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Effect of step width manipulation on tibial stress during running

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ABSTRACT

Narrow step width has been linked to variables associated with tibial stress fracture. The purpose of this study was to evaluate the effect of step width on bone stresses using a standardized model of the tibia. 15 runners ran at their preferred 5 k running velocity in three running conditions, preferred step width (PSW) and PSW ± 5% of leg length. 10 successful trials of force and 3-D motion data were collected. A combination of inverse dynamics, musculoskeletal modeling and beam theory was used to estimate stresses applied to the tibia using subject-specific anthropometrics and motion data. The tibia was modeled as a hollow ellipse. Multivariate analysis revealed that tibial stresses at the distal 1/3 of the tibia differed with step width manipulation ($p=0.002$). Compression on the posterior and medial aspect of the tibia was inversely related to step width such that as step width increased, compression on the surface of tibia decreased (linear trend $p=0.036$ and 0.003). Similarly, tension on the anterior surface of the tibia decreased as step width increased (linear trend $p=0.029$). Widening step width linearly reduced shear stress at all 4 sites ($p<0.001$ for all). The data from this study suggests that stresses experienced by the tibia during running were influenced by step width when using a standardized model of the tibia. Wider step widths were generally associated with reduced loading of the tibia and may benefit runners at risk of or experiencing stress injury at the tibia, especially if they present with a crossover running style.

1. Introduction

62% of people who meet or exceed physical activity guidelines choose running as a form of fitness (Hootman et al., 2002). Unfortunately, the risk for overuse injury in runners is well documented with running related stress fractures accounting for 6–20% percent of running injuries (Bennell et al., 1996; James et al., 1978; Snyder et al., 2006) and affecting as many as 19–21% of athletes and military recruits (Bennell et al., 1996; Cosman et al., 2013; Hoch et al., 2009). The tibia is a common site of stress injury affecting 16–44% of runners and military personnel (Brukner et al., 1996; Plisky et al., 2007; Shaffer et al., 2006; Yagi et al., 2013; Yates and White, 2004).

Running mechanics have been implicated in the development of bone stress. Running mechanics retrospectively linked to stress fracture of the tibia include: greater rearfoot eversion, peak hip adduction, and knee internal rotation angles, and a greater free moment (a surrogate measure of torsion in the lower extremity) (Pohl et al., 2008). In addition, Creaby and Dixon (2008) report a greater medially directed GRF vector during the time of peak active GRF in military personnel with a history of stress fracture.

Running with a step width less than the preferred step width is associated with gait mechanics similar to those reported in people with tibial stress fracture (Brindle et al., 2013; McClay, 1995; Pohl et al., 2006; Williams and Ziff, 1991). For example, reduced step width during running has been linked to an increase in rearfoot pronation (Brindle et al., 2013; Pohl et al., 2006) and greater hip adduction and knee internal rotation (Brindle et al., 2013; Meardon et al., 2012). Pilot work from our lab also suggests that step width during running is negatively associated with the free moment, (Meardon and Derrick, 2008) frontal plane hip, and knee and ankle joint moments (Meardon and Derrick, 2011). Confirming these findings, greater internal peak knee abduction moment and impulse occur with reduced step width (Brindle et al., 2013). Lastly, reduced step width has been shown to increase medial-lateral ground reaction forces during the midstance phase of running when compared to a neutral pattern (McClay, 1995). Taken together, these altered kinematics and greater medial-lateral ground reaction forces, torsional loads, and frontal plane joint moments have the potential to increase stress applied to the tibia during running.

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Traditionally, ground reaction forces and kinematics have been used to estimate the musculoskeletal load during running. However, these measures only capture a portion of the loading environment since contribution from the musculature is not considered. A combination of experimental data collection and musculoskeletal modeling that incorporates muscle forces can be used to provide an estimate of tissue stress or strain. Tibial strains during the stance phase of running have been estimated in a previous study (Edwards et al., 2009). Our results suggest that reduced step length positively influences contact forces and reduces the probability of stress fracture by 3–6%. However, the effect of step width on tibial stress has not been examined.

Therefore, the purpose of this study was to evaluate the effect of step width on bone stress using a standardized model of the tibia. If step width does influence tibial stresses, as expected, manipulation of step width could benefit runners experiencing injury to the tibia.

2. Methods

2.1. Subjects

Prior to the study, the protocol was approved by the Institutional Review Board. All subjects gave written informed consent prior to participation. Fifteen experienced runners volunteered for this study (8 males and 7 females, 23.7 ± 5.4 years, 70.3 ± 9.2 kg, 1.7 ± 0.08 m). All runners were free from injury at the time of data collection and were running on an average of more than 10 miles/week. *A priori* sample size was calculated based on values from a previous study of distal tibia contact forces (Sasimontonkul et al., 2007). 15% difference was used in the calculation with an alpha level of 0.05 and 80% power. Given a clinically relevant difference in the condition means of 15% (Butler et al., 2007; Milner et al., 2007), minimum of 8 subjects were needed to be able to reject the null hypothesis that the condition means are equal.

2.2. Protocol

Anthropometric data (height, weight, thigh length, leg length, malleolar height, foot length, thigh circumference, calf circumference, malleolar width and foot breadth) were collected for use in a musculoskeletal model. Twenty-nine retro-reflective markers were placed on the lower extremity and pelvis. A static trial with the runner in an anatomical neutral alignment was collected.

Subjects ran at a self-selected 5 k running pace over a force platform using a preferred step width (PREFERRED) until 10 successful trials were collected. Running velocity was monitored with motion capture using the horizontal component, in the direction of forward progression, of a marker placed at L5–S1. Trials were considered successful if running velocity was within ± 5% of self-reported 5 K pace and no deviation in running style was observed when crossing the force plate.

Average running velocity and step width data from PREFERRED were used to verify the success of trials during the WIDE and NARROW conditions. Step width was calculated as the mediolateral distance between the right and left heel marker during successive foot strikes when each heel marker was at its minimum vertical position. Participants were verbally cued to run with a "narrower" or "wider" step width. Verbal feedback regarding step width and running velocity was provided after each trial. In general, participants were able to successfully achieve target step widths within 5 practice trials. The order of target step width conditions was randomized. Target step width trials were considered successful if participants ran within ± 5% of their PREFERRED running velocity and step width was at least 5% of leg length greater (WIDE) or less (NARROW) than their PREFERRED step width. Ten successful trials of the right limb were collected for each condition.

Motion capture data were collected with an 8-camera 3D motion capture system (160 Hz, Vicon Nexus, Centennial, CO) and force platform data were collected simultaneously (1600 Hz, AMTI, Watertown, MA). Motion capture data and force platform data were exported to MATLAB (The Mathworks, Natick, MA) for signal processing and analysis.

2.3. Analysis

Motion capture data were interpolated to 1600 Hz using a cubic spline technique. Joint moments and reaction force data were filtered using a 4th order low-pass Butterworth filter with cutoff frequencies corresponding to 95th and 98th percentile frequency of the vertical ground reaction force for the moments and the reaction forces, respectively (Edwards et al., 2011). Static trials were used to estimate joint center locations of the ankle, knee, and hip. Three-dimensional Cardan segment and joint angles were calculated with a flexion/extension, abduction/adduction, and internal/external rotation order of rotations.

Anthropometric measurements were used to calculate segment masses, center of mass location, and moments of inertia (Vaughan et al., 1992) and used to estimate joint moments and reaction forces at the hip, knee, ankle and subtalar joints with inverse dynamics. Custom musculoskeletal software scaled to individual anthropometrics was used to estimate the maximal muscle forces, muscle moment arms, and muscle orientations for 43 lower extremity muscles during the stance phase of running. Muscle and joint definitions were obtained from the Delp leg model (Delp et al., 1990). Dynamic muscle forces were estimated using static optimization with a cost function minimizing the sum of squared muscle stresses (Glitsch and Baumann, 1997). Resulting hip (sagittal and frontal planes), knee (sagittal plane), and ankle (sagittal and frontal planes) torques were constrained equal to those obtained from the inverse dynamics procedure.

The tibia was modeled as a hollow ellipse (Fig. 1) with outer diameters of 23.22 mm and 29.32 mm in the medial-lateral and anterior-posterior directions, respectively, and inner diameters of 10.08 mm and 9.76 mm in the medial–lateral and anterior–posterior directions, respectively (Franklyn et al., 2008). We examined the effect of step width on bone stress at the geometric center of a point located at 67% of the distance from the proximal end of the tibia. This cross-sectional level has the narrowest tibial width and is the general area commonly reported for stress fracture (Milgrom et al., 1989). Muscles crossing this centroid that influenced the loading environment were: soleus, gastrocnemius, tibialis anterior and posterior, flexor digitorum and flexor hallucis, peroneal brevis, longus, tertius, and extensor digitorum and hallucis.

Normal stresses (σ) at four sites of the peripheral surface of the tibia (anterior, lateral, posterior, and medial) were estimated using the following summary equations

$$\sigma_{anterior} = \sigma(-M_{ML}) + \sigma(F_{long})$$

$$\sigma_{lateral} = \sigma(M_{AP}) + \sigma(F_{long})$$

$$\sigma_{posterior} = \sigma(M_{ML}) + \sigma(F_{long})$$

$$\sigma_{medial} = \sigma(-M_{AP}) + \sigma(F_{long})$$

where F_{long} represents the summation of the longitudinal components of the knee reaction force and the muscle forces acting at the centroid, M_{ML} and M_{AP} represent the normal stress contributions from the moments caused by the knee reaction force, the muscle forces and the knee moment.

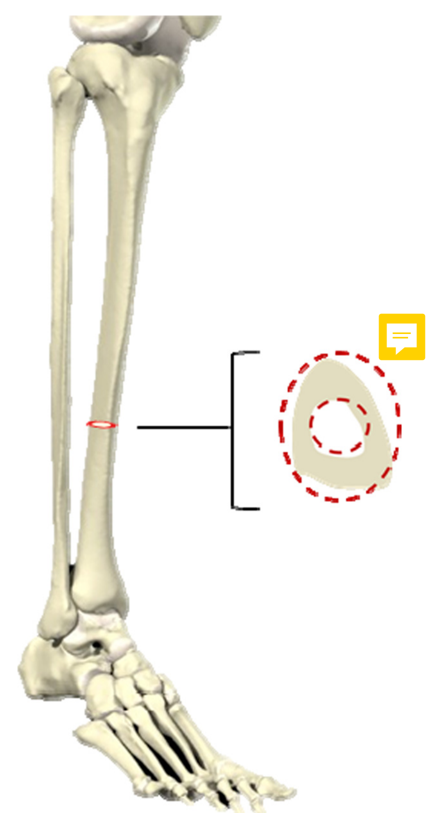

Fig. 1. This figure displays the point along the length of hypothetical tibia in which we estimated tibial stress. A cross sectional image of the tibia with the elliptical model superimposed is displayed. The cross sectional image was replicated from a coronal view MRI of a 24 year old runner.

Shear stress (τ) was estimated at four sites of on the periphery of the tibia using the following summary shear stress equations:

$$\tau_{anterior} = \tau(-M_{long}) + \tau(F_{ML})$$
$$\tau_{lateral} = \tau(M_{long}) + \tau(F_{AP})$$
$$\tau_{posterior} = \tau(M_{long}) + \tau(F_{ML})$$
$$\tau_{medial} = \tau(-M_{long}) + \tau(F_{AP})$$

where F_{ML} represents the medial-lateral components of the knee reaction force and the muscle forces acting at the centroid, F_{AP} represents the anterior-components of the knee reaction force and the muscle forces acting at the centroid and M_{long} represents the torsional moment about the longitudinal axis of the centroid due to the torsional knee moment and muscle forces.

Foot strike index was assessed and calculated as the location of the center of pressure during heel strike relative to the length of the foot (Laughton et al., 2003). This was done in order to assess alterations in foot strike pattern that may have occurred when asking participants to change their step width. Input data to the bone stress model was exported for visual display and included the reaction forces and moments and the muscle forces and moments used in the stress equations as well as the ground reaction forces (Appendix A-C).

2.4. Statistics

Subject condition averages for peak normal stress and peak shear stress at four sites of the periphery of the tibia were exported for statistical analysis (SPSS). Data were analyzed using a repeated measures multivariate analysis of variance ($\alpha = 0.05$) and pairwise comparisons. Polynomial contrasts were used to identify significant linear trends across conditions. Cohen's d effect sizes were calculated to provide insight regarding the magnitude that effect changing step width has on bones stresses (Cohen, 1992).

Table 1
Condition means and 95% confidence intervals are reported for study design variables. A negative step width indicates a running pattern in which one foot crosses over the other. No significant differences between conditions were observed for velocity or strike index. Statistically significant step width comparisons are indicated by superscripts ($p \leq 0.05$).

Step width condition	Velocity (m/s)	Step width (cm)	Strike index (%)
Wide	4.03 [3.68, 4.32]	0.10 [0.08, 0.12] [ab]	26.92 [17.7, 37.9]
Preferred	4.04 [3.68, 4.33]	0.02 [0.00, 0.05] [ac]	31.19 [19.3, 43.2]
Narrow	4.04 [3.69, 4.32]	−0.06 [−0.09, −0.04] [b c]	26.74 [17.3, 37.5]

[a] Wide vs preferred.
[b] Wide vs narrow.
[c] Preferred vs narrow.

3. Results

Table 1 indicates that subjects were able to manipulate their step widths during running ($p < 0.001$) while maintaining a consistent running velocity ($p = 0.896$) and strike index ($p = 0.433$). During the narrow condition, runners demonstrated a crossover running style in which one foot crossed over the other in consecutive foot strikes.

Ensemble averages of normal tibial stress across the stance phase are shown in Fig. 2. Throughout the stance phase, the posterior aspect of the tibia predominantly underwent compressive loading and the anterior tibia underwent tensile loading. Both anteriorly and posteriorly, peak stress occurred in midstance. Although less consistent than anterior and posterior loads, the lateral tibia primarily experienced compression with a small magnitude tensile peak early in stance. Even more variable, loads on the medial tibia were predominantly tensile after the first 20% of stance; however, the medial tibia did appear to experience a compressive peak during early stance. Fig. 3 shows that shear loads at the surface of the tibia are evident during the stance phase of running with peak magnitude occurring midstance at all sites.

A repeated measures MANOVA revealed a significant multivariate effect of step width, Wilk's $\lambda = 0.124$, $p < 0.001$, and partial eta squared = 0.647. The power to detect the effect was 0.999. Thus, the hypothesis that step width would influence bone stress of the tibia was confirmed. Given the significance of the overall test, the univariate main effects for step width were evaluated (Figs. 4 and 5). Significant condition effects on peak normal stress were observed in the anterior, posterior and medial tibia ($p = 0.038$, 0.039 and 0.002, respectively). Increasing step width reduced anterior tension, posterior compression, and medial compression of the tibia (linear trend $p = 0.029$, 0.036 and 0.003). Significant condition effects on peak shear stress were detected at all four sites of the tibia ($p < 0.001$). Increasing step width linearly reduced shear stress at all sites ($p < 0.001$).

Mean differences and effects sizes for paired comparisons are listed in Tables 2 and 3. Normal stress was reduced at the medial tibia when widening step width from runners' preferred running style ($p = 0.010$, $d = -0.449$). Wide step width relative to a cross-over pattern resulted in lower anterior tension, and posterior and medial tibial compression ($p = 0.029$, $d = -0.338$; $p = 0.010$;

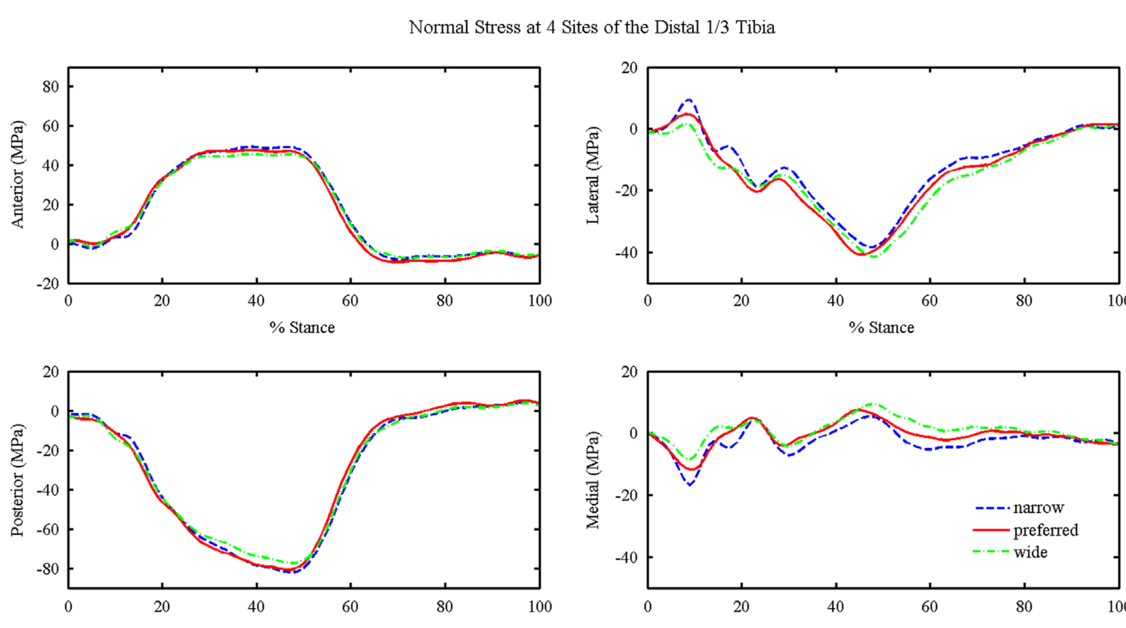

Fig. 2. Ensemble averages for normal stress at 4 sites on the periphery of the tibia are displayed above. Positive values indicate tension and negative values indicate compression.

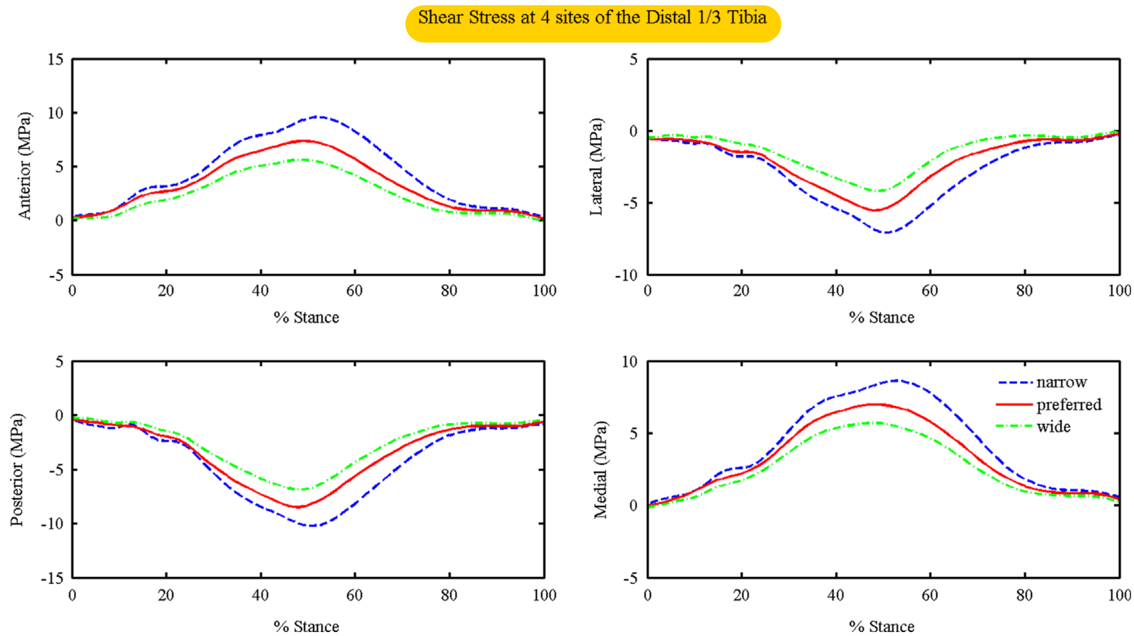

Fig. 3. Ensemble averages for shear stress at 4 sites on the periphery of the tibia are displayed above.

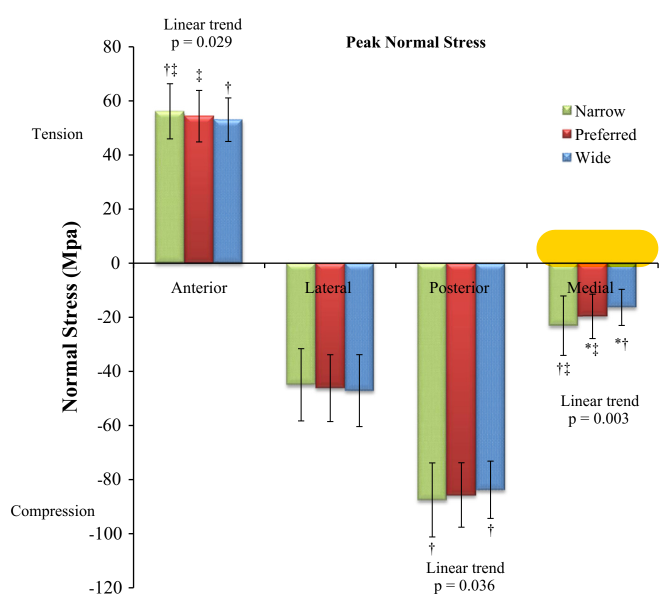

Fig. 4. Regional peak normal stress for each step width condition is illustrated. Error bars indicate the standard deviation associated with each condition. Statistically significant trend analysis results are indicated. Positive values indicate tension and negative values indicate compression. Significant pairwise comparisons ($p \le 0.05$) are indicated as follows: *wide vs. preferred; †wide vs. narrow; ‡preferred vs. narrow.

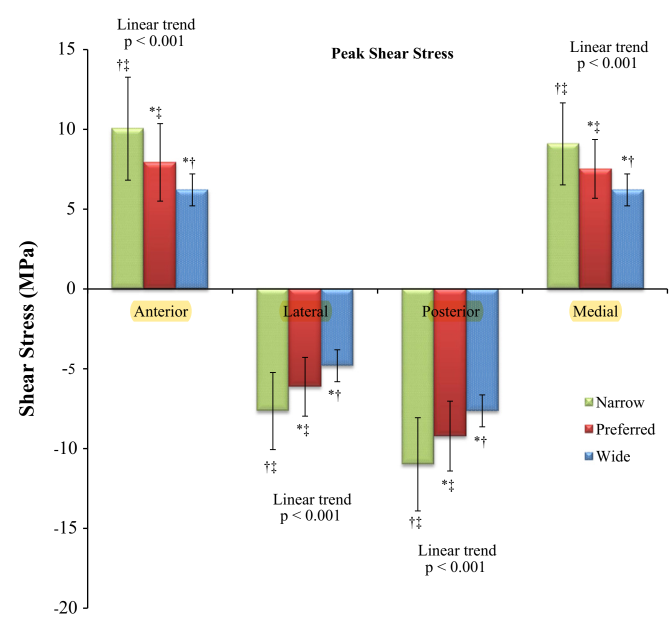

Fig. 5. Regional peak shear stress for each step width condition is shown. Error bars indicate the standard deviation associated with each condition. Statistically significant trend analysis results are indicated. Significant pairwise comparisons ($p \le 0.05$) are indicated as follows: *wide vs. preferred; †wide vs. narrow; ‡preferred vs. narrow.

$d = -0.307$; and $p = 0.003$, $d = -0.744$). Likewise, preferred running step width was associated with less medial tibia tension and medial compression when compared to a crossover pattern ($p = 0.045$, $d = -0.183$ and $p = 0.024$, $d = -0.354$). In terms of shear loading, widening step width tend to reduce tibial stress at all sites and for all condition comparisons ($p < 0.001$ for all comparisons, $d = -0.666$ to -1.282).

4. Discussion

With evolving interest in changing gait mechanics to minimize injury, a better understanding of how gait alterations influence tissue loads is needed. The purpose of this study was to examine how step width, which has been linked to mechanics associated with tibial stress fracture, influences ~~bone stress. Increasing step width reduced both compressive and tensile bone stress on 3 out of 4 sites of the tibia.~~ Moreover, increasing step width systematically reduced shear stresses at 4 sites studied.

Model estimates of bone stress obtained in this study appeared to approximate in-vivo results, although shear stress values were lower than previous reports. in vivo strains on the mid shaft of the medial tibia during over ground running ranged from 394 to $1163\mu\varepsilon$ for tension and -672 to $-2446\mu\varepsilon$ for compression (Burr et al., 1996; Milgrom et al., 2003, 2007). Average in vivo shear stress

Table 2
Pairwise comparisons, 95% confidence limits of the difference, and effect sizes (d) for normal stress are displayed below. Statistically significant differences are indicated by an asterik. The sign of the effect size indicates if increasing step width reduced (−) or increased (+) stress.

	Comparison	|Mean| difference (MPa)	95% confidence interval	p-Value	d
Anterior tension	Wide vs. preferred	1.303	[−4.108, 1.501]	0.334	−0.149
	Wide vs. narrow*	3.105	[−5.829, −0.381]	0.029	−0.338
	Preferred vs. narrow*	1.802	[−3.558, −0.045]	0.045	−0.183
Lateral compression	Wide vs. preferred	0.915	[−3.548, 1.718]	0.241	0.071
	Wide vs. narrow	2.158	[−5.958, 1.641]	0.466	0.162
	Preferred vs. narrow	1.243	[−4.036, 1.550]	0.354	0.093
Posterior Compression	Wide vs. preferred	1.925	[−0.997, 6.478]	0.188	−0.171
	Wide vs. narrow*	3.754	[1.083, 8.642]	0.010	−0.307
	Preferred vs. narrow	1.828	[−0.984, 5.048]	0.245	−0.143
Medial Compression	Wide vs. preferred*	3.326	[0.933, 5.718]	0.010	−0.449
	Wide vs. narrow*	6.759	[2.672, 10.846]	0.003	−0.744
	Preferred vs. narrow*	3.433	[0.523, 6.344]	0.024	−0.354

Table 3
Pairwise comparisons, 95% confidence limits of the difference, and effect sizes (d) for shear stress are displayed below. Statistically significant differences are indicated by an asterik. The sign of the effect size indicates if increasing step width reduced (−) or increased (+) stress.

Site	Comparison	|Mean| difference (MPa)	95% confidence interval	p-Value	d
Anterior shear	Wide vs. preferred*	1.723	[−2.204, −1.242]	< 0.001	−0.666
	Wide vs. narrow*	3.834	[−4.934, −2.735]	< 0.001	−1.282
	Preferred vs. narrow*	1.768	[−2.962, −1.260]	< 0.001	−0.740
Lateral shear	Wide vs. preferred*	1.317	[0.991, 1.643]	< 0.001	−0.668
	Wide vs. narrow*	2.840	[2.056, 3.624]	< 0.001	−1.259
	Preferred vs. narrow*	1.768	[0.871, 2.174]	< 0.001	−0.709
Posterior shear	Wide vs. preferred*	1.581	[1.256, 1.905]	< 0.001	−0.686
	Wide vs. narrow*	3.349	[2.364, 4.334]	< 0.001	−1.250
	Preferred vs. narrow*	1.768	[0.910, 2.626]	< 0.001	−0.685
Medial shear	Wide vs. preferred*	1.316	[−1.665, −0.967]	< 0.001	−0.675
	Wide vs. narrow*	2.883	[−3.773, −1.994]	< 0.001	−1.240
	Preferred vs. narrow*	1.567	[−2.263, −0.870]	< 0.001	−0.701

in a study of 2 participants running over ground was $1444 \pm 141\mu\varepsilon$ (Burr et al., 1996). Using Hooke's Law and a Young's modulus of 20.9 GPa longitudinally and 12.3 transversely (Hoffmeister et al., 2000), the average strains observed on the medial tibia in this study approximated $775 \pm 669\mu\varepsilon$ for tension, $-941 \pm 392\mu\varepsilon$ for compression and $611 \pm 149\mu\varepsilon$ for shear. in vivo work and this modeling study suggest that bone strain varies considerably across subjects. This could be due to variance in the site of strain gauge application or model estimation or inherent subject differences. Bone stress calculations are sensitive to the location of stress measurement relative to the neutral axis of bending and the principal axes. Variations in the location of stress measurement likely influenced results relative to previously published outcomes. Also, we calculated stresses from a standardized bone model. This neglects the influence of bone geometry on the stress environment but does illustrate that stresses are influenced by gait mechanics. Limitations associated with simulating internal loads were characteristics of this study. Modeling a hollow ellipse for an irregularly shaped object like the tibia, which varies across individuals (Franklyn and Oakes, 2012), likely introduced error as seen in Fig. 1 and makes comparison between studies challenging. However since this systematic error is consistent with our conditions, within study comparisons were reasonable (Franklyn and Oakes, 2012). Future studies should incorporate subject specific bone shape and geometries.

The results of this study suggest a reduction in tibial bone stress with increasing step width of up to 5% leg length. Runners self-selected their pace which was held constant within subject across conditions. Rather than forcing runners to run at a fixed speed, we chose to allow runners to run at a comfortable pace in order to minimize the behavioral challenge associated with changing step width. Evaluation of bone stress throughout the stance phase as well as the input data of reaction forces and moments and muscle forces and moments (Appendix A and B) can provide insight to the biomechanical factors contributing to the observed bone stress changes. Normal stress on the anterior and posterior aspects of the tibia is influenced by the longitudinal reaction and muscle forces and the net moments acting about the ML axis of the tibia. The reaction force, summed muscle force, and moment due to the muscles acted to compress the posterior aspect of the tibia while tensioning the tibia anteriorly. Wider step widths tend to be associated with lower net forces and a reduced muscle moment resulting in reduced peak stress both anteriorly and posteriorly. Stress on the medial and lateral aspect of the tibia was influenced by the longitudinal reaction and muscle forces and the net moment acting about the AP axis of the tibia. Graphs of these data reveal a net moment compressing the medial surface of the tibia and tensioning the lateral surface early in stance. Widening step width appeared to largely decrease the moment due to the reaction force in addition to small reductions in the net longitudinal forces. Shear stress is influenced by the ML and AP reaction forces and the moment acting about the vertical axis. Examination of the reaction forces and moments in these planes and axes shows that medial-lateral reaction forces and moments about the vertical axis of the tibia were most influenced by step width changes. Wider step width was associated with smaller ML reaction forces and a smaller moment about the longitudinal axis.

Much of the differences discussed above may be due to differences in the ground reaction forces and moments (Appendix C). Small reductions were noted in the vertical ground reaction forces when running with wider step widths and may have contributed to reduced longitudinal forces. Peak medial ground reaction forces were also reduced with wider step widths. This reduction of the medial ground reaction force with wider step width is consistent with previous work and directs the ankle contact force vector closer to the longitudinal axis of the tibia in the wide running condition (McClay, 1995). Reducing medially directed ground reaction forces may have effectively decreased the bending moment about the AP axis of the tibia subsequently reducing stress on the medial and lateral surfaces of the tibia. Finally, torsion about the longitudinal axis is a primary contributor to shear stress. A systematic reduction in the peak free moment, the moment about a vertical axis resulting from shear forces between the foot and ground, was observed as step width increased. This reduction may account for the consistent differences in shear stress observed across conditions.

The results of our study suggest that frequently cited locations of tibial stress fracture experience larger loads relative to the other surface locations. The posteromedial aspect of the tibia, followed by the anterior tibia, is a common site of stress fracture accounting for 18/20 tibial stress fractures in collegiate track and cross country athletes (Nattiv et al., 2013). In our study, the posterior aspect of the tibia underwent the greatest amount compressive (-85.6 ± 12.1 MPa) and shear stress (-9.3 ± 2.5 MPa). In terms of tension, the greatest amount of tension was observed anteriorly (54.5 ± 9.3 MPa).

Step width modification may be a feasible alternative to reduce bone stress in runners. Consideration must be made to the overall coordination and mechanics of the system and training progression to allow for tissue adaptation to novel loads. Running with a wider step width is associated with an increase in metabolic cost when compared to running with a preferred step width. Arellano and Kram (2011) report that net metabolic power demand increases 9% with ~8% increased step width during running . Unfortunately, we did not measure metabolic costs during our manipulation of step width. Metabolic cost was likely greater in the wider conditions which should be considered prior to intervention. Another factor to consider is the role of step width in postural control during running. Location of the foot under the center of mass may create a more stable stance position with little muscular stabilization required and alterations to limb placement may impact dynamic postural control (Arellano and Kram, 2011).

In conclusion, this research suggests that the stress environment of the tibia is influenced by gait mechanics independent of bone geometry. In general, wider step width conditions were associated with lower stresses; and while not always significant, stresses generally decreased with the increasing step width conditions of this study. These findings in combination with findings from previous studies of injury may suggest that future gait retraining efforts and studies consider step width characteristics, especially if one demonstrates a crossover pattern. However, further studies are needed to ascertain the relationship between step width and injury and to identify the optimal step width for reducing tissue stress during running while minimizing metabolic costs.

Conflict of interest statement

The authors of this manuscript do not have any financial or personal relationships that unduly influenced this work.

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Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at http://dx.doi.org/10.1016/j.jbiomech.2014.04.047.

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