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A biomechanical assessment of running with hallux unstable shoes of different material stiffness

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Purpose: Functional footwear with different unstable profiles has been widely used to mimic barefoot condition and offload plantar loading for pathological or injury prevention. However, little research investigates the effect of unstable structure on particular foot functions. In this study, a prototype of unstable shoe design with unstable element of different stiffness placed at the hallux (a regionalized rocker) was used. The primary objective was to analyse the biomechanical performance of running with hallux unstable shoes, aiming to potentially stimulate and increase the toe gripping function. *Methods*: The lower limb kinematics and plantar pressure distribution were measured to comparatively analyse the soft (*SS*) and hard (*SH*) unstable shoes with flat control shoes (*CS*). *Results*: The *SS* showed increased big toe and reduced forefoot plantar pressure. The *SS* led to similar lower limb kinematics to baseline *CS* except for reduced hip abduction, increased rotation range of motion (ROM), increased peak ankle plantar flexion and ROM. The *SH* presented significantly altered lower limb kinematics across hip, knee and ankle, and laterally distributed plantar pressure. *Conclusions*: Unstable shoes with soft material led to reduced medial metatarsal loading by increasing the support area and modified joint kinematics minimally. Unstable shoes with stiffer material presented compensatory kinematic movements across all joints and laterally shifted plantar loading distribution. These findings may provide implications on toe grip function training for foot pressure off-loading.

Key words: *regional unstable shoes, toes gripping function, kinematics, metatarsal loading*

1. Introduction

The human foot is the most distinct adaptation from evolution of upright bipedal locomotion, characterised by formation of unique foot structure and function, including the arch and toes [1], [11]. Different foot arch types (flat, high and normal) have been reported to present both static and dynamic biomechanical functions for asymptomatic or pathological feet [6], [16]. Important toe functions, both prehensile and ambulatory, have evolved naturally [9], and are believed to increase the load bearing area in the push-off phase of locomotion [13], [14]. However, foot deformities, particularly to the forefoot [14] and toes, like hallux valgus [4], present medially focused plantar loading to the metatarsal region. Relevant muscle and tendon dysfunction or degenerative conditions [7] are reported to contribute to the alteration of foot biomechanical functions [2]. Furthermore, prolonged running activities have led to increased foot and motor system pathology. The redistributed plantar pressure following long distance running presents increased forefoot and decreased toe loading, and more focussed medial-lateral foot loading [17]. These are considered as potential pre-cursor factors before running injuries manifest.

Peripheral neuropathy disorders, especially diabetes mellitus, have been reported with increased risk of plantar ulceration due to repetitive loading impact

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without protective sensation [12]. The success of shoes with protective functionality is not-conclusive, and degenerated intrinsic muscles and deformed natural foot shape still are prevalent. Unstable shoes as one kind of functional footwear have been proposed based on perturbation stimuli from unstable structures or surfaces [18]. Footwear designed to alleviate pressure to specific plantar regions have been evaluated for loading transfer and ulceration protection [5], [10], [15], [22]. Specifically, the Masai Barefoot Technology (MBT) shoe with round outsole shape in the anterior-posterior direction has been investigated on the effect of static postural control, dynamic walking and running gait kinematics alteration, muscles activation, joint loading alleviation and plantar pressure redistribution [18], [23], [24], [25]. Based on the functional concept of unstable shoes, also known as 'rocker shoes', specific structures to increase sole thickness and rigidity and perturb medial to lateral plantar surface have been designed and tested with both genders and different age groups under conditions of standing, walking or running [5], [10], [15], [21].

The gripping function of toes, especially hallux, could perform as actively pushing off the ground in the toe-off phase, thus expanding the load-bearing area, which were concentrated in the forefoot region [9]. It was reported in our previous studies [14], [15] that this function would be highly pronounced among habitually barefoot population, exhibiting altered plantar pressure distribution and ankle joint kinematics during running activities. Despite the multiple studies considering plantar unstable stimuli in rocker shoes, there are no studies that evaluate the effect of regionalized instability integrated with special toe functions. Hence, in this study, a prototype of hallux unstable shoes was manufactured, with an unstable hemisphere structure of different stiffness (Shoes Hard and Shoes Soft) fixed to the outsole of the hallux region to form unstable stimuli. The objective was to measure the immediate lower extremity kinematics, plantar pressure distribution and gait pattern response of healthy young males running with experimental shoes compared to control flat shoes. It was hypothesized that experimental shoes would stimulate the toe gripping function and present modified gait patterns, altered kinematic characteristics and reduced forefoot plantar loading.

2. Materials and methods

Participants

Twenty-five male habitually shod (rearfoot striking) runners were recruited to join the study. The basic demographic information includes, age: 23.6 ± 2.1 years, height: 173 ± 4.6 cm, weight: 68 ± 5.8 kg, foot length: 255 ± 4.8 mm, measured with the use of a 3D foot scanner following previously established protocol [19]. All participants were right-foot dominant, which was defined by kicking football with preferred leg. No history of wearing any unstable shoes, rocker shoes or toning shoes were reported, with no illness and global or regional motor disorders in the prior half year of the experiment. Participants with foot deformities or overpronation, oversupination, high-arched or flat foot were excluded, and all had no history of systematic sports or running training.

This study was approved by the Ethics Committee of the Ningbo University (ARGH20150916). Participants were informed of the purpose, requirements and process of this experiment. Written consent was obtained before the test from each participant.

Test Shoes

All shoes are EU size of 41 (255 mm). The control (baseline) shoes (*CS*) selected were normal flat shoes with canvas cover and soft sole without heel-toe drop and toe spring, which has been used in previously published studies [5], [10], [15]. The control shoes were used as baseline for comparative analysis of the foot and lower limb kinematics, plantar pressure and gait pattern versus experimental unstable shoes.

Fig. 1. The prototype of experimental shoes with rocker structure (a hemisphere 4 cm in diameter and 1 cm in height) to the outsole of hallux region (based on the rocker stiffness, Shoes-*S* and Shoes-*H* are defined)

Unstable structure with a hemisphere of 4 cm diameter and 1 cm height, was fixed to the outsole of the hallux region to form the experimental shoes in this study (Fig. 1). Footprint images from foot scanner were utilised in cooperation with outsole print of control shoes to define the position of unstable structure for experimental shoes, with the spherical vertex locating right under the central pressing point of hallux. Unstable stimuli to the hallux was devised with stiffness variation (Hard & Soft) of rubber elements, creating shoes-Hard (*SH*) and shoes-Soft (*SS*). The material properties were defined from measured elastic moduli (E) of Hard (3.3603 MPa) and Soft (0.3702 MPa) rubbers for unstable structures. The stiffness values are presented using a Hardness scale (H_A) value of shoes-H (67.92) and shoes-S (8.44) calculated from Eq. (1)

$$E = (15.75 + 2.15\, H_A) / (100 - H_A)\ [3]. \qquad (1)$$

Test protocol and procedures

The running tests, including collection of kinematic and plantar pressure data, were conducted in a motion analysis laboratory. Participants ran on an indoor track at self-selected speed. Prior to the test, participants randomly selected control shoes (*CS*) or experimental shoes (*SH & SS*) to perform five minutes' warm-up for familiarization and step adjustment with dominant (right) foot landing on the force platform, to avoid leg dominance effect on biomechanical performance [8]. During the test, participants ran randomly with the three shoe conditions.

The kinematics test utilized an eight-camera Vicon 3D motion analysis system (Oxford Metrics Ltd., Oxford, UK) and Nexus version 1.8.5 software to collect and process lower limb kinematic data at a frequency of 200 Hz. The Plug-In-Gait model [5], [13], using sixteen reflective markers, was used to capture the lower limb kinematic data. Reflective markers (12 mm in diameter) were used to define the joint centre and motion axis, including left anterior-superior iliac spine (LASI), right anterior-superior iliac spine (RASI), left posterior-superior iliac spine (LPSI), right posterior-superior iliac spine (RPSI), left thigh (LTHI), right thigh (RTHI), left knee (LKNE), right knee (RKNE), left tibia (LTIB), right tibia (RTIB), left ankle (LANK), right ankle (RANK), left heel (LHEE), right heel (RHEE), left toe (LTOE) and right toe (RTOE). Participants' height, weight, lower limb length, knee width and ankle width were measured and inserted into the model for the static model. A 3D Kistler force platform (1000 Hz) was used to define stance with a threshold vertical ground reaction force value of 20 N.

An insole plantar pressure measurement system (Novel Pedar system, Germany) was used to record foot pressure, with insole size of 41 (EU) and model W-1644L-1645R, with a frequency of 100 Hz. The insoles were calibrated with a pressure calibration system from 0 to 50 N/cm^2, so as to improve data accuracy and reduce experiment error. The insole was divided into seven anatomical parts (Fig. 2), including heel (H), middle foot (MF), medial forefoot (MFF), central forefoot (CFF), lateral forefoot (LFF), big toe (BT) and other toes (OT). The BT region of insole was consistent with the location of unstable structure to the outsole of experimental shoes (*SS* & *SH*).

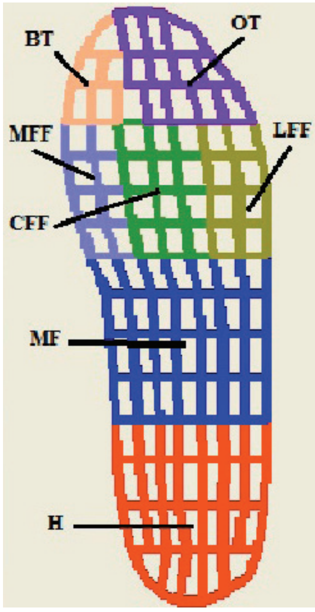

Fig. 2. The seven anatomical regions of plantar insole

Data process and statistical analysis

The kinematics and plantar pressure measurements were synchronously conducted. After five minutes' warm-up and familiarization, participants were required to run on the track with self-selected speed to present natural running characteristics and complete collection of six successful trials of kinematic and plantar pressure data. Stance contact time and running speed were measured to evaluate gait patterns similar to previous studies [5], [6], [8], [13], [16]. Kinematic data collection included mean joint angle (±SD), peak angle, joint range of motion of hip, knee and ankle during stance [5], [13]. Stance was defined from the vertical ground reaction force measured by the Kistler force platform with the threshold of vertical ground reaction force over 20 N. Plantar pressure measurements included maximal force, peak pressure and force time integral for each anatomical part to analyse foot biomechanical function [6], [8], [13], [15], [16]. All six trials of kinematic and plantar pressure data were averaged and normalized to minimize the inter-trials' error.

Contact time, running speed, peak angle value, joint range of motion, maximal force, peak pressure and force time integral were identified for statistical analysis using SPSS 17.0 (SPSS, Chicago, IL, US). The repeated measures ANOVA and post hoc Bonferroni test was conducted on the statistics analysis of kinematic and plantar pressure while running with control shoes (*CS*) and experimental shoes (*SH & SS*). The significance level was set at 0.05.

3. Results

Participants randomly selected *CS*, *SH* and *SS* to perform the running test with self-selected speed. The stance times of the right foot were 0.263 ± 0.019s (*CS*), 0.262 ± 0.023 s (*SH*) and 0.248 ± 0.019 s (*SS*), respectively. There was a significant difference exhibited between *CS* and *SS* (p = 0.024) and *SS* and *SH* (p = 0.031). However, *CS* and *SH* revealed no difference (p = 0.75). The running speed recorded for the three shoe designs were 2.46 ± 0.1 m/s (*CS*), 2.54 ± 0.12 m/s (*SH*) and 2.69 ± 0.11 m/s (*SS*). A significant difference was observed between *CS* and *SS* (p = 0.001) and *SS* and *SH* (p = 0.031), however, *CS* and *SH* again showed no difference (p = 0.217).

Lower limb kinematics

The lower limb kinematics varied significantly under conditions of running with *CS*, *SH* and *SS* (Fig. 3) for joints angles in stance. Peak angles and range of motion (ROM) are presented in the Table 1. Both *SS* and *CS* showed consistent joint motion trends. However, SS showed significantly smaller peak hip abduction angle, larger horizontal ROM, smaller peak knee extension angle, larger peak ankle plantarflexion angle and larger sagittal ROM than *CS*. The *SH* presents significant differences with *CS* and *SS*, as illustrated in Fig. 3 and Table 1.

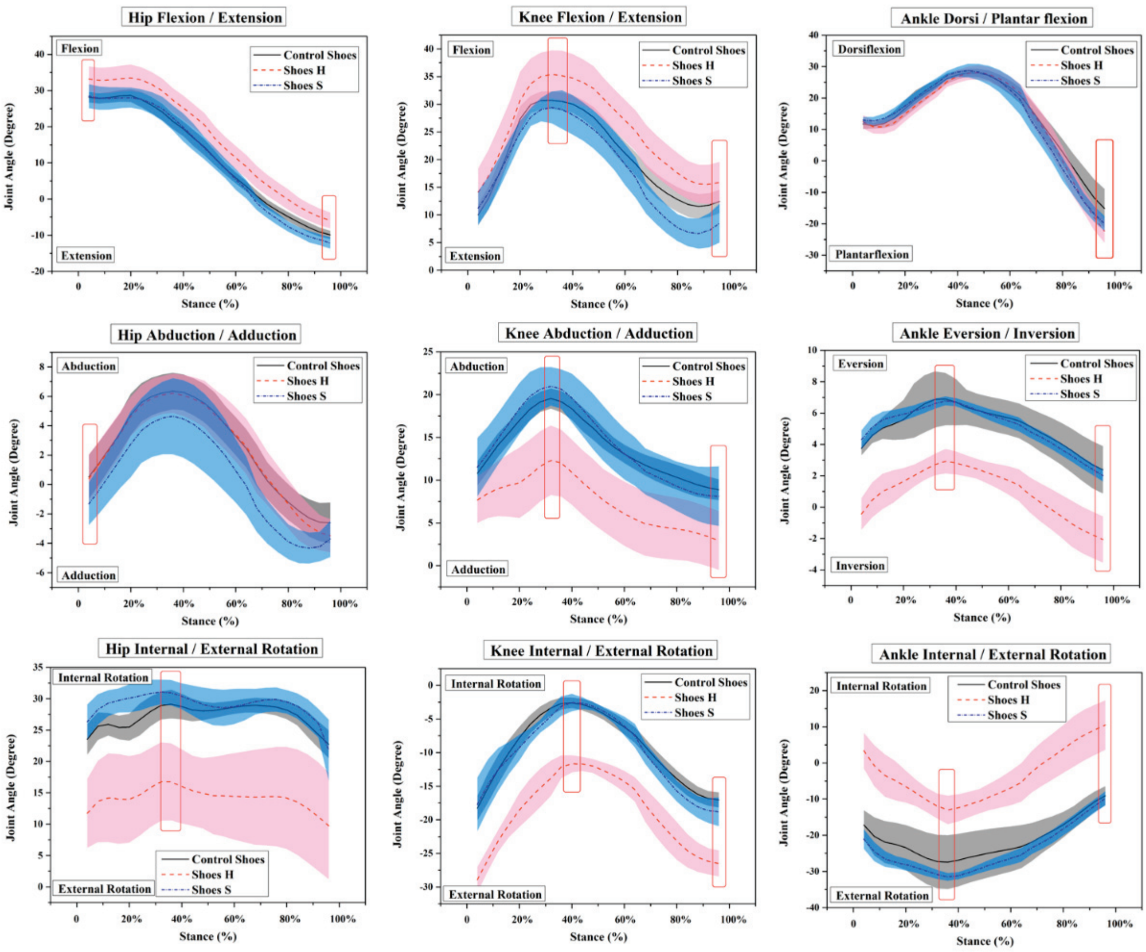

Fig. 3. The lower extremity joints angle curve in a stance (the black solid line represents CS, the red dash line represents SH, and the blue dot line represents SS) with significance highlighted with rectangle

Table 1. The peak angle value and ROM of lower extremity joints ($N = 25$)

Joints	Motion [°]	Control Shoes	Shoes H	Shoes S
Hip	Flex.	28.59(1.77)	33.53(3.63)§¶	28.42(3.31)
	Ext.	–9.88(1.12)	–5.9(2.22)§¶	–11.25(1.53)
	ROM	38.47(2.89)	39.43(5.85)	39.67(4.84)
	Abd.	6.37(1.25)¢	6.21(1.33)¶	4.65(2.59)
	Add.	–2.61(1.39)	–3.49(1.13)	–4.32(1.05)
	ROM	8.98(2.64)	9.7(2.46)	8.97(3.64)
	Int. Rot.	29.14(2.29)	16.78(6.26)§¶	31.08(1.96)
	Ext. Rot.	22.74(2.23)	9.72(8.49)§¶	21.76(6.92)
	ROM	6.4(4.52)¢	7.06(2.23)¶	9.32(4.96)
Knee	Flex.	30.73(1.58)	35.41(4.32)§¶	29.44(2.88)
	Ext.	10(1.91)¢	14.12(4.28)§¶	6.64(2.73)
	ROM	20.73(3.49)	21.29(8.6)	22.8(5.61)
	Abd.	19.54(1.73)	12.33(4.06)§¶	20.97(2.25)
	Add.	8.87(1.21)	2.94(3.44)§¶	8.12(3.49)
	ROM	10.67(2.94)	9.39(3.46)¶	12.85(5.74)
	Int. Rot.	–2.59(0.77)	–11.68(1.27)§¶	–2.55(1.29)
	Ext. Rot.	–18.29(1.82)	–28.91(1.89)§¶	–18.8(2.15)
	ROM	15.7(2.59)	17.23(3.16)	16.25(3.44)
Ankle	Dorsiflex.	28.04(1.23)	28.17(2.23)	28.73(2.16)
	Plantarflex.	–15.15(6.22)¢	–22.39(3.66)§	–19.96(2.37)
	ROM	43.19(7.45)¢	50.56(5.89)§	48.69(4.53)
	Ever.	6.89(1.67)	2.92(0.78)§¶	6.77(0.28)
	Inver.	2.37(1.51)	–2.07(1.48)§¶	2.01(0.38)
	ROM	4.52(3.18)	4.99(2.26)	4.76(0.66)
	Int. Rot.	–9.08(2.62)	10.46(6.82)§¶	–9.81(1.94)
	Ext. Rot.	–27.44(7.45)	–13(3.93)§¶	–31.52(1.08)
	ROM	18.36(4.88)	23.46(2.89)§	21.71(3.02)

Notes: "§", "¶", and "¢" indicates significance ($p < 0.05$), "§" representing significance between *CS* and *SH*, "¶" representing significance between *SH* and *SS*, and "¢" representing *CS* and *SS*.

Plantar pressure

The maximal force (a), peak pressure (b) and force time integral (c), all presented significant differences at the forefoot and toe regions (Fig. 4). In the MFF, *SS* showed reduced maximal force compared to *CS* (p = 0.39) and *SH* (p = 0.27), with significance in peak pressure (p = 0.008 and p = 0.015) and force time integral (p = 0.002 and p = 0.003). In the CFF, the *SS* showed smaller maximal force than *CS* (p = 0.006) and *SH* (p = 0.017), and reduced peak pressure (p = 0.002) and force time integral (p = 0.013), compared to *CS*. The peak pressure of *CS* in CFF is significantly larger than that of *SH* (p = 0.001). For the LFF, maximal force (p = 0.016 and p = 0.015) and force time integral (p = 0.018 and p = 0.015) showed difference between *CS* & *SH* and *SS* & *SH*. The peak pressure of *SH* in LFF was larger (p = 0.031) than that of *SS*. For the BT, the *SS* showed greater maximal force and peak pressure than *CS* (p = 0.000 and p = 0.000) and *SH* (p = 0.003 and p = 0.003). The force time integral in BT while running with *CS* is smaller than *SH* (p = 0.001) and *SS* (p = 0.000). For the OT region, *SH* showed similar behavior with larger maximal force (p = 0.000 and p = 0.000), peak pressure (p = 0.002 and p = 0.000) and force time integral (p = 0.000 and p = 0.005) than *CS* and *SS*, respectively.

4. Discussion

Previous studies have evaluated the biomechanics of unstable or rocker shoes across participants of different age, gender and pathology. This study proposed a novel functional footwear prototype integrating the unstable stimuli with toes gripping function, which is the first design combining toe function with stimuli of varying stiffness. The purpose of this study was to investigate the effect of running with hallux focussed unstable shoes (*SH* & *SS*) compared to baseline control shoes. The *SS* presented altered gait pattern (stance time and speed) and plantar pressure parameters (maximal force, peak pressure and force time integral) while retaining a consistent motion trend with *CS*. What else, the *SH* showed deviated plantar pressure and kinematics from *CS* and *SS*.

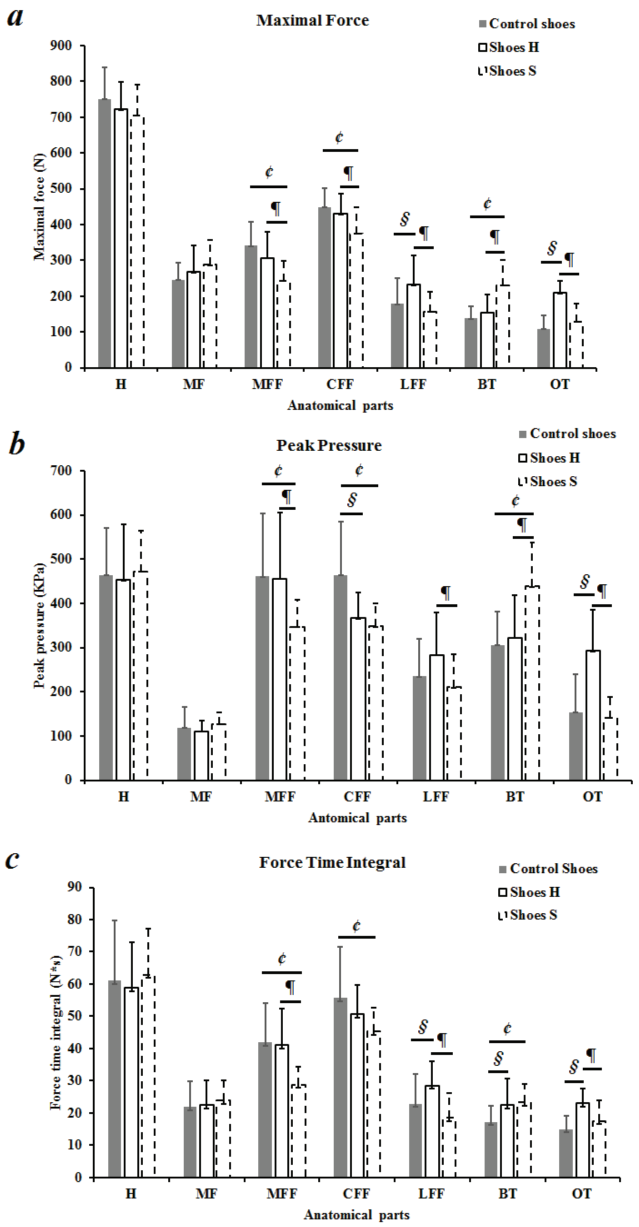

Fig. 4. The maximal force, peak pressure and force time integral of running with *CS*, *SH* and *SS* ("§", "¶", and "¢" indicates significance $p < 0.05$, with "§" representing significance between *CS* and *SH*, "¶" representing significance between *SH* and *SS*, and "¢" representing *CS* and *SS*)

Gait patterns differ among individuals of different age [8], shod and barefoot or striking habits [13], and unstable stimuli conditions [5], but are consistent among different foot types (flat, high and normal arch) [6]. Participants in this study were instructed to run with self-selected speed due to safety risks concerning the use of unstable shoes. *SS* exhibited higher speed and shorter contact time compared to *CS* and *SH*, respectively. These might partly be explained by the soft stimuli from the *SS*, which triggers the gripping function of toes (particular the hallux) during push-off phase. However, further investigation on muscle activities from EMG data may needed. But, the outcome is consistent with previous findings of shorter contact time with active toe function [13], which is in contrast to other studies that show longer contact time with slower speed when experiencing unstable perturbations [5].

The *SS* illustrates consistent lower extremity joint angle trends with the *CS* during the stance phase (Fig. 3). However, the hip rotation ROM in stance and peak knee extension angle in push-off phase of *CS* are greater than *SS*. This may be partly explained by the location of the unstable attached to the *SS*, which has been shown to induce compensatory movements and different muscle activity as with a previous study [5]. Furthermore, the ankle ROM in this study increased with respect to the *CS*, whereas another study showed reduced ankle ROM compared to *CS* [21]. The likely reason may be that in this study we used a regionalized unstable placement to the hallux but the previous study used a whole sole rocker. Moreover, the *SS* presents larger peak ankle plantarflexion [20] (Table 1), which also contributes to increased ROM from the windlass mechanism [14]. Another explanation is that the ankle, which is at a limbs length away from the centre of mass, presents greater kinematic excursion away from baseline, compared to proximal joints near the centre of mass [23].

The *SH* significantly alters the lower limb kinematic characteristics during stance compared to the *CS* and *SS*. During push-off, the *SH* leads to more unstable stimuli to the hallux region than the *SS* due to the higher elastic modulus, and exhibits different joint angles compared to the *CS* and *SS*. For the hip, larger peak flexion and smaller extension angle are presented among the rearfoot striking participants, which is a possible compensatory strategy for the stiff unstable perturbation in the push-off phase. The increased externally rotated position of the hip with the *SH* is the involuntary response of stiffer hallux unstable stimuli. A previous study reported the primary compensatory motion was at the distal joints (ankle and knee) compared to the proximal hip joint [23], however, this study found hip motion compensation was also significant. Consistent with previous studies [5], [23], the knee and ankle with the *SH* showed greater kinematic variations than the *CS* and *SS*, respectively. With *SH* peak knee flexion is greater, whereas peak knee abduction and internal rotation is decreased. During push-off, knee extension is reduced, whereas knee adduction and external rotation is increased. This is consistent with previously reported compensatory mechanisms leading to kinematics changes of the distal joint [5], [23] and potential protective strategies to reduce knee joint loading [10]. The ankle joint is significantly inversed and internally rotated for the *SH*, the likely consequence of passive adjustment to the regionalized unstable stimuli [5], [23], [24].

The plantar pressure distribution has been previously utilized to analyse foot function [6], [15], [16], and widely integrated into injury risk assessment [2], [4], [12], [14], [15], [17]. The plantar pressure showed no difference at the heel and mid foot regions since the unstable rocker structure was only attached to the hallux outsole. As hypothesized, the *SS* presents greatly reduced maximal force, peak pressure and force time integral to the medial forefoot, central forefoot and lateral forefoot regions. In contrast, the plantar loading increased at the big toe region. The toes gripping function may explain the higher pressure to the BT and reduction at the forefoot [9], and decreased loading to the metatarsals [2], [13], [14] due to increased load bearing area. The importance of the toes function is highlighted from deformation [4], [14], fatigue [17] and pathology [12] studies. Hence, the toe function has implications for training using the rocker to address pathology seen in neuropathology and ulceration risk.

With *SH*, the maximal force, peak pressure and force time integral remain consistent to the MFF and CFF regions (apart from the decreased peak pressure in CFF), and increase to the LFF and OT regions. The increased kinetics to the LFF and OT regions is linked with the increased ankle inversion and internal rotation. The force time integral (impulse) to the BT region increased greatly in the *SH* compared to *CS*, possibly indicating that the hallux is passively contacting the sole without active gripping motion. This has implications for risks of ankle sprain with the *SH*.

Several limitations concerning this study should not be ignored. Firstly, all participants were males, which was originally for the purpose of alleviating gender-related locomotion functional difference. Secondly, self-preferred running speed was instructed during the test, which initially aimed at presenting the individual performance from running different shoes without interference from controlled speed. Future studies will focus on the effect of walking and running with the hallux unstable shoes between genders and under controlled speed situations. The longitudinal training effect from hallux unstable shoes shall be considered.

5. Conclusion

This study proposed a novel unstable shoe design with unstable elements of different stiffness placed at the hallux. The primary objective was to analyse the biomechanical performance of running with hallux unstable shoes, aiming to potentially stimulate and increase the toe gripping function. Unstable shoes with soft material led to reduced medial metatarsal loading by increasing the support area and modified joint kinematics minimally. Unstable shoes with stiffer material presented compensatory kinematic movements across all joints and laterally shifted plantar loading distribution.

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Conflict of interests

The authors declare that no conflict of interest exist in this study.

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