Unpredictable shoe midsole perturbations provide an instability stimulus to train ankle posture and motion during forward and lateral gym lunges

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Biomechanical responses to shoe instability during lunge movements

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Unstable footwear may enhance training effects to the lower-limb musculature and sensorimotor system during dynamic gym movements. This study compared the instability of an unstable shoe with irregular midsole deformations (IM) and a control shoe (CS) during forward and lateral lunges. Seventeen female gym class participants completed two sets of ten forward and lateral lunges in CS and IM. Ground reaction forces, lower-limb kinematics and ankle muscle activations were recorded. Variables around initial ground contact, toe-off, descending and ascending lunge phases were compared statistically ($p<.05$). Responses to IM compared to CS were similar in forward and lateral lunges. The IM induced instability by increasing the vertical loading rate ($p< .001$, $p = .009$) and the variability of frontal ankle motion during descending ($p = .001$, $p< .001$) and ascending phases ($p = .150$, $p = .003$), in forward and lateral lunges respectively. At initial ground contact, ankle adjustments enhanced postural stability in IM. Across muscles, there were no activation increases, although results indicate peroneus longus activations increased in IM during the ascending phase. As expected, IM provided a more demanding training stimulus during lunge exercises and has potential to reduce ankle injuries by training ankle positioning for unpredictable instability.

Keywords: footwear; instability; kinematics; electromyography; gym training

Introduction

Instability training devices, such as Swiss balls, wobble boards and foam pads are commonplace in gyms as they enhance core muscle strength and balance (Cosio-Loma et al., 2003; Anderson et al., 2013). Unstable shoes (US) apply this same principle, and cause instability by design features also including a reduced base of support, as well as, softer materials within the midsole. Specifically, they have proven highly marketable for females (Dierick et al., 2017). Proposed neuromuscular training effects from US include increased lower-limb muscle activations and enhanced balance (Nigg et al., 2012). Yet, not all previous studies report increased muscle activations during gait (Sacco et al. 2012; Stöggl, et al., 2010) or balance enhancements after regular wear (Ramstrand et al., 2010).

Exercise classes are a female dominated activity (Apps et al., 2015), which frequently include closed-kinetic chain movements requiring minimal equipment, and train multiple joints and muscle groups (Cordova et al., 1999). These functional exercises are beneficial because they are applicable to daily life and sports, requiring strength, flexibility and balance. Additionally, they allow clinicians to screen for movement control (Cook et al., 2006; Kritz et al., 2009). The difficulty of functional exercises can be adapted to the individual's ability or training aim. For example, lunges can be simplified for populations who may be at risk of falling, such as the elderly (Flanagan et al., 2004). Moreover, the hip or ankle musculature can be specifically targeted by selecting forward or lateral lunge directions (Rieman et al., 2013; Flanagan et al., 2004). Functional exercises on instability devices further destabilise the sensorimotor system, and are incorporated in advanced balance training programmes. This is reported to increase trunk muscle activations (Anderson et al., 2013), which improves core stability with regular training (Cosio-Limo et al., 2003), and increase frontal plane ankle motion variability (Strøm et al., 2016). Nairn and colleagues (2017) highlighted instability training is location dependent, with lower-limb motion and muscle activations changing in response to a distal perturbation but not a proximal perturbation. This concept led to the development of therapeutic US technologies specifically for the rehabilitation of ankle injuries, as they enable more ecological training (Page, 2006; McKeon et al., 2008). Sandals with a hemisphere-shaped sole under the midsole have been used for this purpose. During functional exercises these sandals were reported to increase shank muscle activations (Blackburn et al., 2003) and improve single-leg balance after regular training (Michell et al., 2006). Recently developed devices provide the perturbation underneath the subtalar joint. They similarly increase shank muscle activations in healthy participants whilst walking (Donovan et al., 2014), walking with jumps (Fautrelle et al., 2017) and in participants with ankle instability during functional balance tasks (Donovan et al., 2015). However, short-term enhancements to strength and balance in patients with chronic ankle instability were no different between those who trained with ankle destabilising devices and control shoes (Donovan et al., 2016).

An innovative US with irregular midsole deformations (IM) provided a more demanding training stimulus that required different ankle and knee joint stability whilst walking and running compared to a commercial US (Apps et al., 2016; Apps et al., 2017). It has not been investigated whether IM may provide a beneficial instability-training stimulus during lunges. Therefore, the purpose of this study was to compare the biomechanical and neuromuscular adaptations during forward and lateral lunges in IM compared to a regular shoe in female gym class attendants. Based on previous instability training studies (Behm & Anderson, 2005; Strøm et al., 2016; Blackburn et al., 2003) and our walking and running investigations (Apps et al., 2016; Apps et al., 2017), it was hypothesised:

(1) IM will induce instability, which will result in increased and more varied ground reaction force loading rates and increased lower-limb movement variability.
(2) This will be controlled by kinematic adjustments to enhance stability, particularly around initial ground contact and toe-off, and increasing activation of muscles about the ankle joint.

Methods

Participants

Seventeen healthy female students who regularly attended gym classes for at least one year were recruited from Beijing Sports University (21.6 ± 1.6 years, 166.3 ± 4.2 cm, 55.6 ± 3.5 kg, 20.9 ± 0.9 BMI, gym class experience 3.3 ± 2.0 years, classes 6.3 ± 1.9 hours/week). Liverpool John Moores University research ethics committee approved the study protocol and all participants were informed about procedures prior to signing consent forms. All participants were self-reported injury free for at least 6 months at the time of testing and had Brannock foot size female US 8.0 ± 0.5 (The Brannock Device Co., Syracuse, NY, USA).

Shoe conditions

Two shoe midsole conditions were tested: an irregularly deforming midsole (IM) to provide unpredictable instability and the regular cross training shoe midsole with a flat outsole (Figure 1). An IM was developed to provide unpredictable instability. It was created with three highly flexible rubber bags (hardness: 28 Asker C, thickness: 1.5 mm) and placing freely moving ball bearings (12 mm diameter: stiff material) and cube shapes (height 15 mm, hardness: Shore A 85, TPU material) inside. The length of the rubber bags varied to cover the rearfoot, midfoot and forefoot shoe sole regions at 30%, 30% and 40% of the shoe upper length respectively. Inside the rubber bags over the rearfoot, midfoot and forefoot regions there were 14, 13 and 15 ball bearings and 3, 2 and 2 cube shapes, respectively. This created different irregular midsole deformations during each foot placement (Apps et al., 2016).

The control shoe (CS) midsoles were cut to be the same width and weight as IM (IM: 218g, CS 215g) by attaching aluminium (5g) weights. An advantage of these shoe modifications is the same shoe upper (Li Ning Fengchao TD, Li Ning Co, Beijing, size female US 8.0) stays on throughout testing and the different midsole condition is attached by Velcro. This enabled identical reflective marker placement during testing in all conditions.

Figure 1 near here

Protocol

Participants completed two sets of ten right leg forward lunge repetitions and two sets of ten right leg lateral lunge repetitions in each shoe condition. During all lunges participants placed their hands on their iliac crests, looked straight ahead and kept their trunk erect, as variation of these were reported to affect lunge biomechanics (Farrokhi et al, 2008). For the forward lunge, participants started with legs shoulder width apart and took a right step forward. Then, flexed their right knee until about 90° with the right thigh approximately parallel to the ground, and the back left leg lowered towards the floor (Figure 2). Following knee flexion, they extended their right knee to push back to the starting position. For the lateral lunge participants started in the same position and laterally stepped right. Then, flexed their right knee until the right shank was in a vertical position over the right foot. They were asked to prevent the right knee moving forward anteriorly whilst keeping the left leg extended (Figure 2). After maximal knee flexion, the right knee extended to push back and return to the start position. One lunge step was completed every 3 seconds dictated by a metronome beat at 40 beats per minute to control the frequency (2 beats per lunge). The enforced movement rate resulted in 10 lunges performed per 30 seconds. Participants lunged to their preferred step length and width. The lunging technique was verbally explained and demonstrated to participants. Before each test condition participants sufficiently practiced the lunge technique and speed.

Figure 2 near here

Data acquisition and processing

All biomechanical measurements were synchronised and only collected for participants’ lunge leg. Variables were selected to analyse the preparations for initial ground contact (during the last 100 ms prior to ground contact), the descending phase (initial contact until maximum knee flexion), and the ascending phase (maximum knee flexion until toe-off) of the lunges. Measurements of the distal lower-limb were made based on previous research revealing greater influence occurring in closer proximity to the perturbation stimulus (Nigg et al., 2006; Price et al., 2013; Apps et al., 2016; Nairn et al., 2017).

Ground reaction forces

Ground reaction forces (GRF) from participants’ lunge leg were collected with a force plate (90 x 90 cm, AMTI OR6GT, Watertown, MA, USA) flush with the laboratory floor, sampling at 1500 Hz. Lunge ground contact was determined using a 20 N threshold. The analogue signals were filtered by a 4^{th} order Butterworth filter with frequency cut-off of 50 Hz. Ground reaction forces were normalised to bodyweight. Loading rate was computed as the slope between adjacent frames on the vertical GRF. To assess instability during initial loading, the maximum and variability (coefficient of variation (CV)) of the loading rate during the descending phase was calculated.

Kinematics

A seven-camera motion capture system (Vicon Peak, Oxford, UK), sampling at 300 Hz, recorded three-dimensional kinematic data. Reflective markers attached to the lunge leg were tracked in six degrees of freedom, according to the CAST technique (Cappozzo et al., 1995). The local coordinate system of the thigh was defined at 8 cm medially to the greater trochanter proximally and the mid-point of the femoral epicondyles distally. This definition has been shown to give accurate sagittal knee kinematics (Sinclair et al., 2014). The local coordinate system of the shank was defined as the mid-point of the femoral epicondyles proximally and the mid-point of the malleoli distally. The shoe segment was a virtual segment with the same coordinate system as the shank to ensure the ankle angle was at zero degrees in the static trial. Tracking marker clusters were attached on the lateral side of the right thigh (4 markers) and shank (4 markers) on a rigid plate, and to the shoe at the proximal posterior, distal posterior and the lateral heel counter. Additionally, a marker was placed on the distal posterior heel counter of the left shoe. Due to the exact same marker placement in both shoe conditions (see shoe conditions), neutral positions and orientations of anatomical markers relative to tracking markers were determined from a static trial in the CS only. A global neutral configuration is beneficial because it allows comparing the absolute angular differences between midsole conditions. Marker coordinate data were filtered with a 4^{th} order, zero lag Butterworth digital filter with a 10 Hz cut-off frequency.

To assess adaptations to the overall lunge movements, step length, step width and ground contact time were calculated. Step length was defined as the anteroposterior distance and step width as the mediolateral distance between the distal heel markers at initial contact for the forward lunge. This was switched for the lateral lunge to ensure the stepping direction always corresponded to step length. Posture at initial ground contact and toe-off were measured by the shoe-surface angle, sagittal and frontal ankle angle, and sagittal knee angle. Shoe-surface angle was computed in the sagittal plane for forward lunges and the frontal plane for lateral lunges, to correspond to the movement direction. Lower-limb movement variability (CV) of the sagittal and frontal ankle, and sagittal knee ranges of motion in the descending and ascending lunge phases were computed. Ankle range of motion was calculated using maximum dorsiflexion angle as indicator for separating the descending and ascending phases, not knee flexion angle. This was due to peak dorsiflexion occurring earlier in the ground contact phase (Mean ± SD forward lunge: 48.9±3.6%, lateral lunge: 51.1±3.3%) than peak knee flexion (Mean ± SD forward lunges: 53.5±2.5%, lateral lunges 52.9±2.6%).

Surface electromyography

Surface electromyography of the tibialis anterior, peroneus longus, gastrocnemius medialis muscles was recorded with a wireless telemetric system (TeleMyo DTS, Noraxon Inc., Scottsdale, AZ, USA), sampling at 3 kHz. The electrodes were pre-gelled bi-polar Ag/AgCl circular electrodes (Tian run, Beijing, China) of 10 mm diameter with an inter-electrode spacing of 25 mm. Skin was shaved, abraded and cleaned with an alcohol wipe to reduce impedance. Muscles were located and electrodes placed parallel to the muscle fibres according to SENIAM international standards (Hermens et al., 2000).

The myoelectric signals were digitally band-pass filtered with a bi-directional 4th order Butterworth filter (cut-off frequencies: 10 and 400 Hz) and full wave rectified. A linear envelope was created by applying a 61-point moving average filter after visual inspection of the signals revealed this smoothed the data sufficiently without losing the true peaks and troughs. To reduce inter-subject variation, EMG data for each muscle were normalised to the average peak value, across analysed phases, for each muscle of CS trials for both forward and lateral lunges. This has been applied in previous unstable shoe studies (Romkes et al., 2006; Buchecker et al., 2012) and has been demonstrated good reliability and sensitivity during running, but it is unknown if this also the case during lunges (Albertus-Kajee et al., 2011).

The mean amplitude was calculated in the following periods: pre-activation (the 100 ms before initial contact), the descending phase and the ascending phase. Certain electrode data contained artefacts and were excluded from subsequent analyses. After exclusion, the number of participants (N) per muscle was: gastrocnemius medialis = 15, peroneus longus =15, tibialis anterior = 15.

Statistics

For all variables in both lunge types and shoe conditions, the average magnitude across all 20 lunges were computed for each participant for statistical analyses (SPSS Inc, Chicago, IL, USA). To verify parametric assumptions were met, data were checked using the Shapiro-Wilk test and visually verified with boxplots (Ghasemi & Zahediasl, 2012). Repeated measures multivariate analysis of variance (rMANOVA) tests were performed on the forward and lateral lunge data separately to determine significant differences between shoe conditions. To test hypothesis (1) separate rMANOVA tests were applied to the magnitude and variability of vertical loading rates (2 x 2), the joint range of motion variability in descending phase, as well as, the ascending lunge phase (2 x 6). To test hypothesis (2) separate rMANOVA tests were applied to temporal-spatial parameters (2 x 3), kinematics at initial ground contact (2 x 4), kinematics at toe-off (2 x 3), muscle activations during pre-activation (2 x 3), muscle activations during the descending phase (2 x 3), and muscle activations during the ascending phase (2 x 3). Significant results ($p < .05$) were followed up with simple univariate tests with Bonferroni adjusted p-values to control for multiple comparisons. To further indicate the magnitude of any univariate differences, effect size was calculated using Cohen's d (Cohen, 1988). Values of 0.2, 0.5 and 0.8 are considered as small, moderate and large effect sizes, respectively.

Results

Temporal-spatial characteristics

There was a significant difference to the overall forward lunge movement between CS and IM ($F_{(3,14)} = 4.21$; $p = .026$; $\eta^2 = .47$). Univariate follow-up tests revealed this was caused solely by an increased ground contact time in IM compared to CS (Table 1). There was no overall change observed in the lateral lunges between shoe conditions ($F_{(3,14)} = 2.24$; $p = .128$; $\eta^2 = .33$) (Table 1).

Table 1 near here

Ground reaction forces

The rMANOVA tests revealed increased instability in IM compared to CS from the vertical GRF loading rates in forward ($F_{(2,15)}$ = 41.79; p<.001; η^2 =.85) and lateral ($F_{(2,15)}$ = 16.20, p<.001; η^2 =.68) lunges. Univariate follow-up tests indicated this was due to both an increased maximum magnitude and an increased variability of the vertical GRF loading rate for forward and lateral lunges (Table 2).

Table 2 near here

Kinematics

There were significant posture alterations to the lunge leg at initial ground contact during forward ($F_{(4,12)}$ = 12.01; p < .001; η^2 =.80) and lateral lunges ($F_{(4,12)}$ = 10.40; p = .001; η^2 =.78) (Table 3). Univariate follow-up tests revealed, across lunge type, this was due to reduced shoe-surface angles, reduced ankle dorsiflexion and increased ankle inversion in IM. In addition, during forward lunges there was increased knee flexion in IM. There were also significant posture alterations to the lunge leg at toe-off during forward ($F_{(3,14)}$ = 34.34; p < .001; η^2 =.88) but not lateral lunges ($F_{(3,14)}$ = 0.63; p = .610; η^2 =.118) (Table 3). During the forward lunge, univariate tests indicate this was caused by increased ankle plantarflexion and inversion in IM (Figure 3), as well as, reduced knee flexion. There were no significant univariate test results in the lateral lunge. Participants had a plantarflexed ankle at toe-off in IM and CS in the lateral lunge (Figure 4).

Table 3 near here

Figure 3 near here

Figure 4 near here

There were significant differences in the variability (CV) of joint ranges of motion between shoe conditions for the forward ($F_{(6,11)} = 4.16$; $p = .020$; $\eta^2 = .69$) and lateral lunges ($F_{(6,11)} = 11.47$; $p < .001$; $\eta^2 = .86$). During the descending phase, frontal ankle variability increased in IM during forward and lateral lunges (Table 4). During the ascending phase, frontal ankle variability also increased in the lateral, but not the forward lunges. Sagittal ankle variability increased in IM during forward lunges in the descending phase, but not lateral lunges. No sagittal knee differences were observed.

Table 4 near here

Surface Electromyography

There were no significant differences in shank muscle activation levels in the forward lunges (Table 5) during the pre-activation ($F_{(3,12)}$ = 3.29; p = .058; η^2 =.45), descending ($F_{(3,12)}$ = 0.77; p = .532; η^2 =.16) or ascending phase ($F_{(3,12)}$ = 3.19; p = .063; η^2 =.44) between shoe conditions. No significant shank muscle activation differences were observed either in the lateral lunges (Table 5) during the pre-activation ($F_{(3,12)}$ = 1.33; p = .311; η^2 =.25), descending ($F_{(3,12)}$ = 2.19; p = .142; η^2 =.35) or ascending phase ($F_{(3,12)}$ = 3.36; p = .055; η^2 =.46) between shoe conditions.

During the ascending phase in both lunge types, and during pre-activation in the forward lunges rMANOVA p-values bordered on conventional levels of statistical significance ($0.1< p >.05$). This was due to 12 out of 15 participants having an increased peroneus longus activation in the ascending phase during forward lunges and lateral lunges in IM (Figure 5). Individual analysis revealed greater ankle plantarflexion angle at toe-off was correlated with higher gastrocnemius medialis activation in the ascending phase (Figure 6).

Table 5 near here

Figure 5 near here

Discussion

This study compared the instability caused by an unstable shoe with irregular midsole deformations (IM) to a regular gym shoe (CS) during forward and lateral lunges in female gym class attendants. To assess this, temporal-spatial characteristics, ground reaction forces, lower-limb kinematics and ankle muscle activations were measured. Results confirmed our first hypothesis: IM induced greater instability; observed by greater and more varied vertical GRF loading rates and increased variability of frontal plane ankle motion. The kinematic responses corroborated our second hypothesis; postural adjustments enhanced stability at initial ground contact across lunge types and at toe-off in the forward lunge. Gastrocnemius medialis and peroneus muscle activations tended to increase in the ascending phase, positioning the foot and stabilising the ankle for push-off. These findings have practical implications for training footwear designs for advanced balance training.

Prevalence of ankle sprains is reduced through balance training that progresses to using functional exercises on a balance board (McGuine & Keene 2006). Shortly after initial ground contact, when IM induced increased and varied loading, non-contact ankle sprains injuries often occur (Blackburn et al., 2003; Fong et al., 2009). Thus, learning to control the IM instability through regular training could be incorporated in ankle injury prevention programs. The advantage of US, and particularly the IM tested here, over instability training devices is they are convenient because they do not require certain positioning for users of different abilities. Moreover, they allow continuous rather than intermittent training during walking and other functional movements whilst they are worn. Yet, if training effects of US are enhanced compared to current instability devices is unclear

(Donovan et al., 2015). The IM provides unpredictable perturbations, which is advantages over US that cause predictable perturbations. Thus, training with IM is more likely to reduce ankle injuries because they are caused by unexpected perturbations, although this claim warrants investigation.

Unlike past studies, we did not instruct participants to take a stride as long as comfortable (Escamilla et al., 2008), or impose a specific step length (Riemann et al., 2012). Instead, participants were free to choose any preferred step length, applicable to exercising at a gym. This resulted in step lengths for forward and lateral lunges being shorter in comparison to previous research (Riemann et al., 2013; Escamilla et al., 2008). Despite this, the only difference to the overall lunge movements observed between footwear conditions was a 4 ms longer contact time in IM compared to CS during forward lunges (Table 1). This is important because it suggests participants' ability to perform the functional lunge movements was not inhibited by the IM. Lower-limb kinematic adaptations during forward and lateral lunges suggest a cautious posture was implemented at initial ground contact and toe-off to mediate the effects of the IM stimulus. At initial ground contact participants had a reduced sagittal and frontal shoe-surface angle in IM, during forward and lateral lunges respectively. This strategy has been shown to reduce the risk of losing balance by reducing the braking impulse at the shoe-floor interface (Marigold & Patla, 2002). If this adapted foot position can be learnt and applied to sports with unpredictable instability it would reduce risk of slipping. The plantarflexed and inverted positioning of the ankle in IM were responsible for this flatter foot adaptation.

By optimising the musculoskeletal system mechanical energy expenditure is reduced (Roy & Stefanyshyn, 2006). The cautious posture adopted for initial ground contact in IM is an example of this strategy, as muscle activations were largely similar to CS during pre-activation and the descending phase. However, monitoring mechanical energy expenditure would be needed to support this theory. The peroneus longus muscle activations increased in most participants during the ascending phase across both lunge types (Figure 5), although the rMANOVA result was not significant across all muscle groups. This is in agreement with previous research demonstrating muscles closer to the instability stimulus increase activation level (Nairn et al., 2017). Increased peroneus longus activation helps to control the frontal ankle motion variability and stabilise the ankle for toe-off. In future research, it is recommended to focus on the response of the peroneal muscles with instability training, as weaker evertor muscles are linked causing ankle injury (Willems et al., 2002). Moreover, our female gym class participants were highly trained in performing functional movements. The IM may elicit a greater training effect on the peroneal muscles of participants who are less trained or have weaker ankles. Gabriel et al. (2008) found females had reduced ankle stiffness during push-off, which was related to their reduced strength and proprioception. They recommended females use training programs to improve their contractile capabilities of the ankle during push-off during gait, a purpose IM being suitable for.

Ascending phase gastrocnemius medialis activations increases in IM during forward lunges were related to a plantarflexed ankle at toe-off (Figure 6). This can be assumed as a stability strategy to prevent the centre of pressure moving posteriorly across the unstable objects in IM. In the lateral lunges, ankle plantarflexion occurred at toe-off in both shoe conditions indicating this is a more stable posture and there are reduced margins of stability in this lunge direction.

Figure 6 near here

A limitation of this study is that only the immediate responses to IM whilst performing lunges were measured. Familiarisation was limited to the time taken for participants to adopt the proper lunge technique. However, longer time to accommodate or habitual use of IM during gym classes may result in different adaptations. Furthermore, the haptic sensation of the objects inside the IM bags that could have altered the biomechanical response in the lunge movements and is suggested to be removed from future prototypes.

Conclusion

The shoe with irregular midsole deformations provided a more challenging stimulus during forward and lateral lunges than a regular cross-training shoe. The instability was evident from the increased, varied loading and frontal ankle joint variability. Optimising the musculoskeletal system by adopting a cautious posture at initial ground contact resulted in few muscle activation increases during this phase in the irregular midsole. The irregular midsole may offer additional benefits over current instability devices and footwear used for ankle injury prevention and rehabilitation because these do not provide the unpredictable perturbations that cause them. Future research should investigate the longer-term neuromuscular adaptations of gym exercises in unstable footwear on ankle movement control and peroneal muscle conditioning for injury prevention training.

Disclosure of interest

The authors report no conflict of interest.

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Table 1. Mean (SD) temporal-spatial parameters in the irregular midsole shoe (IM) and control shoe (CS) during forward lunges and lateral lunges

		CS	IM	Effect size	Significance
Forward lunge	Stance time [ms]	1.18 (.08)	1.22 (.10)	.77	IM>CS, p = .018
	Step length [m]	.724 (.10)	.718 (.08)	.19	p > 1.00
	Step width [m]	.100 (.04)	.098 (.04)	.08	p > 1.00
Lateral lunge	Stance time [ms]	1.24 (.13)	1.27 (.11)	.36	p = .489
	Step length [m]	.701 (.08)	0.681 (.08)	.64	p = .075
	Step width [m]	.072 (.05)	0.075 (.04)	.05	p > 1.00

Table 2. Maximum and variability (CV) of the vertical ground reaction force loading rate across participants (Mean (SD)), in the irregular midsole shoe (IM) and control shoe (CS) during forward and lateral lunges

	Vertical load rate	CS	IM	Effect size	Significance
Forward lunge	Maximum (Bw/sec)	15.6 (6.6)	23.8 (11.1)	1.33	IM>CS, $p < .001$
	Variability (CV)	22.7 (5.1)	33.8 (8.2)	1.37	IM>CS, $p < .001$
Lateral lunge	Maximum (Bw/sec)	27.9 (11.0)	35.0 (15.1)	.80	IM>CS, $p = .009$
	Variability (CV)	21.1 (5.5)	29.4 (6.4)	.87	IM>CS, $p = .005$

Table 3. Mean (SD) kinematic posture at initial contact and toe-off in the irregular midsole shoe (IM) and control shoe (CS) during forward and lateral lunges

Lunge, Phase		CS	IM	Effect size	Significance
Forward lunge, Initial contact	Sagittal shoe-surface [°]	36.2 (4.8)	30.2 (4.8)	1.34	CS>IM, $p < .001$
	Sagittal ankle [°]	12.9 (4.9)	9.0 (6.1)	.90	CS>IM, $p = .012$
	Frontal ankle [°]	6.4 (3.3)	7.8 (3.4)	.82	IM>CS, $p = .020$
	Sagittal knee [°]	35.9 (7.2)	38.5 (6.6)	1.03	IM>CS, $p = .002$
Lateral lunge, Initial contact	Frontal shoe-surface [°]	-25.7 (4.6)	-23.3 (3.5)	.84	IM>CS, $p = .018$
	Sagittal ankle [°]	17.2 (4.4)	14.5 (4.6)	1.53	CS>IM, $p <.001$
	Frontal ankle [°]	7.0 (4.8)	9.0 (4.7)	.81	IM>CS, $p = .022$
	Sagittal knee [°]	34.2 (4.7)	33.7 (5.2)	.27	$p > 1.00$
Forward lunge,	Sagittal ankle [°]	3.1 (9.4)	-8.0 (14.6)	1.01	IM>CS, $p = .002$

Toe-off	Frontal ankle [°]	2.5 (2.9)	6.6 (3.8)	1.75	IM>CS, p <.001
	Sagittal knee [°]	39.2 (6.5)	37.0 (6.7)	.82	CS>IM, p = .011
Lateral lunge, Toe-off	Sagittal ankle [°]	-21.5 (10.0)	-23.0 (8.4)	.23	p >1.00
	Frontal ankle [°]	10.7 (3.5)	10.0 (4.4)	.17	p >1.00
	Sagittal knee [°]	31.7 (5.9)	30.7 (5.7)	.23	p >1.00

Positive sagittal joint angles represent flexion and positive frontal ankle angles inversion.

Table 4. Joint range of motion variability (CV), expressed as mean (SD), in the irregular midsole shoe (IM) and control shoe (CS) during forward and lateral lunges

		Phase	CS	IM	Effect size	Significance
Forward lunge	Sagittal ankle [°]	Descending	14.0 (4.5)	17.0 (5.9)	1.00	IM>CS, p=.006
		Ascending	15.3 (5.3)	19.7 (9.1)	.42	p = .600
	Frontal ankle [°]	Descending	21.4 (7.3)	31.0 (6.7)	1.00	IM>CS, p = 0.006
		Ascending	29.8 (12.3)	37.4 (9.6)	.60	p = 0.150
	Sagittal knee [°]	Descending	4.8 (1.6)	5.5 (1.8)	.39	p = .744
		Ascending	5.9 (1.5)	6.0 (0.8)	.07	p > 1.00
Lateral lunge	Sagittal ankle [°]	Descending	16.6 (6.1)	15.1 (4.6)	.22	p > 1.00
		Ascending	13.6 (10.7)	8.3 (7.1)	.47	p = .426
	Frontal ankle [°]	Descending	15.3 (4.2)	31.3 (10.2)	1.53	IM>CS, p < .001
		Ascending	20.6 (6.8)	28.8 (8.6)	.84	IM>CS, p = .018
	Sagittal knee [°]	Descending	4.6 (1.7)	5.0 (1.2)	.27	p > 1.00
		Ascending	5.2 (1.8)	5.6 (1.6)	.17	p > 1.00

Table 5. Mean (SD) normalised muscle activation magnitudes in the irregular midsole shoe (IM) and control shoe (CS) during the forward and lateral lunge phases.

	Phase	Muscle	CS	IM	Effect Size	Significance
Forward lunge	Pre-activation	GM	8.6 (6.4)	11.0 (9.0)	.59	p = .113
		PL	13.8 (11.0)	14.2 (10.1)	.10	p > 1.00
		TA	22.2 (8.8)	21.4 (10.8)	.13	p > 1.00
	Descending	GM	12.0 (4.4)	12.7 (4.8)	.27	.955
		PL	24.4 (7.5)	25.1 (7.4)	.26	.995
		TA	28.7 (6.4)	27.8 (8.5)	.18	p > 1.00
	Ascending	GM	12.0 (6.9)	15.7 (8.8)	.58	p = .121
		PL	19.7 (13.7)	23.8 (13.7)	.83	IM>CS, p = .019
		TA	21.1 (3.8)	20.9 (4.7)	.06	p > 1.00
Lateral lunge	Pre-activation	GM	9.4 (7.3)	10.8 (11.1)	.24	p > 1.00
		PL	13.8 (8.8)	14.5 (8.5)	.21	p > 1.00
		TA	31.4 (12.6)	28.6 (11.5)	.33	p = .669
	Descending	GM	13.0 (5.8)	13.2 (5.8)	.06	p > 1.00
		PL	20.2 (10.3)	21.3 (9.5)	.37	p = .508
		TA	33.6 (8.3)	30.4 (8.2)	.60	p = .106
	Ascending	GM	21.5 (5.7)	24.9 (9.3)	.53	p = .180
		PL	26.1 (8.2)	29.0 (8.1)	.82	IM>CS, p = .020

	TA	21.3 (6.0)	20.1 (5.4)	.34	p = .642

GM = Gastrocnemius Medialis, PL = Peroneus Longus, TA = Tibialis Anterior