Effects of minimal, maximal and conventional footwear on whole-body and muscle metabolic power during running.

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Abstract

The aim of this investigation was to examine running biomechanics in minimal, maximal and conventional running shoes using a musculoskeletal simulation approach to evaluate their effects on whole-body and muscle-level metabolic power. Fifteen healthy male runners completed overground trials in all three footwear conditions at a controlled speed of 4.0 m/s. Kinematic data were collected using an eight-camera motion-capture system, and ground reaction forces were obtained from an in-ground force plate. Musculoskeletal simulations quantified whole-body metabolic power as well as joint and muscle-level mechanical and metabolic outputs. Comparisons between footwear conditions were performed using linear mixed-effects models with repeated measures. Whole-body metabolic power did not differ between maximal (11.35W/kg), minimal (11.07W/kg) and conventional (11.26W/kg) shoes. In contrast, joint mechanics and muscle-level energetics showed significant differences. Knee power generation and absorption were significantly greater in maximal (generation=8.05; absorption=−18.07W/kg) and conventional (generation=8.20; absorption=−17.00W/kg) footwear than in minimal (generation=6.53; absorption=−12.81W/kg). Ankle power generation and absorption were significantly higher in minimal footwear (generation=15.89; absorption=−11.96W/kg) than in conventional (generation=14.23; absorption=−10.40W/kg) or maximal (generation=13.15; absorption=−9.37W/kg) footwear. Stance-limb peak knee extensor power was higher in conventional (6.38W/kg) and maximal (5.85W/kg) than in minimal (5.48W/kg) footwear. Whole-body metabolic power did not differ between footwear conditions, yet distinct changes in joint mechanics and muscle-level metabolic power were observed. In minimal footwear, these changes reflected a shift in the distribution of mechanical and metabolic work from the knee to the ankle.

Keywords: Biomechanics; Energy expenditure; Musculoskeletal simulation; Running; Running shoes

Introduction

Running has garnered widespread participation as a physical activity modality, largely due to its simplicity and ease of access [1]. As a primary mode of aerobic exercise, its significant physiological and psychological health benefits are well recognised [2,3]. Beyond its role in health and fitness maintenance, running constitutes a major competitive sport, with organized events spanning road, cross-country, and trail disciplines that attract both amateur and elite participants [4].

Improving running performance is a primary goal for runners. While physiological adaptations from training are essential, running economy is a critical determinant of competitive success [4]. Endurance performance itself is delineated as the capacity to sustain a specific work output; a process metabolically underpinned by the aerobic re-synthesis of adenosine triphosphate via oxidative phosphorylation [5]. This metabolic process is contingent upon the delivery and utilization of oxygen alongside the availability of fuels in the form of both carbohydrate and fats. Consequently, running economy is operationally defined as the steady-state oxygen consumption required to maintain a given submaximal running velocity [6].

Running economy is strongly associated with running performance [7], often demonstrating superior predictive validity compared to measures of maximal oxygen uptake [8]. A higher running economy indicates that an individual can sustain a given submaximal work output at a reduced fraction of their maximum aerobic capacity, thereby enhancing running performance, as better economy enables faster race times at an equivalent level of fitness [4]. Running economy is influenced by an interplay of physiological, biomechanical, and anthropometric variables [9,10,11]. Due to its established association with improved performance, enhancements in running economy are a primary objective to both runners and researchers alike.

The running shoe represents the primary interface between the body and the ground. Consequently, substantial advancements in its design have been driven by the dual aims of reducing the high incidence of chronic running-related injuries [12] and enhancing performance [13]. This impetus has led footwear manufacturers to develop a wide range of midsole cushioning technologies, giving rise to both minimal and maximal shoe designs [13]. Minimal running shoes are characterized by a low or zero heel-to-toe drop, high midsole flexibility, and reduced mass [14]. In contrast, maximal running shoes, while also often featuring a low heel-to-toe drop, incorporate a significantly greater volume of midsole cushioning along their entire length [13]. The biomechanical implications of running in these contrasting running shoe types have consequently become a prominent focus of contemporary footwear research.

Minimal running shoes are associated with elevated vertical loading rates, increased tibial accelerations, greater effective mass, and augmented vertical limb stiffness during running, relative to both conventional and maximal running shoes [12,15,16,17,18]. These footwear designs typically promote a more anterior footstrike position [19,20,21], and a reduced step length [17,21,22] compared to conventional and maximal footwear. Furthermore, studies indicate that minimal footwear is linked to enhanced gastrocnemius and soleus muscle forces, while conventional and maximal footwear elicit greater quadriceps muscle forces [21]. Several investigations have also demonstrated that minimal footwear reduces patellofemoral joint loading compared to both conventional [22,23,24,25,26] and maximal running shoes [25], leading some to propose a differential role of the knee and ankle joints in impact absorption across footwear conditions [24].

Modifying the mechanical characteristics of footwear has also been hypothesized as a method to enhance running economy [27]. Proponents of minimal footwear often contend that they are more economical [28]. Indeed, studies by Squadrone & Gallozi [29], Perl et al., [30] and Moore et al. [31] showed that running in minimal footwear led to significant improvements in running economy compared to conventional shoes. However, some research, has shown no effect of minimal footwear [11], and others such as that by Sinclair et al. [32], has demonstrated that minimal footwear was associated with diminished running economy.

Running economy is typically assessed using indirect calorimetry, which quantifies metabolic cost by analyzing respiratory gases influenced by the body's metabolism [33]. Although this approach does offer valuable insights for runners, it provides only a global assessment of running economy [34] and has limited clinical applicability due to the requirement for prolonged steady-state ambulation [35,36]. Recent advancements in computational biomechanics present a promising avenue for assessing running economy. Musculoskeletal simulation-based approaches that incorporate metabolic cost models offer a more effective means of estimating metabolic demand [37]. These methods are particularly advantageous in clinical contexts, as they rely on motion capture data rather than prolonged periods of steady-state ambulation. The principal advantage lies in the ability to parse the total whole-body metabolic cost into constituent muscular contributions [37]. This allows for the identification of specific muscle-level adaptations to different conditions and interventions, which a single systemic measure of oxygen consumption would otherwise obscure. Given that minimal, conventional, and maximal footwear have been shown to alter both muscle force distributions in major lower extremity musculature as well as the role of different joints in shock absorption [21,24]; examining muscle-level metabolic power offers a crucial additional dimension to understanding footwear-related biomechanical adaptations. These computational tools demonstrate a near-perfect correlation with indirect calorimetry [38], and may represent a superior methodology for examining the biomechanical influence of footwear during running, although direct comparative investigations of footwear using this detailed metabolic analysis are yet to be published.

Therefore, the aim of the current investigation was to examine running biomechanics in minimal, maximal and conventional running shoes using a musculoskeletal simulation-based approach to observe their effects on whole-body and muscle level metabolic power. An investigation of this nature may provide important information regarding the metabolic effects of minimal, maximal and conventional running shoes.

Methods

Participants

Fifteen males (mean ± SD: age 28.93 ± 6.66 years, height 1.73 ± 0.06 m, body mass 69.53 ± 5.07 kg, BMI = 23.33 ± 2.64 kg/m²) who habitually ran using conventional footwear volunteered for the current investigation. For inclusion into this investigation, participants were required to complete a minimum of 35 km per week of training and be between the ages of 18 and 40. All participants were free from pathology at the time of data collection and provided written informed consent, in accordance with the principles outlined in the Declaration of Helsinki. The procedure utilized for this investigation was approved by a university ethics committee (STEMH 361).

Sample size

Our focused literature search of PubMed, Scopus and Web of Science Core Collection did not identify a published comparison of minimal, maximal and conventional running footwear that used musculoskeletal simulation to quantify both whole-body metabolic power. Therefore, a pragmatic sample size calculation was undertaken using data from our previous work [32]. This showed that with a mean ± SD difference in steady state oxygen consumption of 1.90 ± 2.69 ml/kg/min between conditions; and a correlation between conditions of 0.85, to achieve α = 5% and β = 0.80, that 15 participants would be required for a within subjects' comparison between footwear.

Footwear

The footwear used in this study comprised three commercially available models representing a spectrum of footwear design: New Balance 1260 v2 (New Balance, Boston, Massachusetts, United States; termed conventional running shoes), Vibram FiveFingers ELX (Vibram, Albizzate, Italy; termed minimal), and HOKA ONE ONE Rapa Nui 2 Tarmac Road (HOKA, Goleta, California, United States; termed maximal) (Figure 1). Footwear characteristics were quantified using the minimalist index of

Esculier et al. [14]. The maximal shoe had a mass of 318 g, a heel thickness of 45 mm, and a heel-toe drop of 6 mm, corresponding to a minimalist index score of 18. The minimal shoe had a mass of 167 g, a heel thickness of 7 mm, and a heel-toe drop of 0 mm, yielding a minimalist index score of 92. The conventional shoe had a mass of 285 g, a heel thickness of 25 mm, and a heel-toe drop of 14 mm, with a minimalist index score of 20. To enhance ecological validity, footwear mass was not equalized across conditions.

@@@Figure 1 near here@@@

Procedure

Participants ran at a target speed of 4.0 m/s (±5%) and made contact with an embedded piezoelectric force plate (Kistler Instruments Ltd., Winterthur, Switzerland) with their right (dominant) foot; force data were recorded at 1000 Hz. Running speed was controlled using infra-red timing gates (Newtest, Oy Koulukatu, Finland). Footstrike and toe-off were defined using a vertical ground reaction force (GRF) threshold, with the stance phase identified as the period during which vertical GRF exceeded 20 N [39]. For each footwear condition, participants performed five valid trials. A trial was considered valid if the running speed fell within the prescribed range, the foot fully contacted the force platform, and there was no observable alteration in gait attributable to the testing setup. Footwear condition order was counterbalanced across participants. Kinematic and GRF signals were acquired simultaneously. Kinematics were collected at 250 Hz using an eight-camera motion capture system (Qualisys Medical AB, Goteburg, Sweden). The motion capture system was calibrated prior to each data collection session.

Body segments were quantified with six degrees of freedom using the calibrated anatomical systems technique [40]. Anatomical coordinate systems for the thorax, pelvis, thighs, shanks, and feet were established from retroreflective markers placed on C7, T12, and the xiphoid process, with additional bilateral markers located on the acromion processes, iliac crests, anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanters, calcanei, and the first and fifth metatarsals (Figure 2a). To capture segment motion, carbon-fibre tracking clusters (four non-linear retroreflective markers) were secured to the thigh and shank using double-sided tape and rigid sports tape. The feet were tracked using the calcaneus and first and fifth metatarsal markers, the pelvis using the PSIS and ASIS markers, and the thorax using T12, C7, and xiphoid markers. Static calibration trials were then performed so that the anatomical marker set could be defined relative to the tracking clusters/markers. Intra-rater reliability for the investigator responsible for placing the anatomical markers has previously been reported as excellent [41].

Ankle and knee joint centres were defined as the midpoints between the malleolar markers and between the femoral epicondyle markers, respectively [42,43]. The hip joint centre was estimated using a regression approach based on ASIS marker locations [44]. Segment coordinate axes were defined with the Z (transverse) axis directed vertically from the distal to proximal end of the segment, the Y (coronal) axis oriented from posterior to anterior, and the X (sagittal) axis derived via the right-hand rule and directed from medial to lateral (Figure 2b).

@@@Figure 2 near here@@@

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 step phase. GRF and marker trajectories were low-pass filtered at 50 Hz and 12 Hz, respectively, using a 4th-order zero-lag Butterworth filter. Kinetic and kinematic cut-off frequencies were obtained using residual analysis [45].

Running biomechanics

Joint powers were obtained as the product of the joint torque (quantified via Newton-Euler inverse dynamics) and the joint angular velocity [46], where positive power values indicated power generation and negative values representing absorption [47]. Joint powers were normalized to body mass (W/kg) and indices of peak generation and absorption were obtained from the hip, knee and ankle joints.

In accordance with the protocol of Addison and Lieberman [48], an impulse-momentum modelling approach was used to calculate effective mass (% BW), which was quantified in accordance with the below equation:

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

In the maximal and conventional footwear conditions, the impact peak was identified as the initial peak in the vertical GRF. In the minimal footwear condition, where an impact peak was not consistently evident, the location of the impact peak was assigned at the same relative point in stance as observed in the maximal and conventional shoes, in accordance with the approaches described by Lieberman et al. [49] and Sinclair et al. [18]. Time to impact peak (Δ time) was calculated as the interval from footstrike to the impact peak. The vertical GRF integral over the impact peak period was computed using a trapezoidal function. The change in foot vertical velocity (Δ foot vertical velocity) was defined as the difference in vertical foot velocity between footstrike and the impact peak [50]. Foot velocity was derived in Visual 3D [18] from the vertical component of the foot segment centre of mass.

The strike index was calculated as the position of the centre of pressure location at footstrike, relative to the total length of the foot [51]. A strike index of 0–33% denotes a rearfoot, 34–67% a midfoot and 68–100% a forefoot strike pattern. The step length (m) was determined as the linear anterior distance in the foot centre of mass location at footstrike between ipsilateral and contralateral footfalls in accordance with Sinclair et al., [21].

The duty factor was calculated as the ratio of stance time to 2 times the sum of stance and flight time [52], and the ratio of braking to propulsion was calculated by dividing the braking by the propulsion duration [53]. Braking duration was quantified as the period from footstrike to maximum knee flexion and propulsion from maximum knee flexion to toe-off [53].

Centre of mass oscillation was quantified as the maximum displacement of the model centre of mass (m) in the vertical direction [54]. Limb stiffness (N/kg/m) was calculated from the ratio of the peak normalized vertical GRF (N/kg) to the maximum vertical compression of the leg spring, which was calculated as the change in limb length from footstrike to minimum length [17]. Limb length (m) was quantified as the vertical height of the proximal end of the thigh segment within Visual 3D [17]. Vertical stiffness (N/kg/m) was calculated from the ratio of the peak normalized vertical GRF to the centre of mass oscillation [54].

Musculoskeletal simulation

Subsequently, step-phase data (defined as the interval between ipsilateral (right) and contralateral (left) foot contacts) were exported from Visual 3D and imported into OpenSim 3.3 (Simtk.org). Biomechanical processing was performed using a validated musculoskeletal model that was scaled to each participant's anthropometry. The model comprised 12 segments, 19 degrees of freedom, and 92 musculotendon actuators [55], and was used to estimate lower-limb muscle forces and joint contact forces. Prior to force estimation, a residual reduction algorithm was applied to minimise dynamic inconsistency between the experimental kinematics, measured GRFs, and the model's simulated dynamics [56].

The computed muscle control (CMC) was used to generate muscle-driven simulations of running in each footwear condition for each participant. The CMC simulation results were used as inputs into a modified version of the muscle energetics OpenSim probe proposed by Umberger et al., [37]. Specifically, parameters included in this model include the slow- and fast-twitch fiber composition of each modelled muscle, which we obtained from Johnson et al., [57], Garrett et al., [58] and Alway, [59]; and muscle mass, which was calculated for each muscle as a function of its physiological cross-sectional area, optimal fiber length and length at which muscle fibers generate maximum isometric force [38]. Total metabolic rate for each muscle was required to remain non-negative, reflecting the physiological principle that negative mechanical work does not contribute to ATP synthesis [60]. We modelled physiological cross-sectional area of skeletal muscle at a value of 1059.7 kg/m^3 which is representative of the density of mammalian muscle [37] and adopted the muscle specific tensions utilized by Hamner & Delp, (2013) to compute the maximum isometric force from the measured physiological cross-sectional areas [38].

To compute the normalized whole-body metabolic power (reflective of the sum of metabolic powers from each modelled musculotendon actuator), we added a basal rate of 1.13 W/kg [61] to that provided via the Umberger et al., [37] OpenSim probe. Whole-body metabolic power (W/kg) for statistical analyses was quantified by integrating the resulting whole-body metabolic rate over the step duration using a trapezoidal function [62].

In addition, muscle-level metabolic powers were also calculated to provide further insight into the metabolic demands associated with the different footwear conditions. This was achieved by summing the metabolic powers of muscles in the stance (right) and swing (left) limbs according to their primary anatomical functions [63]. Muscles were grouped as follows: hip flexors (iliacus, psoas major, pectineus, sartorius, tensor fasciae latae), hip extensors (gluteus maximus, biceps femoris long head, semimembranosus, semitendinosus), hip abductors (gluteus medius, gluteus minimus), hip adductors (adductor brevis, adductor longus, adductor magnus, gracilis), hip rotators (piriformis, quadratus femoris, gemellus), knee extensors (rectus femoris, vastus medialis, vastus lateralis, and vastus intermedius), knee flexors (biceps femoris long head), ankle dorsiflexors (tibialis anterior, extensor digitorum longus, extensor hallucis longus, peroneus tertius) and ankle plantarflexors (lateral and medial gastrocnemius, soleus, flexor digitorum longus, flexor hallucis longus, peroneus brevis, peroneus longus, tibialis posterior). Lumbar muscles, which were not directly associated with either limb, were grouped separately and included the erector spinae, internal obliques, and external obliques.

For each muscle group, the peak muscle metabolic power (W/kg) was identified for both the stance and swing limbs and 'overall' muscle metabolic power (W/kg) was quantified by integrating muscle metabolic power over the step duration again using a trapezoidal function [62]. Finally, the metabolic powers from both the stance and swing limb muscle groups were summed and used to derive combined herein described as 'total' muscle metabolic power, from which the overall and peak values were extracted using the aforementioned methods. Finally, to evaluate whether the proportional contributions of muscle level to whole-body metabolic power varied as a function of the experimental footwear conditions; the ratio of overall (i.e. integrated), stance, swing and total muscle powers from each muscle group were divided by whole-body metabolic power.

Statistical analyses

Descriptive statistics of means ± standard deviations (SD) were calculated for each of the experimental variables and footwear conditions outlined above. To contrast these variables between the three different experimental footwear conditions, within-subjects linear mixed-effects models were adopted utilizing compound symmetry and restricted maximum-likelihood methods; with footwear modelled as a fixed factor and participants representing the random intercept [21]. Effect sizes were calculated using Cohen's d, which were interpreted as 0.2 = small, 0.5 = medium, and 0.8 = large [64]. Statistical significance for all analyses was accepted at the $P \leq 0.05$ level, and all statistical analyses were conducted using SPSS v31.0 (IBM, SPSS, Armonk, NY, USA). In the interests of brevity, only statistically significant between-footwear effects are described in the text; complete descriptive statistics (means ± SD) for all variables are provided in Tables 1-3.

Results

Running biomechanics

@@@Table 1 near here@@@

Peak hip power generation was greater in the conventional ($P = 0.006$, $d = 0.83$) and minimal ($P = 0.018$, $d = 0.69$) footwear compared with maximal (Table 1). Peak knee power generation and absorption were both greater in the maximal (generation: $P < 0.001$, $d = 1.34$; absorption: $P < 0.001$, $d = 1.14$) and conventional (generation: $P < 0.001$, $d = 1.28$; absorption: $P < 0.001$, $d = 1.28$) footwear compared with minimal (Table 1). Peak ankle power generation was higher in minimal footwear than in maximal ($P < 0.001$, $d = 1.71$) and conventional ($P < 0.001$, $d = 1.21$), and higher in conventional than in maximal footwear ($P = 0.011$, $d = 0.75$) (Table 1). Peak ankle power absorption was also greater in minimal footwear than in maximal ($P < 0.001$, $d = 1.41$) and conventional ($P = 0.002$, $d = 0.97$), and greater in conventional than in maximal ($P = 0.002$, $d = 0.97$) (Table 1).

Strike index was greater in minimal footwear compared to maximal (P = 0.004, *d* = 0.90) and conventional (P < 0.001, *d* = 1.27) footwear (Table 1). The braking-to-propulsion duration ratio was higher in maximal (P < 0.001, *d* = 1.09) and conventional (P < 0.001, *d* = 1.07) footwear compared with minimal (Table 1). Step length (P = 0.023, *d* = 0.66) was significantly greater in conventional than minimal footwear, whereas cadence was significantly greater in minimal than conventional footwear (P = 0.03, *d* = 0.63) (Table 1).

Musculoskeletal simulation

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@@@Table 2 near here@@@

Total overall (P = 0.006, *d* = 0.84) and swing-limb (P = 0.019, *d* = 0.69) hip flexor metabolic power were greater in maximal than in minimal footwear (Table 2). Swing-limb overall hip extensor metabolic power was higher in conventional than in maximal footwear (P = 0.018, *d* = 0.69) (Table 2). Total overall (P = 0.009, *d* = 0.79) and stance-limb (P = 0.016, *d* = 0.70) hip abductor metabolic power were greater in maximal than in conventional footwear (Table 2). Stance-limb overall knee extensor metabolic power was also higher in conventional than in minimal footwear (P = 0.044, *d* = 0.57) (Table 2).

Total overall ankle dorsiflexor metabolic power was also greater in the conventional than in both maximal (P < 0.001, *d* = 1.62) and minimal (P < 0.001, *d* = 1.45) footwear (Table 2). Stance-limb overall ankle dorsiflexor metabolic power was significantly greater in the conventional (P = 0.003, *d* = 0.92) and minimal (P = 0.037, *d* = 0.60) footwear compared with maximal (Table 2). Swing-limb overall ankle dorsiflexor metabolic power was higher in the conventional (P < 0.001, *d* = 1.61) and maximal (P < 0.001, *d* = 1.48) footwear compared with minimal (Table 2).

@@@Table 3 near here@@@

Peak total (P = 0.001, *d* = 1.05) and swing-limb (P = 0.002, *d* = 1.01) hip flexor metabolic power were greater in maximal than in minimal footwear (Table 3; Figure 4f & 6f). Stance-limb peak hip adductor power was higher in maximal than in conventional (P = 0.032, *d* = 0.61) and minimal (P = 0.029, *d* = 0.63) footwear, and total peak adductor power was also greater in maximal than in conventional (P = 0.018, *d* = 0.69) and minimal (P = 0.013, *d* = 0.74) (Table 3; Figure 4d & 5d). Hip abductor peak power for the stance (P = 0.020, *d* = 0.68), swing-limb (P = 0.021, *d* = 0.67), and total (P = 0.018, *d* = 0.69) were greater in maximal than in conventional footwear. Stance-limb peak knee extensor power was higher in conventional (P = 0.011, *d* = 0.76) and maximal (P = 0.03, *d* = 0.63) than in minimal footwear (Table 3; Figure 4c, 5c & 6c).

Total peak dorsiflexor power was greater in conventional than in maximal (P < 0.001, *d* = 1.56) and minimal (P < 0.001, *d* = 1.23) footwear (Table 3; Figure 4a). Stance-limb peak ankle dorsiflexor metabolic power was greater in conventional (P = 0.003, *d* = 0.91) and minimal (P = 0.019, *d* = 0.69) footwear than in maximal footwear (Table 3; Figure 5a). Swing-limb peak dorsiflexor power was higher in conventional (P = 0.016, *d* = 0.71) and maximal (P = 0.007, *d* = 0.81) than in minimal footwear (Table 3; Figure 6a). Swing-limb peak plantarflexor power was higher in maximal than minimal footwear (P = 0.035, *d* = 0.61) (Table 3; Figure 6b), and peak lumbar power was greater in minimal than conventional footwear (P = 0.037, *d* = 0.60) (Table 3; Figure 4j).

@@@FIGURE 7 NEAR HERE@@@

The total hip flexor contribution to whole-body metabolic power was greater in maximal than in minimal footwear (P = 0.039, *d* = 0.59) (Figure 7). The total hip abductor contribution was also greater in maximal than in conventional footwear (P = 0.037, *d* = 0.60) (Figure 7). The stance-limb ankle dorsiflexor contribution to whole-body metabolic power was greater in conventional than in maximal footwear (P = 0.003, *d* = 0.93) (Figure 7). Swing-limb dorsiflexor contribution was higher in maximal (P < 0.001, *d* = 1.58) and conventional (P < 0.001, *d* = 1.49) than in minimal footwear (Figure 7). The total dorsiflexor contribution was greater in conventional than in maximal (P < 0.001, *d* = 2.32) and minimal (P < 0.001, *d* = 1.46) footwear (Figure 7). The lumbar contribution to whole-body metabolic power was greater in maximal (P =

0.019, *d* = 0.68) and minimal (P = 0.017, *d* = 0.70) than in conventional footwear (Figure 7).

Discussion

The aim of the current investigation was to observe the effects of minimal, maximal and conventional running shoes using a musculoskeletal simulation-based approach to examine their effects on whole-body and muscle level metabolic power. To our knowledge, no prior study has undertaken an exploration of this nature; and may therefore generate important information regarding the metabolic effects of minimal, maximal and conventional running shoes using an innovative modelling approach.

In the current study, running biomechanics were examined to determine the fundamental differences in mechanics between the experimental footwear. In agreement with the observations of Sinclair et al., [19,21,62], the biomechanical analysis revealed that the strike index was significantly greater in minimal footwear. Specifically, the contact position was significantly more anterior in minimal footwear; and when categorized using the strike index, a midfoot strike pattern was adopted in minimal footwear whereas a rearfoot pattern was utilized in both maximal and conventional running shoes [51]. Furthermore, the reduced step length found in minimal footwear also supports previous analyses [21,22,25], and the concept that footstrike and stride length are coupled [65]. As a greater ratio of braking to propulsion duration in running indicates a greater proportion of time spent slowing down the body's momentum [54] as would be expected when a greater step length is being utilized, this observation supports kinetic analyses demonstrating decreased braking forces when a non-rearfoot contact pattern is adopted [66]. More broadly, it is proposed that the above biomechanical observations relate to the absence of cushioning in minimal footwear, causing runners to adopt an anterior foot contact position and shorter step length in order to compensate for the lack of midsole interface in an attempt to attenuate the load experienced by the lower extremities [49].

Although biomechanical differences were observed across footwear conditions, this study found no differences in whole-body metabolic power. In relation to previous analyses using indirect calorimetry this observation aligns with the findings of Sinclair et al. [32], who also reported no metabolic benefit when comparing footwear of equal mass, but contrasts with studies by Squadrone and Gallozzi [29], Perl et al. [30], and Moore et al. [31], which reported improved running economy in minimal footwear. Moore et al. [31] included familiarisation and corrected oxygen consumption for shoe mass, observing benefits only when stride length remained consistent. Squadrone and Gallozzi, [29] used habitual minimal runners and incorporated familiarisation but did not equalise mass. Perl et al. [30] accounted for mass, footstrike, and stride, while Sinclair et al. [32] matched footwear mass but did not include familiarisation. In contrast, our study used non-habitual minimal users and did not control for mass, with the view of prioritising ecological validity over experimental constraint. Participants altered their strike pattern, step length, and cadence in the minimal condition, which may explain the lack of metabolic savings as Gruber et al. [67] proposed that deviating from habitual stride patterns increases metabolic cost, supporting the view that unfamiliarity with minimal footwear may offset any theoretical energetic advantage.

Although whole-body metabolic power did not differ between footwear conditions, clear distinctions emerged in both joint mechanics and muscle-level metabolic powers, indicating the presence of compensatory redistribution of energy production across the lower limb. Consistent with previous work [68,69,70,71], the present findings suggest that running in minimal footwear shifts the locus of mechanical work from the knee to the ankle joint. Greater ankle power generation and absorption observed in the minimal condition corresponded with elevated metabolic power in the dorsiflexor and plantarflexor groups, reflecting increased muscular demand associated with enhanced distal propulsion and braking. In contrast, the reduced knee joint power generation and absorption in minimal footwear were accompanied by lower knee extensor metabolic power, indicating that distal mechanical adjustments alleviated proximal load. Maximal footwear demonstrated the opposite pattern, with higher knee power generation and absorption and reduced ankle involvement, which paralleled increased metabolic demands in the hip flexor, adductor, and abductor groups. These muscles likely contributed to stabilisation and limb control during the longer ground contact times and greater braking-to-propulsion ratios that were observed in maximal footwear.

Differences in muscle group metabolic power across footwear types produced substantial local variations in both overall and peak metabolic power; however, these changes offset one another at the systemic level, resulting in no detectable difference in whole-body metabolic power. This suggests that the central nervous system may regulate mechanical and energetic output globally, distributing muscular effort to maintain overall efficiency despite altered joint and muscle mechanics [72]. This study provides the first integrated analysis linking muscle-level and whole-body metabolic power, showing that footwear alters the internal coordination of muscular metabolic work without affecting aggregate metabolic demand. These findings highlight compensatory redistribution of energy production across the lower limbs and further support the use of musculoskeletal simulation to quantify locomotor metabolic demands, offering detailed, muscle-specific insight beyond the global assessments provided by indirect calorimetry [34], which is limited by the need for prolonged steady-state ambulation [35,36].

A limitation of this study concerns the simplifications inherent in musculoskeletal modelling/ simulation. The simulation model adopted in this study neglects several physiological factors such as the influence of past muscle states [73] and muscle temperature [74], both of which can alter contractile dynamics and may therefore influence the estimation of metabolic cost. Moreover, muscle model parameters are commonly derived from cadaveric data and may not accurately reflect the properties of living muscle tissue in healthy men and women. Another limitation relates to the use of the CMC algorithm to resolve muscle redundancy. CMC minimizes the sum of squared muscle activations at each time step [75], a criterion that produces realistic joint kinematics [61] but may not represent physiologically optimal control. Alternative formulations that minimize energy expenditure [76], fatigue [77], or that consider longer time windows [78] should be developed/ investigated to yield more metabolically plausible solutions. Finally, the exclusion of upper-body muscle energetics introduces a small systematic underestimation of whole-body metabolic power but is unlikely to have influenced the observed trends [38].

Conclusion

This study provides the first integrated evaluation of how minimal, maximal, and conventional running footwear influence both joint mechanics and muscle-level metabolic power using a musculoskeletal simulation framework. Although each footwear type produced distinct biomechanical patterns and substantial shifts in metabolic energy generation within the lower limb musculature, these local alterations did not translate into detectable differences in whole-body metabolic power. The findings indicate that runners reorganise muscular metabolic effort to preserve global energetic output, highlighting a compensatory redistribution of metabolic demand across muscle groups. Minimal footwear increased reliance on the ankle musculature, whereas maximal footwear shifted demand toward the hip and knee, demonstrating that footwear design meaningfully influences the internal coordination of muscular work even when whole-body cost remains unchanged. These results emphasise the value of musculoskeletal simulation for quantifying locomotor energetics, as it provides detailed insight into muscle-specific metabolic demands that cannot be captured through indirect calorimetry alone. This approach offers a potentially powerful tool for understanding how footwear and other interventions shape the energetic strategies underpinning human running.

Credit author statement

Jonathan Sinclair: Conceptualization, Methodology, Investigation, Formal Analysis, Writing – Original Draft, Writing – Review & Editing, Visualization.

Jinluan: Conceptualization, Methodology, Writing – Original Draft, Writing – Review & Editing.

Ethical statement

The research involved human participants and received ethical approval from the Science, Technology, Engineering & Health Ethics Committee at University of Lancashire (STEMH 361). Participation was voluntary. Participants were informed of study objectives and content prior to participation. They were free to withdraw from the studies at any time. Data was collected on the basis of informed consent.

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Figure labels

Figure 1: Experimental footwear (a) conventional running shoes, (b) maximal, (c) minimal.

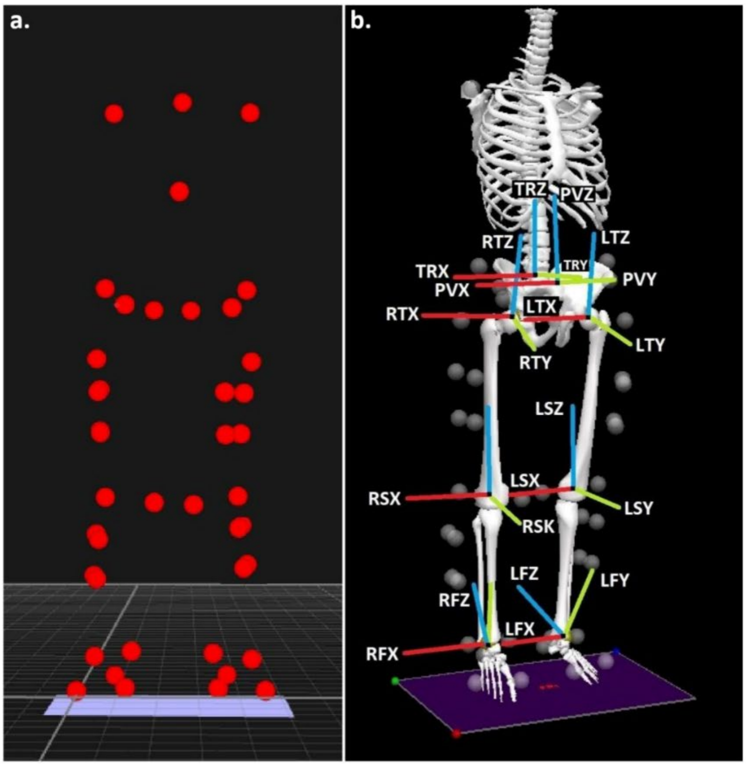

Figure 2: (a) Experimental marker set-up. (b) experimental model with segment co-ordinate axes (red: sagittal, green: coronal and blue: transverse axes).

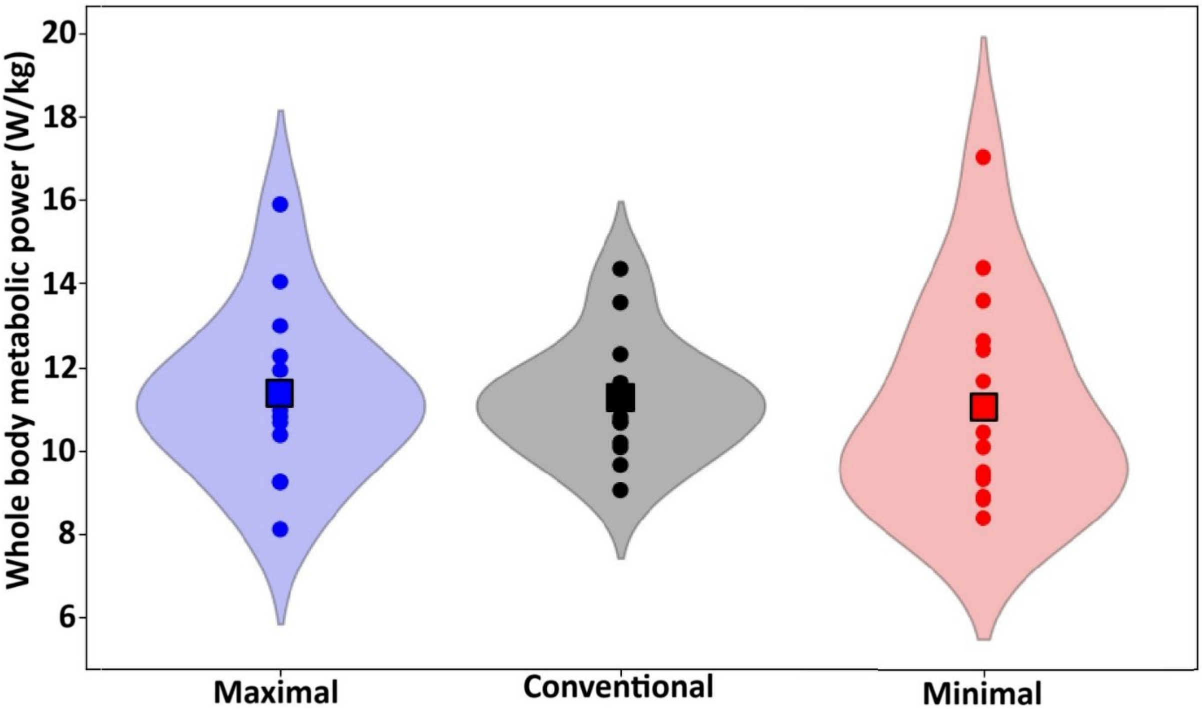

Figure 3: Whole-body metabolic power from each experimental footwear condition. Squares delineate mean values, and coloured circles represent each participant.

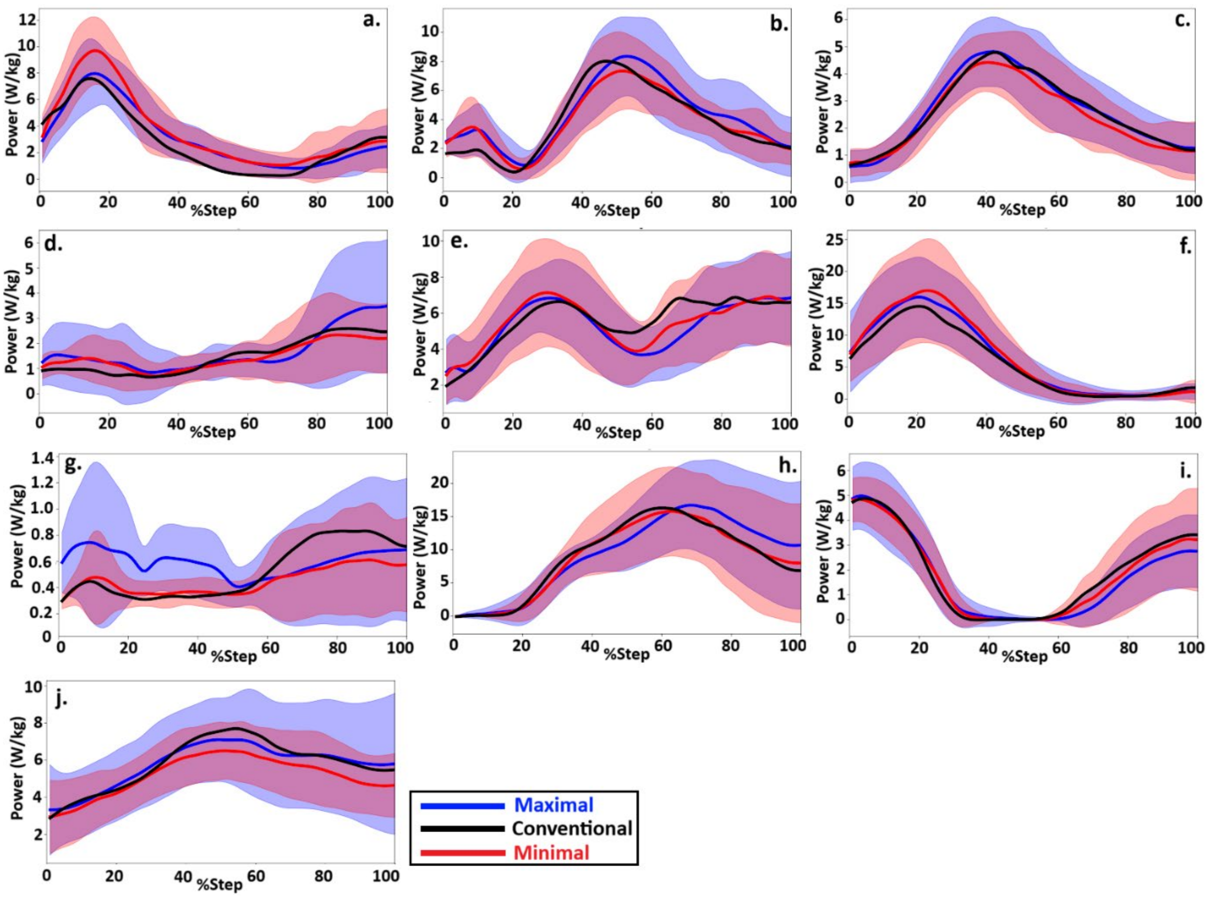

Figure 4: Total muscle metabolic power from each experimental footwear condition. The shaded area represents 1 * standard deviation (a. = ankle dorsiflexors, b. = ankle plantarflexors, c. = hip abductors, d. = hip adductors, e. = hip extensors, f. = hip flexors, g. = hip rotators, h. = knee extensors, i. = knee flexors & j. = lumbar).

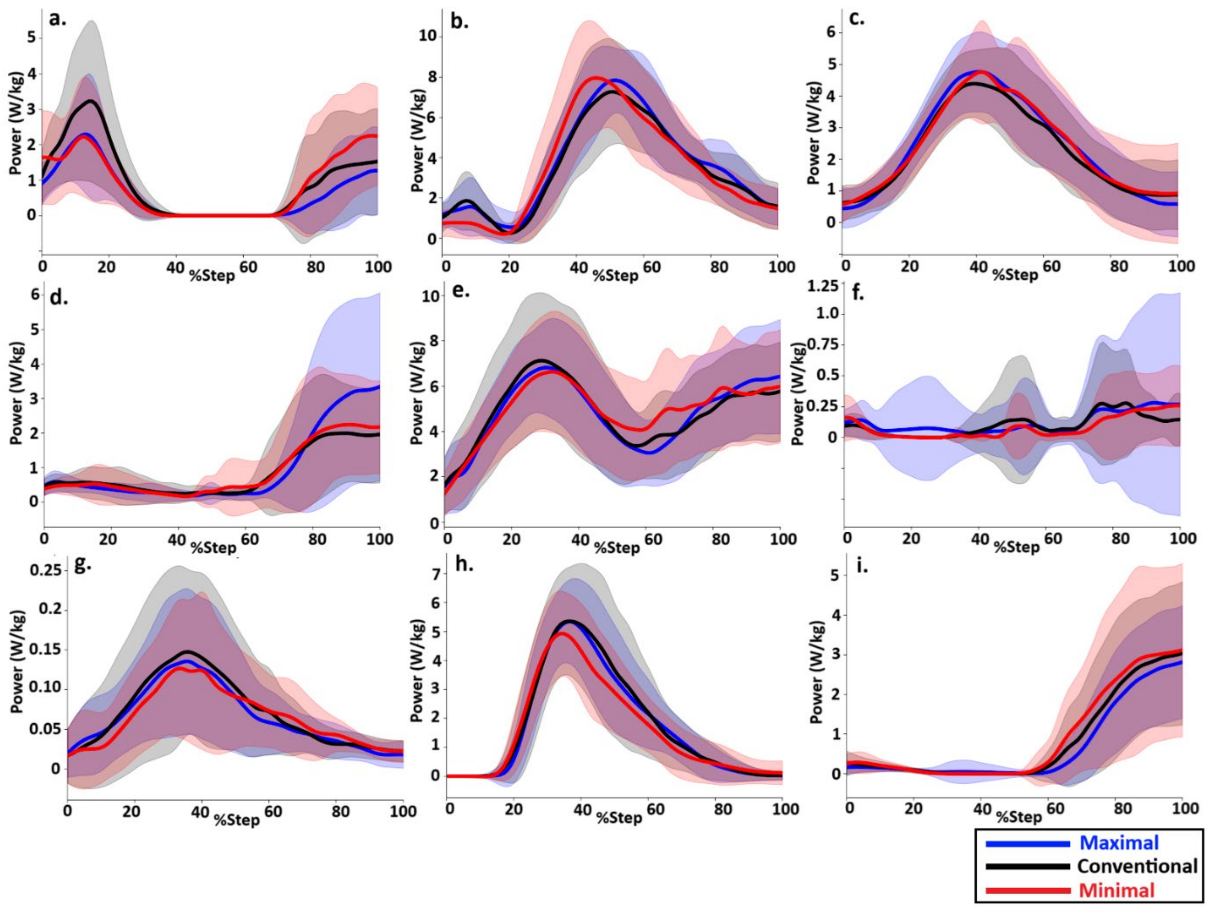

Figure 5: Stance limb metabolic power from each experimental footwear condition. The shaded area represents 1 * standard deviation (a. = ankle dorsiflexors, b. = ankle plantarflexors, c. = hip abductors, d. = hip adductors, e. = hip extensors, f. = hip flexors, g. = hip rotators, h. = knee extensors & i. = knee flexors).

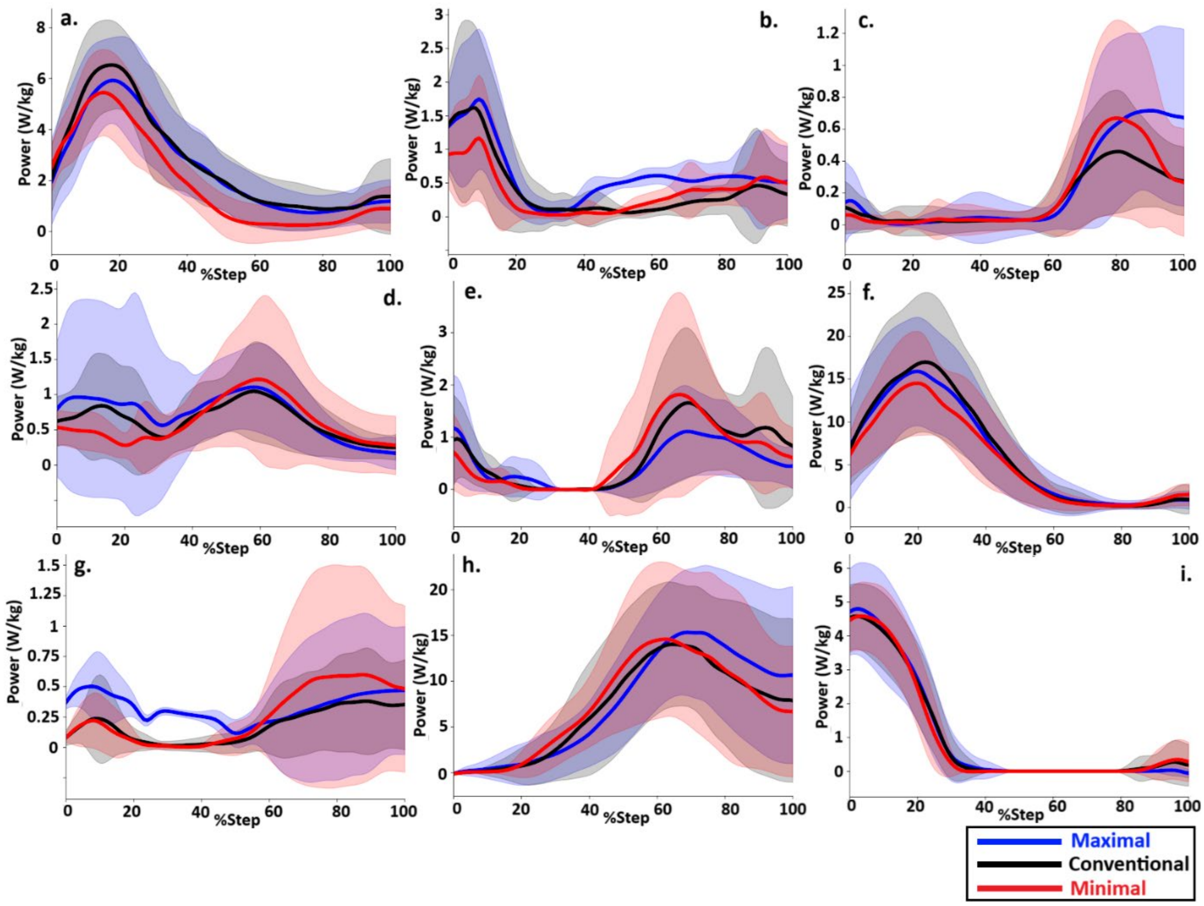

Figure 6: Swing limb metabolic power from each experimental footwear condition. The shaded area represents 1 * standard deviation (a. = ankle dorsiflexors, b. = ankle plantarflexors, c. = hip abductors, d. = hip adductors, e. = hip extensors, f. = hip flexors, g. = hip rotators, h. = knee extensors & i. = knee flexors).

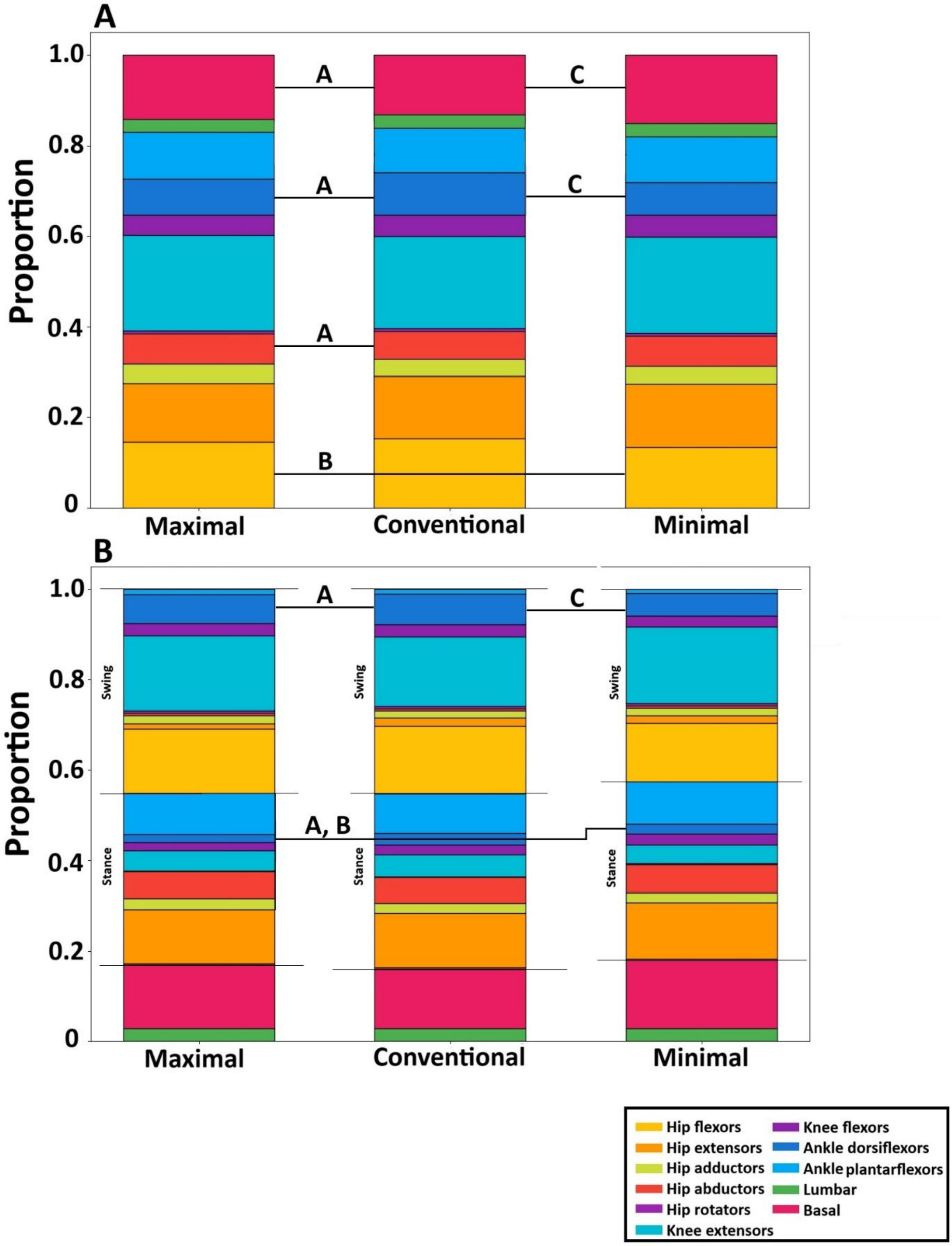

Figure 7: Proportional contribution of (a) total and (b) stance/ swing limb muscle groups to whole-body metabolic power from each experimental footwear condition (A = maximal significantly different from conventional, B = maximal significantly different from minimal & C = conventional significantly different from minimal).

Table 1: Running biomechanical parameters (Mean & SD) as a function of footwear.

	Maximal		**Conventional**		**Minimal**		
	Mean	SD	Mean	SD	Mean	SD	
Peak hip power generated (W/kg)	1.95	1.01	2.64	1.13	2.54	1.50	A, B
Peak hip power absorbed (W/kg)	-3.93	1.82	-4.33	2.34	-4.32	2.44	
Peak knee power generated (W/kg)	8.05	1.59	8.20	1.45	6.53	1.10	B, C
Peak knee power absorbed (W/kg)	-18.07	5.15	-17.00	3.93	-12.81	4.48	B, C
Peak ankle power generated (W/kg)	13.15	1.68	14.23	2.4	15.89	2.78	A, B, C
Peak ankle power absorbed (W/kg)	-9.37	1.89	-10.40	2.42	-11.96	2.27	A, B, C
Effective mass (%BW)	10.89	1.76	11.47	2.61	10.09	2.01	
Strike index (%)	17.85	3.80	14.42	9.30	34.15	19.69	B, C
Duty factor	0.35	0.02	0.34	0.03	0.35	0.03	
Step length (m)	0.90	0.08	0.9	0.08	0.89	0.07	C
Cadence (steps/ minute)	179.35	11.34	175.21	18.09	186.29	17.16	C
COM displacement (m)	0.06	0.01	0.06	0.01	0.06	0.01	
Ratio of braking to propulsion	0.71	0.07	0.72	0.10	0.63	0.07	B, C
Limb stiffness (N/kg/m)	75.72	21.33	73.66	22.42	76.77	19.67	
Vertical stiffness (N/kg/m)	47.75	9.58	46.95	10.83	47.57	10.61	

A = maximal significantly different from conventional, B = maximal significantly different from minimal & C = conventional significantly different from minimal

Table 2: Overall stance, swing and Total (W/kg) muscle metabolic powers (Mean & SD) as a function of footwear.

	Maximal		Conventional		Minimal		
	Mean	SD	Mean	SD	Mean	SD	
Hip flexors							
Stance limb (W/kg)	0.03	0.05	0.03	0.04	0.02	0.02	
Swing limb (W/kg)	1.63	0.43	1.70	0.64	1.46	0.53	B
Total (W/kg)	1.66	0.45	1.73	0.62	1.48	0.52	B
Hip extensors							
Stance limb (W/kg)	1.32	0.21	1.35	0.29	1.38	0.44	
Swing limb (W/kg)	0.14	0.06	0.20	0.12	0.19	0.14	A
Total (W/kg)	1.46	0.22	1.55	0.29	1.57	0.51	
Hip adductors							
Stance limb (W/kg)	0.29	0.15	0.25	0.13	0.26	0.16	
Swing limb (W/kg)	0.19	0.07	0.18	0.07	0.17	0.09	
Total (W/kg)	0.48	0.18	0.43	0.16	0.43	0.20	
Hip abductors							
Stance limb (W/kg)	0.67	0.17	0.64	0.15	0.66	0.18	A
Swing limb (W/kg)	0.07	0.03	0.05	0.02	0.06	0.04	
Total (W/kg)	0.74	0.17	0.69	0.15	0.72	0.19	A
Hip rotators							
Stance limb (W/kg)	0.02	0.01	0.02	0.01	0.02	0.01	
Swing limb (W/kg)	0.08	0.14	0.05	0.04	0.07	0.09	
Total (W/kg)	0.10	0.14	0.07	0.05	0.09	0.09	
Knee extensors							
Stance limb (W/kg)	0.49	0.15	0.53	0.20	0.44	0.12	C
Swing limb (W/kg)	1.96	0.92	1.79	0.93	1.97	1.19	
Total (W/kg)	2.45	0.87	2.32	0.83	2.41	1.12	
Knee flexors							
Stance limb (W/kg)	0.20	0.07	0.25	0.13	0.27	0.21	
Swing limb (W/kg)	0.29	0.10	0.29	0.06	0.27	0.06	
Total (W/kg)	0.49	0.13	0.54	0.15	0.54	0.22	
Ankle dorsiflexors							
Stance limb (W/kg)	0.18	0.06	0.29	0.12	0.25	0.10	A, B
Swing limb (W/kg)	0.71	0.14	0.76	0.13	0.54	0.09	B, C
Total (W/kg)	0.89	0.15	1.05	0.15	0.79	0.16	A, C
Ankle plantarflexors							
Stance limb (W/kg)	1.02	0.13	0.97	0.14	0.98	0.15	
Swing limb (W/kg)	0.15	0.18	0.11	0.05	0.10	0.03	
Total (W/kg)	1.17	0.17	1.09	0.16	1.08	0.13	
Lumbar							
Total (W/kg)	1.61	0.44	1.48	0.27	1.66	0.51	

A = maximal significantly different from conventional, B = maximal significantly different from minimal & C = conventional significantly different from minimal