Reducing flight time during running decreases tibial-fibular strains: a finite element analysis

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Abstract

Purpose: Reducing strains within the tibia and fibula during running may reduce the risk of stress fractures. We examined the effect of reduced flight time during running (i.e., grounded running) on finite-element predicted bone strains within the tibia-fibula complex. **Methods:** Nine physically active males ran on an instrumented treadmill at 2.2 m/s using a preferred and reduced flight time technique in a randomized order. Three-dimensional force and motion capture data were recorded during running and a computed tomography image was subsequently acquired for the participant's left leg. An inverse-dynamics-based musculoskeletal modeling workflow was used to calculate bone-on-bone contact and muscle forces during the stance phase of running. These forces served as inputs to a participant-specific finite-element model to estimate peak bone strains and strained volume (i.e., the volume of bone experiencing strains above a specific threshold) within the tibia-fibula complex. **Results:** Guided attempts to reduce flight time was successful with an 18 ms (95% CI: 12 ms, 25 ms; $p<0.001$) reduction in flight time. Reducing flight time was associated with significant reductions in peak tibial/fibular strains (17% lower; 95% CI: -7.1%, -25.0%; $p=0.002$) and strained volume (35% lower; 95% CI: -13.57%, -50.87%; $p=0.007$). **Conclusion:** We conclude that guided attempts to reduce flight time significantly reduces strains in the tibia and fibula during treadmill running at a fixed speed. These results suggest that grounded running may be a viable technique to reduce musculoskeletal loading and stress fracture risk, particularly in slow runners and those runners coming back from injury.

Keywords: Running, stress fracture, injury, biomechanics, gait retraining

Introduction

Running is one of the most accessible and popular forms of physical activity, annually adopted by over 600 million people worldwide [1]. Unfortunately, running-related injuries are common, and these injuries act as barriers to achieving the many physical and mental health benefits of running. Stress fractures are particularly common amongst runners, with an incident rate of 8-18 injuries per 1000 hours of running [2]. Stress fractures have been associated with a mechanical fatigue phenomenon, wherein cyclic submaximal loading leads to microdamage accumulation and structural failure at loading magnitudes much lower than the bone's monotonic failure strength [3]. The rate of microdamage accumulation and fatigue failure in bone is heavily dependent on the resulting strain from the applied loading magnitude [4,5]. Hence, information regarding bone strain in running is of paramount importance to evaluate the efficacy of training interventions to reduce the risk of stress fractures.

Stress fractures in runners frequently occur at the tibia [6,7], and seminal work quantified tibial bone strains *in vivo* through direct strain gauge measurement. These studies investigated differences in tibial strains in response to changes in running speed [8], grade [9], surface [10], and fatigue [11]. *In vivo* strain gauge measurements are invasive and typically limited to a single anatomical location, which do not characterize the complex strain distribution experienced throughout the bone [12]. For this reason, more recent studies have estimated tibial strains in different running conditions by means of musculoskeletal modeling and finite element analysis [13–16]. In addition to a noninvasive estimation of peak strain, this approach has the added benefit of approximating the entire strain distribution within the bone, and the volume of bone experiencing high strain (i.e., above a presumed yield) is strongly associated with the number of cycles to failure [4].

Several modeling studies over the past few decades have investigated potential interventions to reduce tibial loads and/or strains in running. For instance, the work of Edwards et al. (2009) suggested that running with a reduced stride length may decrease bone strains and consequently, the risk of tibial stress fracture [17]; these findings were later verified through a prospective study in colligate runners [18]. Alternative interventions that may reduce tibial loads and thus the risk of stress fracture include running with a wider step width [19] and the use of advanced footwear technology [20]. Recently, Bonnaerens et al., [21] investigated an intervention that eliminated the flight phase of running (i.e., "grounded running"; Gatesy & Biewener, 1991), which suggested reduced peak muscle and joint contact forces at the lower extremity with this technique. Given the complex relationship between applied load and bone strain, we were interested to further characterize the influence of grounded running on the mechanical environment within the tibia.

The purpose of this study was to investigate the effects of grounded running on bone strains in the tibia-fibula complex using a combination of experimentation, musculoskeletal modelling, and finite element analysis. To this end, an inverse-dynamics based musculoskeletal model was used to estimate lower-extremity muscle forces while participants ran with a preferred or grounded form of running. Participant-specific finite element models based on computed tomography (CT) scans were then used to compute tibial-fibular strains in each condition. Based on previous research, suggesting reduced lower-extremity joint contact forces in grounded running [21], we hypothesized that shortening the flight phase in running (indicative of grounded running) would reduce strains within the tibia-fibula complex when compared to preferred running. In addition, it may be beneficial to know what factors are associated with changes in bone strains in running to be used in future training interventions. For this reason, we

also conducted a correlation analysis between running biomechanics, external and internal forces, and finite element-predicted strains. We hypothesized that changes in muscle forces associated with changes in running technique would be correlated with tibial bone strain.

Methods

Participants

Previous studies comparing the effect of preferred versus grounded running demonstrated an effect size of Cohen's d = 2.75 for external forces [23] and Cohen's d = 1.25 for internal forces (e.g., Soleus force [21]). Given uncertainty in the effect size for the primary outcome of bone strain, an *a priori* sample size of N = 8 was calculated, assuming a cautious effect size of Cohen's d = 1.0 to reach a statistical power of $1\text{-}\beta = 0.8$, at a significance level of $\alpha = 0.05$ (GPower 3.1.9.7). To account for possible dropouts, a convenience sample of ten physically active males were recruited and only nine participants (age: 27.2 ± 1.98 years, height: 1.76 ± 0.09 m, body mass: 75.5 ± 11.5 kg) completed the entire protocol described below. The participants did not have any musculoskeletal injuries, defined as any injury that would limit running activity within the six months prior to data collection. The study protocol was approved by the local ethics committee (REB #19-1845) and all participants provided written informed consent.

Experimental Protocol

Participants visited the lab on three different occasions with a maximum of a one-week between sessions. In the first visit, participants were familiarized with the grounded running technique. Participants ran on an instrumented treadmill (Bertec Corps., Columbus, OH), while the vertical component of the ground reaction force (VGRF) was projected in real time on a

screen facing the runner. Grounded running was defined by the elimination of a flight phase and no double-leg-support [22]. This was accomplished by asking participants to remove the flight phase while keeping the lowest VGRF slightly above zero. An error threshold was identified on the screen by a ribbon signifying a VGRF below 50 N. Participants were given as much time as needed to learn and run comfortably with the grounded running technique. In the second visit, twenty retroreflective markers were used to capture the motion of the participant during running. These markers were attached to the left foot (3 markers), left ankle joint (2 markers), left shin (4 markers), left knee joint (2 marker), left thigh (4 markers), left and right hip joints (2 markers), left and right anterior superior iliac spines, and sacrum. The participants ran on the instrumented treadmill with a preferred and grounded running technique for 5 minutes each in a randomized order; running velocity was held constant at 2.22 m/s for both conditions and the VGRF was projected on the screen in a similar manner to the first visit. Only one speed was investigated because maintaining a grounded running technique at faster speeds was not feasible for all participants. Three-dimensional external forces and moments from the instrumented treadmill were recorded at 2000 Hz and motion capture data were recorded at 200 Hz (Vicon motion Analysis Ltd., Oxford, UK) for 30 seconds at a 2.5-minute mark of each condition. In the third visit, a computed tomography (CT) scan was obtained using a GE Revolution GSI (GE Healthcare, Chicago, IL) with acquisition settings of 120 kVp and 180 mA. The CT scans captured the entire participants' left leg and were reconstructed with an in-plane resolution of 0.486 mm by 0.486 mm and a slice thickness of 0.625 mm. A hydroxyapatite calibration phantom (QRM GmbH, Mohrendorf, Germany) was placed in the field of view of each scan to transform CT attenuation values in Hounsfield units to bone mineral equivalent density (ρ_{HU}).

Musculoskeletal Modeling

We used a previously established inverse-dynamics-based musculoskeletal modeling pipeline to estimate muscle and joint contact forces in the left leg during the running protocol [17,24]. The musculoskeletal model included 44 lower extremity muscle-tendon units based on Arnold et al. [25]. For every 1% of the gait cycle, muscle forces were estimated by minimizing the following cost function [26]:

$$J = \sum_{i=1}^{44} \left(\frac{f_i}{PCSA_i}\right)^2$$

where f_i is the estimated force for the *i*th muscle and $PCSA_i$ is the physiological cross-section of the *i*th muscle. The muscle forces were bound between zero and the maximum dynamic forces, adjusted for length and velocity. The moments produced by the muscles about the lower extremity joints were constrained to match the joint moments predicted by the inverse dynamics analysis. Specifically, the flexion-extension, and abduction-adduction moments at the hip, and the flexion-extension moment at the knee and ankle were used as constraints. The optimization was implemented using the interior-point algorithm in the MATLAB Toolbox (i.e., fmincon function). Bone-on-bone joint contact forces were calculated as the algebraic sum of muscle forces from the musculoskeletal model and intersegmental reaction forces from inverse dynamics.

Finite Element Modeling

Finite element models of the tibia-fibula complex followed the procedures previously published by our group [13,27], which predicted tibial bending and torsion similar to those from experimental bone pin studies [28] and strains from experimental measurements [9]. Briefly, the

CT images were manually segmented and converted into 3D models of the tibia and fibula using the Mimics Innovation Suit (v21, Materialize, Leuven, Belgium). The 3D tibia and fibula models were then meshed using NetGen/NGSolve (https://ngsolve.org). Bones were meshed with quadratic tetrahedral elements with an average of ~300,000 elements for the tibia and ~50,000 elements for the fibula. A convergence study from our previous work illustrated that changing the number of elements from 130,000 to 270,000 changed peak tibial strains by 4% [14], suggesting the finite element mesh was adequately refined. Bone was modeled as an inhomogeneous orthotropic material with bone apparent density (ρ_{app}) being used to assign element moduli according to:

$$E_3 = 6570\ \rho_{app}^{1.37}$$

where E_3 is the elastic modulus in the longitudinal direction and $\rho_{app} = \rho_{\mathrm{HU}}$ /0.626 g/cm^3 [29]. The elastic and shear moduli in the other directions along with the Poisson's ratios were determined according to Rho et al., [30] as reported in our previous work [16]

Finite Element Boundary Conditions and Loading

Boundary conditions were defined similar to Khassetarash et al., [27], where the tibial plateau was fixed in all degrees of freedom and a point on the medial proximal tibia was fixed in the anterior-posterior direction; the ankle joint was fixed in the anterior-posterior and medio-lateral directions. The tibia and fibula were connected by binding adjacent elements of the bones at the proximal and distal side. The anterior and the posterior proximal tibia-fibula ligaments were modelled as spring elements with stiffness according to Marchetti et al. (2017). The ankle joint contact force along with a residual moment [16] were applied to the ankle joint centre. Forces from sixteen muscles along with the patellar tendon were applied as point forces at their insertions on the tibia or fibula. The muscles included: biceps femoris (short and long head),

semimembranosus, semitendinosus, tensor fascia latae, gracilis, soleus, tibialis anterior, tibialis posterior, sartorius, flexor digitorum longus, flexor digitorum brevis, flexor hallucis, flexor tertius, peroneus longus, and peroneus brevis. The quasi-static finite element simulation was performed in Abaqus (version 2021, Dassault System, RI, USA) using muscle forces from the instant of peak ankle joint contact force.

Variables of Interest

We focused our analysis on elements in the tibial and fibular diaphysis [14]. For these elements, we calculated the 90th percentile pressure modified von Misses strain [32] as a measure of peak strain and strained volume (defined as the volume of elements experiencing a pressure modified von Misses strain above 3000 $\mu\varepsilon$), as the latter was shown to correlate well with the fatigue life of whole rabbit tibiae [4]. To quantify changes in biomechanics associated with running technique, secondary dependent variables of interest included spatiotemporal parameters (i.e., flight time, contact time, step frequency, duty factor) and the VGRF.

Statistical Analysis

A Shapiro-Wilk test of normality was performed on all variables of interest. The results were reported in mean ± standard deviation unless otherwise stated. Running technique (i.e., preferred or grounded) was treated as an independent variable while the finite element results (i.e., 90th percentile strain and strained volume) as well as spatiotemporal and VGRF were treated as dependent variables. Paired t-tests were conducted for dependent variables to compare preferred and grounded running conditions. The statistical significance was set at 0.05. An exploratory correlation analysis was also performed between finite element predicted strains, spatiotemporal variables, VGRF, and select muscle or tendon forces (i.e., soleus, patellar tendon). These two muscle and tendon forces were chosen because of their large magnitudes and

significant contributions to the bone strain distribution. A Benjamini-Hochberg [33] false discovery rate correction was applied to the significance values from the correlation analysis; statistical significance was set at 0.05 after the p-values were corrected.

Results

Attempts to decrease flight time through grounded running were successful. Indeed, flight time was reduced from 31 ± 11 ms during preferred running to 13 ± 4 ms during grounded running ($p < 0.001$) (Figure 1). No changes in contact time (345 ± 25 ms in preferred condition vs. 354 ± 21 ms in grounded running; P = 0.21) or step frequency (160 ± 8.8 steps per minute in preferred condition vs. 164 ± 8.9 steps per minute in grounded running; $p = 0.24$) associated with the intervention were observed. Duty factor increased from 0.46 ± 0.16 during preferred running to 0.48 ± 0.06 during grounded running ($p < 0.001$). The VGRF also reduced significantly during grounded running (1605 ± 211 N vs. 1351 ± 240 N; $p = 0.006$).

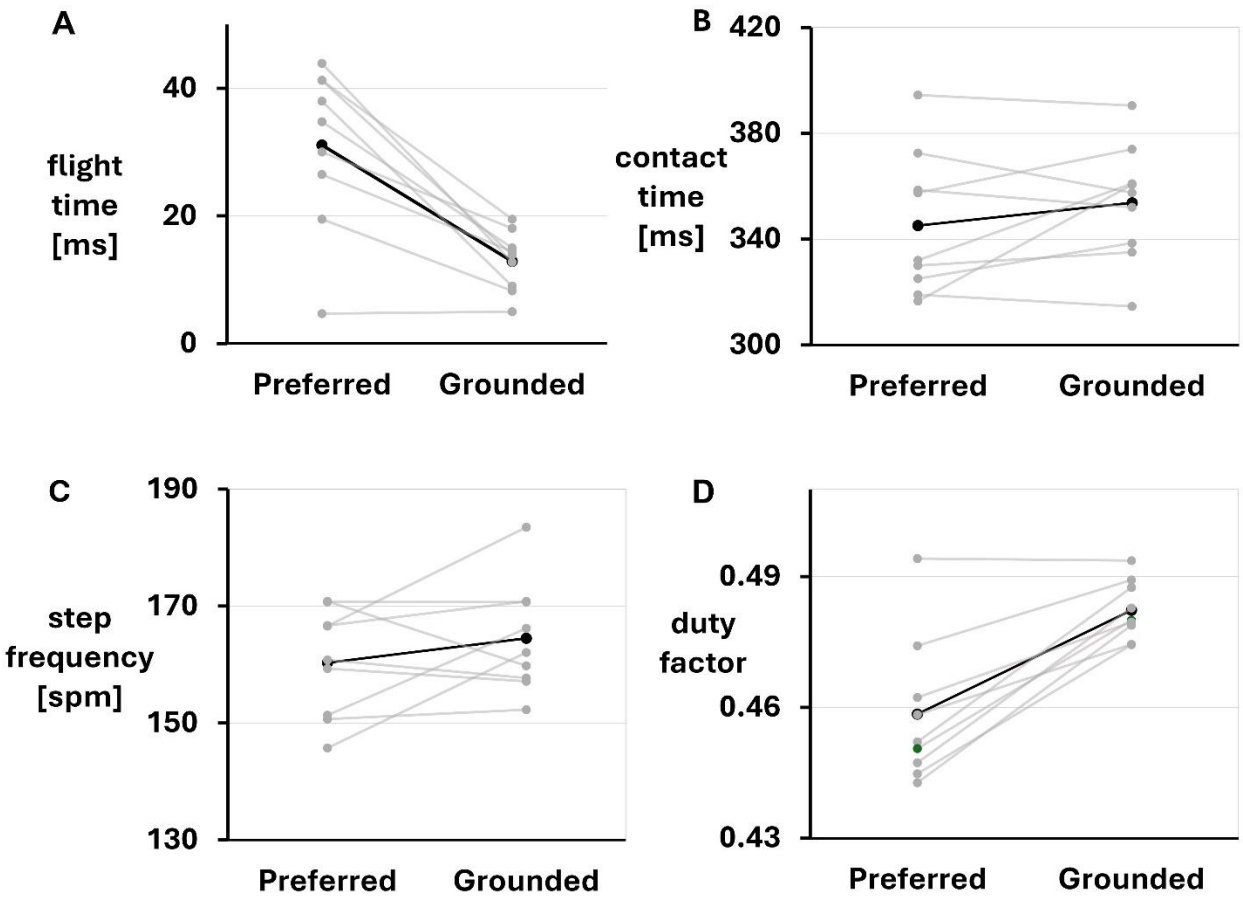

Figure 1**.** Spatiotemporal variables during preferred and grounded running including: A) flight time B) contact time, C) step frequency, and D) duty factor. In both graphs, the thick black line represents the mean of all participants, and the gray transparent lines represent individual participants.

The distribution of pressure-modified von Misses strain for a representative participant is shown in **Figure 2**. In general, the strains on the anterior and posterior aspects of the tibial and fibular midshaft were lower during grounded running when compared to preferred running. **Figure 3** depicts the strained volume for the same participant. In both conditions, most of the volume of bone experiencing over 3000 με (i.e., elements contributing to the strained volume calculation) were on the posterior aspect of the tibia and fibula. Quantitatively, peak strain was significantly lower during grounded running compared to preferred running (**Figure 4**A, 4193 ± 737 με vs. 3498 ± 738 με; $p = 0.002$). Strained volume was also significantly reduced in the

grounded running condition when compared to preferred running (**Figure 4**B, 9680 ± 3159 mm^3 vs. 6305 ± 2905 mm^3; $p = 0.007$).

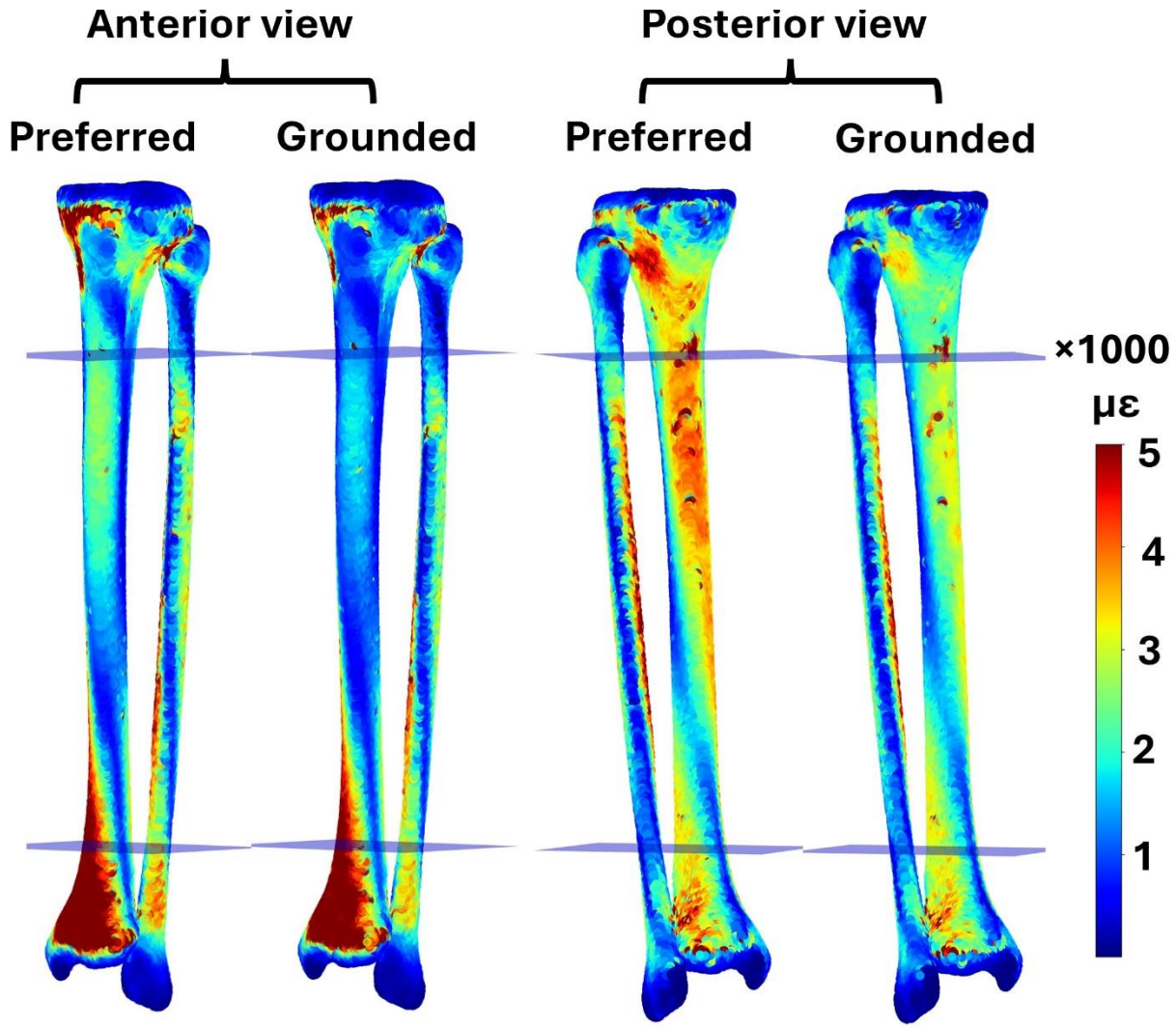

Figure 2. Pressure-modified von Misses strain distribution in the tibia-fibula complex during preferred and grounded running for a representative participant. The elements between the proximal and distal planes shown in the figure were considered for peak strain calculation.

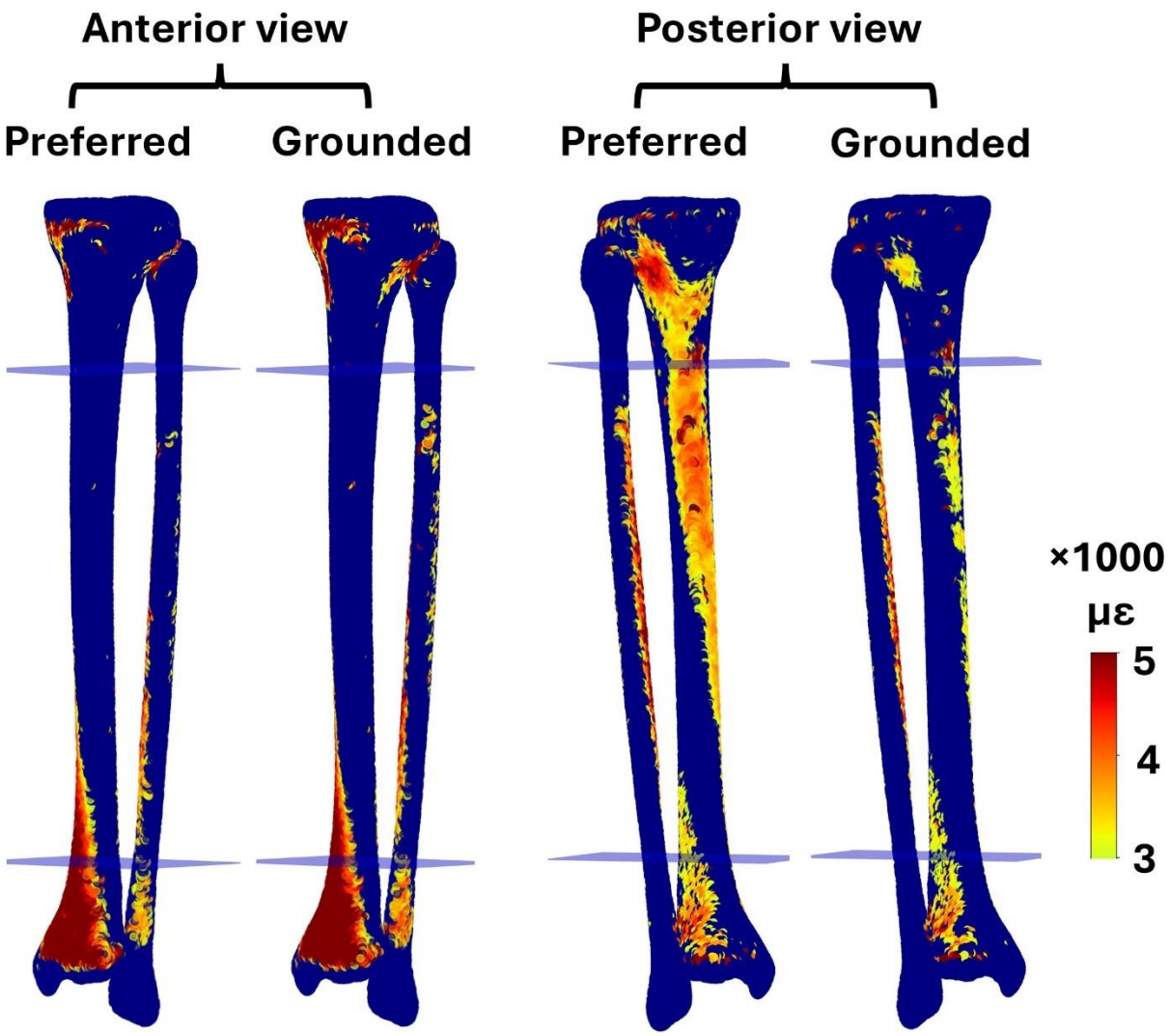

Figure 3. Strained volume (i.e., volume of elements experiencing strains above 3000 με) during preferred and grounded running for a representative participant. The elements between the proximal and distal planes shown in the figure were considered for strained volume calculation.

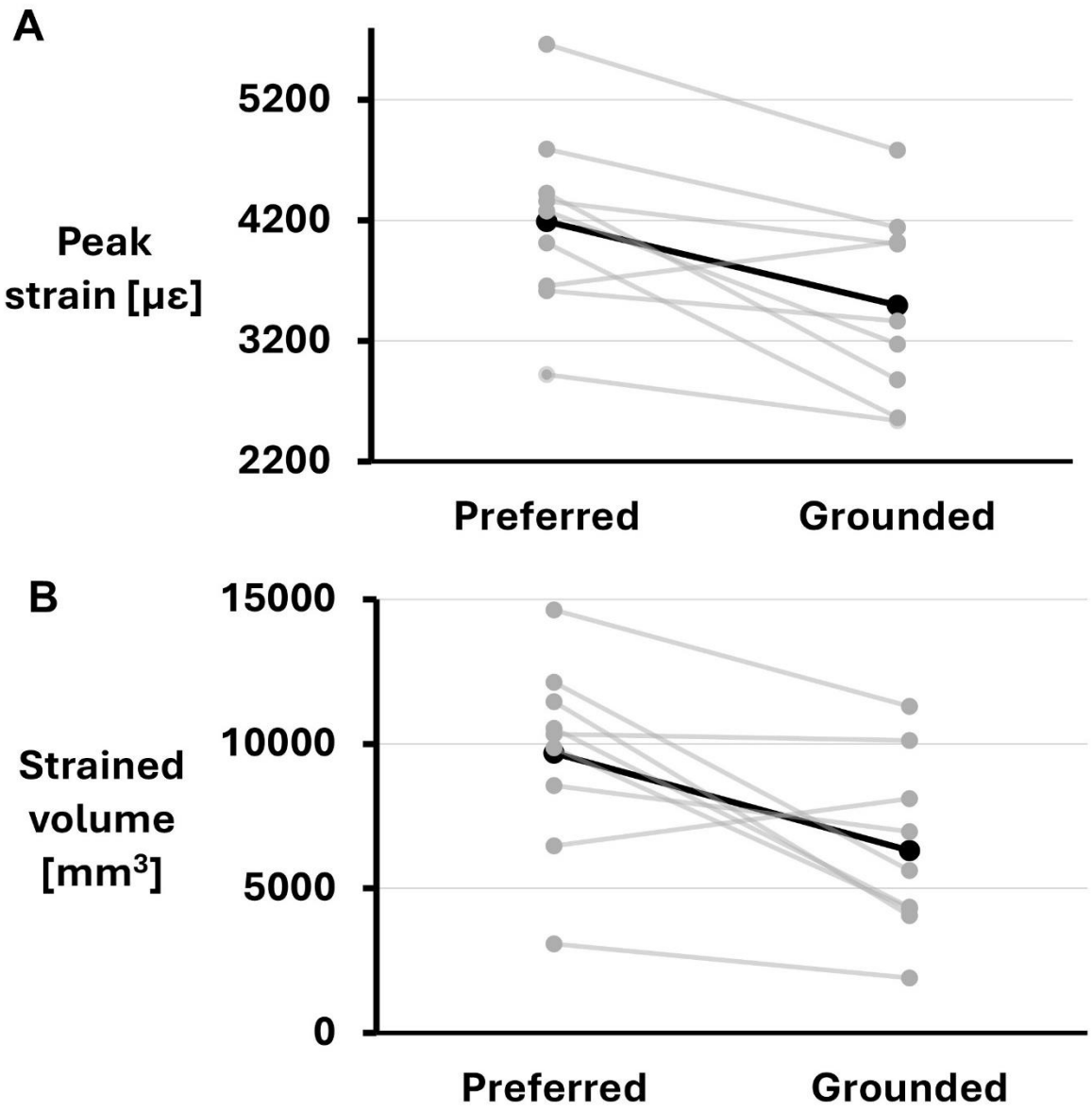

Figure 4. Finite element predicted A) peak strain (i.e., 90th percentile strain within the midshaft of the tibia-fibula complex) and B) strained volume (i.e., volume of bone experiencing strains above 3000 με within the midshaft of tibia-fibula complex). In both graphs, the thick black line represents the mean of all participants, and the gray transparent lines represent individual participants.

Correlation analyses (**Table 1**) revealed that between 70% to 90% of the change in peak strain could be explained by peak patellar tendon force ($p = 0.016$) and peak soleus force ($p < 0.001$), while strained volume was not well-explained by any of the variables explored (all $p \geq 0.19$).

Table 1. Correlation analysis between the finite element outputs and select musculoskeletal loading and running biomechanics variables.

	Variable	**R-squared**	**Corrected *p*-value**
Δ peak strain	Δ VGRF	0.160	0.328
	Δ flight time	0.453	0.468
	Δ duty factor	0.712	0.097
	Δ Soleus force	0.912	**<0.001**
	Δ Patellar tendon force	0.702	**0.023**
Δ strained volume	Δ VGRF	0.233	0.285
	Δ flight time	0.317	0.192
	Δ duty factor	0.388	0.336
	Δ Soleus force	0.517	0.328
	Δ Patellar tendon force	0.5641	0.192

Discussion

Mechanical fatigue is believed to play an important role in the development of stress fracture [3], and the fatigue behaviour of bone is strongly dependent on the resulting strain from the applied cyclic load [34]. In this study, we investigated the effects of grounded running (i.e.,

reducing flight time) on finite element-predicted tibial/fibular strains. We observed a significant reduction in both the peak strain and strained volume in grounded running when compared to a preferred running technique. The reductions in peak strain were explained, in part, by corresponding reductions in muscle forces associated with grounded running. These findings suggest that grounded running may represent a practical gait strategy for lowering the risk of stress fractures.

The reductions in peak strain and strained volume associated with grounded running would be expected to reduce the risk of stress fracture considerably. On average, grounded running reduced the magnitude of peak strain by an average of 695 ± 581 με (95% CI: 316 με, 1076 με) when compared to preferred running. Using a stress-life approach, we can estimate how much longer a participant could hypothetically run by adopting a grounded running technique before bone failure. The relationship between bone stress, or strain, and the number of cycles to failure is well explained by an inverse power-law with an exponent ranging from -5 to -15 [35]. Assuming an exponent of -7, used previously for estimations of bone damage [36,37] and stress fracture risk [17], the ~17% reduction in peak strain associated with grounded running would correspond to a ~256% increase in the number of cycles to failure. In other words, a grounded running technique may allow an individual to run more than three times farther before experiencing a stress fracture. Note that this simplified estimation does not consider the large scatter observed in *ex vivo* cyclic tests of bone [38], or the bone adaptation and bone remodeling response to microdamage and mechanical loading [39].

Prior research investigating grounded running demonstrated a reduction in external forces [23] as well as muscle and joint contact forces [21] when compared to a preferred technique. The current study contributes to this literature by demonstrating these reductions in applied load were

also associated with concurrent reductions in bone strain. Our findings underscore the limitations of relying solely on external loading parameters for predicting bone strain. Indeed, the correlation analysis between finite element outputs and VGRF revealed weak associations (R^2 = 0.16-0.23), indicating that external forces alone are insufficient predictors of peak strain or strained volume. On the other hand, our study found that select muscle-tendon forces explained a substantial portion of the variance in peak strain (R^2 = 0.70-0.91), suggesting their utility as a surrogate measure of bone strain and presumably stress fracture risk. These findings highlight the crucial role of muscle forces in modulating the stress-strain environment within bone, beyond what is captured by external loading variables.

While this study provides insights into the bone strain distribution in grounded running, it is important to acknowledge certain limitations inherent to musculoskeletal and finite element modeling. For example, the estimation of muscle forces relied on an optimization approach with a cost function to minimize the sum of muscle stresses squared. While this cost function has been implicitly tested [40], it lacks explicit validation. Moreover, the muscle parameters used in the model, including origins, insertions, moment arms, and theoretical maximum forces, were based on estimations rather than direct measurements from our participants. These estimations may affect the magnitude of strain but are highly unlikely to affect our interpretation given the repeated measure design of this study. With regard to the finite element model, while the boundary conditions were grounded in physiological principles [16], they may not perfectly replicate the actual joint mechanical behavior *in vivo*. Nevertheless, our prior research [13,27] has demonstrated promising results in predicting bone bending and strains that align with experimental findings from bone pin [28] and strain gauge [9] studies.

It is also crucial to remember that this investigation focused on an acute intervention, and the long-term implications of prolonged grounded running on the musculoskeletal system remain unknown. Future studies should consider the potential for adaptations and changes in bone strains with chronic exposure to grounded running. It was encouraging to see the relative ease with which participants adopted the grounded running technique. Grounded running was accomplished with minimal instruction, but this was facilitated through real-time visual feedback. If grounded running program were to be implemented in practice several laboratory-based gait training sessions would likely be required (e.g., [41]). In this regard, there are several questions left to be answered regarding the length and frequency of gait training sessions for optimal adoption and recall. Although we would not recommend grounded running as a universal strategy, it could represent a viable load-management technique for those runners with a history of overuse injury, particularly in situations where running speed can be sacrificed. It is important to note that one participant did not reduce their tibial strains with grounded running (see Figure 4). This participant had a duty factor of 0.49 during the preferred running condition and was therefore unable to further increase duty factor during grounded running. This highlights the potential need for practitioners to pre-screen those individuals that may benefit most from a grounded running intervention.

Conclusion

This study suggested that reducing the flight phase of running by transitioning to a grounded running technique is an effective intervention for decreasing strains within the tibia-fibula complex in the majority of runners. In this regard, grounded running may represent a viable technique for reducing the risk of stress fractures. Although confirmation through

prospective studies is still needed, reducing the flight phase appears to be a promising strategy, particularly for running at slower velocities.

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Conflict of interest

No conflict of interest to declare.

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