Effects of running with minimal and conventional footwear in habitual and non-habitual users; a musculoskeletal simulation and statistical parametric mapping based approach.

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Abstract

The current investigation examined running biomechanics in minimal and conventional footwear in two groups of runners who either ran habitually in minimal footwear (habitual minimal footwear users) or habitually in conventional footwear (non-habitual minimal footwear users). We studied ten male non-habitual minimal footwear users and ten male habitual minimal footwear users, who were required to complete ≥35 km per week of training. Lower extremity joint loading was explored using a musculoskeletal simulation approach. Differences between conditions were examined using statistical parametric mapping and 2x2 mixed ANOVA. This study revealed via the strike index that minimal footwear caused a more anterior contact position in both groups (habitual: minimal=61.68% & conventional=46.48% /non-habitual: minimal=33.79% & conventional=22.61%), although non-habitual runners still adopted a rearfoot strike pattern. In addition, in non-habitual users minimal footwear increased tibial accelerations (habitual: minimal=6.35g & conventional=7.06g /non-habitual: minimal=9.54g & conventional=8.16g), loading rates (habitual: minimal=105.44BW/s &

conventional=105.97BW/s /non-habitual: minimal=293.00BW/s & conventional=154.36BW/s) and medial tibiofemoral loading rates (habitual: minimal=196.17BW/s & conventional=274.96BW/s /non-habitual: minimal=274.96BW/s & conventional=212.57BW/s). Furthermore, minimal footwear decreased patellofemoral loading in both habitual (minimal=0.28BW·s & conventional=0.31BW·s) and non-habitual (minimal=0.26BW·s & conventional=0.29BW·s) users. Finally, Achilles tendon loading was larger in minimal footwear and in habitual runners (habitual: minimal=0.79BW·s & conventional=0.71BW·s /non-habitual: minimal=0.71BW·s & conventional=0.65BW·s) whereas iliotibial band strain rate was reduced in habitual (minimal=28.32%/s & conventional=30.30%/s) in relation to non-habitual (minimal=42.96%/s & conventional=42.87%/s) users. This study highlights firstly the importance of transitioning to minimal footwear and also indicates that post transition they may be effective in attenuating the biomechanical mechanisms linked to the aetiology of many chronic injuries.

Introduction

Recreational distance running is arguably the most popular aerobic exercise modality (Lee et al. 2014). There is a plethora of evidence indicating that running mediates significant physiological and psychological benefits (Lee et al. 2014). However, despite the physical benefits that it manifests, distance running is also associated with a high susceptibility to chronic injuries; as 19.4-79.3 % of runners will experience a pathology each year (Van Gent et al. 2007). Unfortunately, chronic pathologies are a significant barrier to training adherence in runners and lead to a substantial economic burden (Hespanhol et al. 2016). Specifically, patellofemoral pain, iliotibial band syndrome, tibial stress fractures, medial tibial stress syndrome, Achilles tendinopathy and pain secondary to hip and knee osteoarthritis are commonly experienced in sports medicine clinics (Taunton et al. 2002, Van Ginckel et al. 2009; Winkelmann et al. 2016; Snyder et al. 2006).

The running shoe is the primary interface between the body and surface; as such significant developments in running shoe technology have emerged, in an attempt to mediate the incidence of chronic running pathologies (Sinclair et al. 2013a). However, since the introduction of the conventional running shoe, the rate and location of chronic running injuries has not changed, leading to the notion that technological developments in running footwear have not been successful in influencing running pathologies (Davis, 2014). This has led to the proposal that running in minimal footwear that lacks the cushioning and motion control properties associated with the conventional running shoe, may be associated with a reduced incidence of chronic running injuries (Lieberman et al. 2010; Davis, 2014). Based on this notion, several minimal footwear models are currently available commercially.

Several studies have examined differences in running biomechanics between minimal and conventional running shoes. These investigations have shown that minimal footwear alter spatiotemporal running characteristics, causing runners to adopt a more plantarflexed ankle at footstrike (Sinclair et al. 2013ab, Hollander et al. 2015), mid/forefoot strike pattern (Squadrone et al. 2015; Sinclair et al. 2019), increased stride rate (Warne et al. 2014) and reduced stride length (Sinclair et al. 2016; Sinclair et al. 2019) compared to conventional running shoes. In addition, previous comparisons of conventional and minimal footwear have also shown that minimal footwear are associated with increased vertical loading rates (Sinclair et al. 2013ab), tibial accelerations (Sinclair et al. 2013a), and effective mass (Sinclair et al. 2018a). Finally, previous work examining the effects of minimal footwear on the loads experienced by the lower extremities have revealed that minimal footwear reduces the loads experienced by the patellofemoral joint (Sinclair, 2014; Bonacci et al. 2014), but increase the forces borne by the Achilles tendon (Sinclair, 2014; Sinclair et al. 2019) and the tibiofemoral joint (Sinclair et al. 2018b). However, it is important to recognise that the conclusions drawn from the aforementioned investigations were based upon results obtained from novice users of minimal footwear. Indeed, Tam et al. (2017) proposed that in acute investigations of minimal footwear, runners do not sufficiently alter their running mechanics sufficiently to reduce the vertical loading rate. Therefore, it can be concluded that the overall evidence that minimal footwear is able to attenuate the biomechanical factors linked to the aetiology of chronic pathologies is currently insufficient. As such, with regards to minimal footwear, runners must select footwear based on the findings from acute studies conducted on runners who are unaccustomed to using minimal footwear. Therefore, it can be concluded that further investigation of running biomechanics between minimal and conventional footwear in those who habitually wear minimal and conventional footwear is warranted.

Furthermore, previous analyses concerning the biomechanical differences between minimal and conventional footwear, have adopted inverse-dynamic driven modelling-based approaches to quantify lower extremity musculoskeletal loading (Sinclair et al., 2019). However, joint torques are representative of global indices of joint loading, and therefore are not representative of localized joint loading (Herzog et al. 2003). Substantial developments in musculoskeletal modelling have been made in recent years, allowing indices of skeletal muscle forces; muscle kinematics and joint reaction forces be obtained through musculoskeletal simulation analyses (Delp et al. 2007). This approach may be more effective than traditional inverse-dynamic based methods and allows a more detailed examination of the specific parameters linked to the aetiology of chronic pathologies to be undertaken. Such approaches have not yet been utilized to explore biomechanical differences between minimal and conventional running shoes in runners who run habitually in minimal footwear (habitual minimal footwear users) or conventional footwear (non-habitual minimal footwear users).

There has yet to be a published investigation examining differences in running biomechanics between minimal and conventional footwear in those who habitually wear minimal and conventional footwear. Therefore, the aim of the current investigation was to examine running biomechanics in minimal and conventional footwear in those who habitually wear minimal and conventional footwear, with reference to the biomechanical mechanisms linked to the aetiology of chronic pathologies, using a musculoskeletal simulation-based analysis.

Methods

Participants

Ten male conventional footwear users (henceforth termed non-habitual minimal footwear users) (age 27.67 ± 5.57 years, height 1.71 ± 0.03 m and body mass 68.76 ± 4.78 kg) and ten male habitual minimal footwear users (henceforth termed habitual minimal footwear users) (age 33.50 ± 4.58 years, height 1.75 ± 0.04 m and body mass 71.74 ± 7.74 kg) volunteered to take part in this study. Participants were required to complete a minimum of 35 km per week of training. To be considered a habitual minimal footwear user, volunteers were required to have been training exclusively in minimal footwear for a minimum period of 24 months in footwear scoring ≥75 on the minimalist index described by Esculier et al. (2015). The procedure utilized for this investigation was approved by a university ethical committee (REF 637). All runners were free from musculoskeletal pathology at the time of data collection and provided written informed consent in accordance with the principles outlined in the Declaration of Helsinki.

Footwear

The footwear used during this study consisted of New Balance, 1260 v2 (New Balance, Boston, Massachusetts, United States; henceforth termed conventional) and Vibram Five-Fingers, ELX (Vibram, Albizzate, Italy; henceforth termed minimal) (Figure 1). The conventional footwear had an average mass of 0.285 kg, heel thickness of 25 mm and a heel drop of 14 mm and minimal an average mass of 0.167 kg, heel thickness of 7 mm and a heel drop of 0 mm. The footwear were also scored using the minimalist index of Esculier et al. (2015), and the conventional footwear received a score of 20 and minimal a score of 92.

@@@FIGURE 1 NEAR HERE@@@

Procedure

Participants ran at 4.0 m/s (± 5%), striking an embedded piezoelectric force platform (Kistler Instruments Ltd., Winterthur, Switzerland) with their right (dominant) foot. Running velocity was monitored using infrared timing gates (Newtest, Oy Koulukatu, Finland). The stance phase was delineated as the duration over which 20 N or greater of vertical ground reaction force (GRF) was applied to the force platform. Runners completed five successful trials in each footwear condition. The order that participants ran in each footwear condition was counterbalanced. Kinematic and GRF data were synchronously collected. Kinematic data were captured at 250 Hz via an eight-camera motion analysis system (Qualisys Medical AB, Goteburg, Sweden). Dynamic calibration of the motion capture system was performed before each data collection session.

Body segments were modelled in 6 degrees of freedom using the calibrated anatomical systems technique (Cappozzo et al. 1995). To define the anatomical frames of the thorax, pelvis, thighs, shanks and feet retroreflective markers were placed at the C7, T12 and xiphoid process landmarks and also positioned bilaterally onto the acromion process, iliac crest, anterior superior iliac spine (ASIS), posterior super iliac spine (PSIS), medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, calcaneus, first metatarsal and fifth metatarsal. Carbon-fibre tracking clusters comprising of four non-linear retroreflective markers were positioned onto the thigh and shank segments. In addition to these, the foot segments were tracked via the calcaneus, first metatarsal and fifth metatarsal, the pelvic segment was tracked using the PSIS and ASIS markers and the thorax segment was tracked using the T12, C7 and xiphoid markers. Static calibration trials were obtained in each footwear allowing for the anatomical markers to be referenced in relation to the tracking markers/ clusters.

To measure axially directed accelerations at the tibia, an accelerometer (Biometrics ACL 300, Gwent United Kingdom) sampling at 1000Hz was used. The device was mounted onto a piece of lightweight carbon-fibre material using the protocol outlined by Sinclair et al. (2013a). The accelerometer was attached securely to the distal anterio-medial aspect of the tibia in alignment with its longitudinal axis, 0.08 m above the medial malleolus. Strong non-stretch adhesive tape was placed over the device and leg to avoid overestimating the acceleration due to tissue artefact (Sinclair et al. 2013a).

Processing

Dynamic trials were digitized using Qualisys Track Manager (Qualisys Medical AB, Goteburg, Sweden) in order 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. In addition, the tibial acceleration signal was filtered using a 60 Hz Butterworth zero lag 4th order low pass filter (Sinclair et al. 2013a). Kinematics of the hip, knee and ankle were quantified using an XYZ cardan sequence of rotations (where X is flexion-extension; Y is ab-adduction and is Z is internal-external rotation). In addition, tibial internal rotation kinematics were also calculated in accordance with Eslami et al. (2007). All force parameters throughout were normalized by dividing by bodyweight (BW).

In accordance with the protocol of Addison & Lieberman, (2015), an impulse-momentum modelling approach was utilized to calculate effective mass (% BW), which was quantified in accordance with the below equation:

$$\text{Effective mass} = \text{vertical GRF integral} / (\mathit{\Delta\ foot\ vertical\ velocity} + \mathit{gravity} * \mathit{\Delta\ time})$$

The impact peak was defined firstly in non-habitual runners when wearing conventional footwear, as the first peak in vertical GRF. In habitual runners and non-habitual runners wearing minimal footwear where no impact peak was expected, according to the protocols of Lieberman et al. (2010) and Sinclair et al. (2018a) we defined the position of the impact peak at the same relative position, which was shown to be 11.87 % of the stance phase. The time (ms) to impact peak (*Δ time*) was quantified as the duration from footstrike to impact peak. The vertical GRF integral (BW·ms) during the period of the impact peak was calculated using a trapezoidal function. The change in foot vertical velocity (*Δ foot vertical velocity*) was determined as the instantaneous vertical foot velocity averaged across the 10 frames prior to the impact peak (Sinclair et al. 2018a). The velocity of the foot was quantified using the centre of mass of the foot segment in the vertical direction, within Visual 3D (Sinclair et al. 2018a).

Loading rate (BW/s) was also was also extracted by obtaining the peak increase in vertical GRF between adjacent data points using the first derivative function within Visual 3D and the peak tibial acceleration (g) was extracted as the highest positive acceleration peak during the stance phase. The strike index was calculated as the position of the centre of pressure location at footstrike, relative to the total length of the foot (Squadrone et al. 2015). A strike index of 0–33% denotes a rearfoot, 34–67% a midfoot and 68–100% a forefoot strike pattern. Finally, limb stiffness during running was quantified using a mathematical spring-mass model (Blickhan, 1989). Limb stiffness (BW/m) was calculated from the ratio of the peak normalized vertical GRF to the maximum vertical compression of the leg spring which was calculated as the change in limb length from footstrike to minimum length during the stance phase (Farley & Morgenroth, 1999). Limb length was quantified as the vertical height of the proximal end of the thigh segment within Visual 3D.

Following this, data during the stance phase were exported from Visual 3D into OpenSim 3.3 software (Simtk.org). Two validated musculoskeletal models were used to process the biomechanical data both of which were scaled to account for the anthropometrics of each runner. The first with 12 segments, 19 degrees of freedom and 92 musculotendon actuators (Lerner et al. 2015) was used initially to estimate lower extremity joint forces. As muscle forces are the main determinant of joint compressive forces (Herzog et al. 2003), muscle kinetics were quantified using static optimization in accordance with Steele et al. (2012). Compressive patellofemoral, medial/ lateral tibiofemoral, ankle and hip joint forces were calculated via the joint reaction analyses function using the muscle forces generated from the static optimization process as inputs. Furthermore, patellofemoral stress (KPa/kg) was quantified by dividing the patellofemoral force by the contact area. Patellofemoral contact areas were obtained by fitting a polynomial curve to the sex specific data of Besier et al. (2005), who estimated patellofemoral contact areas as a function of the knee flexion angle using MRI. Finally, Achilles tendon forces were estimated in accordance with the protocol of Almonroeder et al. (2013), by summing the muscle forces of the medial gastrocnemius, lateral, gastrocnemius, and soleus muscles.

In addition, patellofemoral, medial/ lateral tibiofemoral, ankle, hip and Achilles tendon instantaneous load rates (BW/s and KPa/BW/s) were also extracted by obtaining the maximum increase in force/ stress between adjacent data points using the first derivative function in Visual 3D. Finally, the integral of the hip, tibiofemoral, ankle, patellofemoral and Achilles tendon forces (BW·s) and stresses (KPa/BW·s) during the stance phase were calculated using a trapezoidal function.

Running in minimal footwear has been shown to alter step length during running, which increases the number of footstrikes necessary to run a set distance. We therefore estimated the total impulse per kilometre (BW·km) by multiplying these parameters by the number of steps required to run a kilometre. The number of steps required to complete one kilometre was quantified using the step length (m), which was determined by taking the difference in the horizontal position of the foot centre of mass between the right and left legs at footstrike.

The second model also had twelve segments, 23 degrees of freedom and 92 muscle-tendon actuators and was adapted from the generic OpenSim gait2392 model to include the iliotibial band (Foch et al. 2013). The iliotibial band itself was included within the gait2392 model but as a muscle with only a passive contractile component and an optimal muscle fiber length of zero (Foch et al. 2013). Iliotibial band kinematics during the stance phase were calculated via the muscle analyses function within OpenSim and iliotibial band strain (%) was calculated by dividing the change in length of the band during stance and dividing by its resting length at each time frame. In addition, the strain rate (%/s) was calculated as the change in strain between adjacent data points. The resting length of the iliotibial band was determined as its length during the static calibration trial (Hamill et al. 2008). Peak iliotibial band strain and strain rate were measured at the instance of peak knee flexion during stance (Hamill et al. 2008).

Statistical analyses

Following data processing, compressive joint forces (hip, patellofemoral, medial tibiofemoral and lateral tibiofemoral), Achilles tendon loading and three-dimensional kinematics during the entire stance phase were temporally normalized using linear interpolation to 101 data points. Differences across the entire stance phase were examined using 1-dimensional statistical parametric mapping (SPM) with MATLAB 2017a (MATLAB, MathWorks, Natick, USA), in accordance with Pataky et al. (2016), using the source code available at http://www.spm1d.org/. Differences as a function of both FOOTWEAR (FOOTWEAR – conventional or minimal) and GROUP (GROUP - habitual or non-habitual) were examined using paired and independent t-tests (SPM t).

For discrete parameters that could not be examined using SPM (joint integral, joint loading rate, joint integral per kilometre, step length, instantaneous load rate, strike index, limb stiffness, tibial accelerations, iliotibial band strain, iliotibial band strain rate and effective mass), means and standard deviations were calculated for each condition. Differences in discrete biomechanical parameters were examined using 2 (FOOTWEAR – conventional of minimal) x 2 (GROUP- habitual or non-habitual) mixed ANOVAs, Effect sizes were calculated using partial eta^2 (pη^2). In the event of a significant interaction, simple main effects tests were adopted. Discrete statistical actions were conducted using SPSS v25.0 (SPSS Inc., Chicago, USA). Statistical significance was accepted at the $P \leq 0.05$ level.

Results

@@@TABLE 1 NEAR HERE@@@

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Lower extremity external loading, strike index and step length

For effective mass there was a significant FOOTWEAR*GROUP interaction (P=0.01, pη^2 = 0.31). Simple main effects tests showed that effective mass was larger in the conventional running shoes compared to minimal in habitual runners (P=0.01, pη^2 = 0.53) but there were no significant differences between footwear in non-habitual runners (P=0.26, pη^2 = 0.11). In addition, when wearing minimal footwear, effective mass was significantly greater in non-habitual runners compared to habitual (P<0.001, pη^2 = 0.61) but there were no differences between habitual and non-habitual runners when running in conventional footwear (P=0.50, pη^2 = 0.03) (Table 1).

For loading rate there was also a significant FOOTWEAR*GROUP interaction (P=0.002, pη^2 = 0.41). Simple main effects tests showed that loading rate was significantly larger in the minimal footwear compared to conventional in non-habitual runners (P=0.004, pη^2 = 0.63) but there was no significant difference between footwear in habitual runners (P=0.94, pη^2 <0.001). In addition, when wearing minimal footwear, the loading rate was significantly greater in non-habitual runners compared to habitual ($P<0.001$, $p\eta^2 = 0.52$) but there were no differences when running in conventional footwear ($P=0.06$, $p\eta^2 = 0.19$) (Table 1).

For peak tibial accelerations, there was a significant FOOTWEAR*GROUP interaction ($P=0.005$, $p\eta^2 = 0.36$). Simple main effects tests showed that tibial accelerations were significantly larger in minimal footwear compared to conventional in non-habitual runners ($P=0.03$, $p\eta^2 = 0.42$) but there was no significant difference between fotowear in habitual runners ($P=0.09$, $p\eta^2 = 0.29$). In addition, when wearing minimal footwear, tibial accelerations were significantly greater in non-habitual compared to habitual runners ($P<0.001$, $p\eta^2 = 0.57$) but there were no differences between habitual and non-habitual runners in conventional footwear ($P=0.20$, $p\eta^2 = 0.09$) (Table 1).

For limb stiffness there was a significant FOOTWEAR*GROUP interaction ($P=0.04$, $p\eta^2 = 0.21$). Simple main effects tests showed that limb stiffness was greater in minimal compared to conventional footwear in non-habitual runners ($P<0.001$, $p\eta^2 = 0.57$) but there were no differences between footwear when running in conventional footwear ($P=0.20$, $p\eta^2 = 0.09$) (Table 1).

For strike index there was a main effect of FOOTWEAR ($P=0.002$, $p\eta^2 = 0.36$), which showed that the strike position was more anterior in minimal footwear. In addition, there was also a main effect of GROUP ($P=0.007$, $p\eta^2 = 0.34$), which indicated that the strike was also more anterior in habitual runners (Table 1).

For step length there was a significant FOOTWEAR*GROUP interaction ($P=0.04$, $p\eta^2 = 0.20$). Simple main effects tests showed that step length was significantly larger in conventional compared to minimal footwear in habitual runners ($P=0.001$, $p\eta^2 = 0.72$) but there was no difference between footwear in non-habitual runners ($P=0.70$, $p\eta^2 = 0.02$). In addition, when wearing minimal footwear compared to conventional, step length was significantly greater in non-habitual runners ($P=0.02$, $p\eta^2 = 0.28$) but there were no differences between habitual and non-habitual runners when running in conventional footwear ($P=0.11$, $p\eta^2 = 0.14$) (Table 1).

Joint loading

For medial tibiofemoral loading rate there was a significant FOOTWEAR*GROUP interaction ($P<0.001$, $p\eta^2 = 0.76$). Simple main effects tests showed that the loading rate was significantly larger in the conventional compared to minimal footwear in habitual runners ($P=0.001$, $p\eta^2 = 0.91$) but significantly greater in minimal compared to conventional footwear in non-habitual runners ($P=0.005$, $p\eta^2 = 0.61$). In addition, when wearing minimal footwear, medial tibiofemoral loading rate was significantly greater in non-habitual compared to habitual runners ($P=0.02$, $p\eta^2 = 0.26$) but in conventional footwear was significantly greater in habitual compared to non-habitual runners ($P=0.04$, $p\eta^2 = 0.21$) (Table 1).

For the integral of patellofemoral joint force, there was a main effect of FOOTWEAR ($P=0.03$, $p\eta^2 = 0.25$), which was shown to be larger in conventional footwear (Table 1).

For the integral of Achilles tendon force, there was a main effect of FOOTWEAR ($P=0.02$, $p\eta^2 = 0.27$), which was shown to be larger in minimal footwear. In addition, there was a main effect for GROUP ($P=0.002$, $p\eta^2 = 0.42$), which indicated that the Achilles tendon integral was greater in habitual runners (Table 1). For the Achilles tendon integral per kilometre, there was a main effect of FOOTWEAR ($P=0.004$, $p\eta^2 = 0.38$), which was shown to be larger in minimal footwear. In addition, there was a main effect for GROUP ($P=0.002$, $p\eta^2 = 0.41$), which indicated that the Achilles tendon integral was greater in habitual runners (Table 2).

For the ankle integral per kilometre, there was a main effect for GROUP ($P=0.02$, $p\eta^2 = 0.27$), which indicated that the ankle integral was greater in habitual runners (Table 2).

Iliotibial band kinematics

For iliotibial band strain rate, there was a main effect for GROUP ($P<0.001$, $p\eta^2 = 0.52$), which indicated that the strain rate was greater in non-habitual runners (Table 1).

Statistical parametric mapping - joint loading

Minimal footwear was associated with increased Achilles tendon force compared to conventional running shoes in the first 20% of the stance phase in both habitual and non-habitual runners (Figure 2ab).

Statistical parametric mapping - three-dimensional kinematics

Conventional footwear was associated with increased hip flexion compared to minimal from 20-40% of the stance phase in habitual runners (Figure 2c). Conventional footwear was also associated with increased knee flexion compared to minimal from 40-60% of the stance phase in both habitual and non-habitual runners (Figure 2de). In additional, minimal footwear compared to conventional was associated with increased tibial and knee internal rotation during from 20-60% of the stance phase in habitual runners (Figure 3ab). Furthermore, it was revealed that the ankle exhibited increased plantarflexion in minimal footwear from 0-5% of the stance phase in both habitual and non-habitual runners (Figure 3cd). Finally, in conventional footwear compared to minimal, habitual runners were similarly associated with increased plantarflexion from 0-5% of the stance phase (Figure 3e).

@@@FIGURE 2 NEAR HERE@@@

@@@FIGURE 3 NEAR HERE@@@

Discussion

The aim of the current investigation was to examine differences in running biomechanics between minimal and conventional footwear, in those who habitually wear minimal and conventional footwear. To the authors knowledge, this is the first quantitative comparison of these footwear in habitual and non-habitual minimal footwear users using a musculoskeletal simulation and SPM based approach.

The kinematic analysis using SPM of the sagittal plane ankle angle aligned with the discrete analysis of the strike index, supports previous investigations in that minimal footwear transferred the footstrike to a more anterior position in both habitual and non-habitual runners (Squadrone et al. 2015; Sinclair et al. 2019). Furthermore, in support of previous analyses the findings from this study also showed that habitual minimal footwear users similarly were associated with a significantly more anterior footsrike position in relation to non-habitual runners (Larson et al. 2014). It is important to contextualize the strike index values observed in both conditions, as regardless of which footwear condition was utilized non-habitual runners maintained a rearfoot strike pattern and habitual runners adopted a midfoot contact position. This supports proposition of Tam et al. (2017) that in acute investigations non-habitual runners do not sufficiently alter their running mechanics and continue to exhibit a rearfoot strike pattern.

For the indices of external loading, in agreement with previous analyses this investigation showed that tibial accelerations and loading rates were found to be greater in minimal footwear in non-habitual runners (Sinclair et al. 2013ab) and in non-habitual runners when wearing minimal footwear (Lieberman et al. 2010). As non-habitual runners adopted a rearfoot strike pattern when wearing minimal footwear, it was expected that both effective mass and limb stiffness were also increased when non-habitual runners adopted minimal footwear. It is proposed that the increases in external loading indices were mediated by the corresponding changes in effective mass and limb stiffness, which have been shown previously to be positively related to the magnitude of the both tibial accelerations and loading rate (Sinclair et al. 2018a). As tibial accelerations/ loading rates were increased in non-habitual runners using minimal footwear, these observations may be clinically meaningful. Given the proposed association between tibial accelerations/ loading rates and the aetiology of chronic injuries (Davis et al. 2004), this study indicates that non-habitual runners wearing minimal footwear are at increased risk from impact related injuries.

Although no differences were revealed using SPM, the discrete analysis showed that the patellofemoral force integral was significantly larger in conventional footwear in both habitual and non-habitual groups. This finding concurs with those observed previously by Sinclair, (2014), Sinclair et al. (2016) and Bonacci et al. (2014) who showed significant reductions in patellofemoral loading when running in minimal footwear. The discrete and SPM based analyses showed that minimal footwear transferred the footstrike to a more anterior position and also reduced the extent of peak knee flexion in both habitual and non-habitual groups. It is proposed that these observations are responsible for the reductions in patellofemoral loading as previous analyses have shown that the function of the knee joint as an energy absorber is reduced when there is an increased plantarflexion involvement (Sinclair & Selfe, 2015). Importantly, excessive patellofemoral joint loading is considered a key mechanism linked to the aetiology of pain symptoms in active individuals (Ho et al. 2012). Therefore, the findings from the current investigation indicate that in both habitual and non-habitual runners, minimal footwear may be effective in attenuating the biomechanical parameters linked to the aetiology of patellofemoral pain.

In addition, it was revealed via the discrete analysis, that the loading rate at the medial aspect of the tibiofemoral joint was larger in the conventional footwear in habitual runners and in minimal footwear in non-habitual runners. This supports those of Sinclair et al. (2018b) who showed in non-habitual runners, that minimal footwear increased the loading rate at the medial aspect of the knee joint. This observation indicates that the loading rate at the medial tibiofemoral joint was statistically larger when runners performed in their non-preferred footwear condition. Because the loading rate at the medial knee has been cited as important predictor of radiographic knee osteoarthritis, the findings from this investigation indicate that runners are at increased risk when running in their non-preferred footwear condition without habituation (Morgenroth et al. 2014).

Furthermore, this investigation showed using both SPM and discrete analyses that Achilles tendon loading indices were significantly larger in minimal footwear and in habitual runners collectively. This observation concurs with previous investigations (Sinclair, 2014, Sinclair et al. 2019) showing that in non-habitual runners' minimal footwear significantly enhanced

Achilles tendon loading compared to conventional running shoes, although there is no comparative literature examining the mechanics of the Achilles tendon in habitual minimal footwear users. Importantly, the current study also showed that habitual runners were associated with enhanced Achilles tendon loading compared to non-habitual users. It is proposed that the mechanism responsible for these observations is the more anterior footsrike position in minimal footwear and in habitual users, which served to enhance triceps surae muscle forces during the eccentric aspect of the stance phase (Almonroeder et al. 2013). This observation may be clinically important, as the initiation of Achilles tendinopathy is believed to be mediated through repeated and excessive loads experienced by tendon itself without sufficient rest in between loading exposures (Selvanetti et al. 1997). However, Davis et al. (2017) postulate that greater tendon loading in habituated runners may instigate the stimulus required for tendon hypertrophy and enhanced stiffness within the muscle–tendon unit necessary for the storage and release of elastic energy. Anrsten et al. (2017) support this notion as they showed that habitual minimal footwear users were associated with greater tendon cross sectional area and increased stiffness.

Finally, the current study also importantly showed that iliotibial band strain rate was greater in non-habitual runners. This finding may be clinically important as modelling investigations suggest that increased strain rate is the biomechanical risk factor linked to the aetiology of iliotibial band syndrome (Hamill et al. 2008). The main mechanical difference (irrespective of footwear) between groups, was the adoption of a midfoot strike pattern in habitual minimal footwear users compared to non-habitual. Therefore, the findings from this study lend support to the proposition of Lalonde (2013) that a rearfoot landing should be avoided for the prevention of iliotibial band syndrome in runners, although further aetiological investigations are required to substantiate this notion. As such, the current investigation indicates that transitioning to minimal footwear may be beneficial for runners in that they are able to attenuate their risk from iliotibial band syndrome.

A potential limitation to the current study that should be acknowledged is that only male runners were examined. Females have been shown to exhibit distinct external loading kinetics (Ferber et al. 2003), lower extremity kinematics (Sinclair et al. 2012, Ferber et al. 2003), limb stiffness (Sinclair et al. 2016), patellofemoral (Sinclair & Selfe, 2015) and Achilles tendon (Greenhalgh & Sinclair, 2014), parameters compared to male runners. This therefore suggests that further investigation of minimal footwear in habitual users using a female sample is warranted before comprehensive conclusions can be drawn. Furthermore, the efficacy of musculoskeletal simulation analyses depends on the fidelity of the primary neuromusculoskeletal model used to quantify the mechanics of the movement being investigated (Seth et al., 2011). Many assumptions and simplifications are made in the development of musculoskeletal simulation models, which could potentially impact the results from the current investigation (Seth et al., 2011). Therefore, there is considerable scope for future analyses to address and improve upon these limitations, in order to provide more accurate and valid musculoskeletal simulations.

In conclusion, the biomechanics of minimal and conventional footwear have received widespread research attention. However, there has not been quantitative comparison of these footwear in habitual and non-habitual minimal footwear users using a musculoskeletal simulation and SPM based approach. This study revealed that minimal footwear mediated a more anterior contact position in both groups, although non-habitual runners still adopted a rearfoot strike pattern. In addition, minimal footwear increased tibial accelerations, loading rates and medial tibiofemoral loading rates in non-habitual runners and decreased patellofemoral loading in both habitual and non-habitual groups. Finally, Achilles tendon loading indices were larger in minimal footwear and in habitual runners whereas iliotibial band strain rate was reduced in habitual runners. Therefore, this study highlights firstly the importance of transitioning to minimal footwear and also indicates that post transition they may be effective in attenuating the biomechanical mechanisms linked to the aetiology of many chronic injuries.

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Table 1: Discrete biomechanical parameters (mean ± standard deviations) as a function of FOOTWEAR and GROUP.

	Non-habitual				Habitual				
	Conventional		**Minimal**		**Conventional**		**Minimal**		
	Mean	*SD*	**Mean**	*SD*	**Mean**	*SD*	**Mean**	*SD*	
Effective mass (% BW)	9.59	1.93	11.32	1.81	9.06	1.53	7.83	1.01	B, C
Loading rate (BW/s)	154.36	69.86	293.00	126.14	105.97	27.20	105.44	48.95	A, B, C
Peak tibial acceleration (g)	8.16	2.04	9.54	1.90	7.06	1.60	6.35	0.86	B, C
Limb stiffness (BW/m)	63.41	28.52	65.91	22.69	63.46	33.51	48.12	13.97	C
Iliotibial band strain (%)	2.41	2.09	2.44	1.85	2.09	2.18	2.54	1.27	
Iliotibial band strain rate (%/s)	42.87	14.67	42.96	12.71	30.30	7.37	28.32	8.06	B
Patellofemoral integral (BW·s)	0.29	0.08	0.26	0.08	0.31	0.13	0.28	0.10	A
Patellofemoral loading rate (BW/s)	156.50	55.49	154.50	33.75	179.18	48.37	143.14	22.99	
Patellofemoral stress integral (KPa/BW·s)	0.56	0.14	0.52	0.13	0.59	0.21	0.55	0.18	
Patellofemoral stress loading rate (KPa/BW/s)	323.28	125.29	326.40	81.24	375.92	117.61	302.65	59.30	
Achilles integral (BW·s)	0.65	0.07	0.71	0.05	0.71	0.05	0.79	0.12	A, B
Achilles loading rate (BW/s)	153.96	43.34	179.34	67.96	179.34	67.96	148.14	38.75	
Ankle integral (BW·s)	1.21	0.12	1.30	0.12	1.30	0.12	1.33	0.19	
Ankle loading rate (BW/s)	251.82	41.42	281.18	55.95	281.18	55.95	247.14	40.27	
Hip integral (BW·s)	1.34	0.16	1.31	0.09	1.31	0.09	1.25	0.13	
Hip loading rate (BW/s)	276.22	41.21	291.88	82.86	291.88	82.86	260.00	123.79	
Medial tibiofemoral integral (BW·s)	0.86	0.10	0.83	0.06	0.83	0.06	0.85	0.12	
Medial tibiofemoral loading rate (BW/s)	212.57	51.75	274.96	75.23	274.96	75.23	196.17	64.60	C
Lateral tibiofemoral integral (BW·s)	0.44	0.07	0.44	0.05	0.44	0.05	0.41	0.07	
Lateral tibiofemoral loading rate (BW/s)	157.20	63.56	151.07	38.23	151.07	38.23	130.20	36.26	
Strike index (%)	22.61	17.92	33.79	24.69	46.48	21.44	61.68	19.33	A, B

A = main effect of FOOTWEAR
B = main effect of GROUP
C = FOOTWEAR x GROUP interaction

Table 2: Discrete temporal biomechanical parameters (mean ± standard deviations) as a function of FOOTWEAR and GROUP.

	Non-habitual				**Habitual**				
	Conventional		**Minimal**		**Conventional**		**Minimal**		
	Mean	*SD*	**Mean**	*SD*	**Mean**	*SD*	**Mean**	*SD*	
Step length (m)	1.41	0.14	1.4	0.15	1.29	0.17	1.23	0.15	A, B, C
Patellofemoral integral per kilometre m (BW·km)	543.91	163.20	493.74	160.51	646.21	300.30	612.77	253.39	
Patellofemoral stress integral per kilometre (KPa/BW·km)	1048.04	280.21	970.31	268.78	1203.71	494.77	1188.42	439.10	
Achilles integral per kilometre (BW·km)	1196.94	174.21	1328.26	134.48	1446.36	181.08	1697.98	361.64	A, B
Ankle integral per kilometre (BW·km)	2255.40	334.19	2410.35	249.97	2637.91	428.70	2849.76	582.01	B
Hip integral per kilometre (BW·km)	2507.87	504.79	2465.54	403.70	2672.95	391.66	2676.06	452.34	
Medial tibiofemoral integral per kilometre (BW·km)	1608.00	298.37	1561.71	234.69	1694.68	239.98	1826.23	378.21	
Lateral tibiofemoral integral per kilometre (BW·km)	815.75	191.45	826.17	133.47	902.91	173.35	875.81	201.87	

A = main effect of FOOTWEAR
B = main effect of GROUP
C = FOOTWEAR x GROUP interaction

Figure labels

Figure 1: Experimental footwear (A = conventional and B = minimal).

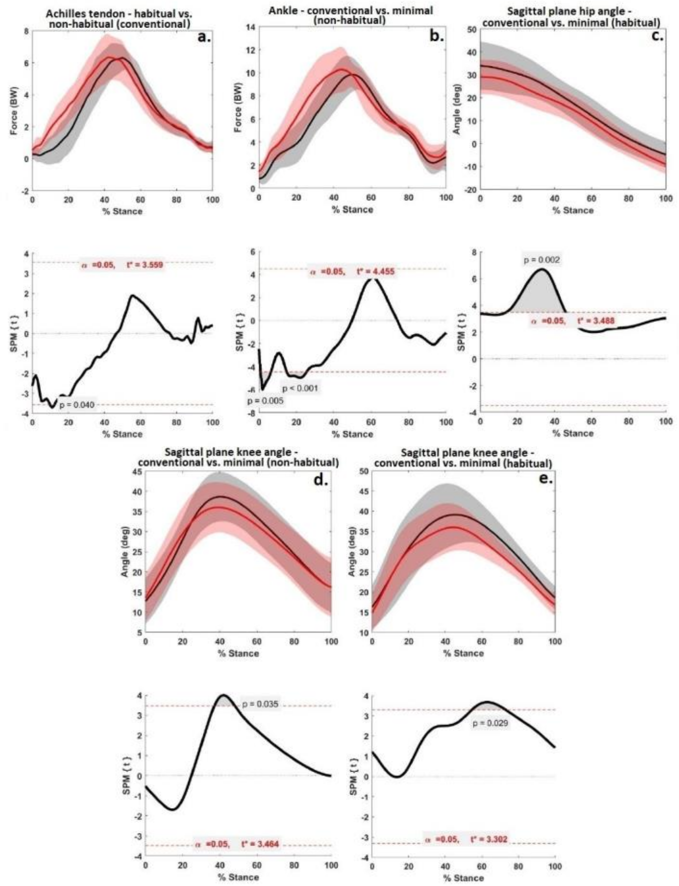

Figure 2: Statistical parametric mapping results of Achilles tendon and ankle forces in addition to hip and knee kinematics (FOOTWEAR: black = conventional/ red = minimal & GROUP: black = non-habitual/ red = habitual).

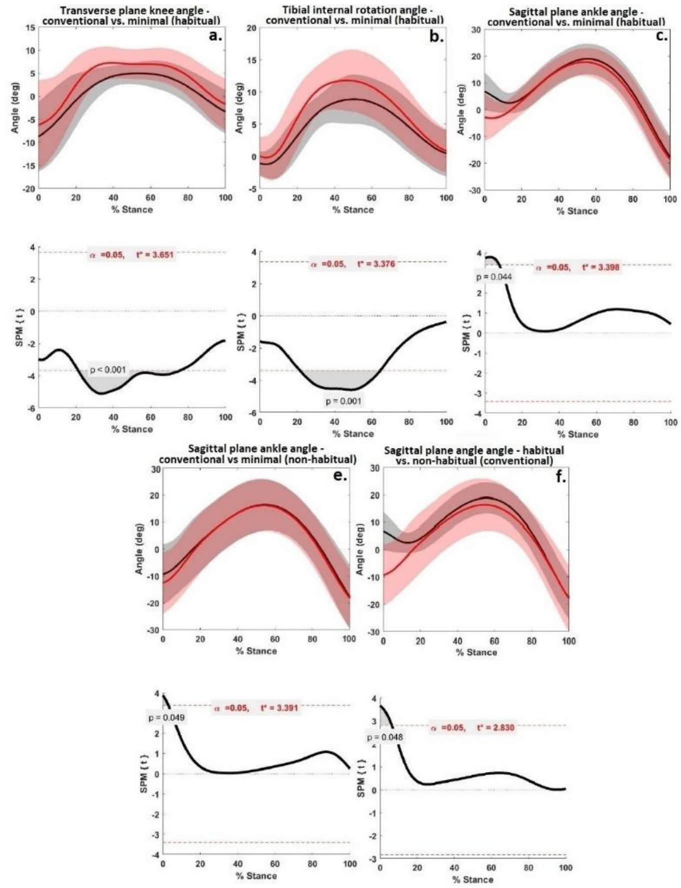

Figure 3: Statistical parametric mapping results of tibial internal rotation, knee and ankle kinematics (FOOTWEAR: black = conventional/ red = minimal & GROUP: black = non-habitual/ red = habitual).