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Effects of running in minimal, maximal and traditional running shoes: a musculoskeletal simulation exploration using statistical parametric mapping and Bayesian analyses

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ABSTRACT

The current study aimed to use a musculoskeletal simulation approach to examine running biomechanics in minimal, maximal and traditional running shoes using a concurrent SPM and Bayesian approach. Thirteen male participants ran over a force platform at 4.0 m/s in minimal maximal and traditional running shoes. Lower extremity joint loading and muscle forces were explored using a musculoskeletal simulation approach. Differences between conditions were examined using statistical parametric mapping (SPM) and Bayesian one-way repeated measures ANOVA. Bayesian analyses showed that traditional running shoes increased vastus intermedius (208.8BW·ms), vastus lateralis (320.2BW·ms) vastus medialis (188.7BW·ms), lateral tibiofemoral (495.9BW·ms) and patellofemoral joint stress (1683.4KPa/BW·s) integrals compared to minimal running shoes (185.0BW·ms, 281.9BW·ms, 167.2BW·ms, 456.5BW·ms & 1524.9KPa/BW·s). Furthermore, SPM showed that minimal footwear increased glutaeal, medial tibiofemoral and hip forces during the first 10% of the stance phase and Achilles tendon forces from 20 to 40% stance compared to traditional running shoes, whereas Bayesian analysis showed that minimal footwear increased loading rates (366.9BW/s) compared to maximal and traditional running shoes. (186.5BW/s) and traditional running shoes (161.5BW/s). Finally, SPM also showed that maximal footwear enhanced ankle eversion from 10 to 30% of stance compared to both minimal and traditional running shoes. This study therefore shows that minimal footwear may place runners at increased risk from impact related chronic injuries yet attenuate risk from patellofemoral and lateral tibiofemoral pathologies compared to traditional running shoes. In addition, owing to increases in ankle eversion, maximal running shoes may enhance risk to the aetiology of medial tibial stress syndrome compared to minimal and traditional running shoes.

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Footwear; biomechanics; simulation; minimal; maximal; running

Introduction

Engagement with distance running is unequivocally associated with a range of physiological benefits (Lee et al., 2014). However, despite mediating clear physical improvements, running is also associated with a high incidence of chronic pathologies with 19.4–79.3% experiencing an injury each year (van Gent et al., 2007). Chronic injuries prevent runners from engaging in training/competition and place significant fiscal demands on the healthcare system (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 (Snyder et al., 2006; Taunton et al., 2002; Van Ginckel et al., 2009; Winkelmann et al., 2016).

As the interface between foot and ground, running shoes have been proposed as an important mechanism by which the biomechanical factors linked to the aetiology of chronic injuries may be influenced (Sinclair, 2014). However, since the introduction of the modern running shoe in the 1970s, the rate and location of chronic running injuries has not changed, leading some to speculate that traditional running footwear has not been successful in influencing running pathologies

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(Davis, 2014). Based on this notion, footwear manufacturers have introduced running shoes with varied levels of midsole cushioning, offering both minimal and maximal running shoe options (Sinclair, Fau-Goodwin, et al., 2016).

Minimal running shoes feature a low/zero heel-toe drop, high levels of midsole flexibility and low mass (Esculier et al., 2015), whereas maximal running shoes whilst also incorporating a low heel-toe drop, feature a much larger amount of midsole cushioning throughout the entire length of the shoe. There has been considerable research interest into the biomechanics of running in minimal and maximal running shoes with regards to the biomechanical parameters linked to the aetiology of injury. Minimal running shoes have been shown to be associated with increased vertical loading rates, tibial accelerations, effective mass and vertical limb stiffness when running compared to both traditional (Sinclair, Greenhalgh, et al., 2013; Sinclair, Hobbs, et al., 2013; Sinclair, Fau-Goodwin, et al., 2016; Sinclair, Atkins, et al., 2016; Sinclair et al., 2018) and maximal running shoes (Hannigan & Pollard, 2020; Sinclair et al., 2015) although no differences in impact loading parameters has been found between maximal and traditional running shoes (Chan et al., 2018; Hannigan & Pollard, 2020; Sinclair, Fau-Goodwin, et al., 2016). Furthermore, ankle eversion/tibial internal rotation has been shown to be greater in minimal and maximal conditions compared to traditional running shoes (Sinclair, 2014; Hannigan & Pollard, 2020). In addition, minimal footwear have also been shown to be associated with enhanced Achilles tendon forces compared to traditional (Sinclair, 2014; Sinclair et al., 2019) and maximal running shoes (Sinclair et al., 2015) and greater medial tibiofemoral compartment loading compared to traditional running shoes (Sinclair et al., 2018). However, across several investigations minimal footwear has been shown to reduce patellofemoral joint loading compared to both traditional (Bonacci et al., 2018; Sinclair, 2014, Sinclair, Richards, et al., 2016; Sinclair et al., 2019; Yang et al., 2019) and maximal running shoes (Sinclair, Richards, et al., 2016) conditions.

Furthermore, whilst previous analyses have examined the effects of minimal, maximal and traditional running shoes on the risk factors linked to the aetiology of chronic running pathologies, these biomechanical parameters have habitually been explored through discrete point analyses. For time normalised biomechanical parameters, statistical parametric mapping (SPM) may represent a more effective statistical process, as it is capable of examining an entire time-based data sequence and thus reduces type II error by removing the need for multiple tests (Pataky et al., 2013). Similarly, Bayesian analyses have become considerably more prevalent and practicable in the last several years (Pullenayegum & Thabane, 2009). However, despite their prospective benefits (Ashby, 2006) and the excess of statistical literature advocating their adoption, their use in biomechanical analyses remains limited. To date there has yet to be any biomechanical investigation which has examined the effects of minimal, maximal and traditional running shoes on running biomechanics using a concurrent SPM and Bayesian approach.

However, previous analyses examining biomechanical differences between minimal, maximal and traditional running shoes, have all used musculoskeletal modelling-based approaches driven by inverse dynamics, to quantify lower extremity joint loading linked to the aetiology of injury. Recently, substantial developments in musculoskeletal simulation have been made, allowing indices of skeletal muscle forces; muscle kinematics and joint reaction forces be obtained (Delp et al., 2007). This approach may offer a more informative modality by which to contrast the effects of different footwear on running biomechanics as through muscle driven indices of joint loading, the ability to examine iliotibial band kinematics alongside muscle forces it allows a more detailed and accurate examination of the specific parameters linked to the aetiology of chronic pathologies to be undertaken (Herzog et al., 2003). However, such approaches have not yet been used to explore biomechanical differences between minimal, maximal and traditional running shoes.

Therefore, the current study aimed to use a musculoskeletal simulation approach to examine running biomechanics in minimal, maximal and traditional running shoes using a concurrent SPM and Bayesian approach. An investigation of this nature may provide important information regarding the efficacy of minimal, maximal and traditional running shoes.

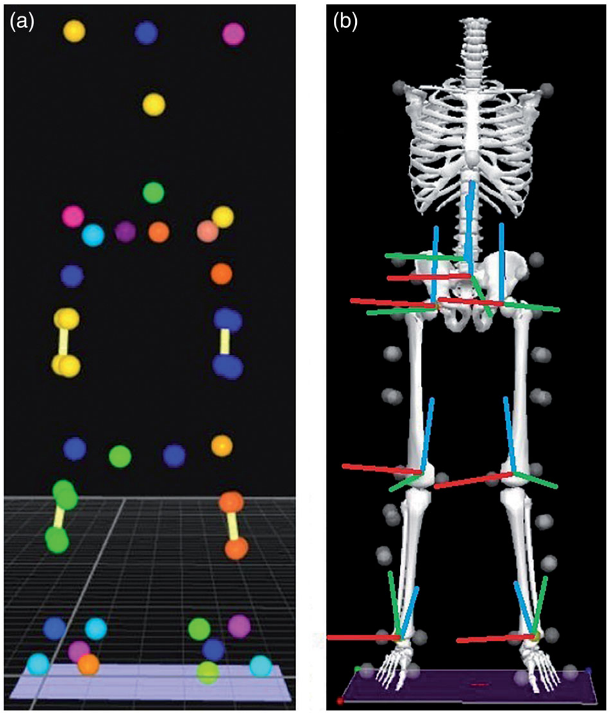

Figure 1. (a) Experimental marker set-up. (b) experimental model with segment co-ordinate axes (red = sagittal, green = coronal and blue = transverse axes).

Methods

Participants

Thirteen males (age 28.82 ± 5.22 years, height 1.71 ± 0.04 m and body mass 70.75 ± 4.39 kg) volunteered to take part in this study. Participants were required to complete a minimum of 35 km per week of running. The procedure used for this investigation was approved by a university ethics committee. All participants 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; termed traditional running shoes), Vibram Five-Fingers, ELX (Vibram, Albizzate, Italy; termed minimal) and HOKA OneOne Rapa Nui 2 Tarmac Road (HOKA Goleta, California, United States; termed maximal) (Figure 1). The footwear were scored using the minimalist index of Esculier et al. (2015), and their details are shown in Table 1.

Table 1. Experimental footwear characteristics.

	Maximal	Minimal	Traditional
Mass (g)	318	167	285
Heel thickness (mm)	45	7	25
Heel-toe drop (mm)	6	0	14
Esculier et al. (2015) minimalist index	18	92	20

Procedure

Participants ran at 4.0 m/s (± 5%), striking an embedded piezoelectric force platform (Kistler Instruments Ltd., Winterthur, Switzerland) sampling at 1000 Hz, with their right (dominant) foot. Running velocity was monitored using infra-red 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. Participants completed five successful trials in each footwear condition. A successful trial was defined as one within the specified velocity range, the foot made full contact with the force platform and with no evidence of gait modifications due to the experimental conditions. 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 (Figure 2(a)). Carbon-fibre tracking clusters comprising of four non-linear retroreflective markers were positioned via double sided tape and rigid sports tape onto the thigh and shank segments. In addition to these, the foot segments were

Figure 2. Experimental footwear (a) = traditional running shoes, (b) = maximal, (c) = minimal.

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. The sagittal, coronal and transverse axes of each segment were defined in accordance with Bennett et al. (2020) (Figure 2(b)).

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. 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). All force parameters throughout were normalized by dividing by bodyweight (BW).

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

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

The impact peak was defined in the maximal and traditional running shoes as the first peak in vertical GRF. In the minimal footwear where there was not a consistent impact peak, according to the protocols of Lieberman et al. (2010) and Sinclair et al. (2018), we defined the position of the impact peak at the same relative position as in the maximal and traditional running shoes. The time to impact peak (Δ *time*) was quantified as the duration from footstrike to impact peak. The vertical GRF integral 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 change in vertical foot velocity between the instances of footstrike and the impact peak (Chi & Schmitt, 2005). 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., 2018).

Loading rate (BW/s) was also extracted by obtaining the peak increase in vertical GRF between adjacent data points using the first derivative function within Visual 3D. 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, vertical limb stiffness during running was quantified using a mathematical spring-mass model (Blickhan, 1989). Vertical 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 optimisation 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 optimisation process as inputs. Furthermore, patellofemoral stress (KPa/BW) 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·ms), stresses (KPa/BW·ms) and muscle forces (BW·ms) during the stance phase were calculated using a trapezoidal function.

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 fibre 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).

Analyses

Following data processing, compressive joint forces (hip, patellofemoral, ankle, 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. In agreement with Pataky et al. (2013), SPM was implemented in a hierarchical manner, analogous to one-way repeated measures ANOVA with post-hoc *t*-tests. Therefore, the entire data-set was examined first, and if statistical significance was reached then post-hoc tests comparing individual footwear conditions were conducted on each component separately. For discrete parameters that could not be examined using SPM (joint integral, muscle force integral, joint loading rate, instantaneous load rate, strike index, vertical limb stiffness, 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 Bayesian one-way repeated measures ANOVA with default prior scales using JASP software 0.10.2 (Wagenmakers et al., 2018). Bayesian factors (BF) were used to explore the extent to which the data supported the alternative (H_1) hypothesis. Bayes factors were interpreted in accordance with the recommendations of Jeffreys (1961), with values above 3 indicating sufficient evidence in support of H_1. In the

Table 2. Discrete biomechanical parameters (mean ± standard deviations) as a function of footwear.

	Maximal		Minimal		Traditional	
	Mean	SD	Mean	SD	Mean	SD
Effective mass (% BW)	11.0	1.9	9.9	2.3	11.4	2.7
Loading rate (BW/s)	186.5	46.1	366.9[a,b]	133.2	161.5	63.1*
Vertical limb stiffness (BW/m)	74.6	23.9	76.0	22.1	72.9	24.3
Iliotibial band strain (%)	3.6	1.1	3.4	1.8	3.7	1.2
Iliotibial band strain rate (%/s)	45.1	15.1	44.1	16.4	47.7	16.9
Patellofemoral integral (BW·ms)	852.0	222.6	769.6	199.7	863.5	152.4
Patellofemoral loading rate (BW/s)	180.1	48.4	159.7	58.8	175.2	44.5
Patellofemoral stress integral (KPa/BW·ms)	1658.0	307.4	1524.9	283.7	1683.4[c]	175.2*
Patellofemoral stress loading rate (KPa/BW/s)	393.1	91.3	345.0	133.4	369.9	79.6
Achilles integral (BW·ms)	699.5	61.6	724.2	55.7	699.5	45.2
Achilles loading rate (BW/s)	122.5	19.2	134.3	24.9	131.5	20.4
Ankle integral (BW·ms)	1341.2	93.0	1333.7	68.5	1331.2	87.7
Ankle loading rate (BW/s)	199.5	25.5	203.3	39.5	207.8	22.0
Hip integral (BW·s)	1346.6	94.1	1367.1	69.0	1371.2	103.2
Hip loading rate (BW/s)	232.5	65.8	224.4	92.7	210.9	64.7
Medial tibiofemoral integral (BW·ms)	915.9	86.4	895.5	69.4	923.2	95.8
Medial tibiofemoral loading rate (BW/s)	212.5	48.9	196.6	53.7	183.9	43.3
Lateral tibiofemoral integral (BW·ms)	493.0[c]	52.5	456.5	62.7	495.9[c]	51.7*
Lateral tibiofemoral loading rate (BW/s)	120.2	29.4	103.0	28.7	119.5	32.2
Strike index (%)	17.4	4.2	31.9[a]	22.2	13.7	9.9*

*Bayesian main effect. [a]Larger than Trainer. [b]Larger than maximal. [c]Larger than minimal.

event of a main effect, post-hoc Bayesian paired t-tests were conducted between each footwear condition (Wagenmakers et al., 2018).

Results

External loading and strike index – discrete parameters

For the loading rate there was decisive evidence of a main effect of footwear (BF = 92162.49). Post-hoc analyses showed that instantaneous loading rate was larger in the minimal compared to the traditional running shoes (BF = 73.03) and maximal (BF = 91.18) footwear (Table 2). For strike index there was very strong evidence of a main effect of footwear (BF = 46.84). Post-hoc analyses showed that the strike index was larger in the minimal compared to the traditional running shoes (BF = 10.09) (Table 2).

Three-dimensional kinematics – statistical parametric mapping

Maximal footwear was associated with increased knee flexion from 40 to 70% the stance phase compared to minimal running shoes (Figure 3(a)). In the minimal footwear the ankle was shown to exhibit increased plantarflexion from 0 to 10% of the stance phase compared to the traditional running shoes and increased dorsiflexion from 30 to 50% of the stance phase in relation to the maximal running shoes (Figure 3(b,c)). Finally, maximal footwear exhibited increased ankle eversion compared to traditional running shoes from 10 to 30% of the stance phase and from 10 to 20% of the stance phase in relation to minimal running shoes (Figure 3(d,e)).

Joint loading – discrete parameters

For the lateral tibiofemoral integral there was substantial evidence of a main effect of footwear (BF = 5.47). Post-hoc analyses showed that the lateral tibiofemoral integral was larger in the traditional (BF = 3.05) and maximal (BF = 3.02) compared to minimal running shoes (Table 2). For the patellofemoral force integral there was substantial evidence of a main effect of footwear (BF = 3.20). Post-hoc analyses showed that the patellofemoral force integral was larger in the traditional (BF = 3.39) compared to minimal running shoes (Table 2). Finally, for the patellofemoral stress integral there was substantial evidence of a main effect of footwear (BF = 3.11). Post-hoc analyses showed that the patellofemoral stress integral was larger in the traditional (BF = 3.72) compared to minimal running shoes (Table 2).

Muscle forces – discrete parameters

For the vastus intermedius integral there was substantial evidence of a main effect of footwear

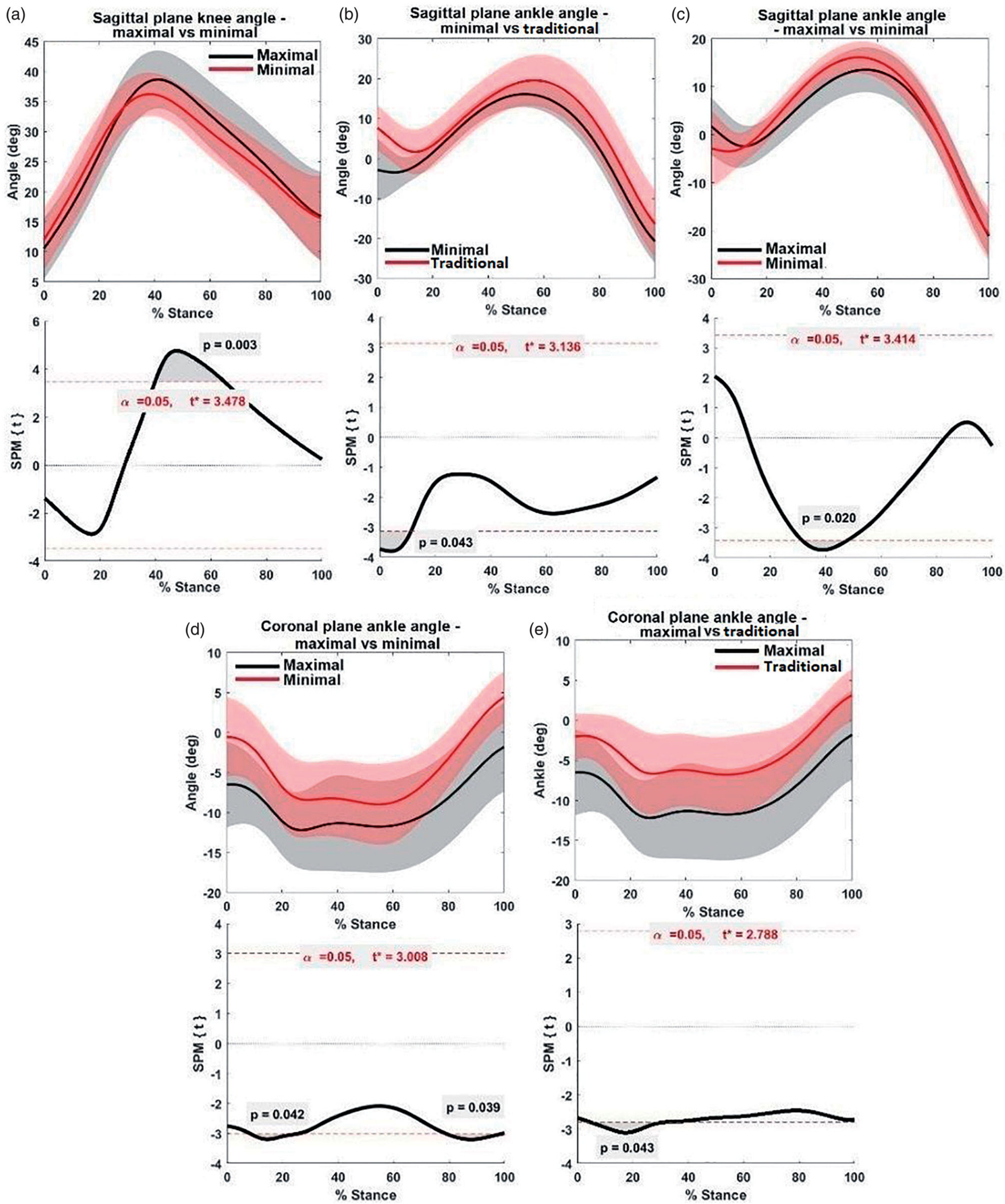

Figure 3. SPM analyses for joint loads.

(BF = 9.04). Post-hoc analyses showed that the vastus intermedius integral was larger in the traditional (BF = 5.07) compared to minimal running shoes (Table 3). For the vastus lateralis integral there was strong evidence of a main effect of footwear (BF = 10.70). Post-hoc analyses showed that the vastus lateralis integral was larger in the traditional (BF = 4.04) and maximal (BF = 4.14) compared to minimal running shoes (Table 3). For the vastus medialis integral there was substantial

Table 3. Discrete muscle force parameters (mean ± standard deviations) as a function of footwear.

	Maximal		Minimal		Traditional	
	Mean	SD	Mean	SD	Mean	SD
Biceps femoris LH integral (BW·ms)	15.6	8.7	19.0	11.3	24.0	12.9
Biceps femoris SH integral (BW·ms)	38.8	14.2	39.4	14.7	35.2	11.8
Extensor digitorum longus integral (BW·ms)	36.6	19.2	40.7	21.0	39.0	20.1
Extensor hallucis longus integral (BW·ms)	22.0	8.6	20.4	9.4	20.7	8.4
Glutaeus maximus integral (BW·ms)	142.3	50.8	140.2	50.9	160.3	62.7
Glutaeus medius integral (BW·ms)	277.0	34.9	276.9	36.9	279.1	40.2
Glutaeus minimus integral (BW·ms)	122.1	22.0	120.8	22.6	122.7	18.2
Lateral gastrocnemius integral (BW·ms)	71.3	19.8	74.5	19.7	71.2	18.5
Medial gastrocnemius integral (BW·ms)	146.0	17.1	164.8	28.0	154.0	15.8
Tibialis anterior integral (BW·ms)	38.2	19.8	28.3	14.5	34.2[a]	16.7*
Rectus femoris integral (BW·ms)	288.9	77.4	284.0	52.1	272.9	61.5
Semimembranosus integral (BW·ms)	31.5	18.2	29.7	11.7	30.8	12.4
Semitendinosus integral (BW·ms)	6.2	2.0	6.2	2.6	6.6	2.2
Soleus integral (BW·ms)	482.3	54.6	485.0	33.4	474.3	42.3
Vastus intermedius integral (BW·ms)	201.2	39.9	185.0	43.5	208.8[a]	29.9*
Vastus lateralis integral (BW·ms)	308.0	59.1	281.9	67.5	320.2[a]	44.4*
Vastus medialis integral (BW·ms)	182.0	37.6	167.2	40.5	188.7[a]	28.4*

*Bayesian main effect. [a]Larger than minimal.

evidence of a main effect of footwear (BF = 7.16). Post-hoc analyses showed that the vastus medialis integral was larger in the traditional (BF = 4.70) compared to minimal running shoes (Table 3). For the tibialis anterior integral there was substantial evidence of a main effect of footwear (BF = 3.31). Post-hoc analyses showed that the tibialis anterior integral was larger in the traditional (BF = 6.98) compared to minimal running shoes (Table 3).

Joint loading – statistical parametric mapping

Minimal footwear was associated with increased hip force compared to the traditional in the first 5% of the stance phase (Figure 4(a)). In addition, minimal footwear was associated with increased medial tibiofemoral force in first 5% of the stance phase compared to the traditional and maximal running shoes (Figure 4(b,c)). Furthermore, the traditional running shoes were associated with increased patellofemoral force and stress from 40 to 45% of the stance phase (Figure 4(d,e)). Finally, minimal footwear exhibited increased ankle force in the first 5% of the stance phase and increased Achilles tendon force from 20 to 40% of the stance phase compared to the traditional running shoes (Figure 4(f,g)).

Muscle forces – statistical parametric mapping

The traditional running shoes were associated with increased vastus lateralis force from 40 to 45% of the stance phase compared to minimal running shoes (Figure 5(a)). In addition, minimal footwear exhibited increased glutaeus medius and glutaeus minimus force from 5 to 10% of the stance phase compared to the traditional running shoes (Figure 5(b,c)).

Discussion

The current study aimed using musculoskeletal simulation, to examine running biomechanics in minimal, maximal and traditional running shoes using SPM and Bayesian approaches. To the authors knowledge this represents the first quantitative comparison of these footwear conditions using a musculoskeletal simulation-based approach.

Firstly, the kinematic analysis of the sagittal plane ankle angle using SPM as well as the discrete investigation of the strike index showed that the minimal footwear mediated a more anterior footstrike position in relation to the traditional running shoes. This observation supports previous analyses (Sinclair, Fau-Goodwin, et al., 2016; Sinclair et al., 2019; Squadrone et al., 2015). However, the lack of differences between minimal and maximal footwear oppose those of Sinclair, Fau-Goodwin, et al. (2016) yet support the observations of Hannigan and Pollard (2020). It could be speculated that the lack of conformity between studies relates to the divergence in footwear characteristics, as Hannigan and Pollard (2020) used customized midsole thicknesses (rather than different footwear models) to differentiate between minimal and maximal conditions. However, Sinclair, Fau-Goodwin, et al.

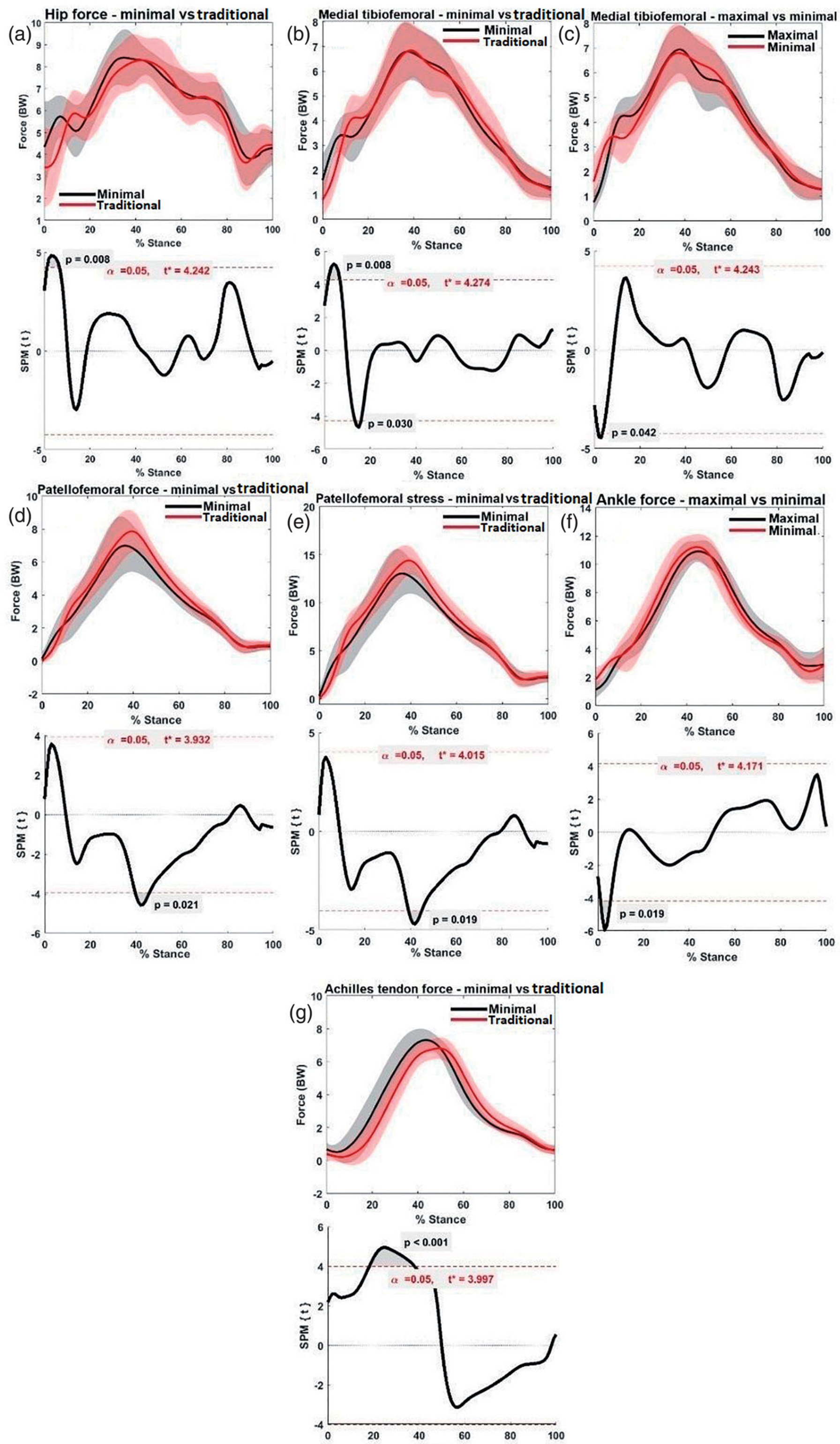

Figure 4. SPM analyses for joint kinematics.

(2016) used identical footwear to those examined in this study, indicating that this may be an inconsistent effect. It is nonetheless important to contextualize the strike index values observed in all of the experimental footwear conditions; as in support of previous findings (Sinclair et al., 2019; Squadrone et al., 2015), regardless of which footwear condition was used, a rearfoot strike pattern was adopted. The aforementioned observations support those of both Tam et al. (2017) and Sinclair

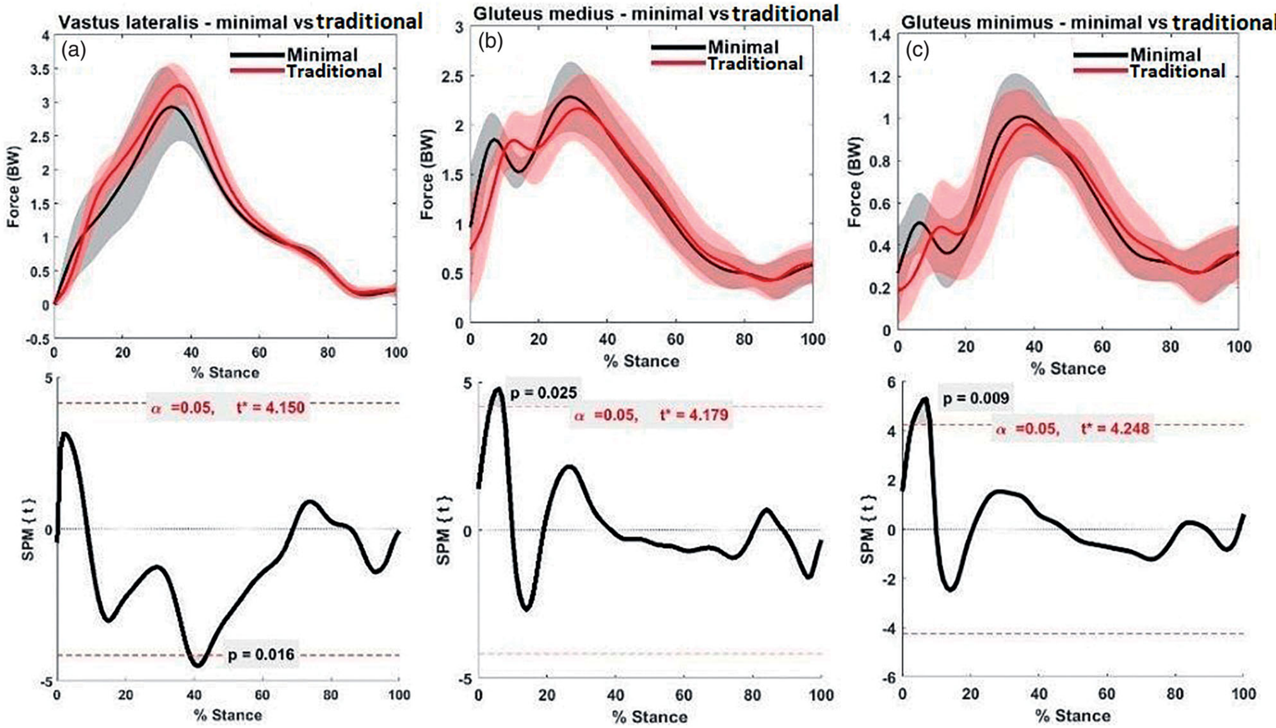

Figure 5. SPM analyses for muscle forces.

et al. (2019) that in acute investigations non-habitual runners do not sufficiently alter their running mechanics and continue to exhibit a rearfoot strike pattern.

In addition, in accordance with previous investigations (Hannigan & Pollard, 2020; Sinclair, Fau-Goodwin, et al., 2016), this study showed that minimal footwear were associated with enhanced loading rates compared to both traditional and maximal running shoes. As runners adopted a rearfoot strike pattern in each of the three experimental footwear conditions it was expected that external loading indices would be enhanced in the minimal footwear, owing to the reduced midsole cushioning in this footwear condition. As loading rates were increased when wearing minimal footwear, these observations may be clinically important. Given the proposed association between excessive loading rates and the aetiology of chronic injuries (Davis et al., 2004), this study indicates that wearing minimal footwear may place runners at increased risk from impact related injuries such as tibial stress fractures.

From a kinematic perspective it was revealed using SPM that maximal footwear were associated with increased eversion in relation to both minimal and traditional running shoes. This observation agrees partially with those found previously by Hannigan and Pollard (2020) who showed that peak eversion was greater in maximal and minimal footwear compared to traditional running shoes. Although the relationship between eversion characteristics and running injuries remains unclear (Dudley et al., 2017), this finding may be clinically meaningful as some have linked excessive eversion to the aetiology of medial tibial stress syndrome (Becker et al., 2018). Although further clarity regarding the role of eversion and maximal running shoes in chronic running injuries is certainly warranted before proposals regarding maximal footwear can be substantiated.

From a musculoskeletal simulation perspective, both the discrete and SPM based analyses showed that patellofemoral joint loading was larger in the traditional running shoes compared to minimal footwear. This observation concurs with those observed previously by Sinclair (2014), Sinclair, Richards, et al. (2016) and Bonacci et al. (2018) who showed significant reductions in patellofemoral loading when running in minimal footwear compared to the traditional running shoes. However, the lack of difference between minimal and maximal running shoes in patellofemoral joint loading was not evident in previous analyses

(Sinclair, Richards, et al., 2016). This lack of agreement between investigations is evident despite the footwear models being identical and indicates that reduced loading at the patellofemoral joint may not be a consistent observation between minimal and maximal running shoes. Minimal footwear transferred the footstrike location to a more anterior position, which has led many to propose that the role of the knee joint as a shock absorber reduces when the footstrike position moves anteriorly, mediating reductions in patellofemoral joint loading (Sinclair et al., 2019). This proposition is supported by the analysis of muscle forces as both SPM and discrete indices showed that quadriceps muscle forces that govern patellofemoral joint loading (Mason et al., 2008) were greater in the traditional compared to minimal running shoes. Importantly, excessive patellofemoral joint loading is considered the biomechanical 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 compared to the traditional condition, minimal running shoes may be effective in attenuating the biomechanical parameters linked to the aetiology of patellofemoral pain.

It was also revealed using SPM that compressive hip forces during the early stance phase were larger in the minimal in comparison to the traditional running shoes. It is likely that this was mediated through increases in glutaeal forces which through SPM were similarly enhanced during early stance in minimal running shoes (Neumann, 2010). This investigation is the first to contrast compressive hip joint loading in minimal, maximal and traditional running shoes so cross study comparisons are not possible; however, the findings support those of Sinclair (2018) who showed that running barefoot increased hip joint forces compared to traditional running shoes. It was also revealed using SPM that medial tibiofemoral loading was greater during early stance yet via discrete analyses that the lateral tibiofemoral integral was greater in minimal compared to both maximal and traditional running shoes. This observation supports those of Sinclair et al. (2018) who showed via SPM and the discrete knee adduction moment, that minimal footwear enhanced the extent of medial knee loading compared to traditional running shoes although there has yet to be any investigation comparing lateral tibiofemoral loading in minimal, maximal and traditional running shoes. Therefore, as the aetiology of degenerative hip and tibiofemoral joint pathologies are influenced by compressive joint loading (Johnson & Hunter, 2014), it is possible that minimal footwear may enhance the risk of chronic hip and medial tibiofemoral pathologies. However, as these differences were primarily observed in early stance when the magnitude of the forces were far from the maximal and (in the (tibiofemoral joint) at the expense of increasing laterally directed loading; further epidemiological research is required concerning the potential clinical influence of running in minimal footwear on joint health.

Furthermore, this investigation showed using SPM that Achilles tendon loading was significantly larger in minimal footwear during midstance in relation to the traditional running shoes. This observation concurs with previous investigations (Sinclair, 2014; Sinclair et al., 2015; Sinclair et al., 2019) indicating that minimal footwear significantly enhanced Achilles tendon loading compared to traditional running shoes. Once again, the lack of difference between minimal and maximal running shoes in Achilles tendon loading was not evident in previous analyses, despite identical footwear between utilized in both investigations (Sinclair et al., 2015), suggesting that reduced Achilles tendon kinetics may not be a uniform finding between minimal and maximal footwear conditions. This finding indicates that minimal footwear may increase the likelihood of Achilles tendinopathy, as tendinopathy is linked to excessive forces experienced by the tendon itself (Selvanetti et al., 1997). However, it has also been advocated that enhanced tendon loading facilitated by minimal running shoes/running barefoot mediates improvements in tendon stiffness characteristics required for effective storage and release of elastic energy (Histen et al., 2017). Thus, future longitudinal analyses of runners transitioning to minimal footwear are necessary in order to examine the effects of transitioning to minimal footwear on the physiological characteristics of the Achilles tendon.

Importantly, the efficacy of musculoskeletal simulation modelling approaches is dependent on the accuracy and fidelity of the underlying model being used to quantify the kinetics of the movement and conditions being investigated (Sinclair et al., 2020). A range of assumptions and mathematical simplifications are made in the development of musculoskeletal models for simulation analyses, which may have impacted the results from the current investigation. As such there is scope for future developments to address and improve upon these limitations, in order to generate more accurate and valid musculoskeletal simulations in relation to the effects of different running shoes on the mechanisms linked to the aetiology of injuries.

The biomechanics of minimal, maximal and traditional running shoes have received widespread research attention. The novel application of musculoskeletal simulation analysis showed via Bayesian analyses that traditional running shoes increased vasti, lateral tibiofemoral and patellofemoral stress integrals and using SPM vastus lateralis forces during midstance compared to minimal running shoes. Furthermore, SPM showed that minimal footwear increased glutaeal, medial tibiofemoral and hip forces during early stance and Achilles tendon forces during midstance compared to traditional running shoes and Bayesian analysis showed that minimal increased loading rates and lateral tibiofemoral impulse compared to maximal and traditional running shoes. Finally, SPM also showed that maximal footwear enhanced ankle eversion during midstance comapred to both minimal and traditional running shoes. This investigation therefore indicates that minimal footwear may increase risk from the biomechanical parameters associated with impact related chronic pathologies yet attenuate risk from patellofemoral pain compared to traditional running shoes. In addition, owing to increases in ankle eversion, maximal running shoes may enhance risk to the aetiology of medial tibial stress syndrome compared to minimal and traditional running shoes.

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