Effects of Footwear and Stride Length on Metatarsal Strains and Failure in Running

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Abstract

Background: The metatarsal bones of the foot are particularly susceptible to stress fracture owing to the high strains they experience during the stance phase of running. Shoe cushioning and stride length reduction represent two potential interventions to decrease metatarsal strain and thus stress fracture risk.

Methods: Fourteen male recreational runners ran overground at a 5-km pace while motion capture and plantar pressure data were collected during four experimental conditions: traditional shoe at preferred and 90% preferred stride length, and minimalist shoe at preferred and 90% preferred stride length. Combined musculoskeletal – finite element modeling based on motion analysis and computed tomography data were used to quantify metatarsal strains and the probability of failure was determined using stress-life predictions.

Findings: No significant interactions between footwear and stride length were observed. Running in minimalist shoes increased strains for all metatarsals by 28.7% (SD 6.4%; $p < 0.001$) and probability of failure for metatarsals 2-4 by 17.3% (SD 14.3%; $p \leq 0.005$). Running at 90% preferred stride length decreased strains for metatarsal 4 by 4.2% (SD 2.0%; $p \leq 0.007$), and no differences in probability of failure were observed.

Interpretations: Significant increases in metatarsal strains and the probability of failure were observed for recreational runners acutely transitioning to minimalist shoes. Running with a 10% reduction in stride length did not appear to be a beneficial technique for reducing the risk of metatarsal stress fracture, however the increased number of loading cycles for a given distance was not detrimental either.

34 **Key Words:** Running injury; stress fracture; minimalist shoe; musculoskeletal model; finite
35 element model

1. Introduction

Stress fractures are considered overuse injuries associated with the mechanical fatigue of bone.

Long periods of repetitive loading, such as that incurred during running, leads to the formation of bone microdamage, or microcracks (Burr, 1997). If the rate of microdamage accumulation exceeds the rate of bone remodeling, these microcracks may propagate to a critical length (O'Brien et al., 2003), thereby increasing the risk of stress fracture. The formation of microdamage in bone, and therefore the number of repetitive loading cycles it may endure prior to failure, is a strong function of strain magnitude (Carter et al., 1976). For the physiological loads relevant to human locomotion, a 10% reduction in strain magnitude can correspond to a 100% increase in the number of cycles to failure (Carter and Caler, 1985). Thus, any mechanism that can be adopted by an individual to reduce strains may also reduce their risk of stress fracture.

Long distance runners experience stress fractures at a relatively high rate compared to other athletes (Brukner et al., 1996), presumably due to the continuous and repetitive nature of mechanical loading associated with running activity. In fact, 14% to 18% of all stress fractures observed in active populations occur in the metatarsals (Brukner et al., 1996; Kiuru, 2005; Matheson et al., 1971), second only to the tibia (Brubaker and James, 1974). Approximately 80% of these fractures occur in the second and third metatarsals (Fetzer and Wright, 2006), presumably because of their long, narrow diaphyses and the larger bending loads they experience during the stance phase of gait (Stokes et al., 1979). Metatarsal stress fractures have been observed frequently in minimalist shoe runners (Cauthon et al., 2013; Salzler et al., 2012), and have been speculated to stem from increased metatarsal loading caused either by the footwear

itself or from changes to running biomechanics associated with the footwear (Firminger and Edwards, 2016).

Multiple studies have reported specific biomechanical alterations in runners when wearing minimalist versus traditional footwear (Bonacci et al., 2013; McCallion et al., 2014). For example, running in minimalist footwear has been associated with increased forefoot plantar pressure (Bergstra et al., 2014) as well as increased ankle and metatarsophalangeal (MTP) joint moments (Firminger and Edwards, 2016). Metatarsal loading tends to occur earlier in stance in minimalist footwear users, as a more anterior foot strike pattern is often displayed (Greenhalgh et al., 2014). Several studies have also reported that minimalist footwear users run at a reduced stride length (Bonacci et al., 2013; Squadrone and Gallozzi, 2009). This finding is important because a reduction in stride length has been associated with an increase in shock attenuation and a decrease in mechanical loading at several lower extremity locations (Derrick et al., 1998; Edwards et al., 2009; Firminger and Edwards, 2016; Heiderscheit et al., 2011), and thus may also directly reduce metatarsal loading independent of footwear selection. To date, the systematic effects of stride length reduction and shoe type on metatarsal loading during running are unknown.

The purpose of this study was to investigate the influence of minimalist footwear and stride length reduction on metatarsal strains and the probability of failure (i.e., stress fracture) in running. To this end, three-dimensional motion analysis was conducted while participants ran in traditional and minimalist footwear at preferred stride length (PSL) and 90% PSL, and combined musculoskeletal – finite element modeling and stress-life predictions were used to quantify

metatarsal strains and the probability of failure. A 10% reduction in stride length was investigated in accordance with our previous work focused on tibial stress fracture (Edwards et al., 2009). We hypothesized that running at 90% PSL would lower metatarsal strains and the probability of failure relative to PSL, and that running in minimalist footwear would increase metatarsal strains and the probability of failure relative to traditional footwear.

2. Methods

2.1. Participants

Fourteen male recreational runners (age: 26.2 (4.2) years; height: 178.4 (5.4) cm; body mass: 75.6 (5.6) kg, mean (SD)) participated in this study. Participants ran 10 km/week, had no lower limb injuries within the previous 3 months of the study, were self-reported rearfoot strikers, and had no prior experience running in minimalist footwear. The university research ethics board approved the study and all participants provided written informed consent prior to data collection.

2.2. Motion capture and analysis

Participants ran overground on a 23 m runway while data were collected during four running conditions, the order of which was counterbalanced to reduce bias. The conditions were: 1) traditional shoe at PSL, 2) traditional shoe at 90% PSL, 3) minimalist shoe at PSL, and 4) minimalist shoe at 90% PSL. The traditional shoe (New Balance 890 v5) weighed 234.8 g and had heel and toe heights of 19.0 and 11.0 mm, respectively. The minimalist shoe (New Balance Minimus Zero v2) weighed 167.3 g and had heel and toe heights of 12.8 and 12.0 mm, respectively. Participants were instructed to run at a speed they would select for a 5-km run.

Running speed was recorded using a pair of timing lights (Banner Multi-Beam; Minneapolis, USA) positioned 1.9 m apart. PSL during warm-up trials was calculated as follows:

$$PSL = \frac{t_{n \text{ strides}} * v}{n_{strides}} \quad [1]$$

where, $t_{n \text{ strides}}$ is the time taken to run a given number of strides ($n_{strides}$), and v is the running speed calculated from the timing lights. The time for five consecutive strides was measured using a stopwatch as the participant ran down the runway. These strides were taken near the middle of the runway length to ensure that the participant was not accelerating or decelerating. Values for running speed ($\pm 5\%$) and PSL determined from the warm-up trials in the traditional shoe were used for all running conditions. After each participant's PSL was determined, markers were attached to the floor of the runway at PSL and 90% PSL to ensure participants ran at the correct stride lengths (Edwards et al., 2009). Additional warm-up trials were provided prior to each of the four running conditions to ensure that participants were not actively targeting the stride length markers. Foot strike pattern was determined by calculating foot strike index (Cavanagh and LaFortune, 1980).

Ten dynamic trials were completed at each running condition during which kinematic data were recorded for the right lower limb at 240 Hz using an eight-camera motion capture system (Motion Analysis Corporation; Santa Rosa, USA). Nineteen retro-reflective markers were placed on the right lower-limb, however for the purpose of this study only the markers placed on the foot and ankle are described. Two markers were placed on the medial/lateral right malleoli and five markers were placed on the right shoe at the heel, first/fifth metatarsal heads, dorsal foot (proximal to the MTP joint), and toes. These markers separated the right foot into two anatomical segments (truncated foot and toes) joined by the MTP joint. The anterior-posterior

axis of the toe coordinate system was created by subtracting a virtual toe marker from the MTP joint centre (defined as the average between the first and fifth metatarsal markers). The vertical axis of the toe coordinate system was created by crossing a vector between the fifth and first metatarsals with the anterior-posterior axis. Finally, the medial-lateral axis of the toe coordinate system was created by crossing the anterior-posterior axis with the vertical axis. The anterior-posterior axis of the truncated foot coordinate system was created by subtracting the MTP joint centre from a virtual heel marker. The medial-lateral axis was created by crossing the anterior-posterior axis with a vector created by subtracting the dorsifoot marker from the heel marker. Finally, the vertical axis was created by crossing the medial-lateral axis with the anterior-posterior axis. The virtual toe and heel markers were both defined by their respective toe and heel markers, but given the vertical position of the MTP joint centre in the truncated foot/toe coordinate systems. A static motion-capture trial was collected in each footwear condition to establish static anatomical coordinate systems for each segment. Kinematic data were filtered using a zero-lag, low-pass fourth order Butterworth filter with a cutoff frequency of 20 Hz (Edwards et al., 2010).

Plantar pressure data were captured concurrently in the right shoe at 200 Hz using a Pedar-X pressure-sensing insole (Novel; Minneapolis, USA). The Pedar-X insole contains 99 cells ranging in area from 164 to 187 mm², and measures pressure perpendicular to the insole surface. A non-functioning Pedar-X insole was placed in the left shoe to compensate for contralateral heel/toe height discrepancies.

2.3. Image acquisition and analysis

Two separate clinical CT scans of the right foot were obtained for each participant while they wore either the traditional or minimalist shoe (Discovery 610, General Electric Healthcare; Wauwatosa, USA). Scanning protocol settings were as follows: 220 mA tube current, 120 kVp peak voltage, pitch = 1, 0.39 x 0.39 mm in-plane resolution, and 0.63 mm slice thickness. During scanning, a custom jig was used to align and maintain the limb in a static, neutral position, and a phantom that contained known calcium hydroxyapatite equivalent concentrations was placed in the field of view (QRM GmbH; Moehrendorf, Germany). Metatarsal bones were manually segmented from the CT image stacks using Mimics software (Materialise; Leuven, Belgium). Local coordinate systems were established for each metatarsal by creating a unit vector between the centroids of the maximum cross-sectional areas in the metatarsal base and head. Static sagittal metatarsal angles within the minimalist and traditional shoes were defined as the sagittal orientation of each metatarsal local coordinate system with respect to horizontal.

2.4. *Musculoskeletal modeling*

Metatarsal head forces acting parallel (axial) and perpendicular (shear) to the bone long axis, in the sagittal plane, were calculated over the entire stance phase of running using a musculoskeletal model of the MTP joint (Stokes et al., 1979) (Fig. 1). Briefly, the model assumed that: 1) the MTP joints were frictionless; 2) the inertial forces were negligible; 3) no extensor muscles of the phalanges were activated during metatarsal head ground contact; and 4) the flexor muscles crossing the MTP joint, i.e., the long toe flexors (representing flexor digitorum longus for the second through fifth rays and flexor hallucis longus for the first ray) and the short toe flexors (representing both the plantar aponeurosis and flexor digitorum brevis) shared an equal proportion of the load. The original model developed by Stokes et al. (1979)

used a pressure mat-like system to estimate vertical forces acting on the metatarsal and phalangeal plantar surfaces while walking barefoot, however for this study a Pedar-X pressure sensing insole was used as participants were shod. Following the methodology of Putti et al. (2007), the Pedar-X insole was segmented into first – fifth metatarsal, hallux, second toe, and third – fifth toe regions. The other input into the musculoskeletal model was the sagittal-plane metatarsal angle, calculated by adding the sagittal angle of the truncated foot coordinate system to the static sagittal angle of the local metatarsal coordinate systems. Potential differences in MTP joint moments, or flexor activity, between conditions were inherently captured by the forces acting at the phalangeal plantar surfaces.

2.5. Finite element modeling

Segmented bones from CT images were converted into geometric meshes using 3-Matic software (Materialise; Leuven, Belgium). Meshes were comprised of 10-node tetrahedral elements with a maximum edge length of 3 mm and median element edge lengths from 1.0 to 1.5 mm. This resulted in FE models ranging from 7,873 to 28,660 elements with 12,527 to 41,764 degrees-of-freedom, depending on bone size. Hounsfield units of voxels comprising each element were averaged and converted to units of apparent density using calibration equations based on the CT phantom. Elements were subsequently assigned isotropic, linear-elastic material properties as a function of bone apparent density (Rho et al., 1995). These specific FE modeling parameters were initially validated in the linear elastic range using cadaveric experimentation (see Appendix A. Supplementary Data for details). The FE models were able to predict experimentally measured periosteal strains on the metatarsal diaphysis with an $r^2 = 0.98$ and an X = Y type of relationship.

All nodes proximal to the maximum cross-sectional area of the metatarsal base were fixed in three-dimensions. The forces acting on the metatarsal head were calculated at 10% intervals from 30% to 80% of stance when the pressure-sensing insole provided non-zero readings in the metatarsal regions. A kinematic couple was used to distribute and apply loads to all nodes more than 3 mm distal to the maximum cross-sectional area of the metatarsal head. The spatial definition of this boundary condition was selected to approximate the contact area of the metatarsal head *in vivo*. All FE models were solved using Abaqus/Standard v6.13 (Dassault Systèmes Simulia Corp.; Providence, USA).

2.6. Data Reduction

Von Mises equivalent strains (ϵ_{vm}) were computed to obtain a single scalar representation of strain magnitude for each element (Fig. 2):

$$\epsilon_{vm} = \sqrt{\frac{2}{3}(\epsilon_1^2 + \epsilon_2^2 + \epsilon_3^2)} \quad [2]$$

where ϵ_1 , ϵ_2 , and ϵ_3 are the principal strains of the Green-Lagrange strain tensor. The strains along each metatarsal diaphysis were extracted, because the majority of metatarsal stress fractures occur in this region (Harrast and Colonna, 2010). These strains were non-normally distributed, thus we obtained the median and 95th percentile strains at the instant of peak perpendicular metatarsal force, as this time point was associated with the largest diaphyseal strains.

The probability of failure, or stress fracture, for each metatarsal was calculated using a Weibull analysis that accounted for stressed volume (see Appendix A. Supplementary Material for

details). Theoretical development and sensitivity of the approach have been previously described elsewhere (Taylor, 1998; Taylor et al., 2004; Taylor and Kuiper, 2001). Briefly, the analysis calculated the likelihood of failure after a given number of loading cycles according to:

$$P_f = 1 - e^{\left[-\left(\frac{V_s}{V_{so}}\right)\left(\frac{\Delta\epsilon}{\Delta\epsilon^*}\right)^m\right]} \quad [3]$$

where P_f is the probability that a volume of bone V_s will fail at strain ranges up to $\Delta\epsilon$. The reference strain range $\Delta\epsilon^*$ is a measure of the bone's fatigue strength at reference volume V_{so} , while exponent m represents the variability in the material's fatigue-life measurements. The P_f was calculated over a range of cumulative running distances between 10 and 40 km, and similar relative differences between conditions were observed in this range (Fig. 3). For the statistical analysis presented below, we examined P_f for a cumulative distance of 40 km, representing two 5 km runs per week for four weeks. This distances was chosen simply because the majority of stress fractures occur after 3-4 weeks of beginning a new training regime (Milgrom et al., 1985).

2.7. *Statistical Analysis*

The dependent variables (i.e., median and 95th percentile strains, and P_f) were trial averaged within conditions and statistical analysis was performed using SPSS software (SPSS Inc., Chicago, USA). A 2×2 repeated measures multivariate analysis of variance (MANOVA) was used to test for the main effects of footwear and stride length. Repeated measures univariate analysis of variance (ANOVA) were then run for each metatarsal to test for individual contributions to main effect differences. A Bonferroni post-hoc correction was used to account for experimentwise error with the criterion alpha level set to $\alpha = 0.01$, or $\alpha = 0.05/(5$ metatarsals). In order to investigate potential causes of metatarsal strain changes between running conditions, differences in metatarsal angles and metatarsal forces were examined using

Cohen's *d* effect sizes and a 2×2 repeated measures MANOVA. Effect sizes were calculated using the estimated marginal means, and were subsequently averaged across metatarsals.

3. Results

Participants ran at 3.8 (SD 0.5) m/s with a PSL and 90% PSL of 1.39 (SD 0.17) m and 1.25 (SD 0.15) m, respectively. Peak perpendicular force, the timepoint used to calculate metatarsal strain, occurred at 52.2% (SD 4.1%) of stance. All participants displayed a rearfoot strike pattern when running in the traditional shoe at both stride lengths, however two participants adopted a midfoot strike pattern during minimalist shoe conditions. No significant interactions between footwear and stride length were observed for multivariate analyses examining median strains, 95th percentile strains, and P_f across metatarsals ($0.348 \leq p \leq 0.452$). There was a significant main effect of footwear for median strain, 95th percentile strain, and P_f ($p \leq 0.005$), however no main effects of stride length were observed ($0.103 \leq p \leq 0.439$, Table 1).

No significant interactions between footwear and stride length were observed for univariate analyses examining dependent variables within each metatarsal ($0.074 \leq p \leq 0.978$, Table 2). The median and 95th percentile strains were 28.7% (SD 6.4%) lower in the traditional shoe for all metatarsals ($p < 0.001$), while P_f was 17.3% (SD 14.3%) lower in the traditional shoe for the second, third, and fourth metatarsals ($p < 0.005$). Running at 90% PSL significantly reduced median and 95th percentile strains for the fourth metatarsal by 4.3% (SD 2.0%; $p \leq 0.006$) and 4.2% (SD 2.0%; $p \leq 0.007$), respectively (Table 2). No main effect of stride length was observed for univariate tests of P_f (Table 2, Fig. 4). Metatarsal angle was significantly greater ($p < 0.001$) in the traditional footwear (35.5° (SD 4.3°), 38.2° (SD 3.4°), 33.3° (SD 3.8°), 27.9° (SD 3.5°),

21.1° (SD 3.4°) for metatarsals 1-5, respectively) than the minimalist footwear (29.0° (SD 2.9), 32.7° (SD 2.4°), 28.2° (SD 2.3°), 22.9° (SD 1.8°), 15.4° (SD 2.2°) for metatarsals 1-5, respectively). An effect size of $d = 1.95$ and $d = 0.19$ was observed for metatarsal angle and force magnitude differences between footwear conditions, respectively. An effect size of $d = 0.26$ and $d = 0.41$ was observed for metatarsal angle and force magnitude differences between stride length conditions, respectively. Positive effect sizes represent larger forces/angles in the traditional shoe and at PSL. A post-hoc sensitivity analysis of the FE model indicated that metatarsal force and strain varied proportionately (i.e., a 50% increase in force resulted in a 50% increase in strain), whereas a decrease in metatarsal angle of 1° increased strains by 1.7%.

4. Discussion

The purpose of this study was to examine the influence of minimalist footwear and stride length reduction on metatarsal strains and the corresponding probability of failure in running. Relative to a traditional running shoe, the minimalist shoe increased strains in all metatarsals, while the probability of failure increased in the second, third, and fourth metatarsals. Relative to preferred stride length, running with a 10% reduction in stride length lowered strains in the fourth metatarsal only, and no changes in the probability of failure were observed in any bone. These findings have important implications for individuals looking to reduce their likelihood for metatarsal stress fracture through practical intervention.

Metatarsal stress fractures have been observed in minimalist shoe runners (Cauthon et al., 2013; Salzler et al., 2012), although it must be noted that prospective evidence linking an increased incidence of metatarsal stress fracture to minimalist shoe running does not exist. A recent

biomechanical study has demonstrated increased forefoot loading and plantar pressure when running in minimalist versus traditional footwear (Bergstra et al., 2014), and our work further illustrates that running in minimalist shoes is also associated with increased tissue-level strains at the metatarsal diaphysis. Post-hoc investigation into the cause of these increased strains was conducted by examining differences in the two main inputs to the musculoskeletal model: sagittal metatarsal angle and load acting on the metatarsal head (Firminger and Edwards, 2017). Whereas the mean sagittal metatarsal angle at the instant of peak perpendicular force was 17.9% (i.e., 5.6°) smaller in the minimalist shoe, the mean loads were only 1.6% greater in the minimalist shoe. Thus, it can be concluded based on the large effect size and post-hoc sensitivity analysis results that the lower sagittal metatarsal angle at peak load during minimalist shoe conditions resulted in increased metatarsal bending and strain caused by a less axially-aligned load. The application of these results to shoe design suggest that it may be more pertinent to increase heel-to-toe drop rather than create a more cushioned shoe in order to lower metatarsal strains. Future work investigating the effect of heel-to-toe drop, in otherwise similarly constructed shoes, on metatarsal strains and probability of failure would be a logical next step.

The prediction of subject-specific metatarsal strains allowed us to further examine the probability of failure for a cumulative distance of running. The second metatarsal had the highest probability of failure, which is consistent with epidemiology literature (Harrast and Colonno, 2010). Running in minimalist shoes increased the probability of failure by 33.8%, 8.3%, and 9.9% for the second, third, and fourth metatarsals, respectively. The probability of failure measures were based on the well-established Weibull analysis (Weibull, 1951), which may be considered an overestimate of stress fracture risk because it does not include the known effects of bone repair

and adaption. Accounting for these effects has been shown to lower the probability of failure (Taylor et al., 2004), but act as an effect modifier rather than a confounder such that the relative differences between conditions would be preserved. From a practical standpoint, higher P_f values in the minimalist shoe indicate that traditionally-shod runners should exercise caution when switching to minimalist footwear. Literature on the method for transitioning to minimalist shoes is inconsistent, with recommendations of minimalist shoe use ranging from 3% to 33% of the total running distance in the first week (Giandolini et al., 2013; Ridge et al., 2013). One study showed that transitioning to minimalist shoes over a 10-week period resulted in significantly more metatarsal bone marrow edemas (precursors to stress fracture) compared to runners who ran in traditional footwear (Ridge et al., 2013). A conservative approach implementing a reduction in running volume and intensity or more frequent but shorter bouts of moderate training over several months may provide adequate time for bone remodeling and promote bone adaption such that the risk of metatarsal stress fracture is minimized.

It is interesting to note that strains were higher for all metatarsals during minimalist shoe conditions, while the probability of failure was only higher in the second, third, and fourth metatarsals. Whereas the median and 95th percentile strain measures represent discrete points of the strain distribution, the probability of failure is a multifactorial measure dependent on the metatarsal's strain range distribution. Inherent to the probabilistic model is that bones experiencing the highest peak strains will not necessarily fail when the corresponding stressed volume is lower. The strain distribution of the first and fifth metatarsals were more positively skewed than that of the other metatarsals, which may have resulted in a similar probability of failure measure between footwear conditions despite having differences in bone strain

magnitudes (Fig. 5). In a practical sense, this suggests that stress fracture risk should be investigated by examining the entire strain distribution of a bone, rather than a single peak strain magnitude.

In contrast to our original hypothesis, the results from this study suggested that, overall, a 10% reduction in stride length did not significantly lower strain magnitudes or the probability of failure (although strains were slightly lower for the fourth metatarsal at 90% PSL). A reduction in stride length is associated with an increased number of loading cycles for a given distance, and it is interesting to observe that this did not manifest as an increased risk of injury. Our previous research indicates that a 10% reduction in stride length is advantageous for decreasing strains and the probability of stress fracture at the tibia (Edwards et al., 2009). Similarly, running at a reduced stride length was associated with decreased loads at the ankle and knee joints (Firminger and Edwards, 2016). Thus, for the lower-extremity, reductions in stride length may represent a beneficial and practical intervention for reducing overall injury risk in runners, especially because reductions in stride length of 10% do not appear to significantly increase the metabolic cost of transport (Hamill et al., 1995).

There are several limitations that should be noted when interpreting the results of this work. This study included only recreational runners who exhibited a rearfoot strike pattern with no prior experience running in minimalist footwear. Although this inclusion criterion clearly limited the study's broad applicability, stress fractures are more likely to occur shortly after changes in training regimen (Harrast and Colonna, 2010; Milgrom et al., 1985), and habitual rearfoot strikers have been shown to continue using a rearfoot strike pattern weeks after transitioning to

minimalist footwear (Willson et al., 2014). Secondly, although the CT scans were extremely accurate at determining the static metatarsal angles in the shoe, the dynamic metatarsal angles throughout stance were based on a rigid truncated foot coordinate system derived from external shoe markers. We see this as the main limitation of this study, because the tarsometatarsal joint is not rigid and each metatarsal may move differentially within each shoe. Theoretically, we would expect that the lower flexibility and greater arch support offered by the traditional shoe would cause less tarsometatarsal motion during stance, thereby minimizing sagittal tarsometatarsal motion as the truncated heel segment plantarflexes. On the other hand, the flexibility and lack of arch support in the minimalist shoe may allow for more sagittal plane movement in the tarsometatarsal joints, which would correspond to a lower metatarsal angle to that estimated herein as the truncated heel segment plantarflexes. According to our post-hoc investigation, this would result in even higher strains in the minimalist footwear as the metatarsals would undergo a larger bending moment. Based on the observed 5.6° average decrease in metatarsal angle for the minimalist footwear conditions, we believe that a meaningful difference between shoe conditions was in fact captured. However, a more accurate description of metatarsal angles in running would require either invasive bone pin measurements (Lundgren et al., 2008) or dual fluoroscopy (Campbell et al., 2016).

5. Conclusion

Significant increases in metatarsal strains and the probability of stress fracture were observed for recreational runners acutely transitioning to minimalist shoes. These findings help to explain, at least in part, why metatarsal stress fractures are observed in minimalist shoe runners. Running with a 10% reduction in stride length was not a beneficial technique for reducing the risk of

381 metatarsal stress fracture, however it does not seem to be detrimental either. Further
382 investigation into footwear characteristics such as arch height, heel-toe drop, and insole stiffness
383 is warranted, as these parameters may alter metatarsal angles during ground contact, and thus
384 have the potential to influence metatarsal strains and stress fracture risk.

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391 **Conflict of Interest**

392 None.

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Table 1: Multivariate Repeated-Measures ANOVA Results

	Shoe Type	Stride Length	Shoe Type and Stride Length Interaction
Median Strain	$F(5,9) = 54.920, p < 0.001^*$	$F(5,9) = 2.218, p = 0.141$	$F(5,9) = 1.206, p = 0.379;$
95th Percentile Strain	$F(5,9) = 48.269, p < 0.001^*$	$F(5,9) = 2.571, p = 0.103$	$F(5,9) = 1.484, p = 0.285$
Probability of failure at 40 km (P_f)	$F(5,9) = 10.613, p = 0.001^*$	$F(5,9) = 1.097, p = 0.425$	$F(5,9) = 0.852, p = 0.547$

**Significant effect ($p < 0.01$).*

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Table 2: Univariate ANOVA Results for Median Strain, 95th Percentile Strain, and P_f (mean (SD))

		Metatarsal					
		1	2	3	4	5	
Estimated Marginal Means	Median Strain (μɛ)	Control	613 (119)	1393 (463)	1203 (539)	1072 (329)	699 (213)
		Minimalist	797 (152)	1937 (452)	1525 (509)	1345 (333)	912 (254)
		p-value	< 0.001*	< 0.001*	< 0.001*	< 0.001*	< 0.001*
		PSL	712 (146)	1697 (480)	1389 (512)	1235 (318)	836 (247)
		90% PSL	699 (121)	1634 (423)	1339 (535)	1182 (340)	775 (228)
		p-value	0.261	0.107	0.044	0.006*	0.051
		Interaction p-value	0.147	0.488	0.22	0.761	0.57
	95th Percentile Strain (μɛ)	Control	1610 (333)	3110 (954)	2654 (1179)	2505 (917)	1532 (520)
		Minimalist	2040 (382)	4341 (917)	3350 (1119)	3090 (988)	1829 (565)
		p-value	< 0.001*	< 0.001*	< 0.001*	< 0.001*	< 0.001*
		PSL	1837 (385)	3803 (992)	3054 (1119)	2858 (920)	1753 (569)
		90% PSL	1813 (322)	3648 (846)	2950 (1175)	2737 (969)	1608 (528)
		p-value	0.355	0.088	0.039	0.007*	0.015
		Interaction p-value	0.099	0.596	0.261	0.924	0.833
P _f (%)	Control	0.48 (0.91)	23.33 (28.70)	12.37 (26.67)	13.33 (22.00)	0.945 (2.95)	
	Minimalist	2.17 (3.34)	57.10 (29.54)	20.68 (28.96)	23.22 (27.31)	2.19 (6.44)	
	p-value	0.024	< 0.001*	0.005*	0.002*	0.207	
	PSL	1.53 (2.75)	41.91 (28.40)	16.84 (26.88)	18.60 (23.80)	1.70 (4.85)	
	90% PSL	1.11 (1.53)	38.52 (27.54)	16.21 (28.19)	17.96 (25.05)	1.43 (4.55)	
	p-value	0.279	0.230	0.602	0.537	0.098	
	Interaction p-value	0.246	0.978	0.074	0.305	0.504	

*Significant effect ($p < 0.01$).

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Fig. 1: Overview of methodology. Motion capture and CT data are combined inputs to the MTP joint musculoskeletal model. In the musculoskeletal model, P_L and P_S represent forces in the long and short toe flexors respectively, separated by angle β . F_t is the toe force measured from Pedar-X insole. F_m and F_j are the metatarsal and MTP joint forces acting on the metatarsal head, which are rotated into axial and shear components using the metatarsal angle α . These forces are then used as boundary conditions for a finite element model to calculate metatarsal strains during stance.

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497 Fig. 2: Representative von Mises equivalent strain distribution from finite element analysis.

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Fig. 3: Representative probability of failure (± 1 SD) of the second metatarsal for both shoe types and stride lengths. A significant main effect of shoe type ($p < 0.01$) was observed for all cumulative running distances between 10 and 40 km, however no significant difference in stride length was observed.

Fig. 4: Representative median von Mises equivalent strains on the metatarsal diaphyseal surface as a function of stance.

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510 Fig. 5: First metatarsal strain distributions (with medians and 95th percentiles overlaid) for a
 511 representative subject in traditional and minimalist footwear.

Appendix A. Supplementary Data

Electronic Supplementary Material

Effects of Footwear and Stride Length on Metatarsal Strains and Failure in Running

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Verification and Validation of Subject-Specific Finite Element Modeling

Periosteal strain predictions using the subject-specific finite element (FE) modeling approach outlined in the manuscript were directly compared to strain gauge measurements from cadaveric experimentation. Three fresh, frozen cadaveric feet, one from a 78-year-old female and two from an 85-year-old male, were obtained through the University of Calgary's Body Donation Program. Each foot was thawed at room temperature and scanned using computed tomography (CT) as described in *Image acquisition and analysis*. The second metatarsals were subsequently excised, cleaned of all soft tissue, and the proximal most 2 cm of bone were potted into a cylindrical aluminum tube. Rectangular strain gauge rosettes (UFRA-1-350-11-1L, Tokyo Sokki Kenkyujo Co., Ltd.; Tokyo, Japan) were adhered with cyanoacrylic glue to the dorsal and plantar periosteal surface of each second metatarsal diaphysis after the surface was sanded and thoroughly cleaned with isopropyl alcohol. A FaroArm (FARO; Lake Mary, USA) was used to digitize select bony landmarks and strain gauge positions so that these locations could be later identified in the CT/FE coordinate system through a transformation matrix.

The potted metatarsals were fixed to a custom apparatus attached to the base of a materials testing machine (858 Mini Bionix II, MTS, Inc.; Minneapolis, USA). The apparatus allowed for a vertical displacement to be applied to the palmar surface of the metatarsal head (Figure 1). Each metatarsal was positioned at an angle of $\alpha = 35^\circ$ in the sagittal plane relative to the horizontal, similar to the mean metatarsal angle at peak perpendicular force measured during dynamic gait trials ($35.4 \pm 4.0^\circ$). A ramped displacement up to 2 mm was applied to the metatarsal head at a rate of 1 mm/s. Periosteal strains were measured at 500 Hz using a high-precision data acquisition system (System 8000, Vishay Precision Group – Micro-

Measurements; Raleigh, USA) collected synchronously with the applied displacement and reaction force data. Three repeat trials were performed and the maximum and minimum principal strains at 50 N were averaged for model comparison. A load of 50 N was chosen to ensure strains were within the linear elastic range.

Figure 1: Experimental test set-up. A displacement/force was applied to the palmar surface of the metatarsal (denoted by arrow F) while the metatarsal was positioned at angle α in the aluminum pot. Periosteal strain gauges are visible on the plantar/dorsal surfaces of the diaphysis.

Metatarsal geometry was segmented from CT scans and converted to 10-node tetrahedral element meshes using the Materialise Mimics Innovation Suite (Materialise; Leuven, Belgium). A maximum mesh edge length of 3 mm was chosen in accordance with a preliminary mesh convergence analysis examining maximum element edge lengths ranging from 2 to 8 mm. Decreasing the maximum edge length from 4 to 3 mm changed displacements and principal stresses and strains by less than 5%. Elements were assigned material properties based on the following equation:

$$E = 0.51\rho_{\text{app}}^{1.37} \quad [1]$$

where E is the modulus of elasticity (MPa) and ρ_{app} is the apparent density (kg/m^3) (Rho et al., 1995); the Poisson's ratio was assumed to be 0.3 throughout (Wirtz et al., 2000). Partial volume effects at the periosteal surface were mitigated by applying Equation 1 to an eroded mask (1 voxel from the surface) of the FE model. Any elements outside of this mask were assigned a representative surface modulus value.

The FE analyses were performed in ABAQUS/Standard (Dassault Systèmes Simulia Corp.; Providence, USA) with boundary conditions matching those of the mechanical testing protocol. Surface nodes representing the proximal potting were fixed in translation and a distributed load of 50 N was applied to the base of the metatarsal head. The three-dimensional strains (Figure 2) occurring at each strain gauge location were transformed into a local coordinate system with a unit normal to the model exterior surface. The model predicted maximum and minimum principal strains occurring along the surface plane were then calculated and directly compared to experimental measurements using linear regression, root mean squared error (RMSE), and maximum error.

Figure 2: Minimum principal strain distribution predicted using FE modeling.

Model-predicted strains were highly correlated with experimental measurements with an $r^2 = 0.98$ (Figure 3). The slope and intercept of the linear regression were not different from unity and zero, respectively, indicating an X=Y type of relationship between model predictions and experimental measurements. The RMSE and maximum error between strain predictions and measurements were $92.4 \mu\epsilon$ and $192.5 \mu\epsilon$, respectively. As a percentage of the highest measured strain (i.e., $846.8 \mu\epsilon$), the RMSE was 10.9% and the maximum error was 22.7%.

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597 *Figure 3: Measured vs. predicted maximum and minimum principal strains on the dorsal and plantar surface of*
 598 *second metatarsal bones. All strain values correspond to an applied force of 50N.*

Probability of Failure Calculation using Weibull Analysis

Due to the inherent microstructural variability of bone – variability that cannot be measured through imaging – two bones that appear identical in macrostructure will inevitably exhibit differences in their fatigue behavior. The Weibull analysis accounts for this natural variability and assigns a likelihood that failure will occur after a specific number of loading cycles. The number of cycles to failure N_f , is related to strain range $\Delta\epsilon$ ($\mu\epsilon$), which for *in vivo* loading is equal to strain magnitude through the standard fatigue equation:

$$N_f = C\Delta\epsilon^{-n} \quad [2]$$

where C and n are empirically derived constants from fatigue tests. We used values of $C = 3.8 \times 10^{29}$ and $n = 6.6$ (Carter and Caler, 1985). If a series of bone samples are cyclically loaded to failure, their measured fatigue strength will vary according to:

$$P_f = 1 - e^{\left[-\left(\frac{V_s}{V_{so}}\right)\left(\frac{\Delta\epsilon}{\Delta\epsilon^*}\right)^m\right]} \quad [3]$$

where P_f is the probability that a volume of bone V_s will fail at strain ranges up to $\Delta\epsilon$. Exponent m represents the degree of scatter observed in the experimentally measured fatigue-life data. The reference strain range $\Delta\epsilon^*$ is a measure of the bone's fatigue strength for a reference volume V_{so} (a larger volume of bone will illustrate lower fatigue strength due to the increased probability of finding weak regions), defined as the $\Delta\epsilon$ at which the probability of failure is 63% for a specific number of loading cycles. We used constants of $m = 8$, $V_{so} = 64 \text{ mm}^3$, and $\Delta\epsilon^* = 5307 \mu\epsilon$, obtained from an endurance run of 10^5 cycles (Taylor, 1998). Note that the $\Delta\epsilon^*$ is a function of N_f , related to $\Delta\epsilon$ in Equation 3 by $\Delta\epsilon^* = 1.067(\Delta\epsilon)$ (Taylor, 1998). Thus, for this study, $\Delta\epsilon^*$ was a function of stride length and cumulative running distance.

As the strain magnitude varies throughout the entire bone, a different P_f can be calculated for each individual element from the FE model. If there are k elements, then P_f for the whole metatarsal is the probability that any one element will fail:

$$P_f = 1 - (1 - P_1)(1 - P_2)(1 - P_3) \dots (1 - P_k) \quad [4]$$

Elements experiencing similar strain magnitudes were stratified into twelve bins, each with a corresponding V_s equal to the sum of element volumes within each group. This volume was then doubled to account for the contralateral metatarsal in the P_f calculation.

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