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Keywords

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1. Introduction

Running represents an easily accessible means of physical activity, linked to a range of physiological [1] and psychological [2] advantages. Nevertheless, running is also correlated with a notably elevated occurrence of chronic pathologies [3], with the annual prevalence of these conditions reaching up to 20–80% of runners [4]. Amongst chronic running-related injuries, bone stress fractures are one of the most frequently reported, constituting as many as 30% of all musculoskeletal injuries associated with running [5]. The tibia has traditionally been identified as the most vulnerable site for stress fractures [6, 7], with around 74% of such injuries manifesting at this location [8]. Stress fractures present distinctive challenges due to their prolonged recovery duration and heightened likelihood of recurrence [9].

Running, as a repetitive cyclic loading activity, consistently subjects the skeletal system to stress, inducing bone loading and potentially initiating bone fatigue [10]. Strain is considered the most accurate representation of actual structural bone damage [11].

In-vivo strains during running tend to be significantly below the bone's ultimate strength, leading to the perception of stress fractures as manifestations of mechanical fatigue [12], often described through an inverse power law relationship [13]. Stress fractures result from the accumulation of microscopic damage within the bone matrix [14]. Incorporating sufficient rest intervals between running sessions allows time for bone remodelling, potentially enhancing bone integrity [15]. However, if the pace of damage accumulation exceeds the capacity for remodelling and adaptation, minor cracks may develop in the bone matrix, progressing into stress fractures [16]. Notably, when the tibia experiences low strain levels, damage accumulation diminishes, providing more time for the tissue to repair microcracks. Conversely, elevated strains lead to a surplus of damage that outpaces the repair and adaptation process [17]. Consequently, identifying tibial loading patterns that minimize strain levels during running may contribute to the prevention of stress fracture injuries [11].

As the primary connection between the foot and the ground, running shoes have been suggested as a crucial factor that can potentially influence the biomechanical elements associated with the development of chronic injuries [18]. Running shoe manufacturers have incorporated reduced levels of midsole cushioning into their footwear designs [19]. Minimal running shoes are distinguished by a low or zero heel-toe drop, high levels of midsole flexibility, and diminished mass [20].

The loading rate of the vertical ground reaction force (GRF) and tibial accelerations and are frequently employed as surrogate indicators for tibial loading and have been long hypothesized as being connected to the initiation and progression of tibial stress fractures [21]. There has been substantial research attention directed toward assessing the impact of minimal running shoes on vertical loading rates and tibial accelerations. Minimal running shoes have been shown to mediate significant increases in vertical loading rates and tibial accelerations compared to conventional footwear [19, 22–23]. However, previous observational biomechanical assessments investigating indices of tibial loading were carried out in runners who were not habitual users of minimal footwear. Tam et al. [24] suggested that investigations involving non-habitual minimal footwear users may not provide a representative picture and that an adjustment/ habituation period is essential to adapt to such novel footwear stimuli. Consequently, additional research on minimal and traditional footwear is needed for runners who habitually utilize these respective footwear modalities.

Crucially, recent evidence indicates that surrogate measures such as tibial acceleration and loading rates of the vertical ground reaction force (GRF) do not accurately represent tibial bone loading in running [21]. Finite element modelling has demonstrated a capacity for more realistic estimates of in vivo tibial bone strains [25, 26], directly tied to the development of stress fractures [11]. This suggests that this technique can be employed for well-informed predictions regarding the potential for tibial damage. Previously, only acute observational analyses of minimal and conventional footwear were feasible, leaving the long-term effects on the initiation and progression of tibial stress fractures unknown. Substantial progress in finite element analyses in recent years now enables computational probabilistic modelling of the tibia [25, 26], facilitating the quantification of the probability of tibial stress fractures in runners employing different footwear modalities. However, neither finite element nor probabilistic modelling of the tibia has been applied to examine differences between minimal and conventional running shoes in individuals who regularly use minimal footwear during running.

Hence, the objective of this study is to assess the impact of minimal and conventional footwear on tibial strains and stress fracture probability in both habitual and non-habitual minimal footwear users, utilizing a combined approach of finite element analysis and computational probabilistic modelling. The outcomes of this research will provide novel insights into the effects of minimal and conventional footwear on tibial strains during running, as well as longitudinal stress fracture probability. The hypothesis posits that minimal footwear will elevate tibial strains and increase the likelihood of tibial stress fractures in both habitual and non-habitual minimal footwear users, with no discernible differences between the two groups.

2. Materials and methods

2.1. Participants

Ten male participants who regularly ran in traditional running shoes (age = 27.67 ± 5.57 years, stature = 1.71 ± 0.03 m, body mass = 68.76 ± 4.78 kg, body mass index = 22.99 ± 1.16 kg/m2, running experience = 6.87 ± 1.52 years and weekly running volume = 45.99 ± 3.26 km) and another group of ten males who regularly used minimalist running shoes (age = 33.50 ± 4.58 years, stature = 1.75 ± 0.04 m, body mass = 71.74 ± 7.74 kg, body mass index = 23.33 ± 1.96 kg/m2, running experience = 7.03 ± 2.09 years, weekly running volume = 43.60 ± 5.46 km and duration of habitual minimal footwear utilization = 34.94 ± 7.80 months) volunteered to participate in this research. Comparisons between groups were undertaken for age, mass, stature, body mass index, running experience and weekly running volume and were all non-significant (P = 0.13–0.85). To be eligible for this study, all participants had to complete at least 35 kilometres of running training per week and be between the ages of 18–40. The participants in the habitual minimalist footwear group needed to have exclusively trained in such shoes for a at least 24 months, in footwear scoring of at least 75 on the minimalist index developed by Esculier et al. [20]. All participants were free from injury at the time of data collection and provided written informed consent, following the principles outlined in the Declaration of Helsinki. The research procedure for this study received approval from a university ethics committee (REF 637).

2.2. Experimental footwear

The footwear examined in this investigation were 1. Conventional footwear (New Balance, 1260 v2, New Balance, Boston, Massachusetts, United States); and 2. Minimal footwear (Vibram Five-Fingers, ELX, Vibram, Albizzate, Italy) (Figure 1ab). Both footwear' were scored using the minimalist index of Esculier et al. [20] (Table 1).

Table 1. Characteristics of experimental footwear

	Minimal	Conventional
Mass (g)	167	285
Heel thickness (mm)	7	25
Heel-toe drop (mm)	0	14
Esculier et al., (2015) minimalist index	92	20

Figure 1. Experimental footwear (a. = conventional & b. = minimal)

2.3. Procedure

Volunteers ran at a self-selected velocity through a 22-meter biomechanics laboratory, contacting an embedded piezoelectric force platform (Kistler Instruments Ltd., Winterthur, Switzerland) that recorded data at a frequency of 1000 Hz, with their right limb, (which was dominant in all participants). The stance phase of running was defined as the period during which the vertical GRF exceeded 20 N [19]. Each participant successfully completed five trials for each type of footwear, meeting the criteria of maintaining the specified velocity range, achieving complete foot contact with the force platform, and displaying no visible alterations in their gait due to the experimental conditions. Furthermore, to ensure that a consistent running velocity was obtained, the difference in the braking and propulsive acceleration portions of the anterior-posterior GRF were examined using a custom Matlab program (MATLAB, MathWorks, Natick, USA) and trials were rejected when this difference was more than 10% of the total rectified braking-propulsive GRF [27]. The order in which participants ran with each type of footwear was counterbalanced. Simultaneous kinematic and GRF data were collected, with kinematic data recorded at a rate of 250 Hz using an eight-camera motion analysis system (Qualisys Medical AB, Goteburg, Sweden.) Prior to each data collection session, a dynamic calibration process for the motion capture system was performed.

The body segments were modelled with six degrees of freedom using the calibrated anatomical systems technique, as detailed by Cappozzo et al. [28]. To establish the anatomical frames for the thorax, pelvis, thighs, shanks, and feet, retroreflective markers were positioned at key bony landmarks, which included C7, T12, xiphoid process, iliac crest, anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, calcaneus, first metatarsal, and fifth metatarsal. Tracking clusters, constructed from carbon fiber and equipped with four non-linear retroreflective markers, were securely attached to the thigh and shank segments using rigid sports tape. The foot segments were tracked using markers on the calcaneus, first metatarsal, and fifth metatarsal, the pelvic segment was tracked with the PSIS and ASIS markers, and the thorax segment was tracked using markers positioned at T12, C7, and the xiphoid process. To establish references for anatomical markers in relation to tracking markers/clusters, static calibration trials were conducted. The centers of the ankle and knee joints were determined as the midpoints between the malleoli and femoral epicondyle markers, whereas the center of the hip joint was calculated using a regression equation based on the positions of the ASIS markers. Each segment's Z (transverse) axis was oriented vertically from the distal segment end to the proximal segment end. The Y (coronal) axis was oriented within the segment from posterior to anterior. Finally, the X (sagittal) axis orientation was determined using the right-hand rule and directed from medial to lateral.

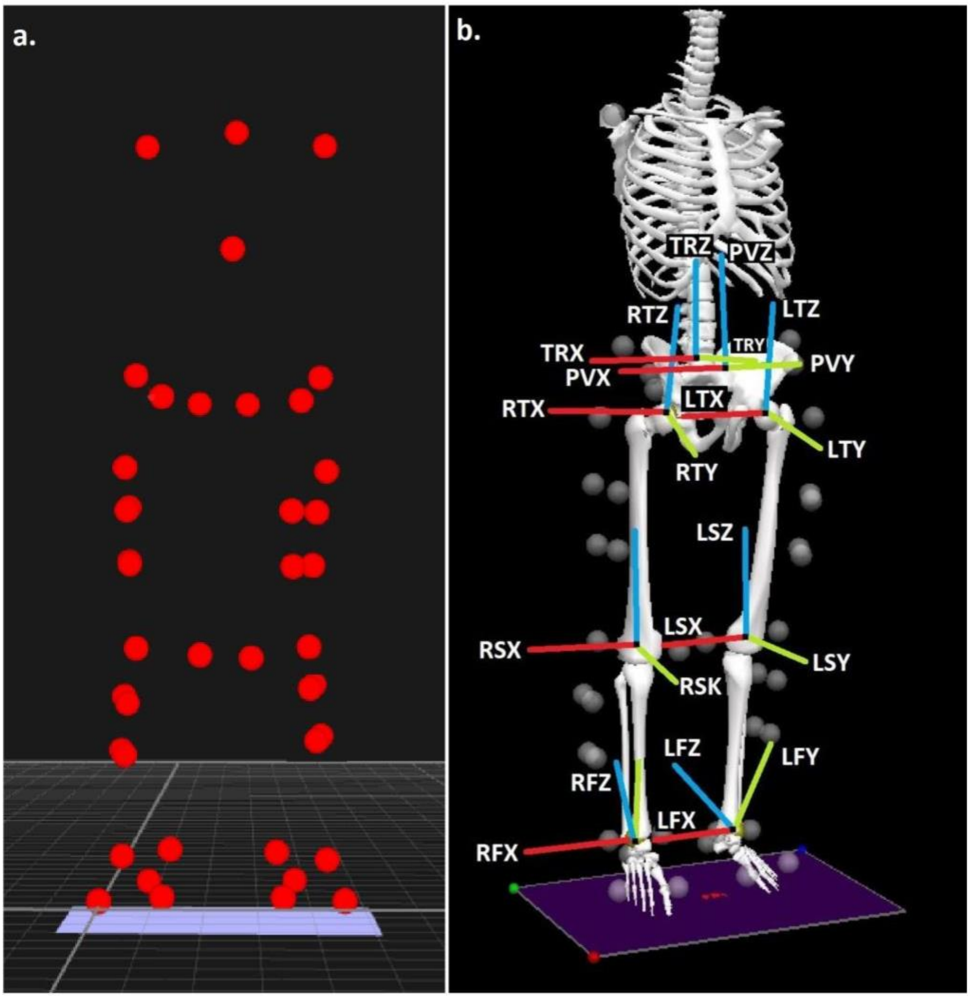

Figure 2. Experimental retroreflective marker positions (a) and (R = right & L = left, TR = trunk, P = pelvis, T = thigh, S = shank & F = foot, X = sagittal plane, Y = coronal plane & Z = transverse plane).

2.4. Processing

Dynamic trials were digitized using Qualisys Track Manager (Qualisys Medical AB, Goteburg, Sweden) to identify anatomical and tracking markers, then exported as C3D files to Visual 3D (C-Motion, Germantown, MD, USA). All data were linearly normalized to 100% of the stance phase. GRF data and marker trajectories were smoothed with cut-off frequencies of 50 Hz at 12 Hz respectively, using a low-pass Butterworth 4th order zero lag filter.

2.5. Running biomechanics

In accordance with the methods described by Addison and Lieberman [29], an impulse-momentum modelling technique was utilized to calculate the effective mass (%BW), i.e., the proportion of the bodyweight that comes to a full stop during the impact phase. This parameter was evaluated using the following formula:

Effective mass = Integral of the vertical GRF / (Δ foot vertical velocity + gravity × Δ time)

The impact peak was delineated in conventional running shoes as the first discernible peak in the vertical GRF. However, in the case of minimal footwear, where a consistent impact peak is not always apparent, we adhered to the criteria set forth by Lieberman et al. [30] and Sinclair et al. [31]. Accordingly, we positioned the impact peak in minimal footwear at the same relative location as observed in conventional running shoes. The time it took to reach the impact peak (referred to as Δ time) was measured as the duration from footstrike to the occurrence of the period impact peak. The integral of the vertical ground reaction force was calculated during the impact peak period using a trapezoidal function. Additionally, the change in foot vertical velocity (Δ foot vertical velocity) was determined as the difference in vertical foot velocity between the moments of footstrike and the impact peak, following the methodology of Chi and Schmitt [32]. Foot velocity was assessed by quantifying the vertical velocity of the foot segment's center of mass within Visual 3D [31].

The strike index, which serves as an indicator of the foot strike pattern, was calculated by considering the position of the center of pressure at footstrike in relation to the entire length of the foot. This calculation followed the procedures delineated by Squadrone et al. [33]. A strike index (%) falling within the range of 0–34% indicated a rearfoot strike pattern, >34–67% signified a midfoot strike pattern, and >67–100% represented a forefoot strike pattern. Furthermore, the step length (m) was determined as the horizontal position of the foot's centre of mass between the right and left limbs at footstrike, in accordance with the methodology provided by Sinclair et al. [34].

Lower extremity kinematics were quantified using an XYZ Cardan sequence (where X is flexion–extension, Y is ab/adduction, and is Z is internal–external rotation). The following hip, knee and ankle three-dimensional kinematic measures of peak angle and range of motion (representing the angular displacement from footstrike to peak angle) were extracted. Running velocity (m/s) was also quantified within Visual 3D, using the linear velocity of the model centre of mass in the anterior direction during the stance phase [34].

2.6. Musculoskeletal simulation

Data associated with the stance phase were exported from Visual 3D to OpenSim 3.3 software (Simtk.org, Stanford, CA). To cater to the unique anthropometric characteristics of each participant, a validated musculoskeletal model was used, which underwent scaling to accommodate the individual anthropometric characteristics of each participant (Figure 3). It featured 12 segments, 19 degrees of freedom, and a total of 92 musculotendon actuators [35]. This model was employed to estimate muscle and joint contact forces in the lower extremities. Initially, a residual reduction algorithm, as detailed by Delp et al. [36], was applied to address any dynamic inconsistencies between the kinematics derived from the measured GRF and the model. Subsequently, muscle kinetics were determined through static optimization procedures, following the methods described by Steele et al. [37].

As muscle forces represent the primary factor influencing joint contact forces [38], subsequent to the static optimization procedure, three-dimensional ankle joint contact forces, presented in the tibial reference frame, were computed through the joint reaction analysis function within OpenSim. This process utilized the muscle forces derived from the static optimization procedure as input. The resulting ankle joint contact force was determined using three-dimensional Pythagorean theorem, and normalized ankle joint contact forces (BW) were extracted for each anatomical axis (anterior-posterior, axial, and mediolateral) at the moment of the peak resultant load.

From the aforementioned static optimization procedures, the normalized muscle forces (BW) with tibial attachment (including biceps femoris long head, biceps femoris short head, extensor digitorum longus, extensor hallucis longus, flexor digitorum longus, flexor hallucis longus, gracilis, rectus femoris, sartorius, semimembranosus, semitendinosus, soleus, tensor fasciae latae, tibialis anterior, tibialis posterior, vastus intermedius, vastus lateralis, and vastus medialis) were extracted at the moment of the peak resultant ankle joint contact force. Furthermore, muscle forces from other muscles crossing the ankle joint, namely medial gastrocnemius, lateral gastrocnemius, peroneus brevis, peroneus longus, and peroneus tertius, were also obtained at the same relative time point.

Finally, the attachment points of each of the aforementioned muscles (with tibial attachment) were extracted using the OpenSim plugin developed by van Arkel et al. [39] (https://simtk.org/projects/force_direction). Using the same plug-in, anatomically directed muscle forces onto the tibia at their attachment points, for each muscle were calculated at the instance of the peak resultant ankle joint contact force in all three anatomical directions. Positive values represent anterior, upwards and laterally directed forces onto the tibia.

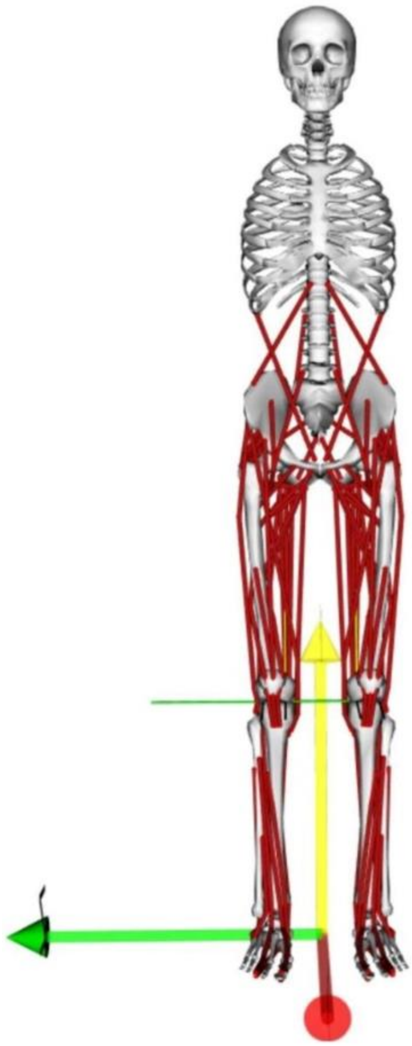

Figure 3. OpenSim musculoskeletal simulation model.

2.7. Finite element analyses

FEBio software (developed by Musculoskeletal Research Laboratories, Salt Lake City, Utah) was utilized to conducting the finite element analysis necessary for calculating tibial strains. The construction of the tibial surface and trabecular model involved employing the statistical shape modelling source code developed by Keast et al., [40] (accessible at https://simtk.org/projects/ssm_tibia). The resulting model comprised 33,004 quadratic tetrahedral elements (Figure 4). Material properties were designated based on those previously adopted by Edwards et al., [25], with an elastic modulus of 17.0 GPa for cortical bone and 1.0 GPa for trabecular bone. Both components were attributed a Poisson's ratio of 0.3 [25].

Each model had boundary conditions imposed, involving complete constraints applied at the tibial plateau [41] (Figure 4a). Net three-dimensional ankle joint contact forces, derived from the musculoskeletal simulation analyses, were then applied to the distal aspect of the tibia [41] (Figure 4b). Additionally, anatomically directed net muscle forces, were applied at every muscle attachment point on the tibia, utilizing the forces obtained from static optimization (Figure 4c). Given that certain bi-articulating muscles, such as the gastrocnemius, generate substantial forces during running without direct insertion points onto the tibia itself [41], their contribution to tibial strain was accounted for by calculating a residual ankle joint moment following the approach outlined by Haider et al., [41]. This residual moment was applied to the distal tibia (Figure 4d). The 50th and 90th percentile von Mises strain ($\mu\varepsilon$) were then extracted for subsequent analysis [42].

Figure 4. Depiction of finite element model mesh with loading and boundary conditions. The tibial plateau was fully constrained (a.). Ankle joint contact forces were applied to the distal tibia (b.), muscle forces (not all shown here) were applied as concentrated forces at their insertion point onto the tibia (c.) and residual moments were applied at the distal tibia (d.)

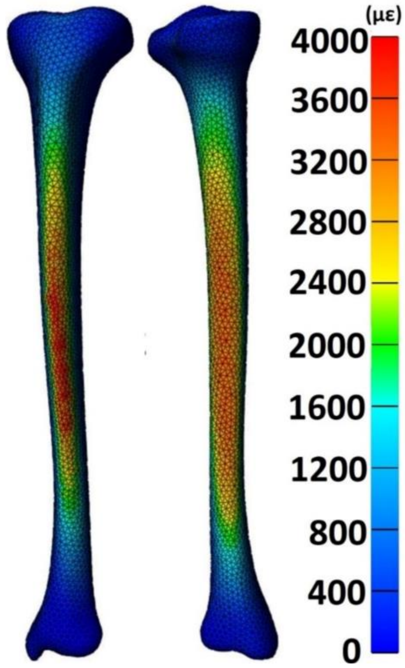

Figure 5. Representative tibial strain distribution on the tibia

2.8. Probabilistic stress fracture model

We determined the probability of stress fracture for each participant in each footwear condition firstly by accounting for the daily running distance which was included into the model as runners completing 5.0 km/day for 100 consecutive days [25, 26, 43]. The number of loading cycles/ footfalls per day in each footwear condition was quantified by dividing the modelled daily running distance i.e. 5.0 km by the stride length in each footwear outlined in section 2.4.1 [43].

The probability of tibial stress fracture was assessed through a probabilistic model considering bone damage, repair, and adaptation, aligning with methodologies from prior analyses [25, 26, 43]. The fatigue life of the tibial bone was modelled based on the standard fatigue equation [44]:

$$\mathbf{FLT = C\Delta\varepsilon^{-n}}$$

In the above equation, FLT represents the number of loading cycles to failure and $\Delta\varepsilon$ denotes the strain range obtained from finite element analysis. Since strain magnitude is zero for certain modelled tibial elements, the maximum strain magnitude from the finite element analysis was utilized to denote the strain $\Delta\varepsilon$. The variable 'n' signifies the slope of the stressed-life curve of bone, and 'C' is a constant. Carter & Caler [44] found a slope of n = 6.6 for fatigue damage of bone at strain magnitudes corresponding to human locomotion [25, 26, 43].

As bone adaptation is mediated as a function of applied loading, there is a concurrent increase in bone cross-sectional area, leading to a gradual attenuation of tibial strains over time. Considering a maximum deposition of lamellar bone accumulation at 4 μm/day on the periosteum membrane [45], an adaptation function was quantified using beam theory-based equations [25, 26, 43]. This adaptation function was calculated as the ratio of strain following bone accumulation to strain with the initially modelled bone geometry. The product of this adaptation function and $\Delta\varepsilon$ was employed to determine alterations in tibial strains due to bone adaptation. An equivalent strain ($\Delta\varepsilon$AD) for each element, accounting for adaptation, was then computed, where tT represents the total modelled duration over which bone adaptation occurred (i.e., 100 days) [46]:

$$\Delta\boldsymbol{\varepsilon}_{AD} = (1/tT \int^{Tt}_0 \Delta\boldsymbol{\varepsilon}^n dt)^{1/n}$$

Given the significant variation in the fatigue life of bone, a widely employed technique in fatigue mechanics to assess the probability of bone failure with adaptation (PfA) is the Weibull approach [47]. Hence, a modified Weibull function was employed, taking into account stressed volume [46]:

$$Pf_A = 1 - \exp [- (V_s / V_{so}) (t/t_f)^w]$$

The variables for the above equation were derived from literature on experimental fatigue testing. This enables the calculation of PfA for a specimen with the stressed volume Vs (obtained from finite element meshes) over the time interval from zero to t. Here, Vso represents the reference stressed volume, tf is the reference time until failure at the applied strain range and number of loading cycles per day, and w signifies the degree of scatter in the material.

As ΔεAD varies across the entire tibial body, PfA exhibits differences from one element to another. Through the finite element analysis, unique PfA indices could be determined for each element. If there are k total elements, the PfA for the entire tibial body represents the probability of failure for any single element [48]:

$$Pf_A = 1 - (1- P_1)(1- P_2)(1- P_3)\ldots(1-P_k)$$

Elements with comparable strain magnitudes were clustered together, following the approach of Taylor & Kuiper [48], who established that eight element groups could be employed without substantial error. The stressed volume (Vs) for each of the eight groups was computed by adding up the element volumes within each group. Subsequently, using the strain values from each group, the aforementioned formulae were applied to calculate a singular PfA for the entirety of the tibia.

Similar to the variability observed in the fatigue life of bone, there is significant diversity in the duration required for the repair of bone microcracks. As estimated by Taylor et al. [46], this repair time is approximately 18.5 ± 12.5 days. Consequently, the cumulative probability of bone repair (PR) was determined by employing a second Weibull function [46]:

$$P_R = 1 - \exp [- (t/t_r)^m]$$

where tr is the reference time for repair and m articulates the degree of scatter in repair time.

Lastly, by calculating the probability that bone will not undergo repair (1-PR) and multiplying it by the instantaneous probability of PfA, integration over time resulted in the cumulative probability of tibial bone failure expressed as a percentage (%) with repair and adaptation (PfRA).

2.9. Statistical analyses

Descriptive statistics of means and standard deviations (SD) were calculated for each of the experimental variables outlined above. Differences in participant characteristics (age, mass, stature, body mass index, running experience and weekly running volume) were examined using between groups linear mixed-effects models. Biomechanical parameters were contrasted using four separate comparisons adjusting each model for running velocity, which was modelled as a continuous fixed covariate [34]. For comparisons 1 & 2 within subjects linear mixed effects models were adopted; (1) non-habitual minimal footwear users in minimal footwear vs. non-habitual minimal footwear users in conventional footwear and (2) habitual minimal footwear users in minimal footwear vs. habitual minimal footwear users in conventional footwear. For comparisons 3 & 4 between groups linear mixed effects models were adopted (3) habitual minimal footwear users in minimal footwear vs. non-habitual minimal footwear users in minimal footwear and (4) habitual minimal footwear users in conventional footwear vs. non-habitual minimal footwear users in conventional footwear. Running velocity was also contrasted using the same four comparisons outlined above only the models did not require covariate adjustment. Linear mixed effects models were undertaken using the compound symmetry and restricted maximum-likelihood methods; with footwear modelled as a fixed factor for within subjects' comparisons and group modelled as a fixed factor for between group comparisons and participants representing the random intercept in all cases [34]. Effect sizes were calculated using Cohen's *d*, which were interpreted as 0.2 = small, 0.5 = medium, and 0.8 = large [49]. Statistical significance for all analyses was accepted at the $P \leq 0.05$ level, and all statistical analyses were conducted using SPSS v28 (IBM, SPSS, Armonk, NY, USA). In the interests of conciseness and clarity, only variables that presented with statistical significance are presented in the text of the Results section.

3. Results

3.1. Running biomechanics

3.1.1. Spatiotemporal and loading indices

In the habitual minimal footwear group, effective mass (P = 0.012, d = 1.00), step length (P = 0.011, d = 1.51) and running velocity (P = 0.021 d = 0.88) were significantly greater in the conventional footwear, and both strike index (P = 0.031 d = 0.75) and daily loading cycles (P = 0.004, d = 1.64) were significantly greater in minimal footwear (Table 2). In addition, in conventional footwear the strike index (P = 0.002, d = 1.21) was significantly greater in the habitual minimal footwear group (Table 2). In minimal footwear the strike index (P = 0.005, d = 1.26) and daily loading cycles (P = 0.034, d = 1.22) were significantly greater, whereas the step length was significantly lower (P = 0.038, d = 1.17) in the habitual minimal footwear group (Table 2).

Table 2. Running biomechanics (mean ± standard deviations)

	Habitual				**Non-habitual**				
	Minimal		**Conventional**		**Minimal**		**Conventional**		
	Mean	SD	**Mean**	SD	**Mean**	SD	**Mean**	SD	
Effective mass (%BW)	7.83	1.01	9.06	1.53	11.32	1.81	9.59	1.93	***A***
Strike index (%)	61.68	19.33	46.48	21.44	33.79	24.69	22.61	17.92	***A, C, D***
Step length (m)	1.23	0.15	1.29	0.17	1.40	0.15	1.41	0.14	***A,***
Daily loading cycles	2061.92	210.44	1958.68	211.27	1805.16	211.76	1788.81	179.28	***A, D***
Running velocity (m/s)	3.36	0.57	3.43	0.54	3.76	0.40	3.68	0.45	***A***

Notes: %BW = percentage bodyweight
A = minimal & conventional footwear significantly different in the habitual footwear group
B = minimal & conventional footwear significantly different in the non-habitual footwear group
C = habitual & non-habitual footwear groups significantly different in conventional footwear
D = habitual & non-habitual footwear groups significantly different in minimal footwear

3.1.2. Lower extremity kinematics

In the habitual minimal footwear group, peak hip flexion (P = 0.002, d = 1.45), transverse plane hip range of motion (P = 0.011, d = 1.09), peak knee flexion (P = 0.007, d = 1.07), and sagittal plane knee range of motion (P = 0.003, d = 1.18) were significantly greater in conventional footwear, whereas sagittal plane ankle range of motion (P = 0.01, d = 0.85), peak ankle external rotation (P = 0.013, d = 1.15) and transverse plane ankle range of motion (P = 0.011, d = 0.95) were significantly greater in minimal footwear (Table 3). In addition, in the non-habitual minimal footwear group peak knee flexion (P = 0.009, d = 1.18), was significantly greater in conventional footwear and sagittal plane ankle range of motion (P = 0.011, d = 1.04), was significantly greater in minimal footwear (Table 3). Finally, in both conventional (P < 0.001, d = 1.63) and minimal (P = 0.004, d = 1.04) footwear the sagittal plane ankle range of motion was significantly greater in the habitual minimal footwear group (Table 3).

Table 3. Lower extremity kinematics (mean ± standard deviations)

		Habitual				Non-habitual				
		Minimal		Conventional		Minimal		Conventional		
		Mean	SD	Mean	SD	Mean	SD	Mean	SD	
						Hip				
Sagittal plane	Peak flexion	29.44	7.90	34.68	10.09	31.03	6.09	32.90	4.83	***A***
(+ = flexion/ - = extension)	Range of motion	0.36	0.31	0.68	1.83	0.87	1.26	1.03	1.21	
Coronal plane	Peak adduction	8.00	4.51	6.58	3.80	10.68	6.12	11.29	5.90	
(+ = adduction/ - = abduction)	Range of motion	7.25	2.56	6.13	3.36	5.51	3.68	5.25	4.12	
Transverse plane	Peak external rotation	-8.83	9.43	-10.43	7.26	-6.04	8.51	-8.16	5.66	
(+ = internal/ - = external)	Range of motion	5.83	6.27	8.03	6.86	4.65	4.42	7.43	5.45	***A***
						Knee				
Sagittal plane	Peak flexion	36.41	5.70	40.17	6.35	36.61	6.11	39.15	6.45	***A, B***
(+ = flexion/ - = extension)	Range of motion	21.51	4.66	23.93	4.36	23.77	4.56	26.43	5.11	***A***
Coronal plane	Peak abduction	-6.23	5.28	-6.10	6.24	-6.47	5.24	-7.10	5.79	
(+ = adduction/ - = abduction)	Range of motion	7.36	4.31	6.98	4.75	3.89	3.11	4.70	3.70	
Transverse plane	Peak internal rotation	8.50	3.89	6.15	2.52	8.25	3.66	9.63	6.00	
(+ = internal/ - = external)	Range of motion	14.73	9.49	14.88	8.31	12.57	6.04	13.03	7.09	
						Ankle				
Sagittal plane	Peak dorsiflexion	16.34	10.06	17.02	9.76	18.14	5.28	17.60	4.40	
(+ = dorsiflexion/ - = plantarflexion)	Range of motion	28.95	6.70	26.43	7.94	21.01	8.41	12.59	9.02	***A, B, C, D***
Coronal plane	Peak eversion	-7.24	3.62	-7.04	3.22	-7.96	3.76	-7.90	2.50	
(+ = inversion/ - = eversion)	Range of motion	12.71	3.02	13.38	2.18	11.35	4.74	11.55	3.79	
Transverse plane	Peak external rotation	-10.63	5.19	-8.11	4.49	-8.77	3.31	-7.69	2.28	***A***
(+ = internal/ - = external)	Range of motion	12.29	2.93	10.79	3.94	9.53	3.94	8.01	2.81	***A***

Notes: A = minimal & conventional footwear significantly different in the habitual footwear group
B = minimal & conventional footwear significantly different in the non-habitual footwear group
C = habitual & non-habitual footwear groups significantly different in conventional footwear
D = habitual & non-habitual footwear groups significantly different in minimal footwear

3.2. Musculoskeletal simulation

3.2.1. Ankle joint contact forces

In both habitual (P = 0.011, d = 0.99) and non-habitual (P = 0.017, d = 0.97) groups, medial ankle joint contact forces were significantly larger in the minimal footwear (Table 4).

3.2.2. Muscle forces

In the habitual minimal footwear group, rectus femoris (P = 0.022, d = 0.76) forces were significantly greater in conventional footwear and semitendinosus (P = 0.002, d = 1.31) and medial gastrocnemius (P = 0.006, d = 1.14) forces significantly greater in minimal footwear (Table 3). In the non-habitual group flexor digitorum longus forces (P = 0.04, d = 0.74) were significantly greater in minimal footwear (Table 4).

Table 4. Joint contact and muscle forces (mean ± standard deviations)

	Habitual				Non-habitual				
	Minimal		Conventional		Minimal		Conventional		
	Mean	SD	**Mean**	SD	**Mean**	SD	**Mean**	SD	
Posterior tibial load (BW)	3.09	0.62	3.00	0.63	3.14	0.33	2.99	0.57	
Axial tibial load (BW)	11.14	2.32	10.44	2.43	11.17	1.61	10.65	1.63	
Medial tibial load (BW)	1.27	0.38	1.09	0.27	1.36	0.37	1.10	0.44	***A, B***
Biceps femoris long head (BW)	0.02	0.03	0.08	0.18	0.14	0.18	0.17	0.19	
Biceps femoris short head (BW)	0.02	0.03	0.05	0.08	0.03	0.04	0.02	0.03	
Extensor digitorum longus (BW)	0.08	0.12	0.07	0.05	0.11	0.10	0.10	0.14	
Extensor hallucis longus (BW)	0.02	0.01	0.04	0.02	0.03	0.02	0.03	0.03	
Flexor digitorum longus (BW)	0.02	0.01	0.03	0.03	0.02	0.02	0.01	0.01	***B***
Flexor hallucis longus (BW)	0.07	0.07	0.04	0.03	0.07	0.09	0.05	0.11	
Gracilis (BW)	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	
Rectus femoris (BW)	1.36	0.60	1.65	0.65	1.55	0.45	1.70	0.34	***A***
Sartorius (BW)	0.03	0.05	0.03	0.05	0.04	0.05	0.04	0.06	
Semimembranosus (BW)	0.41	0.17	0.32	0.17	0.34	0.18	0.39	0.19	
Semitendinosus (BW)	0.15	0.10	0.08	0.06	0.13	0.13	0.11	0.11	***A***
Soleus (BW)	4.83	1.10	4.73	1.25	5.16	0.74	4.66	0.98	
Tensor fasciae latae (BW)	0.43	0.07	0.44	0.07	0.44	0.07	0.44	0.09	
Tibialis anterior (BW)	0.01	0.02	0.04	0.06	0.05	0.07	0.01	0.01	
Tibialis posterior (BW)	1.23	0.65	1.34	0.56	1.22	0.57	1.45	0.70	
Vastus intermedius (BW)	1.65	0.44	1.63	0.51	1.60	0.33	1.43	0.40	
Vastus lateralis (BW)	2.25	0.58	2.25	0.67	1.49	0.30	1.34	0.38	
Vastus medialis (BW)	1.55	0.42	1.53	0.49	2.23	0.43	1.96	0.50	
Lateral gastrocnemius (BW)	0.46	0.23	0.35	0.18	0.58	0.34	0.44	0.31	
Medial gastrocnemius (BW)	2.02	0.39	1.43	0.38	1.69	0.89	1.64	0.71	***A***
Peroneus brevis (BW)	0.17	0.12	0.22	0.21	0.13	0.18	0.27	0.28	
Peroneus longus (BW)	1.19	0.34	1.18	0.54	0.96	0.64	1.13	0.44	
Peroneus tertius (BW)	0.01	0.01	0.02	0.02	0.01	0.01	0.01	0.01	

Notes: BW = bodyweight
A = minimal & conventional footwear significantly different in the habitual footwear group
B = minimal & conventional footwear significantly different in the non-habitual footwear group
C = habitual & non-habitual footwear groups significantly different in conventional footwear
D = habitual & non-habitual footwear groups significantly different in minimal footwear

3.2.3. Finite element analysis

No significant differences were observed between footwear or between habitual or non-habitual minimal footwear groups (Table 5).

Table 5. Finite element analysis outcomes (mean ± standard deviations).

	Habitual				Non-habitual			
	Minimal		Conventional		Minimal		Conventional	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
90th percentile Von Mises strain (με)	3894.45	694.9	3691.7	748.14	4047.03	656.35	3787.73	848.85
50th percentile Von Mises strain (με)	2166.59	386.33	2053.72	415.9	2251.81	365.17	2107.17	471.72

Notes: με = microstrain

3.3. Stress fracture probability

No significant differences were observed between footwear or between habitual or non-habitual minimal footwear groups (Table 6; Figure 6).

Table 6. Probability of failure (mean ± standard deviations)

	Habitual				Non-habitual			
	Minimal		Conventional		Minimal		Conventional	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Probability of failure (%)	9.81	6.69	10.62	7.33	12.08	14.67	13.63	22.08

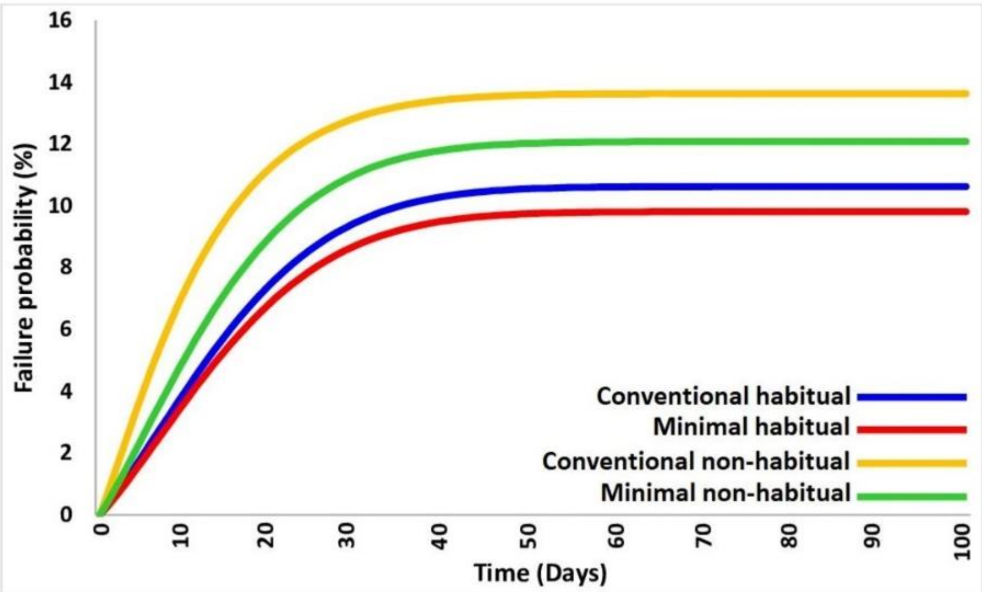

Figure 6. Average probabilities of failure (PFRA) for each footwear

4. Discussion

This study aimed to assess the impact of minimal and conventional running shoes on tibial strains and tibial stress fracture probability in individuals who habitually and do not habitually run in minimal footwear. Notably, this is the first investigation to utilize finite element and probabilistic modelling of the tibia to examine the effects of both minimal and conventional footwear within these distinct groups of runners. Consequently, this study holds the potential to offer novel insights into the influence of these footwear types on tibial strains and the probability of stress fractures among runners. The hypotheses tested in this study propose that minimal footwear will lead to increased tibial strains and a higher probability of tibial stress fractures in both habitual and non-habitual minimal footwear users, with no discernible differences between these two groups.

In the current study, running biomechanics were examined to determine the fundamental differences in mechanics between minimal and conventional footwear, across habitual and non-habitual minimal footwear users. In agreement with previous analyses [19, 22–23], this investigation revealed that sagittal plane ankle joint kinematics and the strike index differed significantly between footwear in the habitual minimal footwear group and also between groups in both minimal and traditional footwear. The strike index showed a more anterior foot contact position in minimal footwear in the habitual group, and according to the previously outlined footstrike definitions [33]; irrespective of footwear, runners in the habitual group exhibited a midfoot strike pattern whereas those in the non-habitual group demonstrated a rearfoot contact position.

Furthermore, the reduced step length and hip flexion characteristics found in minimal footwear also support previous analyses [19, 50], and the concept that foot strike and stride length are coupled [51]. The reductions in step length logically mediated the correspondingly increased number of daily loading cycles that were also found in minimal footwear. It is proposed that the midfoot contact position was responsible for the significant reduction in effective mass that was noted for minimal footwear in the habitual group [31]. This observation concurs with previous analyses of both habitual and non-habitual minimal footwear users [29, 31], and indicates that a greater proportion of body mass was decelerated during the impact component of the stance phase in conventional footwear.

In agreement with our hypotheses, musculoskeletal simulation analyses showed that although there were no differences between habitual and non-habitual runners, in both habitual and non-habitual minimal footwear users, ankle joint contact forces in the medial direction were significantly greater in minimal footwear. To our knowledge, this study is the first to scrutinize footwear differences in three-dimensional ankle joint contact loading in both habitual and non-habitual users of minimal footwear; however, this does concur with previous studies that have adopted external indices as pseudo measures of tibial loading [19, 22–23]. Importantly, in line with previous analyses, this investigation showed increased medial gastrocnemius forces in minimal footwear [50]. It is proposed that this observation was mediated as a function of the more anterior foot contact, sagittal ankle joint position and decreased peak knee flexion in this footwear condition, as minimal footwear has been shown to increase the role of the ankle and decrease the role of the knee as a shock absorber [52]. As muscle forces are considered the primary determinants of joint contact loading [38], it is proposed that greater medially directed forces that were observed in the minimal footwear condition, were mediated as a function of the statistical increases in the forces of muscles crossing the medial aspect of the ankle joint.

Stress fractures are representative of a mechanical fatigue phenomenon, whereby bone strains initiate microscopic damage in the bony matrix [16]. In support of our hypothesis there were no differences in tibial strains between habitual and non-habitual minimal footwear users, yet in opposition to our hypotheses there were also no significant differences between minimal and conventional footwear. It appears that although statistical differences in medially directed ankle joint contact forces were observed in minimal footwear across both participant groups, as medial joint contact forces are small relative to those in the axial and posterior directions, this was not sufficient to mediate alterations in tibial strains. This observation concurs with those noted previously, where despite statistically significant alterations in ankle joint contact forces, no corresponding differences in tibial strains were shown [26, 53]. In support of the conclusions of Khassetarash et al., [53] highlighting the complex relationship between joint contact loads, muscle forces and resulting bone strain; it is proposed that the increased ankle joint contact loads were counterbalanced by increased muscle forces, which served to neutralize each other's effect on net tibial strains [26]. This investigation shows that regardless of minimal footwear user experience; minimal and conventional footwear do not appear to differ in terms of damage potential to the tibia during running [11]. Therefore, (owing to the importance of strain in terms of representation of structural bone damage) as the current investigation showed no statistical differences in tibial strain magnitudes between footwear, it is also to be expected that there were no differences in tibial stress fracture probability, despite increases in the number of daily loading cycles in minimal footwear. The findings from this study indicate that in experienced runners, neither habitual and non-habitual minimal footwear users nor minimal and conventional footwear appear to be any more effective than the other from the standpoint of attenuating runners' likelihood of developing a tibial stress fracture.

Importantly, considering the experimental running velocities, ankle joint contact forces and strains experienced at the tibia observed in the current investigation were similar to other analyses using musculoskeletal simulation and finite element-based approaches [43]. In relation to the observed tibial stress fracture probabilities, the values noted in this study are firstly coherent with other probabilistically obtained failure indices obtained from the scientific literature at similar running speeds [25, 43]; and also, the progression of failure risk over the first 40-days is importantly in line with the epidemiologic literature in runners experiencing tibial stress fractures [54]. A limitation of the current investigation is that musculoskeletal simulation and finite element analyses were adopted to quantify muscle forces, joint contact loads and tibial strains. Musculoskeletal modelling, simulation and finite element procedures are fundamentally based around mathematical assumptions to simplify internal musculoskeletal parameters and although considerable advances have been made in the preceding decade; there remains much more work to be undertaken before the in vivo tibial environment can be examined completely. Furthermore, that this investigation was undertaken using an indoor laboratory environment may represent a limitation to this study as it compromises ecological validity in comparison to an unrestricted outdoor situation. It is not currently feasible to utilize the methodologies adopted within the current study in a non-laboratory scenario, however future technological advancements may allow such approaches to be utilized with enhanced degrees of ecological validity.

5. Conclusions

In summary, a comprehensive comparison of conventional and minimal footwear in habitual and non-habitual minimal footwear users, utilizing cumulative finite element and probabilistic analyses of tibial stress fractures, is notably absent in existing literature. This study contributes to the current clinical and scientific understanding of footwear biomechanics by investigating the impact of the aforementioned footwear on tibial stress fracture probability during running. Notably, in minimal footwear, there were significantly greater medially directed ankle joint contact forces compared to conventional footwear in both habitual and non-habitual users. However, no significant differences were observed in tibial strains and tibial stress fracture probabilities between footwear types or between habitual and non-habitual minimal footwear groups. Consequently, the findings of this study suggest that, in experienced runners, neither habitual and non-habitual minimal footwear use nor the choice between minimal and conventional footwear appears to significantly influence the probability of developing a tibial stress fracture.

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