and coincident with the rotational axis of the torsional spring. A small displacement of the distal metatarsals heads by $\delta$ that dorsiflexes the metatarsals about the respective midfoot joints, rotates the three torsional springs by a small angle $\delta/L$, leading to an elastic energy storage of,

$$E_b = 3 \times \frac{k_0}{2} \left( \frac{\delta}{L} \right)^2 .$$  \hspace{1cm} (5)

However, the transverse arch curvature $R_T$ causes the neighboring metatarsal articulation axes to be misaligned by an angle $\alpha = \ell_0/R_T$. Consequently, the aforementioned displacement of the metatarsal heads splays them apart by a distance $2\delta \sin(\alpha/2)$. The short transverse ligaments connecting the distal metatarsal heads act as springs with rest length $\ell_0$ and spring constant $k_s$. Stretching the two distal metatarsal springs, because of the splaying, leads to an additional elastic energy storage of

$$E_s = 2 \times \frac{k_s}{2} \left[ 2\delta \sin \left( \frac{\alpha}{2} \right) \right]^2 .$$  \hspace{1cm} (6)

The restoring force exerted by the structure is $F = d(E_s + E_b)/d\delta$. For small $\alpha$ corresponding to small transverse curvature, $\sin(\alpha/2) \approx \alpha/2$, and the stiffness of the foot $k_f = F/\delta$ is,

$$k_f = k_{f0} \left( \frac{R_0}{R_T} \right)^2 + 1 ,$$  \hspace{1cm} (7)

where $R_0 = \sqrt{\frac{2k_s}{3k_0}}\ell_0 L$ and $k_{f0} = \frac{3}{L^2}k_0$.

The contribution of the longitudinal arch and plantar fascia to the foot stiffness is $k_{f0}$, and the transverse arch mediated stiffness due to the distal transverse ligaments is $(R_0/R_T)^2k_{f0}$. Therefore, two parameters govern foot stiffness in our model: the zero transverse curvature stiffness $k_{f0}$ and a normalization scale for the radius of curvature $R_0$. Curvature-induced stiffening is predicted to be significant only when $R_T \ll R_0$. The curvature normalization scale $R_0$ increases with the transverse stiffness $k_s$ such that $R_0 \propto \sqrt{k_s}$. When the transverse curvature is small enough to be significant, i.e. $R_T \ll R_0$, we find using equation (7) that the foot stiffness $k_f \approx k_{f0}(R_0/R_T)^2$, i.e. $k_f \propto k_s$. Therefore, all else held equal (i.e. nearly constant $k_0$), and if the transverse arch truly dominates, the overall foot stiffness is predicted to increase upon stiffening the distal transverse metatarsals. Externally reinforcing the forefoot, using elastic sports tape for example, increases $k_s$. An increase in $k_s$ would in turn increase the overall foot stiffness if the hypothesis about the transverse arch is valid. We test this prediction experimentally by measuring how the midfoot height rises upon toe extension, depending upon presence of the transverse tape.

Table 1: Summary statistics of the estimated relative stiffness. Mean difference between variables is not always the same as the differences in their mean as a result of using pairwise differences.

| $\eta_t$ | $\eta_n$ | $(\eta_t - \eta_n)$ | $\hat{\eta}_t$ | $\hat{\eta}_n$ | $(\hat{\eta}_t - \hat{\eta}_n)$ |
|----------|----------|---------------------|----------------|----------------|----------------------|
| mean (in %) | 53 | 7 | 46 | 45 | 12 | 33 |
| 95% CI (in %) | (15, 91) | (-9, 24) | (3, 88) | (26, 63) | (-9, 34) | (4, 61) |
| Cohen’s $d$ | - | - | 1.07 | - | - | 1.13 |
Figure 3: Taping the distal metatarsal region stiffens the foot. (a) Upon extending the toe ($\Delta \theta$), the midfoot height increases ($\Delta h$) for all three conditions, but the least for the taped condition (♦), and the same amount (statistically indistinguishable) for the control (●) and no-tape (■) conditions. (b) Summary statistics showing the incremental rise in the midfoot height per degree of toe extension normalized by the control case according to equation (2). The mean $\eta$ for the taped condition is significantly greater than control ($p < 0.05$), with an effect size (Cohen’s $d$) of 1.07. Whiskers show the 95% confidence interval of the mean, illustrating that the taped condition is indeed stiffer than the control trials. Solid markers are male subjects, and open markers female.
3.2 Estimated stiffness and its experimental manipulation

The midfoot height increases ($\Delta h$) upon toe extension ($\Delta \theta$) for all experimental conditions by several millimeters (Fig. 3a), which is sufficient in magnitude for reliable kinematic measurement. For every subject, the ratio $\Delta h/\Delta \theta$ was the smallest when the distal metatarsal region was tightly wound with a transverse elastic tape (the taped condition, representative trial shown as red diamonds in Fig. 3a), i.e. transverse splay was stiffened. A loosely applied tape (control condition, green circles in Fig. 3a) was no different from having no tape (no tape condition, blue squares in Fig. 3a). To statistically compare the results, we use the dimensionless variable $\eta$ as shown in equation (2), using the notation $\eta_t$ for the taped condition and $\eta_n$ for the no-tape condition. The dimensionless variable $\eta$ quantifies the stiffness of the experimental condition (taped or no-tape) relative to the control condition, as seen from equation (A.15), and the main statistical findings are summarized in table 1 and Fig. 3b. There is statistically significant difference between $\eta_t$ and $\eta_n$, or the taped and untaped conditions ($p = 0.016$), whereas no significant differences between the left and right feet ($p = 0.2$). Cohen’s d was found to be 1.07. Stiffening upon taping is observed for every single measurement (left column of Fig. 3b). By contrast, the no-tape condition is indistinguishable from the control (right column of Fig. 3b).

3.3 Normalizing foot stiffness estimate by the transverse curvature

Anatomical differences in the size and shape of the foot could influence the mechanical coupling between metatarsal bending and splaying. By using the ratio between the experimental condition (taped or untaped) and the control condition (loose tape), we control for most differences other than those in the curvature of the transverse arch (appendix A and equation (A.15)). In order to normalize for variations between subjects that arise from differences in the transverse arch, we approximate the transverse arch as a parabola (Fig. 4a). Under this approximation (appendix B), we find that $\eta \propto (k_{\text{tape}}/C)(L/w)^5$, where $k_{\text{tape}}$ is the stiffness of the elastic tape, $C$ is a parameter that depends on the Young’s modulus and thickness of the plantar fascia, $L$ is the lever length of the foot, and $w$ is the medio-lateral width of the tarsometatarsal articuar region. Assuming that the material properties of the plantar fascia and the elastic tape are consistent across subjects, we define a rescaled version of $\eta$ to incorporate a first approximation for variations in the transverse arch, and is given by,

$$\hat{\eta} = \eta \left(\frac{w/w_{\text{avg}}}{L/L_{\text{avg}}}\right)^5.$$ (8)

The average lever length and tarsal width across the subjects tested are $L_{\text{avg}}$ and $w_{\text{avg}}$, respectively. For the average foot, therefore, $\hat{\eta} = \eta$. We found $\hat{\eta}_t$ and $\hat{\eta}_n$ to be significantly different from each other ($p = 0.028$), with no significant differences between the left and right feet ($p = 0.45$). This normalization reduced variability in the data, evident from the increase in Cohen’s d (effect size) to 1.13, versus 1.07 for the unscaled response $\eta$ (table 1). Rescaling had the greatest effect on subjects 1 and 4, who are outliers in their anatomical ratio (Fig. 4b) and are also outliers in their unscaled response to taping ($\eta_t$, Fig. 3b).

4 Discussion

We have shown that dorsiflexion in the sagittal plane and medio-lateral (transverse) splay of the distal metatarsal heads are mechanically coupled. Furthermore, multiple lines of evidence suggest that this coupling underlies the stiffness of the foot, and is a direct consequence of the transverse arch. Externally increasing the stiffness against splay of the metatarsals increased the stiffness of the entire foot by over 50%. By comparison, the plantar fascia acting through
Figure 4: Normalization of $\eta$ by transverse radius of curvature. (a) Medio-lateral projection of proximal metatarsal heads, overlaid with a half parabola of width $w$ and height $h$. The parabola has a radius of curvature $R_T \propto w^2/h$. (b) Plot of the normalization parameter $\left(\frac{w/w_{avg}}{L/L_{avg}}\right)^5$ for each subject, which governs the contribution of transverse taping to overall foot stiffness (equation B.6). Subjects 1 and 4 are outliers on this normalization parameter, and the same subjects are outliers in Fig. 3b (high values of $\eta_t$) (c) Descriptive statistics and raw data for $\hat{\eta}$, i.e. the rescaled version of Fig. 3 using the normalization parameter in (b). Mean $\hat{\eta}$ for the taped condition is significantly greater than the untaped condition ($p < 0.05$). Whiskers show the 95% confidence interval of the mean, illustrating that the taped condition is indeed stiffer than the control condition. Solid markers are male subjects, and open markers female.
the longitudinal arch contribute only around 25% to the foot’s stiffness \cite{Ker1987, Huang1993}. The stiffness increase due to the transverse tape is over above the windlass mechanism’s contributions, because our experimental design explicitly engages the windlass. These present clear evidence that the mechanical coupling between transverse splay and dorsiflexion affects the foot’s stiffness significantly. The mathematical analyses elucidate how the mechanical coupling arises from the transverse arch. Presence of transverse curvature in the foot, which is well-known and present in healthy human feet \cite{Morton1922, Jones1941, Drapeau2013, Venkadesan2017preprint} kinematically couples sagittal plane kinematics with the orthogonal transverse direction. The strength of this mechanical coupling depends on the ratio $R_0/R_T$ (equation (7)), and the strong effect of transverse taping suggests that indeed $R_T \ll R_0$ for the human foot. This ratio quantifies the kinematic coupling much as a gear ratio for geared transmissions. The ratio amplifies ($R_0/R_T > 1$) or attenuates ($R_0/R_T < 1$) the transmission of any external stiffness applied in the transverse direction (distally) to the sagittal plane dynamics of the midfoot. Our evidence suggests that $R_0/R_T \gg 1$ in the healthy foot, i.e. it substantially amplifies. Measuring the exact transmission ratio in the foot requires a direct measurement of the stiffness of the foot against transverse splay ($k_s$). One way to estimate $k_s$ is to transect the distal transverse metatarsal ligaments in cadaver feet and measure the decrease in overall stiffness using three-point bending tests, analogous to past experiments that quantified the longitudinal elements of the foot \cite{Ker1987, Huang1993}. Finally, accounting for the variations in the transverse arch based on the mathematical model of curvature-induced stiffness led to a decrease in the variability in the estimated stiffness increase. This provides additional evidence that the simple mechanical model of the foot captures the central elements underlying the mechanical coupling that we have discovered.

Our study calls for expanding the current toolbox of foot arch measurements by incorporating measures of the transverse arch. The Arch Height Index (AHI) is one of the most prevalent measures of the foot arch \cite{Williams2000, Butler2005}, and is defined as the mid dorsum height normalized by length of the foot between calcaneus and the first metatarsal head. It focuses solely on the longitudinal arch, and does not quantify curvature of the transverse arch. Undoubtedly, the height of the medial longitudinal arch and the mid-dorsum height correlate with the transverse arch curvature. However, by solely considering ratios of the foot dimensions within the sagittal plane, the transverse curvature is not quantified. As an example, consider two feet with the same mid-dorsum height and of equal lengths, but with different tarsal widths. The transverse radius of curvature is proportional to the square of the width, and would therefore be quadratically larger for the wider foot. This would imply a diminished effect of the transverse arch for the wider foot, despite having equal AHIs. Our study provides some candidate measures of the transverse arch (equations (B.5), (B.6)). However, these measures should be treated as preliminary candidates and their validity needs more extensive studies that compare radiographic measurements of the transverse arch against these measures.

The mathematical analyses suggest that the externally visible geometric curvature is less important than the orientation of the articular axes of adjacent metatarsals. The elastic energy stored in the stretch of distal transverse ligaments as a result of splaying is proportional to the angle of misalignment between the articular axes of adjacent metatarsals (equation (6)). This in turn governs the relative importance of the transverse splay vis-à-vis the longitudinal arch and longitudinally oriented tissues. For example, although a foot may appear to be devoid of any transverse arch, the preferred direction of flexion for adjacent metatarsals may be misaligned by virtue of the articular surface geometry and ligament layout at the tarso-
metatarsal joints. We would term such a foot, one that mimics a transversally curved foot despite appearing geometrically flat, as being functionally curved. Geometrically flat, yet functionally curved appendages are known to occur in the pectoral fin of rayed fishes such as the mackerel (Nguyen et al., 2017). The strength of the effect in our experiments suggest that either one of the two forms of curvature are active in the foot. But, without more detailed studies, we are unable to distinguish between the two forms of curvature.

Experimental assessment of foot stiffness remains a challenging problem because of the relative rigidity of the human foot. This is especially important in the context of a clinical setting, where quick and repeatable measures of foot stiffness form an essential tool for assessing the progression of diseases such as diabetes mellitus (Rogers et al., 2011; Hastings et al., 2016). Existing foot stiffness assessment methods like the Arch Rigidity Index (ARI) compare the loaded AHI to seated AHI (Rabbito et al., 2011; Miller et al., 2014). However, it is well-known that muscle activity in the intrinsic and extrinsic foot muscles substantially affects its structural integrity (Bojsen-Møller, 1979; Blackwood et al., 2005; Kelly et al., 2014, 2015), and are clearly different between sitting and standing. Our experimental protocol of engaging to windlass mechanism provides a safe and controlled perturbation to the foot while being loaded, and therefore with comparable muscle activity. Others have used the windlass mechanism induced deformation of the midfoot to assess its structural stiffness (Gelber et al., 2014; Hastings et al., 2014). However, these previous studies did not control for the applied external load magnitude, the center of pressure, and the external friction as the foot tries to deform. For example, it was our experience from pilot studies (unpublished data) that measurement repeatability needs a low-friction slider beneath the ball of the foot, and a low-friction interface between the plantar surface of the toes and the apparatus.

Our study raises questions that suggest future experiments. We see subject-to-subject variability in our data (Fig. 3b) that is not accounted for by the transverse arch normalization, and may arise from several factors. The transverse tape tension was controlled manually, and may have introduced variability in its stiffness despite customizing the tape to the circumference of each subject’s foot and maintaining the same strain to our best ability. Despite this, the effect of the transverse arch was evident in our experimental measures. Improving the experimental design to incorporate calibrated external transverse springs will reduce the uncertainty from this factor, and could further strengthen the effect of the transverse arch. Following our comment on the observation on the role of external friction, we note that friction in the medio-lateral direction under the forefoot may have contributed to additional variability between trials. This is because medio-lateral friction may contribute to resisting the metatarsal heads splaying apart. We expect this to be a minor effect because the extent to which a healthy foot splays is small, and usually accommodated by the soft pad of the sole (Fig. 1b). Finally, our current measures of the transverse curvature using external markers could be improved upon using radiographic scans.

Our study impacts multiple fields, including podiatric, sports and evolutionary biomechanics. The study of foot pathologies arising from diabetes and other flatfoot disorders may benefit from a renewed examination of the transverse arch in those contexts. Our analyses also provide a functional connection between the arched morphology of the foot, and its mechanical stiffness. Such a functional interpretation of form has important implications for the study of human evolution.

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**Competing interests**

The authors declare no competing financial interests.

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**Author contributions**

SM and MV conceived of the study; SM and MV developed the mathematical model of the foot, with assistance from MLB and AY; LK and AY developed and conducted the human subject experiments; AY analyzed the data and performed statistics in consultation with MV; SM, MV and AY wrote the manuscript.
Figure 5: (a) Sketch of the measurement foot with heel fixed and forefoot on a sliding plate, with the toe extended by $\Delta \theta$. The center of pressure (C) is located under the forefoot (b) Free body diagram of the foot showing all external forces acting on it. The plantar fascia is modeled as a linear spring of stiffness $k_{pf}$ spanning the bottom of the foot. External load on the toe is represented as a force $F_{ext}$ at a height $h_{toe}$ and a moment $M_{ext}$ at the base of the toe F. The foot is loaded with bodyweight W and tension in the Achilles tendon $T_{ach}$. (c),(d) Free body diagrams of the toe and the hindfoot respectively. Normalizing by the control case, we can remove dependence on subject specific foot parameters and the arch height rise can be used as an estimate of foot stiffness.

Appendices

A Estimation of foot stiffness using the Windlass mechanism

Direct load-displacement measurements on cadaver feet have shown foot stiffness to be about 500N/mm (Ker et al., 1987). As a result, in live subjects who weight 600–700 N, safe external loads are unlikely to produce deformations large enough to be measured reliably with external fiducial markers. As our model predicts, transverse taping increases overall foot stiffness, making it even harder to produce measurable deformations. We use an alternative method to indirectly estimate foot stiffness, based on the well-known windlass mechanism of the foot (Hicks, 1954). Extension of the toe stretches the plantar fascia and leads to a rise in the height of the midfoot. We expect this rise to be inversely proportional to foot stiffness, and hence use it as a proxy for stiffness. Here we model the foot to delineate dependence of this proxy measurement on foot parameters.

The foot is modeled as three rigid bodies: toe, metatarsus and tarsus, connected by frictionless hinges. Note that for simplicity, effective stiffness of the foot $k_f$ is replaced by a torsional spring at the tarso-metatarsal joint S, with stiffness $\kappa_f$

$$\kappa_f = L_f^2 k_f$$ (A.1)

Truncated foot length, $L = L_f + L_h$, where $L_f$ and $L_h$ are forefoot and hindfoot lengths respectively. External load on the toe is represented as a force $F_{ext}$ at a height $h_{toe}$ and a moment $M_{ext}$ at the base of the toe F. Ground contact is modeled as a frictionless hinge at the heel H and a frictionless hinge on a roller at forefoot F. $F_{ext}$ extends the toe by $\Delta \theta$ and causes the arch to rise by $\Delta h$. Static force balance on the foot and moment balance
about **H** gives:

\[
F_{\text{ext}} + R_{Fx} + R_{Hx} = 0 \quad \text{(A.2)}
\]
\[
-W + R_{Fy} + R_{Hy} = 0 \quad \text{(A.3)}
\]
\[
(W + T_{\text{ach}})d - R_{Fy}L - F_{\text{ext}}h_{\text{toe}} - M_{\text{ext}} = 0 \quad \text{(A.4)}
\]

Moment balance on the *whole* body about center of pressure at **F** gives

\[
R_{Hy}L - F_{\text{ext}}h_{\text{toe}} - M_{\text{ext}} = 0 \quad \text{(A.5)}
\]

Noting that \(R_{Fx} = 0\) because of the frictionless forefoot support, we can solve for ground reaction forces as follows

\[
R_{Hx} = -F_{\text{ext}} \quad \text{(A.6)}
\]
\[
R_{Hy} = \frac{F_{\text{ext}}h_{\text{toe}} + M_{\text{ext}}}{L} \quad \text{(A.7)}
\]
\[
R_{Fy} = W - \frac{F_{\text{ext}}h_{\text{toe}} + M_{\text{ext}}}{L} \quad \text{(A.8)}
\]

Moment balance on the toe due to external loads and the tension in the plantar fascia \(T_{pf}\) about **F**, we get

\[
F_{\text{ext}}h_{\text{toe}} + M_{\text{ext}} = T_{pf}r_t \quad \text{(A.9)}
\]

A toe extension of \(\Delta \theta\) stretches the plantar fascia and produces tension \(T_{pf}\)

\[
T_{pf} = r_t^2 k_{pf} \Delta \theta \quad \text{(A.10)}
\]

In order to find the moment acting on the torsional spring \(M_s\), we can write moment balance for the hindfoot about **S**, and use \(\text{(A.4)}\) and \(\text{(A.9)}\) to obtain

\[
M_s = WL_f - T_{pf}(r_t \frac{L_f}{L} - r_t + h) - F_{\text{ext}}h \quad \text{(A.11)}
\]

Assuming \(r_t \ll h, L_f/L \approx O(1)\), and \(F_{\text{ext}} \ll T_{pf}\), and \(M_s\) may be approximated as

\[
M_s = WL_f - r_t^2 k_{pf} \Delta \theta h \quad \text{(A.12)}
\]

This moment produces a change in the arch angle \(\Delta \phi\) at **S** such that for small \(\Delta h\)

\[
\Delta h = -L_f \Delta \phi = -L_f \frac{M_s}{\kappa_f} \quad \text{(A.13)}
\]

Assuming the change in arch height is small \((\Delta h \ll h)\), the drop in arch height depends on the body weight \(W\) and the toe-extension angle \(\Delta \theta\) as

\[
\Delta h = \frac{-WL_f^2 + r_t^2 k_{pf} \Delta \theta h L_f}{\kappa_f} \quad \text{(A.14)}
\]

Without extending the toe, the arch height falls by an amount \(\Delta h_0 = -WL_f^2/\kappa_f\) due to body weight. As before, subscripts \(c, t,\) and \(n\) denote control, tape, and no-tape experiments,
respectively. The dependence on subject-specific foot parameters (i.e., $W$, $r_t$, $k_{pf}$, $h$, and $L_f$) may be reduced by normalization with the control as

\[
\frac{\left(\frac{\Delta h - \Delta h_0}{\Delta \theta}\right)_c}{\left(\frac{\Delta h - \Delta h_0}{\Delta \theta}\right)_{t,n}} - 1 = \frac{\left(\frac{1}{\kappa_f}\right)_c}{\left(\frac{1}{\kappa_f}\right)_{t,n}} - 1,
\]

(A.15)

**B Approximate curvature normalization**

In the absence of a direct measure of transverse curvature, we estimate a parameter to normalize data across subjects in Fig. 4, using externally measured parameters of their feet; namely truncated length $L$, and tarsal width $w$. Using equation 7, we can isolate the effect of adding tape of stiffness $k_{tape}$ on overall foot stiffness.

\[
k_{fc} = \frac{2\ell_0^2}{R_T^2} k_s + \frac{3}{L^2} k_0
\]

(B.1)

\[
k_{ft} = \frac{\ell_0^2}{R_T^2} (2k_s + k_{tape}) + \frac{3}{L^2} k_0 = k_{fc} + \frac{\ell_0^2}{R_T^2} k_{tape}
\]

(B.2)

\[
k_{ft} - 1 = \frac{k_{tape} \ell_0^2}{k_{fc} R_T^2}
\]

(B.3)

where $k_{f}$ is the stiffness of the foot and the subscripts $t$, $b$ and $c$ denote taped, no tape and control conditions respectively.

Therefore, from equation A.15

\[
\frac{\left(\frac{\Delta h}{\Delta \theta}\right)_c}{\left(\frac{\Delta h}{\Delta \theta}\right)_{t,n}} - 1 = \frac{k_{tape} \ell_0^2}{k_{fc} R_T^2}
\]

(B.4)

Equation B.4 shows that the factor $\ell_0^2/R_T^2$ governs the increase in foot stiffness due to taping. The distal metatarsal head separation $\ell_0$ can be expected to vary as $\ell_0 \propto L$, assuming an isometric foot. In Fig. 4b, assuming that the transverse arch of the foot is a half parabola of width $w$ and height $h$, the radius of curvature of the parabola can be written as $R_T = \frac{w^2}{2h}$.

Therefore,

\[
\frac{\ell_0^2}{R_T^2} \propto \frac{L^2 h^2}{w^4}.
\]

(B.5)

In the absence of any quantitative measure of $k_{fc}$, it can be approximated as being entirely due to plantar fascia and other plantar soft tissues. \cite{Ker1987, Huang1993, Venkadesan2017} shows that contribution of the plantar fascia to foot stiffness is approximately equal to that of other longitudinal plantar tissues. If plantar fascia have an elastic modulus, $E$, and thickness $t$, then their stiffness is $E w t / L$ where $w$ and $L$ are foot width and length respectively. This stiffness is equivalent to a torsional midfoot stiffness $E w t h^2 / L^3$ and linear stiffness of $k_{fc} \propto E w t h^2 / L^3$ at the forefoot \cite{Venkadesan2017}. Substituting for $k_{fc}$ we get

\[
\frac{\left(\frac{\Delta h}{\Delta \theta}\right)_c}{\left(\frac{\Delta h}{\Delta \theta}\right)_{t,n}} - 1 \propto \frac{k_{tape} L}{E t w^5}
\]

(B.6)
We assume that the Young’s modulus of the plantar fascia modulus $E$ and thickness $t$ remain constant across subjects. The factor $(L/w)^5$, divided by $\left( \frac{L_{avg}}{w_{avg}} \right)^5$, therefore controls for anatomical variation in foot proportions relative the mean foot among the subjects tested.
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