Biomechanical locomotion adaptations on uneven surfaces can be simulated with a randomly deforming shoe midsole

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ABSTRACT

Background: A shoe with unsystematic perturbations, similar to natural uneven terrain, may offer an enhanced training stimulus over current unstable footwear technologies. This study compared the instability of a shoe with unpredictably random midsole deformations, an irregular surface and a control shoe-surface whilst treadmill walking and running.

Methods: Three-dimensional kinematics and electromyography were recorded of the lower limb in 18 active males. Gait cycle characteristics, joint angles at initial ground contact and maximum values during stance, and muscle activations prior to initial contact and during loading were analysed. Perceived stability, injury-risk and energy consumption were evaluated. Instability was assessed by movement variability, muscular activations and subjective ratings.

Results: Posture alterations at initial contact revealed active adaptations in the irregular midsole and irregular surface to maintain stability whilst walking and running. Variability of the gait cycle and lower limb kinematics increased on the irregular surface compared to the control across locomotion types. Similarly increased variability (coefficient of variation) were found in the irregular midsole compared to the control for frontal ankle motion (walk: 31.1 and 14.9, run: 28.1 and 11.6), maximum sagittal knee angle (walk: 7.6 and 4.8, run: 2.8 and 2.4), and global gait characteristics during walking only (2.1 ± 0.5 and 1.6 ± 0.3). Tibialis anterior pre-activation reduced and gastrocnemius activation increased in the irregular midsole compared to the control across locomotion types. During running, peroneus longus activation increased in the irregular midsole and irregular surface.

Conclusions: Results indicate random shoe midsole deformations enhanced instability relative to the control and simulated certain locomotion adaptations of the irregular surface, although less pronounced. Thus, a shoe with unpredictable instability revealed potential as a novel instability-training device.

Keywords: footwear; instability; kinematics; electromyography; lower-limb

1. Introduction

A relatively new concept of training footwear, termed unstable shoes are designed to create instability with the aim of providing functional benefits, such as increasing muscle activations and improving balance. Innovative shoe technologies developed for this purpose include rocker soles, balance pods and midsoles of multiple densities (Price, Smith, Graham-Smith, & Jones, 2013). The concept behind unstable footwear is similar to traditional instability training devices, such as Swiss balls, BOSU balls and wobble boards. Such equipment reduces the base of support causing instability, which can be observed through an increase of movement variability (Cimadoro, Paizis, Alberti, & Babault, 2013). The neuromuscular system has to make alterations to maintain stability and regular use is proposed to enhance balance and train the lower-limb muscles. A limitation of instability training devices is they are only utilised during restricted, isolated exercises and not during functional movements. Unstable shoes in contrast, may allow habitual training during walking, running or aerobic exercises.

Increased muscle activation is one of the acute responses of wearing unstable footwear. For the most frequently tested unstable shoe, Masai Barefoot Technology (MBT), increased tibialis anterior and peroneus longus activations have been commonly reported whilst standing (Buchecker, Pfusterschmied, Moser, & Müller, 2012; Nigg, Hintzen, & Ferber, 2006; Landry, Nigg, & Tecante, 2010) and increased gastrocnemius medialis activations whilst walking (Price et al., 2013; Romkes, Rudmann, & Brunner, 2006). As would be predicted, long-term wear strengthens and conditions the ankle muscles (Kaelin, Segesser, & Wasser, 2011). However, other studies report no significant increases in muscle activation whilst walking (Horsak & Baca, 2013; Nigg et al., 2006; Sacco et al., 2012; Stöggl, Haudum, Birklbauer, Murrer, & Müller, 2010) or running in unstable shoes (Sobhani et al., 2013).

Improved balance is another suggested training effect of regularly wearing unstable footwear. Increased centre of pressure excursion during static two-legged standing reduced over 6-weeks in healthy adults aged between 40 to 70 years old (Landry et al., 2010). The authors suggested this demonstrated improved static balance performance, but results from a dynamic systems perspective suggest this may not always be the case (van Emmerik, & van Wegen, 2000). A better determinant of the postural system’s ability could be assessing reactive balance after an external perturbation. Females older than 50 years old did improve their reactive balance over 8-weeks, but not significantly compared to a control group (Ramstrand, Thuesen, Nielsen, & Rusaw, 2010).

The inconsistent findings, particularly during dynamic locomotion may be due to the different number of participants, amount of pre-exposure to the unstable footwear and evaluation analyses applied. Another potential reason is the majority of previous research included active participants who were less likely to be affected by the unstable shoe instability during locomotion. Perhaps a more challenging unstable shoe construction would have a more pronounced effect. One such shoe design, Reflex Control has a thin sole bar along the longitudinal foot axis, compared to the Masai Barefoot Technology (MBT) shoe that has an anteroposterior sole rocker. Compared to barefoot walking, Reflex Control increased shank muscle activation, but no effect was found in MBT during walking (Schiemann, Lohrer, & Nauck, 2015). In addition, reactive balance during one-legged standing improved after a training program in Reflex Control, but not in MBT (Turbanski, Lohrer, Nauck, & Schmidtbleicher, 2011).

Moreover, although movement variability initially increases whilst walking in MBT shoes, this variability reduces after a 10-week training period (Stöggl et al., 2010). This suggests that instability becomes predictable, due to the cyclic repetitions during gait with the same fixed outsole stimulus of the MBT. Furthermore, Blair, Lake and Sterzing (2013) found initially increased vastus medialis activation whilst walking in an unstable shoe reduced to a similar level to a stable shoe after one hour, but tibialis anterior activation further increased. Trunk acceleration in the unstable shoe also tended to reduce after the hour walking. This suggests neuromuscular adaptations are learnt quickly and benefits of further training reduce over time.

Uneven natural terrain surfaces may provide a superior training modality by creating a continually changing and unpredictable instability. Increased muscle activations, a cautious gait pattern and increased movement variability has been found whilst walking (Gates, Wilken, Scott, Sinitski, & Dingwell, 2012; Marigold, & Patla, 2008; McAndrew, Dingwell, & Wilken, 2010; Sterzing, Apps, Ding, & Cheung, 2014a; Thies, Richardson, & Ashton-Miller, 2005; Voloshina, Kuo, Daley, & Ferris, 2013) and running on irregular surfaces (Sterzing, Apps, Ding, & Cheung, 2014b; Voloshina & Ferris, 2015). However, irregular surfaces are often not accessible in urban areas for convenient and frequent use. An alternative and novel solution would be to develop footwear that causes irregular and unpredictable instability. Kim and Ashton-Miller (2012) constructed experimental sandals with medial and lateral flaps in the sole, which could be deployed at random times to assess response to an unpredictable perturbation. Although the sandals were controlled electronically, which made them unsuitable for use by the general public. Consequently, we developed a training shoe with random irregular midsole deformations. The purpose of this study was to investigate the locomotion instability induced by this shoe compared to an irregular surface during walking and running.

Based on previous research, it was hypothesised the irregular midsole and an irregular surface would provide a similar, higher level of instability compared to a regular shoe-surface. This would be indicated by an increase in movement variability of the global spatial-temporal gait cycle characteristics and at the joint level, although this does not necessarily represent loss of stability. Moreover, there will be postural adjustments and increases in muscle activations to maintain balance. These hypothesises were applicable to both walking and running.

2. Methods

2.1. Participants

Eighteen active male sports science students, who were regular runners participated in this research (22.7 years ± 1.7, 177.2 cm ± 3.8, 69.1 kg ± 5.7). All participants had been injury free for at least 6 months prior to testing and had Brannock foot size male US 10.0 ± 0.5 (The Brannock Device Co., Liverpool, NY, USA). Liverpool John Moores University research ethics committee approved the study protocol and participants gave their written informed consent prior to testing.

As no previous data were available in the irregular midsole condition, a priori power analysis was performed on results of a previous study that compared the irregular treadmill surface condition to the regular treadmill surface (Sterzing et al., 2014a; Sterzing et al., 2014b) in G*Power software (Faul, Erdfelder, Lang, & Buchner, 2007). Kinematic variability of maximum sagittal and frontal ankle and sagittal knee angles during stance phase of walking and running (as used in this study) were tested. Across results a maximum of 13 participants were required to obtain an effect size of 0.75 (p value = .05, β = .20). Along with previous unstable footwear studies, this sample size was deemed appropriate for this study.

2.2. Shoe-surface Conditions

Three shoe-surface conditions were tested on a treadmill during walking and running:

1. A shoe with irregular midsole deformations and a regular treadmill surface (IM)
2. A regular shoe midsole and an irregular treadmill surface (IS)

3. The regular shoe midsole and regular treadmill surface as a control condition (CC)

Both shoe conditions had the same upper (Li Ning Fengchao TD, Li Ning Co, Beijing, size male US 10.0) while the two different midsole modifications were attached. The irregular midsole was created using three highly flexible rubber bags (hardness: 28 Asker C, thickness: 1.5 mm) attached to the shoe upper by Velcro at the rearfoot, midfoot and forefoot at 30%, 30% and 40% shoe length respectively. The segregation of foot regions is based upon previous biomechanical research (Cavanagh & Ulbrecht, 1994). The heel to toe offset was 10 mm unweighted, but due to the deformable bag material this reduced when wearing the IM shoe. In total, 51 ball bearings (12 mm diameter) and 10 cube shapes (height 15 mm, hardness: 85A Shore, TPU material) were placed inside the rubber bags and moved freely during swing phase of the gait cycle, creating a different shoe-surface profile at every ground contact and thus unpredictable perturbations. The ratio of ball bearings was 15:15:21 and cube shapes were 4:3:3 inside the rearfoot, midfoot and forefoot bags respectively.

The regular shoe midsole condition was developed with the midsole of the original shoe and used in CC and IS trials. The medio-lateral midsole shape was cut to identical dimensions of the IM shoe. Aluminium weights (5g) were glued evenly to replicate the weight of the IM bags (Fig 1). The regular shoe midsole weighed 234g and the irregular midsole shoe weighed 233g. Thus, weight and shape midsole differences were minimised. The heel to toe offset of the regular midsole was 10 mm.

Figure 1 near here

All walking and running trials were performed on a treadmill (Pro XL, Woodway Inc.,WI, USA). The treadmill belt slats were covered with Velcro strips (700mm x 58mm), which served as the regular surface. The irregular treadmill surface (IS) was created by randomly fixing 4 types of EVA dome shaped inserts (Ø: 140mm) of different height (10 and 15 mm) and hardness (40 and 70 Asker C) to the treadmill belt by Velcro attachment (Fig 2), as used in previous research (Sterzing et al., 2014a, 2014b). To eliminate visual targeting of foot placements, participants were instructed to look straight ahead. This was monitored by investigators, ensuring participants could not predict what they were to land on.

Figure 2 near here

2.3. Protocol

The treadmill speed was set at 5 km/hr for walking trials, as used in previous unstable footwear research (Nigg et al., 2006; Stöggl et al., 2010), and 8 km/hr for running trials. The slow run speed was selected to improve the level of comfort, as previously tested on IS (Sterzing et al., 2014b; Voloshina & Ferris, 2015). The order of shoe-surface conditions was arranged so CC trials were always first to avoid potential crossover effects from IM and IS, whose order was alternated between participants. Walking trials preceded running trials in the same shoe-surface condition. Before data collection participants were briefed about the testing conditions. After 60 seconds of walking and running in each shoe-surface condition to allow participants to get into a regular locomotion rhythm, biomechanical data were collected for 30 seconds. Surface EMG and lower limb kinematics were recorded synchronously from the subjects' left leg.

2.4. Kinematics

Kinematics were captured by a seven-camera motion analysis system at 300 Hz (Vicon Peak, Oxford, UK). Reflective markers were attached to the following locations to define the left thigh, shank and foot segments: The greater trochanter, medial and lateral femoral epicondyles, the lateral and medial malleoli, on the tip of the shoe and dorsal metatarsal heads 1 and 5. Tracking markers clusters were attached on the lateral side of the thigh (5 markers) and shank (4 markers), and to the shoe at the proximal posterior, distal posterior and lateral heel counter. Position and orientation of anatomical markers relative to tracking markers were determined from a static trial in the anatomical position in the regular shoe only, similar to the CAST technique (Cappozzo, Catani, Della-Croce, & Leardini, 1995). The same shoe upper was kept on throughout all trials allowing identical marker placement in all conditions, ensuring kinematic differences observed cannot be attributed to different marker location. Utilising a global neutral configuration is advantageous because the absolute angular differences between midsole conditions can be compared, which are not influenced by changes in the sole configuration.

After digitising, raw marker co-ordinate data were filtered using a low pass fourth order zero-lag Butterworth filter with cut-off frequencies of 10Hz for walking and 20Hz for running, based on visual inspection of the power spectrum. Stance phase was determined by ground contact algorithms which matched well against pilot data measurements with a foot switch placed inside the shoe-conditions on a treadmill and verified with a force plate. Vertical velocity change of the midpoint between the heel and toe markers identified gait events during walking (O'Connor, Thorpe, O'Malley, & Vaughan, 2007) and the vertical acceleration of the heel and tip of shoe markers was used during running (Maiwald, Sterzing, Mayer, & Milani, 2009). Some kinematic data were not collected successfully due to technical issues and are excluded from subsequent analyses. Kinematic results are based on 16 participants for walking and 17 for running.

Characteristics of the gait cycle were derived from ground contact times. Positive sagittal knee and ankle angles reflect joint flexion, and positive frontal ankle angle represents eversion. To show preparatory posture adaptations shoe-surface and joint angles were calculated at initial contact. We expected the unpredictable instability to have a greater effect during loading occurring in the first half of stance. Therefore, maximum joint angles and ankle ranges of motion between initial contact and maximum positive angles during stance were determined. The single largest ankle inversion angle of all steps between initial ground contact and maximum eversion angle was recorded to indicate any outliers that were obscured when looking at the variability through the standard deviation.

2.5. Surface Electromyography

Surface electromyography (EMG) was recorded the left gastrocnemius medialis, peroneus longus, tibialis anterior, bicep femoris, vastus medialis and vastus lateralis muscle activations using a wireless telemetric system (TeleMyo DTS, Noraxon Inc., Scottsdale, AZ, USA) at 3 kHz. Pre-gelled bi-polar Ag/AgCl circular electrodes (Tian run, Beijing, China) of 10mm diameter and inter-electrode spacing of 25mm were positioned according to international recommendations (SENIAM). To reduce impedance, hair was shaved and skin cleaned with ethanol. The analogue signal was converted to a digital signal by a 16-bit transmitter data acquisition system. Certain electrode data contained artefacts and were excluded from subsequent analyses. After exclusion the number of subjects per muscle for walking and running respectively contained: gastrocnemius medialis (N=14, 15), peroneus longus (N=12, 13), tibialis anterior (N=9, 10), bicep femoris (N=14, 15), vastus medialis (N=13, 15) and vastus lateralis (N=11, 16).

The EMG data were processed in Visual 3D software together with the kinematic data (C-Motion, Rockville, MD, USA). The raw signal was digitally band-pass filtered using a bidirectional 4th order Butterworth filter with cut-off frequencies of 10 and 300Hz, full wave rectified and smoothed using an 11-point root mean square moving average filter. In subsequent analysis, EMG values were normalised to the average peak value of each muscle during the gait cycle of CC trials of the same locomotion type. The normalised mean value was calculated in a pre-activation phase (150ms before initial contact) and a loading phase (from initial contact until maximum knee flexion) to supplement kinematic variables.

2.6. Subjective Perception Assessment

Immediately after biomechanical data collection, subjective perception of the level of stability, injury risk, and energy consumption were collected while participants were still walking or running on the treadmill. Prior to data collection, variables were defined to participants, with the instructor explaining their perceived level of magnitude (low, high) rather than their interpretation (good, bad) was being assessed. Participants assessed all variables verbally from a large 9-point Likert scale (1-very low, 3-low, 5-moderate, 7-high and, 9-very high, with other numbers not denominated) mounted in front of the treadmill, (Fig 2) (adapted from Au & Goonetilleke, 2007; Lam et al., 2013; Sterzing et al., 2014c). This method is advantageous because participants can think solely about the perception variable whilst walking and running (Sterzing et al., 2014a; Sterzing et al., 2014b).

Statistics

All steps (41.0 ± 2.6 for running and 28.6 ± 1.5 for walking) were analysed to compute the mean of all variables for each participant. Variability of gait cycle and kinematic variables were calculated with the coefficient of variation (CV). The CV was calculated by dividing the standard deviation by the mean and multiplying by 100. The CV can be useful for determining the relative magnitude of variability when there are differences in mean readings, but is limited if the mean value is close to zero (James, 2004).

All statistical processing was performed in SPSS (v22, SPSS Inc, Chicago, IL, USA). Normality of data were checked using the Shapiro-Wilk test and visually verified for outliers with boxplots. Most variables followed parametric assumptions and a one-way repeated measures ANOVA, with Bonferroni adjusted post hoc tests were applied to define differences between shoe-surface conditions for walking and running ($p<.05$). The non-parametric Friedman test with Bonferroni adjusted Wilcoxon post hoc tests were applied to the variables with outliers ($p<.05$). Missing data were deleted listwise, as it were the only option available in SPSS. This meant always the same number of participant mean variables were compared.

3. Results

3.1. Kinematics

Differences to mean kinematic results were generally small between conditions but consistent across participants whilst walking (Table 1) and running (Table 2). The gait cycle in IM was characterised by shorter, thus more frequent steps. Variability of gait was significantly increased in IS compared to CC, with the difference being greater in running (Fig 3). During walking IM had rather higher variability similar to IS (26 ± 14% > CC), whereas during running IM had rather lower level of variability similar to CC (3 ± 2% > CC).

Figure 3 near here

At initial ground contact, knee flexion increased in IM compared to IS and CC whilst walking and running. Shoe-surface angle was flattest in IM during walking, and flatter in IM and IS compared to CC during running. Variability of parameters at initial ground contact tended to be greatest in IS across participants and locomotion (Fig 4). Ankle angle variability could not be computed due to mean values ranging around zero. Therefore, the standard deviation is reported separately in Supplementary Table 1 and 2, to give an indication of ankle angle variability.

Figure 4 near here

During stance, maximum ankle eversion reduced in IM whilst walking and running (Fig 5). Sagittal ankle range of motion reduced whilst walking and frontal ankle range of motion reduced whilst running in IM compared to CC and IS. The largest ankle inversion angles recorded were no different between IM and IS during locomotion. During walking, CC had a significantly reduced maximum inversion angle compared to IM and IS ($p = .005$; IM = 11.5 ± 6.1°, IS = 10.1 ± 7.1, CC = 5.9 ± 3.1) but no different during running ($p = .008$; IM = 11.1 ± 4.8, IS = 9.3 ±4.7, CC = 8.7 ± 3.4). Variability of parameters during stance were largely more variable in IS, with IM having similar variability levels of frontal ankle range of motion (walk: 109% > CC, run 143% > CC) and knee flexion (walk: 60%> CC, run: 19% > CC) (Fig 4, Fig 5) across locomotion.

Figure 5 near here

Table 1 near here

Table 2 near here

3.2. Electromyography

Electromyography results showed differences mostly occurred in the shank muscles for both walking (Table 3) and running (Table 4). Tibialis anterior activation significantly reduced during pre-activation and loading in IM whilst walking compared to CC and IS. During pre-activation whilst running, tibialis anterior activation significantly reduced in IM and IS compared to CC. Peroneus longus activation significantly increased during loading in IM and IS compared to CC, and during pre-activation in IS compared to CC whilst running. The gastrocnemius medialis had significantly greater pre-activation in IM than CC during walking and running.

Table 3 near here

Table 4 near here

3.3 Perception

Subjective ratings results showed IM was perceived the least stable, with IS less stable than CC for walking and running. Injury risk level was perceived greatest in IM and greater in IS than CC for walking and running. Energy requirement was perceived greater for IM and IS than CC during walking and running (Table 5).

***Table 5 near here**

4. Discussion

This study compared the instability caused by both a shoe and surface exhibiting irregular perturbations during treadmill walking and running. Biomechanical instability were assessed by changes in movement variability of the spatial-temporal gait cycle and lower limb kinematics, as well as, muscle activations. Whether participants could also perceive changes to instability were also assessed. Results confirmed our hypothesis that the irregular midsole shoe (IM) and irregular surface (IS) increased biomechanical and subjectively perceived instability compared to a regular shoe-surface (CC). Similarly increased variability of frontal ankle motion and maximum knee flexion for both walking and running were found between IM and IS, indicating a comparable, higher level of instability compared to CC. This suggests IM could provide an enhanced training shoe to active consumers, over current unstable footwear technologies, by creating instability in an unpredictable manner similar to IS. Other adaptations were dependant on the type of locomotion or the different stimuli of IM or IS.

Consistent with previous research on uneven surfaces, IM trials triggered increased stride frequency and reduced step length (Marigold & Patla, 2008; McAndrew et al., 2010; Voloshina et al., 2013), reduced shoe-surface angle (Marigold & Patla, 2002; Menant et al., 2008) and increased knee flexion (Gates et al., 2012; Thomas & Derrick, 2003) at initial contact in both walking and running. Shorter steps and a reduced sagittal shoe-surface angle reduce the risk of slipping by decreasing the shear forces and consequently reducing the friction coefficient at the shoe-floor interface (Menant et al., 2008). Increased knee flexion would help to lower the centre of mass, increasing stability (MacLellan & Patla, 2006). These active posture adaptations at initial contact in IM suggest a cautious locomotion pattern was adopted (Menant et al., 2008; Marigold & Patla, 2002). Stability was subjectively perceived lowest in IM, giving further evidence the level of instability was enough to induce these cautious posture alterations. Similar cautious kinematic adaptations at initial contact were found in IS during running, but not walking. This may be due to injury risk of the IS stimuli being subjectively perceived greater in running than walking, and enough to induce a cautious gait strategy.

The higher maximum ankle inversion across all steps and more variable frontal ankle motion in IM and IS compared to the control (Fig 5) were caused by the size, shape and hardness of the materials imposed between the shoe-surface interfaces. This may have caused the greater perceived instability and injury risk. However, this does not mean they were more dangerous to participants. Increased ankle inversion is not a risk factor for ankle sprain in healthy participants whilst running (Willems, Witvrouwa, Delbaere, De Cock, & De Clercq, 2005). Also, the maximum ankle inversion angle was within the normal range of frontal ankle motion (Ottaviani, Ashton-Miller, Kothari, & Wojtys, 1995). Keeping ankle range of motion within this safe range is an advantage of the IM shoe compared to a natural irregular terrain that imposes a greater risk and could cause injury. Thus, the irregular midsoles provide a similar stimulus to an IS, which is not always available or safe to use, and offer a viable alternative.

The increased gait cycle variability in IM and IS during walking, and IS during running is an indicator of instability and has been linked to risk of falling (Moe-Nilssen & Helbostad, 2005; Thies et al., 2005). Previous research also found increased variability of step length and step time on IS (Gates et al., 2012; Marigold & Patla, 2008; McAndrew et al., 2010; Thies et al., 2005; Voloshina et al., 2013; Voloshina & Ferris 2015). However, the increased gait cycle variability does not necessarily represent loss of balance, but rather active alterations to maintain stability to the unpredictable perturbations, allowing the acquisition of more flexible locomotion patterns. The reason for variability being higher in IM during walking than running is related to the reduced shoe-surface angle (walking = 16.6°, running = 12.4°). Reducing the angular displacement of the shoe to the ground likely reduced the perturbation effect whilst running in IM, enabling a more regular locomotion pattern. How to increase the variability whilst running in IM to a similar level as the IS should be considered in the design of future prototypes.

The increased lower-limb kinematic variability in IS and IM has also been reported previously on irregular surfaces during walking (Gates et al., 2012; Sterzing et al., 2014a; Voloshina et al., 2013) and running (Sterzing et al., 2014b; Voloshina & Ferris 2015) and, walking in unstable shoes (Stöggl et al., 2010). According to Dynamics Systems Theory, opposed to the more global movement level, increasing variability at the joint/segment level is associated with functional benefits and not necessarily related with reduced stability (Li, Haddad, & Hamill, 2005). Performance can be achieved consistently through a variety of movement pathways, increasing adaptability to perturbations (Davids et al., 2006; Latash, 2012; Wilson, Simpson, van Emmerik, & Hamill, 2008). There is some evidence to suggest this also reduces the risk of chronic overuse injuries in running because the stresses are spread more evenly over the soft tissues (Hamill, van Emmerik, & Heiderscheit, 1999). In this respect, we propose IM offers wearers another training benefit, in addition to those discussed already, of improving the level of this functional joint variability. Whether the level of functional variability remains high, or reduces to the level of a regular shoe, as reported previously (Stöggl et al., 2010), warrants further investigation.

Electromyography results revealed few common activation strategies to the irregular shoe-surfaces. One prevalent approach to IS and IM was to increase the peroneus longus activation during the loading phase of running. The peroneal muscles are the main muscles to provide eccentric control to protect against lateral ankle sprains (Ashton-Miller, Ottaviani, Hutchinson, & Wojtys, 1996). Therefore, it appears the increased peroneus longus activation was a mechanism to control the increased inversion and more variable frontal ankle motion of IM and IS. With training, this would increase the peroneus muscle strength and reduce the risk of ankle sprains, as found in conventional unstable shoes (Kaelin et al., 2011). The perceived risk of injury and energy requirement were lower walking compared to running in IM and IS, similar to previous research on IS (Sterzing et al., 2014a; Sterzing et al., 2014b). This may relate to the lack of increased peroneus longus activation during walking in IM and IS compared to running. However, some participants increased the peroneus longus activation whilst walking in IM and IS, suggesting individual adaptation strategies for coping with the constraints occurred, as referred to previously (Apps, Ding, Cheung, & Sterzing, 2014). The other common finding was a reduced tibialis anterior activation on the irregular shoe and surface conditions, particularly in IM whilst walking. This result supports previous observations on irregular surfaces (Hettinga, Stefanyshyn, Fairbairn, & Worobets, 2005; Voloshina et al., 2013), and in unstable shoes (Nigg et al., 2006) and is associated with the reduced shoe-surface angle at initial contact.

This research is subject to certain limitations. The use of set speeds on a treadmill, has been shown to affect variability compared to when subjects run at their preferred speed (Sekiya, Nagasaki, Ito, & Furuna, 1997) and overground (Wheat, Milner, & Bartlett, 2004). However, we do not expect that this would have affected any of the conditions differently and confounded our conclusions. The time to accommodate to the shoe-surface conditions was limited to 60 seconds, so the results reported only apply to the acute responses. It is likely adaptations would change after the initial accommodation period, as previously reported (Stöggl et al., 2010; Blair et al., 2013). Furthermore, although the irregular treadmill surface developed did provide continuous unpredictable perturbations, it was limited by the size, hardness and shape of inserts attached and would not have provided the same variety of perturbations as a natural uneven terrain. In IM trials, participants could perceive the objects inside the rubber bags under the plantar sole which may have caused the kinematic adaptations, rather than the instability. Future prototypes should aim to reduce this haptic sensation.

5. Conclusion

In conclusion, we have created a novel shoe that provides continuously random perturbations. The motivation for developing such a shoe was to have a more challenging stimulus than existing unstable footwear, thus providing greater functional training benefits. This shoe successfully increased biomechanical and perceived instability relative to a stable shoe and simulated certain adaptations of an unpredictable irregular surface during walking and running. An additional training benefit of the irregular midsole, of increasing the functional level of joint kinematic variability is proposed, which aligns with the dynamics systems perspective. Future studies should confirm these suggested training advantages over unstable shoes, by assessing the adaptability to unpredictable perturbations after regular use.

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Table 1. Mean (SD) gait cycle parameters and kinematics during walking across participants.

Walking	Variable	CC	IM	IS	ANOVA p-value	Post hoc result
Gait cycle	Stance time [secs]	.63 (.04)	.62 (.04)	.65 (.02)	.010	IS > IM
	Swing time [secs]	.38 (.02)	.36 (.02)	.38 (.02)	<.001	IS, CC > IM
	Step length [m]	.87 (.05)	.86 (.05)	.90 (.03)	.010	IS > IM
	Stride frequency [stride/min]	59.4 (3.2)	61.3 (3.1)	58.5 (1.9)	<.001	IM > IS, CC
Kinematics at initial contact	Shoe-surface [°]	24.7 (4.3)	18.6 (4.8)	22.8 (5.0)	.001	CC, IS > IM
	Ankle dorsiflexion [°]	0.9 (3.0)	-1.1 (4.0)	-0.4 (3.6)	.161	---
	Ankle inversion [°]	-3.3 (3.1)	-3.9 (3.0)	-2.5 (3.9)	.028	IM > IS
	Knee flexion [°]	14.5 (5.7)	20.1 (7.1)	16.9 (5.6)	<.001	IM > CC, IS
Kinematics during stance	Ankle dorsiflexion MAX [°]	7.0 (3.1)	8.0 (3.1)	7.8 (3.7)	.248	---
	Ankle eversion MAX [°]	7.3 (2.1)	5.3 (5.5)	8.7 (3.6)	.005	IS > IM
	Sagittal ankle ROM [°]	17.6 (4.5)	12.5 (4.8)	18.7 (4.0)	<.001	CC,IS >IM
	Frontal ankle ROM [°]	10.6 (3.7)	10.6 (3.3)	11.8 (2.2)	.128	---
	Knee flexion MAX [°]	31.2 (7.5)	33.6 (8.4)	32.1 (7.2)	.038	---

MAX = maximum, ROM = Range of motion

Table 2. Mean (SD) gait cycle parameters and kinematics during running across participants.

Running	Variable	CC	IM	IS	ANOVA p-value	Post hoc result
Gait cycle	Stance time [secs]	.35 (.02)	.34 (.01)	.34 (.02)	.014	CC >IS
	Swing time [secs]	.39 (.04)	.38 (.04)	.40 (.04)	.018	IS > IM
	Step length [m]	.77 (.04)	.75 (.03)	.75 (.04)	.011	CC> IS
	Stride frequency [stride/min]	82.2 (3.5)	84.3 (4.4)	82.2 (3.8)	.001	IM > CC, IS
Kinematics at initial contact	Shoe-surface [°]	16.4 (2.5)	12.5 (3.0)	12.9 (3.8)	< .001	CC > IM, IS
	Ankle dorsiflexion [°]	6.7 (3.1)	6.1 (0.4)	5.0 (3.9)	.017	CC > IS
	Ankle inversion [°]	-5.7 (3.4)	-6.1 (3.4)	-4.6 (4.4)	.530	---
	Knee flexion [°]	22.2 (4.2)	28 (4.4)	26.9 (4.1)	< .001	IM > IS > CC
Kinematics during stance	Ankle dorsiflexion MAX [°]	13.6 (2.9)	16.2 (4.0)	13.5 (3.5)	< .001	IM > CC, IS
	Ankle eversion MAX [°]	9.2 (3.5)	4.1 (7.5)	9.6 (5.2)	< .001	CC, IS > IM
	Sagittal ankle ROM [°]	16.6 (1.9)	17.0 (2.0)	17.2 (2.3)	.439	---
	Frontal ankle ROM [°]	14.9 (3.0)	11.1 (4.3)	14.4 (2.9)	.001	CC,IS > IM
	Knee flexion MAX [°]	48.6 (4.3)	48.4 (4.8)	49.7 (4.7)	.006	IS > CC, IM

MAX = maximum, ROM = Range of motion

Table 3: Normalised mean (SD) electromyography data during pre-activation and loading phases across participants during walking

Muscle	Phase	CC	IM	IS	ANOVA p-value	Post hoc result
Gastrocnemius Medialis	Pre-activation	1.8 (1.2)	4.4 (3.4)	3.1 (3.2)	.008	IM>CC
	Loading	4.2 (2.1)	5.2 (2.6)	4.2 (1.9)	.263	---
Tibialis Anterior	Pre-activation	18.7 (5.8)	11.8 (5.6)	15.0 (6.9)	.004	CC, IS>IM
	Loading	19.2 (3.8)	9.1 (4.1)	18.2 (6.0)	<.001	CC, IS>IM
Peroneus Longus	Pre-activation	4.7 (2.0)	5.1 (2.6)	6.4 (2.5)	.113	---
	Loading	9.3 (3.7)	14.0 (5.8)	13.5 (5.6)	.062	---
Bicep Femoris	Pre-activation	27.2 (3.6)	23.0 (9.2)	22.9 (5.5)	.005	CC>IS
	Loading	12.1 (4.7)	13.4 (7.4)	12.1 (4.9)	.484	---
Vastus Medialis	Pre-activation	14.4 (6.5)	14.1 (7.6)	13.6 (6.8)	.843	---
	Loading	28.2 (5.4)	28.8 (9.7)	29.9 (8.4)	.699	---
Vastus Lateralis	Pre-activation	10.2 (4.5)	9.0 (5.2)	8.8 (4.7)	.307	---
	Loading	29.1 (5.6)	23.5 (7.5)	23.8 (6.9)	.030	CC>IS

Table 4: Normalised mean (SD) electromyography data during pre-activation and loading phases across participants during running

Muscle	Phase	CC	IM	IS	ANOVA p-value	Post hoc result
Gastrocnemius Medialis	Pre-activation	2.3 (1.7)	3.5 (3.2)	2.9 (2.6)	.039	IM>CC
	Loading	21.3 (4.8)	20.8 (6.4)	19.2 (5.6)	.234	---
Tibialis Anterior	Pre-activation	24.1 (3.5)	10.6 (8.2)	12.6 (5.6)	<.001	CC>IM,IS
	Loading	10.4 (4.2)	10.4 (7.0)	15.5 (15.8)	.301	---
Peroneus Longus	Pre-activation	4.3 (1.5)	7.0 (5.2)	6.9 (3.8)	.018	IS>CC
	Loading	24.0 (5.4)	30.8 (10.0)	34.6 (22.2)	.023	IM,IS>CC
Bicep Femoris	Pre-activation	24.3 (5.3)	24.1 (12.2)	21.4 (7.9)	.420	---
	Loading	10.6 (5.2)	10.5 (6.6)	9.9 (3.8)	.803	---
Vastus Medialis	Pre-activation	8.7 (2.9)	8.8 (2.9)	8.8 (2.6)	.963	---
	Loading	31.8 (3.2)	28.3 (6.0)	31.5 (7.3)	.069	---
Vastus Lateralis	Pre-activation	6.6 (3.2)	6.6 (2.4)	6.9 (3.7)	.752	---
	Loading	29.5 (4.9)	26.4 (8.3)	29.6 (15.4)	.144	---

Table 5: Subjective perception scores (Mean (SD)) during walking and running across participants

Variable	Locomotion	CC	IM	IS	ANOVA p-value	Post hoc result
Stability	Walk	5.6 (1.2)	2.9 (1.2)	4.2 (1.4)	<.001	IM<IS<CC
	Run	5.4 (1.6)	2.7 (1.2)	3.8 (1.6)	<.001	IM<IS<CC
Injury risk	Walk	3.2 (1.3)	6.3 (1.1)	5.8 (1.5)	<.001	IM>IS>CC
	Run	3.7 (1.3)	6.8 (1.4)	6.0 (1.6)	<.001	IM>IS>CC
Energy Consumption	Walk	3.1 (1.4)	4.6 (1.5)	4.7 (1.4)	<.001	IM, IS>CC
	Run	4.9 (0.9)	6.5 (1.3)	6.3 (1.4)	<.001	IM, IS>CC

Fig 1. The regular and irregular shoe midsoles. The regular midsole (left, top) was removed from the original shoe upper and cut into same width as IM bags (left, middle), weights attached (left, bottom). The irregular midsole shoe (right, top), the rubber midsole bags (right middle) with cubes and ball bearings placed inside (close up: bottom right). © 2013. All rights reserved (Sterzing et al., 2013 (Li Ning Sports Goods Co. Ltd, China)).

Fig 2. The regular and irregular treadmill surface. The regular treadmill surface covered with strips of Velcro (top left) and the irregular treadmill surface, created by attaching 4 kinds of EVA inserts to the belt via Velcro (top right). Data collection of an IS run trial, the large 9-point Likert scale allowed scores to be taken whilst participants were still on the treadmill (bottom).

Fig 3. Variability (CV) of gait cycle parameters across participants. 1 = significantly greater than CC, 2 = significantly greater than IM, 3 = significantly greater than IS ($p<.05$). Notice IM has higher values similar to IS during walking and lower values similar to CC during running.

Fig 4. Variability (CV) of joint/segment angles at initial contact (IC) and during stance across participants. ROM = range of motion. 1 = significantly greater than CC, 2 = significantly greater than IM ($p<.05$).

Fig 5. Joint angle plotted against stance phase during walking and running across subjects. Solid thick lines represent mean sagittal ankle angle (top), frontal ankle angle (middle) and sagittal knee angle (bottom). CC illustrated by the black line, IM the lighter line and IS the lightest line (mostly overlaid by CC). Shaded areas (CC, IM) and dotted lines (IS) illustrate mean intra-subject variability at each percentage of stance phase from 0% at heel-strike to 100% at toe-off.