THE BIOMECHANICAL AND NEUROMUSCULAR RESPONSES TO SHOE-SURFACE INSTABILITY

Charlotte Lucy Apps

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ABSTRACT

Unstable shoes are a unique category of footwear that provide a training stimulus during casual wear. Reported therapeutic benefits have led to their commercial success, although enhancing muscle activations and balance, normally associated with instability training devices like wobble boards, are not as effective. Therefore, the aim of this thesis was to develop and evaluate an unstable shoe that provides continually unpredictable perturbations. The notion was that a more challenging, and varied instability, alike uneven terrain, would provide an enhanced training stimulus.

The first study investigated if the developed shoe with irregular midsole deformations (IM) was more demanding, alike uneven terrain, it was compared to an irregular surface (IS) and a stable shoe-surface during treadmill walking and running. Generally, IM increased instability compared to the control, demonstrated by a more cautious gait pattern and posture at initial contact, and subjectively perceived as less stable whilst walking and running. Certain responses in IM were similar to IS, such as the increased variability of frontal ankle motion and maximum sagittal knee angle in stance phase. This is proposed to benefit IM wearers by improving the level of functional joint variability. The second study assessed if the varied instability of IM would be more de-stabilising than an unstable shoe (US) by comparing joint stability strategies during walking and running. Results revealed joint stiffness reorganisations between the ankle and knee in IM and US during loading. Further joint stiffness re-distributions and increased ankle co-contraction were found in IM compared to US, indicating additional adaptations are required for stability. The third study revealed IM may similarly be beneficial in gym training, as adaptations were also triggered in forward and lateral lunge movements to maintain stability. Increased gastrocnemius medialis and peroneus longus activations were required for the push-off phase in lunges, which related to ankle position. Investigating short-term training effects after regular IM

wear was not feasible with the prototype developed, so instead a 6-week IS walking intervention was undertaken, as similar responses were found to IM during locomotion in the first study. Results showed no significant improvements to ankle muscle strength and postural balance compared to a control group, who did not undertake a training intervention.

This research revealed IM provided an innovative stimulus that increased instability compared to an US and simulated certain responses to an IS. Unpredictable instability provided by footwear may have potential use for injury prevention and rehabilitation interventions, but future work needs to assess which populations it benefits.

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CHAPTER 1.

Introduction

Since running became a popular pastime in the 1970's, biomechanics has helped to develop and evaluate functional footwear. Various shoe constructions and material can be modified and their effect on performance, injury and comfort evaluated (Sterzing et al., 2012). In order for sporting goods manufacturers to be competitive, they strive to keep developing innovative products that benefit and appeal to consumers. Early footwear prototypes need to be tested to validate they are providing the desired functions. Biomechanical and subjective perception assessments are direct approaches that are used to reveal the effects on human performance. Prior to the development of prototypes, fundamental scientific studies enable an in-depth and advanced understanding of human movement, which provides the logical reasoning behind inventing a new footwear technology (Sterzing et al., 2012). This basic research often takes longer than permitted to company researchers who are occupied with routine footwear testing of commercial prototype cycles. Collaborating with academic institutes has offered a solution to this problem. This project is an example of such collaboration, where biomechanical research underpins an original footwear concept.

Traditionally footwear is developed with a flat, level outsole to provide stability. The stabilising role of the shoe results in the small ankle muscles being under-utilised and possibly even weakened (Nigg 2005; Landry et al., 2010). Conditioning all muscles enables joint stability to be maintained across different positions and helps to prevent imbalances, which may be a risk factor for injury (Nigg et al., 2009; Baumhauer et al., 1995). Instability training devices, such as wobble boards, are used to train the stabilising muscles which reduces injury rates (McGuine & Keane, 2006; Hrysomallis, 2007). This concept led to the development of a new type of functional footwear: unstable shoes. Unlike traditional footwear, they are specifically designed to induce instability. Rather than having to devote specific time for training, like instability training devices, unstable footwear are beneficial because they allow training through casual wear in daily activities.

Masai Barefoot Technology (MBT) was the first brand of unstable footwear developed in 1996. The shoe is characterised by an anteroposterior rocker shaped outsole and soft, flexible material under the rearfoot. The company claim they give the feel of walking on natural, soft, uneven ground and promote instability. Positive therapeutic effects associated with MBT include reducing joint pain (Armand et al., 2014; Nigg et al., 2006b), changing posture (New & Pearce, 2007), and increasing blood circulation (Sousa et al., 2012). This has led to the brand becoming increasing popular, with over 1 million pairs sold per year across 20 countries reported in 2008 ("MBT shoes", 2015). Other proposed training effects include conditioning the lower limb musculature and enhancing balance, similar to instability training devices (Nigg et al., 2012). However, not all studies demonstrate MBT increase lower-limb muscle activations during gait (Nigg 2006a, Stöggl et al., 2010; Sacco et al 2012), or improve balance after regular wear (Turbankski et al., 2011).

The success of MBT resulted in other footwear companies developing different models of unstable footwear. Various innovative shoe technologies have been developed to create instability: rocker soles, balance pods and midsoles of multiple densities (Price et al., 2013). Yet, unlike MBT, which has many published studies providing evidence of their effectiveness, other unstable shoe brands lack peer-reviewed research (Price et al., 2013). Specifically, New Balance, Skechers and Reebok have been sued over false claims their toning shoes increase muscle activations, burning more calories (Hines, 2012). This signifies the importance of establishing footwear functions with scientific evidence. A possible reason for this is the instability technologies do not provide a challenging enough stimulus for healthy participants who do not have muscle weaknesses or balance deficiencies. Alternatively, adaptations to unstable footwear may be learnt quickly after initial wear and any additional muscular activations or kinematic adjustments are not required to maintain balance.

Thus, a gap was identified for an innovative footwear modification, to have unpredictable instability. This is based on the notion that current unstable footwear becomes predictable due to the fixed outsole and does not de-stabilise the sensorimotor system (Stöggl et al., 2010). A shoe-surface interface with varied perturbations should therefore provide a greater training stimulus. The first step to validating this concept and create a new footwear product was to establish how humans move over irregular surfaces, which exhibits such continual unpredictable perturbations. Once this basic research had validated the concept, our focus shifted to how to incorporate this effect into footwear. The journey of developing a surface and novel footwear design with continual unpredictable instability is detailed in this thesis. The biomechanical and neuromuscular adaptations during human movement are explored in the experimental chapters. This evidence is used to advise if such a technology may provide superior training effects over existing unstable footwear.

CHAPTER 2.

Literature Review

2.1. Introduction

This chapter reviews the literature of biomechanical and neuromuscular adaptations to instability at the shoe-surface interface. The review is not exhaustive but aims to give an overview of each area. The layout is organised into three main sections:

1. Ways in which instability at the shoe-surface interface has been used as a training modality are reviewed, including: instability training devices and unstable footwear. Despite advantageous training effects identified, there is a gap in current commercial instability training devices that provides a challenging enough stimulus for the healthy population during daily activities. This provided rationale for the development of a shoe with unpredictable instability with the intention of simulating natural irregular surfaces.
2. Increased movement variability is identified as one of the outcomes of using instability training devices. The second section discusses the functional role of movement variability, which allows adaptations to shoe-surface perturbations. There is a lack of instability training research that quantifies the different types of variability, providing insight into the neuromuscular control. This dynamics systems perspective is applied to the developed irregular midsole shoe in Chapter 5.
3. The biomechanical and neuromuscular strategies for coping with shoe-surface instability are discussed. Although humans adapt to irregularities at the ground the vast majority of previous biomechanical research has been on level stable surfaces. Only recently have advances been made as to how we negotiate instabilities at the shoe-surface interface, and there is still a lack of knowledge of how stability is maintained at the joint level. Information of these control mechanisms will improve understanding of potential training effects with shoe-surface instability.

The aim and objectives of the thesis are stated, based on the current limitations and lack of knowledge identified in instability training.

2.2. Instability Training Devices

Constraints define the margins which human motion must function within (Glazier & Davids, 2009) and movement is selected to seek the best solutions within those boundaries. Environmental constraints include the surrounding surface interfaces that we come into contact with (Newell, 1986). Manipulating constraints can allow or even require new movement patterns to be acquired. The devices we interact with can therefore be modified to change our biomechanics to promote performance, reduce injury risk or enhance rehab.

Instability training devices have been developed as a type of environmental constraint. They utilise unstable support surfaces that have either a reduced base of support (top row: Figure 2.1), or a soft deforming material (middle row: Figure 2.1), or a combination of both (bottom row: Figure 2.1). This destabilises the postural control system which has to find alternative movement solutions to maintain stability. This process of advanced self-organisation can be observed by an increase in movement variability and muscle activation (Cimadoro et al., 2013).


Figure 2.1. Example instability training devices. A: wobble board, B: balance board, C: foam roller, D: dyna disc, E: BOSU ball dome side up, F: balance pods, G: BOSU ball dome side down, H: swiss ball.

These devices are increasingly popular and are commonplace in gyms due to the superior training effects achieved during functional aerobic exercises, without needing external mechanical loads. The most popular reason for training with instability training devices is to enhance core muscle strength and coordination (Anderson et al., 2013). This has been shown to improve trunk stability (Stanton et al., 2004). Improved performance is then achieved when returning to stable conditions, for greater speed, power and strength can be utilised (Behm & Colado, 2012).

The increased muscle activations of the lower-limb muscles reported is also proposed to condition these muscles. For example, Behm and Anderson (2005) found higher soleus muscle activations performing a squat on unstable discs, compared to a stable squat condition. On the other hand, Wahl and Behm (2008) found no lower-limb or trunk muscle activations differences between an unstable dyna disc and a stable floor whilst performing several lower limb exercises, including forward and lateral lunges. This was likely because the participants were highly experienced resistance trainers and made subtle kinematic compensations to maintain stability. A greater degree of instability was suggested for trained individuals. The ankle evertors muscles can also be trained, helping to prevent lateral ankle sprains. Higher peroneus longus activation helps control the increased frontal plane ankle variability on various instability training devices (Strøm et al., 2016). McGuine and Keene (2006) found reduced lateral ankle injuries in school sports teams that completed a balance training program compared to those that did not. The intervention progressed to incorporating sports movements (throwing, catching a ball) and squats whilst on a balance board.

One limitation with instability training devices is they are only used during restricted, isolated exercises, and not during routine movement patterns. Unstable shoes in contrast, can be worn habitually during walking, running, or aerobic exercises. Rather than needing to position external equipment, unstable shoes are convenient to apply to closed-kinetic chain functional movements.

2.3. Unstable footwear

2.3.1. Introduction

Unstable footwear (US) mimics the concept of instability training devices and provides ‘fixed instability’ from the uniform outsole shape. Innovative shoe technologies have been introduced to induce instability including: rocker soles, balance pods and midsoles of multiple densities (Price et al., 2013). Figure 2.2. illustrates some of the various brands.

Specific design features predispose different models to affect sagittal or frontal plane stability. For example, the US brand, Masai Barefoot Technology (MBT), has an anteroposterior rocker shaped outsole, affecting sagittal plane motion (Figure 2.2: A).

The aim of this section is to assess the research evidence of the claimed training effects. Biomechanical adaptations to aid stability in unstable footwear designs are addressed in Section 2.5.2. As recognised, a key benefit of unstable footwear is they allow casual training through daily activities, and therefore the effects during dynamic movements will be the focus.

Figure 2.2. Examples of unstable shoe brands. A: MBT, B: Skechers Shape Ups, C: New Balance True Balance, D: FitFlop, E: Reebok Easy Tone, F: Li Ning Bubble Gym, G: Reflex Control Schuh, H: ApoSTherapy and I: Balance sandals.

2.3.2. Lower-limb muscle activations

Increasing muscle activations is one of the claimed benefits of US. This acute (immediate) response, is proposed to strengthen the muscles with regular wear, but only one conference abstract has demonstrated this for the ankle muscles (Kaelin et al., 2011). If US do condition muscles this would be beneficial because wearing regular stable footwear may weaken the foot muscles over time because they are not needed to be activated as much to maintain stability (Nigg et al., 2006a). The most common method of assessing muscle activation is surface electromyography (EMG). One of the problems with this technique is that only the larger superficial muscles can be recorded. Nigg and colleagues (2012) proposed that MBT shoes have a greater training effect on the smaller ankle joint muscles. This has only been demonstrated for standing balance using a specially designed EMG circumferential linear array and quantifying the muscle activations by wavelet intensities (Landry et al., 2010). Results found, even after regularly wearing MBT for 6-weeks, flexor digitorum longus and the anterior compartment group muscle activations remained increased compared to a control shoe and barefoot condition.

Other research studies on locomotion have relied on EMG to measure activations in the larger lower-limb muscles. There is conflict in the literature as to whether US increases the activation level of these muscles. The majority of papers have investigated MBT, with certain studies reporting no significant increases in muscle activation whilst walking or running in a rocker shoe (Table 2.1). On the other hand, increased gastrocnemius, vastus medialis, vastus laterialis, rectus femoris (Romkes et al., 2006) and peroneus longus activations (Price et al., 2013) has been found whilst walking in MBT in healthy participants during parts of the gait cycle (Table 2.1). Buchecker and colleagues (2012a) found increased vastus lateralis and gastrocnemius medialis activations in MBT during later phases of stance in overweight males. One common muscular activation strategy whilst walking in MBT is a reduced tibialis anterior activation (Romkes et al., 2006; Forghany et al., 2014; Sacco et al., 2012), often around heel strike.

Table 2.1. Studies that investigated lower-limb muscle activations in unstable footwear during locomotion. Those that found no increases in muscle activations effect in grey.

Author (year)	(n)	Unstable shoe(s)	Walk/ Run	EMG Variables	Results (p<.05)
Nigg et al. (2006a)	6	MBT	Walk	Wavelet analysis	N/A
Romkes et al. (2006)	12	MBT	Walk	RMS during phases of stance	Tibialis anterior↓ in loading, Gastrocnemius↑ 1st half of stance, Vastus medialis/lateralis↑ 2nd half, Rectus femoris↑ midstance,↓IC
Stöggl et al. (2010)	12	MBT	Walk	RMS, integral and frequency during stance	N/A
Buchecker et al. (2012a)	10	MBT	Walk	Mean amplitude during phases of stance	Vastus lateralis↑ mid/terminal stance, Gastrocnemius↑ terminal stance
Sacco et al. (2012)	25	MBT	Walk	Peaks and integrals	Tibialis anterior↓
Horsak & Baca (2013)	12	Reebok Easy Tone	Walk	Mean amplitude during phases of stance	N/A
Sobhani et al. (2013)	16	Modified rocker shoe	Walk/ Run	Peaks	Walk: Tibialis anterior↑ Run: N/A
Price et al. (2013)	15	MBT, FitFlop, Reebok Easy-Tone, Skechers Shape-Ups	Walk	RMS during phases of stance	MBT: Tibialis anterior↓, Peroneus Longus and Gastrocnemius↑ loading, All other shoes: Peroneus Longus↑ pre-swing
Forghany et al. (2014)	20	MBT, modified rollover shoe	Walk	Peaks and integrals	MBT/rollover: Tibialis anterior IC peak↓ Soleus/Gastrocnemius↑ in loading
Balbinot et al. (2015)	10	Reebok Easy Tone	Walk	Mean amplitude during 1st and 2nd half of stance	Rectus femoris ↑in 1st and ↓in 2nd, Vastus lateralis ↓ in 1st and ↑ in 2nd

↑/↓ = significant increase/decrease compared to a control shoe. RMS = Root mean Square. IC = initial contact.

Only very recent research has investigated muscle activation patterns in other US brands. Price and colleagues (2013) compared EMG of lower-limb muscles in several US models whilst walking. In contrast to MBT, FitFlops, Reebok Easy Tone and Sketchers Shape-Ups increased peroneus activity during pre-swing. This is related to these US having medio-lateral instability from shaped or softer forefoot materials which influence push-off (Figure 2.2: D-F). Two studies investigated the effect of walking in Reebok Easy Tone, with one reporting changes in muscle activations (Balbinot et al., 2015) and the other not (Horask et al., 2013). These contrasting results make it difficult to determine if US do increase muscle activations during locomotion. To the authors knowledge, only one study has investigated EMG whilst running in a modified rocker shoe, which found no effect on the magnitude of lower-limb muscle activations (Sobhinai et al., 2013). Whether there is an effect in commercial US is unknown and warrants investigation.

An important observation was that US research is limited to standing, walking and running. As US utilise the same concept as instability training devices, they may also enhance functional movements of gym exercise routines. Balance sandals are used by clinicians to improve ankle function after injury. They are designed similarly to wobble boards, with a hemisphere shape under the midfoot (Figure 2.2: I). Blackburn and co-workers (2003) compared shank muscle activations between balance sandals and regular training shoes in participants who had used the sandals in rehabilitation whilst performing functional movements: single leg stance, side-steps and high knees. Increased peroneus longus, soleus and tibialis anterior activations were found across movements in the balance sandals. This confirmed why balance sandals were effective in rehab. Whether US that can be worn habitually has the same effect is investigated in Chapter 7.

Reasons for the inconsistent US findings may be due to the different number of participants, number of steps measured, and data analyses (Table 2.1). Furthermore, different models of unstable shoes tested with varied severity of outsole rocker effect postural stability differently, which may require different neuromuscular control (Buchecker et al., 2012b). Additionally, there has not been a standardised control shoe across studies. The control shoe should be identical in features apart from the instability technology of the US. MBT footwear tends to be heavier than typical footwear, and added mass has been shown to influence energy expenditure and therefore likely muscle activation (Divert et al., 2008). However, Foghany and colleagues (2014) found energy expenditure between a lighter control shoe, a weight matched control shoe and an US did not differ. Moreover, wider footwear dimensions may affect medio-lateral stability. For instance, lateral heel-flair has been shown to alter frontal ankle kinematics and loading characteristics (Nigg et al., 1988).

The majority of research has been conducted on healthy participants who were likely not as affected by US instability during locomotion and do not need to increase muscle activations to maintain stability. Perhaps a more challenging unstable shoe construction would have a more pronounced effect on muscle activation patterns in the healthy population. One such developed unstable shoe, the Reflex Control Schuh (Figure 2.2: G), creates instability along the longitudinal foot axis by a thinner sole bar. A study by Schiemann and co-workers (2015) found significantly increased overall shank muscle activations in the Reflex Control, but not in an MBT, compared to being barefoot.

2.3.3. Balance

Another proposed benefit of US is that they improve balance. During static conditions, balance has been considered the ability to maintain the centre of mass with the base of support (Hof et al., 2005). However, in dynamic conditions this definition is not always true and it is more difficult to determine. The extrapolated centre of mass concept has been used to assess balance in dynamic situations, which is calculated by the position of the centre of mass added with its velocity and multiplied by the angular frequency of the inverted pendulum model (Hof et al., 2008).

In US research the centre of pressure has been used to assess balance. Landry and colleagues (2010) investigated the effect of regularly wearing MBT on static balance during double-legged standing in participants over 50 years old. Three footwear conditions were tested: MBT , barefoot and in a control shoe. After 6-weeks, antero-posterior and medio-lateral centre of pressure excursions reduced in the MBT shoes, indicating improved static balance, but no differences were found in the other footwear conditions. A better determinant of the postural system's ability in dynamic situations could be assessing the time to stabilise after an external perturbation. Ramstrand and colleagues (2008) found improved dynamic balance whilst barefoot after regularly wearing MBT shoes in children with a range of developmental disorders and females older than 50 years old (2010), indicated by a reduced time to stabilise the centre of pressure after a sagittal perturbations of the force plate. In healthy participants, a specific MBT training intervention did not improve reactive one-legged standing balance whilst barefoot after 6-weeks, but Reflex Control Schuh did (Turbanski et al., 2011). Thereby, providing more evidence the more demanding US enhances training adaptations.

Additionally, movements that are more dynamic may provide a greater training stimulus and challenge to the neuromuscular system. Michell and colleagues (2006) conducted an 8-week balance training intervention that included the lateral lunge and squat. Participants with functionally unstable and stable ankles were randomly assigned to undertake a training programme with unstable balance sandals or athletic trainers. Improved barefoot, single-leg static balance performance was reported in the antero-posterior direction in the balance sandals group and across both groups medio-laterally, but to a greater extent in the balance sandals group. A limitation identified was the lack of control group.

Quantifying any increased demand on the sensorimotor system during locomotion in US has been assessed with principal components analysis (PCA). PCA is a powerful analysis tool to extract the main source of variance in the data set from numerous dimensions. To calculate the principle components, the mean values are subtracted from each data point (similar to SD) of all the waveform data (in biomechanical analysis this is typical the ensemble averaged joint angle/moment curves). The eigenvalues are then extracted from the covariance matrix, the largest representing principle component 1, explaining where most variance occurs within the data set. An in depth explanation of PCA calculation is documented elsewhere (Robertson et al., 2014). Federolf et al., (2011) computed the principal components during walking in MBT compared to regular footwear and barefoot. The eigenvalues of the principle components were the same order between all conditions, suggesting similar movements were performed to maintain balance. PCA was not applied in this research because the unstable shoe-surface conditions did not aim to de-stabilise the overall movement patterns (Section 3.2).

The Lyapunov Exponent (LyE), is another calculation for estimating stability of human movement, which simply explained calculates the logarithmic rate of divergence between 2 close points within the reconstructed state space (Dingwell et al., 2001). It can be calculated on any kinematic spatial-temporal data variable. Federolf and colleagues (2011) found the first 2 principle components were less stable whilst participants walked in MBT compared to barefoot and a control shoe. However, only acute responses were assessed and whether balance improves with regular use was not investigated. Reasons why LyE were not applied in this thesis are given in Section 2.4.5.

Variability of movement has also been used to assess sensorimotor control (Section 2.4). Stöggl and colleagues (2010) conducted a 10-week training intervention of wearing MBT to assess biomechanical variability. Preliminary wear increased variability of joint kinematics and localised plantar pressure areas during walking compared to regular shoes by 35%. However, the variability induced by MBT had regressed back to the same level as the regular footwear after the training period, indicating that the fixed instability stimulus had become predictable. This suggests the sensorimotor system was not making additional postural balance adjustments. To promote further training benefits, the authors suggested regularly changing the US design, or having an insole construction that can be manually altered. This study also lacked a control group.

Further evidence US do not cause long-term instability, are studies that track changes over the first hour of initial wear. Blair and colleagues (2013) found initially increased vastus medialis activation whilst walking in US (Bubble Gym shoe) reduced to a similar level to a stable shoe after one hour, but tibialis anterior activation further increased. Trunk acceleration also tended to reduce after the hour walking. This suggests neuromuscular adaptations are learnt quickly and benefits of further training reduce over time. Moreover, Yap co-workers (2013) found increased energy expenditure in Sketchers Shape-Ups compared to participants' own weight matched control shoe in the first 5 minutes of walking, but no different after 40 minutes. Therefore, adaptations to US are learnt even during the first hour of wear, signifying a more challenging instability constraint is required to increase movement variability, which requires postural alterations to maintain balance. Thus, the familiarisation period to US would also appear critical to the findings of muscle activations and kinematic markers of instability. Accommodation periods from a maximum of 2 minutes to 4 weeks (Table 2.2) have been reported, which may explain the different results reported.

Table 2.2. Different familiarisation periods in unstable footwear studies

Author	Habituation Period
Price et al., 2013	Max 2 minutes
Stoggl et al., 2010	3 minutes
Plom et al., 2014; Landry et al., 2010	5 minutes
Sousa et al., 2014	10 minutes
Schiemann et al., 2015	15 minutes
Tateuchi et al., 2014	20 minutes
Taniguchi et al., 2012	30 minutes
Nigg et al., 2006b	4 days
Nigg et al., 2006a; Buchecker et al., 2013; Horsak & Baca., 2013	2 weeks
Romkes et al., 2006	4 weeks

A solution to the short-lived US instability would be to develop a shoe with continuously changing, unpredictable instability, similar to natural uneven terrain. Although not recognised as an instability training device, uneven surfaces are anecdotally supported to provide functional training benefits to runners (BBC, 2016). Ueven surfaces also create instability during walking and running that requires postural alterations and increases muscle activations (Section 2.5), but do not allow training during daily activities. Kim and Ashton-Miller (2012) did develop a shoe with unpredictable perturbations but it was controlled electronically by the investigator. Therefore, it was identified scope for developing a novel shoe with continuous mechanical perturbations that are unpredictable to the wearer.

2.3.4. Knee and low-back pain

Reducing joint pain is a proposed therapeutic benefit of US (Nigg et al., 2012). Since 2004, AposTherapy has been manufactured, designed with two convex shaped elements attached at the outsole to the forefoot and rearfoot (Figure 2.2: H). The mechanism for reducing joint pain is by reducing the ground reaction moment arm, which reduces loading, and with continual use promote improved gait function (Elbaz et al., 2010). The shoe is calibrated individually to patients by trained physiotherapists and verified by gait analysis. A training intervention with MBT also reduced pain in patients with low back pain (Armand et al., 2014). In contrast, no significant difference in the level of pain was found in patients with knee osteoarthritis compared to a control group (Nigg et al., 2006b). The mechanism by which MBT reduces joint pain is unknown, it is postulated to be due to increased activation of the smaller muscles that reduce internal joint loading (Nigg et al., 2012). Alternatively, the softer material in the heel region of the midsole in MBT may reduce the angular ankle velocity during the initial loading period of stance. This is because increased joint velocity has been linked with knee pain (Radin et al., 1991).

2.3.5. Summary

- US are unique instability training devices that allow training through daily activities. They are suggested to destabilise the locomotor system requiring increased muscle activations and require kinematic adjustments to maintain stability.
- With regular wear, these adjustments have been shown to improve balance and one conference abstract suggests they may strengthen the ankle muscles.
- Nevertheless, there is a lack of consistency and evidence in US research. Main reasons for this relate to: the abilities of participants, severity of instability from the unstable and the control shoes tested, and time for familiarisation.
- Specifically, healthy young participants do not have muscle strength or balance deficits, and improvements have been found more frequently in the more destabilising unstable footwear models. Perhaps, as well, more dynamic movements such as aerobic gym exercises are better for promoting training effects (Chapter 7).
- The instability effects of US reduce over time, possibly in the first hour of wear and thus these quick adaptations likely limit any training effects.

- This review revealed the potential for developing a more challenging US technology that provides unpredictable instability, similar to uneven surfaces (Chapter 3).

2.4. Variability

2.4.1. Introduction

From a biomechanical perspective, movement variability describes the typical variation of kinematics when the same task is repeated (Stergiou et al., 2006). Instability training devices and US, as mentioned in Sections 2.2 and 2.3, cause an increase in movement variability. Recent developments in motor control suggest it is rather simplistic to relate this with reduced stability, and it can actually demonstrate increased movement control dependent upon the level assessed. This section discusses the theories and different types of movement variability. Then, the notion instability training may improve performance or reduce injury-risk is discussed. The different analyses for quantifying movement variability were reviewed to identify appropriate methods for this research.

2.4.2. Motor Control Theories

Traditionally, increased movement variability was considered to be detrimental to performance and should therefore be eliminated as much as possible. This concept is derived from the generalised motor program theory (Schmidt et al, 1979). Since skilled performers complete tasks with high consistency, it was envisaged that the movement patterns must also have low variability (Newell & Corcos, 1993).

The more recent perspective, dynamical systems theory, supports the viewpoint that movement variability is a natural phenomenon among human movement and has functional benefits (Davids et al., 2006; Latash, 2012). Dynamical systems are signals that constantly change and evolve over time (Davids et al., 2003), and this viewpoint has been adopted as a framework to study human movement, for variation in kinematic waveforms is an inherent in biological data. Studying the variation of movement using this framework reveals underlying control strategies of self-organisation and indicates how stable the system is (Latash et al., 2002).

Unlike the traditional viewpoint, dynamics systems theory considers variability to be advantageous because it allows flexibility to perturbations and is required to change co-ordination pattern (Van Emmerick & Van Wegan, 2002). Moreover, variability allows adaptable responses to external perturbations, which can indicate balance is maintained (Section 2.4.3). Once the system has self-organised to find a stable motor pattern, variability reduces, but this can lead to the system becoming rigid and loosing of balance if faced with instability (Stergiou et al., 2006). To improve this motor flexibility, inducing instability into the system forces exploration of new motor patterns and self-organisation to find stable solutions. Instability training devices and US are useful in this respect (Section 2.2-2.3). Monitoring variability using the dynamics systems framework is important in this research to understand how shoe-surface instability challenges and changes movement responses.

2.4.3. Types of Variability

Previous research has segregated the analysis of movement variability into two levels (Hamill et al., 2012):

1. Whole body/global level or end-point variability.
2. Individual joints/segments that contribute to the global movement.

High end-point variability is detrimental to successfully completing the task (Bertenthal, 1999). Increased variability of the individual joints or segments that contribute to the overall global movement characteristics can be either beneficial or damaging to performance. Bernstein (1967) was the earliest scientist to document these types of variability whilst studying a hammering task. He noted the movement trajectory of a hammer varied between trials, despite the nail consistently being hit, demonstrating there was a redundancy. This redundancy was termed the 'degrees of freedom problem': how the motor system should organise various segments to choose the best movement solution. An alternative, more recent viewpoint has since been proposed: the numerous solutions of controlling the degrees of freedom are not a problem, but rather a luxury of the human locomotor system (Latash et al., 2012). This supports dynamical system theory because it suggests there is greater flexibility to changing constraints.

Movement variability changes during skill learning. During the early learning phases the strategy is to freeze the redundant joint/segment variability. As greater control is learnt, the degrees of freedom are released which has been observed as an increase in joint level variability (Vereijken et al, 1992). Skilful performances have been shown to have reduced levels of end-point variability and increased joint/segmental variability. This enables the task goals to be achieved consistently through a variety of movement pathways, allowing adaptability to changing environments. This relationship has been reported in elite marksmen (Arutyunyan,1968), triple jumpers (Wilson et al., 2008) , gymnasts (Hiley et al., 2013) sprinters (Bradshaw et al., 2007) and elite race walkers (Preatoni et al, 2010).

2.4.4. Instability training to optimise variability

Environmental constraints are used to impose additional perturbations required to develop and maintain the higher level of joint level variability (Schöllhorn et al., 2009). Instability training devices, US or perturbing surfaces are used for this purpose. However, limited research monitors the different types and amount of variability with these environmental constraints.

MBT, as mentioned, initially increases movement variability whilst walking, which reduces with regular use indicating improved balance. Stöggl and colleagues (2010) also found an interaction effect in the regular shoe for some variables; there was an increase in joint variability after regularly wearing MBT for 10-weeks. This is a positive training effect, for greater variability at the joint level was achieved in the regular shoe.

However, quantifying if regularly wearing unstable shoes improves walking performance, from the movement variability, is more difficult than in athletic sports where specific outcome measures, such as speed or accuracy can be used. A goal in locomotion that could be more easily tracked is the maintenance of stability. During gait, increased variability of stride, stance and swing time, as well as, step length and width have been linked to controlling balance and fall risk (Sekiya et al., 1997; Thies et al., 2005; Hausdorf et al., 2005) and are analogous to end-point variability. These variables are defined as global gait parameters. Therefore, when investigating non-performance related tasks, variability at both joint and whole body level will give greater insight of neuromuscular control.

It has been hypothesised that too much or too little movement variability also increases susceptibility to injury. Joint level variability, as mentioned, allows flexibility to perturbations and therefore reduces risk of acute injuries. Anterior cruciate ligament (ACL) deficient patients were found to have reduced sagittal knee joint variability, from the LyE, indicating a loss of complexity (Moraiti et al., 2007). Brown and colleagues (2012) studied movement variability in single-leg jumping tasks in chronic ankle instability patients. There was reduced hip and knee variability prior to initial contact in the patient group, but no differences at the ankle. Thus, forcing the neuromuscular system to deal with additional instability training devices may be useful for injury prevention (McGuine & Keene, 2006).

Reduced levels of co-ordination variability between the lower-limb joints (Section 2.4.5) has been reported in chronic overuse injuries, such as patellofemoral pain (Hamill et al.,1999; Heiderscheit et al., 2002), tibial stress fractures (Hamill et al., 2005) and low-back pain (Seay et al., 2011). The patellofemoral pain sufferers also displayed increased end-point variability of stride frequency (Heiderscheit et al., 2002). Therefore, the chance of a more severe injury from loss of balance control in these patients was also indicated. The mechanism of the chronic pain is speculated to be due to the forces being spread less evenly across the underlying tissues, hence the reduced joint variability (James, 2004). This may explain the reduced knee and low-back pain experienced after wearing US (section 2.3.4). However, these studies have exposed such changes retrospectively and it remains unknown whether resultant changes to variability are the cause or effect of the injury (Silvernail et al., 2015).

2.4.5. Methodologies for measuring movement variability

In statistics, variability measures the spread of data about the central tendencies. The standard deviation (SD) is the most common way to calculate the magnitude of variability (Riley & Turvey, 2002). The calculation involves taking the deviance from the mean of each data point, squaring these values, these are then summed and the square root of this is the SD. An advantage of the SD measure is that all data points are accounted for (Vincent, 1999). The coefficient of variation (CV) is the most common method for normalising the variability. It is calculated by dividing the SD by the mean ((SD/mean) *100). The CV can be useful for determining the relative magnitude of variability when there are differences in mean readings (James, 2004). For larger quantities tend to have larger variability, but the relative difference may not be representative. A limitation of the CV is if the mean value is close to zero the results can give infinitely large numbers that are not valid.

The magnitude of variability has been explored at both discrete points and continuously during the movement task. Discrete methods quantify variability using the chosen statistic at the time points or distinct event of interest. Continuous methods track the variability over the whole movement or during sub-periods, and have been considered advantageous for understanding the entire movement (James, 2004). Typically, data is presented using the mean ensemble curve and the SD is either displayed at each time point across the curve, or averaged throughout the movement period so appears the same throughout (Figure 2.3).

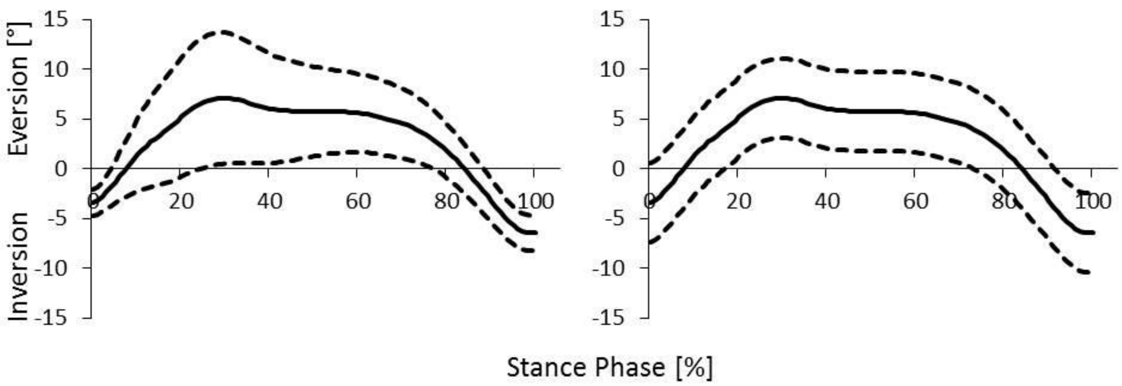

Figure 2.3. Mean ensemble frontal ankle angle curve (solid line) during stance phase treadmill walking of 30 steps, normalised to 101 data points. Variability (dotted line) is expressed by the standard deviation about the mean, using the point by point method (left) and average SD during stance (right). Notice how the variability fluctuates during stance in the point by point method.

Variability of the relative motion between joints/segments has been used to assess coordination. This has been quantified on angle-angle plots using a vector coding technique (Sparrow et al., 1987), where the relative angle between the joint/segments (coupling angle) to the horizontal is computed for each data point (Heiderscheit et al., 2002). Another method, the continuous relative phase (CRP) calculates the difference between the phase portraits (angle/velocity curve) of two segment/joints (Hamill et al., 1999; Hamill et al., 2005; Seay et al., 2011). The variability of the CRP could be qualitatively evaluated as the SD at each data point. However, these continuous methods typically quantify one value for variability, the average SD across the movement cycle or its sub-phases (Hamill et al., 2000). Therefore, how variability changes during a movement is masked similarly to the discrete methods (Figure 2.3). This limitation identifies scope for developing a new assessment method which quantifies how variability changes during the movement.

Rather than assessing the magnitude of variability (linear methods), non-linear methods (such as LyE) quantify the temporal structure. These methods have been suggested to quantify stability; defined as the ability to 'offset' a perturbation and not variability (Stergiou & Decker, 2011). For the research conducted in this project non-linear methods were not applied. A review of the non-linear methods suggested they are suitable for local perturbation experiments, for example stride-stride fluctuations. However, these authors also stated the magnitude and timing needs to be consistent to analyse stability, and therefore they are not appropriate for random external perturbation experiments, such as continuous irregularities at the shoe-surface interface (Bruijn et al., 2013). As the shoe and surfaces tested in this research aimed to create unpredictable, random perturbations non-linear analyses were not suitable. Moreover, non-linear techniques do not consider how stability of the joint/segments may differ from the more global/end point parameters. Nonetheless, the measures that quantify the magnitude of variability present how the different levels of the system respond to their constraints, which was the focus of this research.

2.4.6. Summary

- Movement variability is functional, enabling adaptability in changing environments. Desirable levels of low end-point and high joint/segment variability can be trained.
- Instability training devices impose perturbations, provoking increased variability during routine movement patterns. Although, there is a lack of research on how these effect the different types of variability.
- Effects of instability training on performance during everyday locomotion can be monitored; from the variability of end-point parameters, such as global gait parameters.
- Increased joint level variability is associated with fewer acute/chronic injuries. Thus, instability training may be utilised as an injury prevention strategy.
- Measuring the magnitude of variability across sub-periods and at both the joint and whole body level will give greater insight of neuromuscular control to shoe-surface instability.

2.5. Stability Mechanisms

2.5.1. Introduction

In physics, stability is defined as the likelihood of falling following a perturbation. A stable system is more likely to return to the same equilibrium state than an unstable system. The definition in the analysis of human movement is the susceptibility to perturbations, both external (environmental) and internal (noise from sensorimotor system) (Dingwell et al., 2001). Over recent years a lot of analysis techniques have been developed to quantify stability during human locomotion, but there is still not a gold standard (Bruijn et al., 2013). Example variables that have quantified stability during locomotion across uneven surfaces are summarised in Table 2.3. Regardless of the analysis, research consistently reports unstable surfaces reduce stability. US and instability training research relates increased muscle activations and lower limb adaptations to responses to reduced stability.

This section evaluates the biomechanical and neuromuscular adaptations identified for maintaining balance when encountering unstable shoe-surface interfaces. Kinematic and kinetic strategies to US and irregular surfaces (IS) are reviewed, as well as, joint stiffness and muscle co-contraction adaptations. One benefit of US, identified in Section 2.3, is they allow training through dynamic movements. Therefore, locomotion studies are evaluated.

Table 2.3. Variables used to quantify stability during locomotion across uneven surfaces

Stability Measure	Studies
Variability of the global gait parameters	Thies et al. 2005; MacLellan & Patla, 2006; McAndrew et al. 2010; Voloshina et al. 2013; Gates et al. 2012; Marigold and Patla, 2008; Voloshina et al. 2013; Voloshina and Ferris 2015
Variability of the margin of stability	MacLellan and Patla, 2006; Gates et al. 2013
Centre of mass acceleration	Marigold and Patla, 2008
Head and trunk acceleration	Menz et al. 2003a

2.5.2. Lower-limb kinematics and kinetics in unstable footwear

US designs target either anteroposterior or medio-lateral stability, which affects muscles groups differently (Section 2.3.2), but also influences ankle kinematics. Changes can be active adaptations to increase stability or passive alterations imposed by the shoe design. The thick, rounded outsole of MBT and Sketchers Shape-ups (Figure 2.1: A) likely causes compensations at initial contact and a different foot-rollover motion. This design causes the increased ankle dorsiflexion reported in MBT during the first half of stance (Romkes et al., 2006; Nigg et al., 2006a; Landry et al., 2012; Taniguchi et al., 2012), but reduces dorsiflexion peaks angles (Romkes et al., 2005; Forghany et al., 2014). The anteroposterior rocker also causes a more anterior the centre of pressure (COP) position at initial contact (Price et al. 2013; Forghany et al., 2014; Stewart et al., 2007).

Regarding other US brands, Skechers Shape-Ups similarly increase initial dorsiflexion (Talaty et al., 2016). FitFlops are reported to both increase (Price et al., 2013) and decrease peak dorsiflexion angle (James et al., 2015). The reason suggested for this different finding is the use of the single static barefoot calibration trial to compute neutral positions of joint angles in the study by James et al., (2015). While, Price and colleagues (2013) captured a static trial for each shoe condition, making it hard to compare relative ankle angles with footwear of different heel-toe offsets. Therefore, the use of a single static calibration allows a better comparison between shoe conditions with different heel-toe offsets. It should also be noted, it may not be possible to do this accurately if markers cannot be placed on the foot, but arranged at slightly different locations between shoe conditions. Both these studies also examined COP distribution in Fitflops. James and colleagues (2015) found COP shifted anteriorly in early stance and medially in late stance, suggesting the Microwobbleboard technology affects sagittal plane ankle control initially and frontal plane control during push-off. Price et al. (2013) similarly found reduced medio-lateral COP range in Fitflops, but only significantly compared to Reebok Easy Tones. This likely reflects the authors COP analysis covering all of stance, rather than sub-periods or different plantar regions.

It appears MBT does not change ankle kinematics in the frontal plane or coronal plane (Romkes et al., 2006; Nigg et al., 2006a), but are often not reported. This is possibly due to the antero-posterior rocker outsole design only influencing the sagittal plane. Landry and co-workers (2012), on the other hand, did find a reduced ankle ab/adduction range of motion with a principal components analysis. James and colleagues (2015) found frontal ankle range of motion and eversion peak increased in FitFlop compared to a sandal, but reduced compared to a regular shoe. This likely relates to the reduced shoe stiffness and increased sandal stiffness. How other US brands affect frontal plane kinematics is unknown.

Other lower-limb kinematic alterations in US during gait are less consistent. MBT are reported to reduce the range of motion at the knee (Romkes et al., 2006; Buchecker et al., 2013; Forghany et al., 2014) and hip (Romkes et al., 2006; Buchecker et al., 2013; Landry et al., 2012; Forghany et al., 2014), however other studies found no differences (Talaty et al., 2016; Price et al., 2013; Nigg et al., 2006a). The discrepancies may relate to the global gait parameters. For example, Romkes and colleagues (2006) reported a slower gait speed and reduced step length in MBT that resulted in reduced hip and knee range of motion. Other studies found no change in joint range of motion when gait speed was similar (Price et al., 2013) or when walking speed was controlled (Nigg et al., 2006a). These differences signify the importance maintaining a similar gait speed between US and control shoe conditions. This will ensure lower-limb kinematics are not influenced by different speeds, although imposing a non-preferred speed may affect variability.

As pointed out, EMG measurements are limited to the large superficial muscles, but is has been suggested US have greater effects on the smaller, deep ankle muscles. Activating these muscles is advantageous for joint stability because they react faster to unstable shoe perturbations (Nigg et al., 2012). To evaluate the contribution of all muscles about a joint, the internal moments can be calculated through inverse dynamics. Joint moment strategies during US locomotion have also been investigated with contrasting results. A summary of kinetic findings is provided in Table 2.4. Most studies report walking in MBT or rollover footwear reduces ankle dorsiflexion and plantarflexion, knee flexion and hip flexion moment peaks. Similarly, FitFlops reduce sagittal ankle and knee moment peaks, but increase hip extension moment compared to a regular sandal (James et al., 2015).

Frontal plane joint moments in MBT footwear are less well documented, but reduced peak knee adduction moments have been reported in overweight males (Buchecker et al., 2012a) and no different in women with osteoarthritis (Tateuchi et al., 2014). A modelling study shows activating smaller muscles closer to the joint axis reduces internal loading (Nigg et al., 2005). Thus, the reduced sagittal ankle moments suggest greater activations of the small muscles are utilised to maintain stability and reduce internal loading. This assumption does not account for muscle co-contraction increases (Section 2.5.5).

The only research conducted on running in US has used MBT and rollover footwear. This may be due to the fact that MBT shoes were originally designed for standing and walking, although the company has recently developed running specific models. Findings conclude similarities to walking studies: rocker outsoles reduce dorsiflexion and plantarflexion moments, but in contrast have limited effects on knee and hip moments (Boyer & Andriacchi., 2009; Sobhani et al., 2013; Sobhani et al., 2016).

Increased initial ground reaction force (GRF) loading rates can be an indicator of reduced stability, which is reported whilst walking in MBT. This has been related to reduced ankle plantarflexion angle and capability to absorb shock (Sacco et al., 2012; Forghany et al., 2014). In addition, initial vertical GRF peaks has been found to reduce in males (Taniguchi et al., 2012), but increase in females (Sacco et al., 2012). Research on Reebok Easy Tone is scarce, with no substantial alterations to kinematics, or kinetics during stance reported for the lower limbs, but an increased vertical ground reaction force loading rates (Horsak & Baca, 2013). These results suggest US may cause instability during the loading period of stance.

Studying the GRF with joint moments, will allow further insight into stability strategies. For reduced joint moments could be a sign of reduced GRF loading or a reduced moment arm, the latter may be utilised to keep the GRF vector closer to the joint axis and prevent injury. US studies have not addressed this notion, but have identified an anterior COP shift in MBT and rollover footwear, linked to the reduced ankle dorsiflexion moments (Price et al. 2013; Forghany et al., 2014; Stewart et al., 2007).

Table 2.4. Overview of studies that investigated lower-limb kinetics in unstable footwear during locomotion.

Author (year)	n	Unstable shoe(s)	Walk/ Run	Variables	Results (p<.05)
Nigg et al. (2006)	6	MBT	Walk	Resultant joint moment angular impulses	N/A
Boyer et al. (2009)	19	MBT	Run	GRF, 3D lower-limb joint moments	GRF: ↓1st medial peak and 2nd anteroposterior peaks Ankle moment: ↓dorsiflexion, plantarflexion and inversion
Buchchecker et al. (2012a)	10	MBT	Walk	Lower-limb joint loading, sagittal moments, knee adduction moment	Knee: ↓ 1st adduction
Landry et al. (2012)	23	MBT	Walk	Principal components analysis: 3D lower-limb moments	Ankle moment:↓ dorsiflexion, plantarflexion, ↑inversion, abduction, Knee moment:↓ adduction , flexion Hip moment: ↑extensor, abductor
Sacco et al. (2012)	25	MBT	Walk	GRF	GRF: ↑ vertical peak and loading rate
Taniguchi et al. (2012)	14	MBT	Walk	GRF, lower-limb joint moments	GRF: ↓ vertical components, Ankle moment: ↓dorsiflexion and platarflexion Knee moment: ↑ flexion
Buchchecker et al. (2013)	22	MBT	Walk	Lower-limb sagittal moments joint , knee adduction moment	Ankle moment:↓ dorsiflexion, plantarflexion, Knee moment: ↓ flexion, extension, adduction (1st peak) Hip moment: ↓hip flexion
Horsak & Baca (2013)	12	Reebok Easy Tone	Walk	lower-limb joint moments	GRF: ↑first vertical peak Ankle moment:↑dorsiflexion Hip moment:↑ extension
Sobhani et al. (2013)	16	Rocker shoe	Walk/ Run	lower-limb joint moments	Both locomotion Ankle moment: ↓ plantarflexion
Forghany et al. (2014)	20	MBT, rollover shoe	Walk	lower-limb joint moments, GRF peaks and average loading rate	GRF Loading rate: MBT>rollover>control Both shoes: Ankle moment: ↓ dorsiflexion, plantarflexion Knee moment: ↓ flexion, extension
Tateuchi et al. (2014)	17	MBT	Walk	Knee flexion and adduction moment, GRF	Knee moment: ↓flexion
James et al. (2015)	25	Fitflop	Walk	3D ankle moments, sagittal knee and hip moments	Ankle moment: ↓ dorsiflexion, plantarflexion, adduction Knee moment: ↓ flexion, extension Hip moment:↑ flexion, extension
Talaty et al. (2016)	10	MBT, Skechers Shape-Ups	Walk	lower-limb joint moments	Both shoes: Ankle moment:↓ dorsiflexion, plantarflexion Knee moment: ↓ flexion Hip moment: ↓ flexion
Sobhani et al. (2016)	16	Rocker shoe	Run	lower-limb joint moments	Ankle moment: ↓ plantarflexion Knee moment: ↑ extension

↑/↓ = significant increase/decrease compared to a control shoe. Sagittal plane joint moment variables unless otherwise stated

2.5.3. Lower-limb kinematics on irregular surfaces

Previous research investigating IS generally relate adaptations more directly to improving whole-body stability than US research. Adaptations of global gait parameters on IS indicate a conservative approach to increase stability. Increasing step width whilst walking across IS has been reported to increase the base of support (Marigold & Patla, 2008; Menz et al., 2003a; Gates et al., 2013; Menant et al., 2008). This strategy has often been more apparent in populations who are known to have increased risk of falls, such as older adults (Marigold & Patla, 2008; Thies et al., 2005), who likely have reduced balance ability and require a more cautious gait pattern to maintain stability. Similarly, a longer step length has been found during walking on a compliant (MacLellan & Patla, 2006) and IS (Menz et al., 2003a). Other than increasing the base of support, longer step lengths are proposed to be beneficial by reducing the number of foot contacts and allowing longer to prepare limb posture for the next step (Menz et al., 2003a). Contrasting this, other studies report shorter step lengths whilst walking on uneven (Voloshina et al., 2013) or oscillating surfaces (McAndrew et al., 2010). In older adults this is associated with reduced walking speed (Menant et al., 2008; Marigold & Patla, 2008). This has also been suggested as a cautious adaptation that ensures trunk and head stability (Menz et al., 2003b).

Posture alterations also improve stability on IS. Reducing the shoe-surface angle, as well as step length, at initial ground contact reduces the risk of falling by decreasing the shear forces and consequently friction coefficient at the shoe-floor interface (Marigold & Patla, 2002; Menant et al., 2008). Additionally, lowering the position of the centre of mass increases stability by reducing the ground reaction force moment arm and increasing the force required to induce a fall (MacLellan & Patla, 2006). This has been utilised on IS by increasing knee and hip flexion, to lower the centre of mass, at initial ground contact whilst walking (Gates et al., 2012). Thomas and Derrick (2003) found increased knee angle increased the impact shock attenuation at the head whilst running on an uneven treadmill surface. This was suggested to be through reducing the effective mass at initial ground contact to cope with the variable impacts on the IS, but they did not investigate whether centre of mass was lowered.

2.5.4. Leg, ankle and knee joint stiffness strategies

Joint stability has been described as the ability to retain kinematic movement within the desired pathway (Gabriel et al., 2008). Adjusting levels of joint stiffness, therefore, has been recognised as a control strategy to maintain joint stability (Hogan, 1984). It has been postulated there is an optimal amount of stiffness to enhance performance and reduce risk of injury (Brazier et al., 2014). A stiffer joint will have a stronger resistance to external net moment perturbations, possibly preventing excessive joint motion and injury (Riemann et al., 2002; Wang et al., 2015). On the other hand, Kim and Lockhart (2012) found reduced ankle joint stiffness was associated with improved limb stability during stable level walking, after an 8-week balance training intervention in elderly adults. Moreover, increased leg stiffness associates with greater impact peaks and loading rates (Butler et al., 2003), which is linked with repetitive overuse injuries whilst running (Hreljac et al., 2000). Retrospectively, a stiffer knee joint has been associated with increased low back pain (Hamill et al., 2009) and tibial stress fractures whilst running (Milner et al., 2006), and knee osteoarthritis in walking (Dixon et al., 2010; Zeni & Higginson, 2009), but there is a lack of prospective studies to confirm this association. The ability to increase joint stiffness acutely in response to external perturbations, but display lower levels with repetitive loading therefore seems favourable to reduce the likelihood of injury.

Previous studies have investigated stiffness adaptations on surfaces with varied hardness. Reduced vertical leg displacements increased leg stiffness on softer surfaces, and increased leg displacements decreased stiffness on harder surfaces during running. This mechanism allowed the centre of mass motion, and vertical stiffness to remain constant on all surface compliances (Ferris et al., 1998). Similarly, ankle and knee joint stiffness both increase during hopping on softer surfaces (Farley et al., 1998) and whilst running in softer midsole shoes (Baltich et al., 2015), compared to a harder surface and shoe condition. Although the shoe-surface interface used in these studies was flat, the soft conditions may produce some instability which is controlled through increasing stiffness levels

Joint stiffness strategies have also been observed regarding unpredictable perturbations. For example, hand stiffness increased in response to upper limb unpredictable perturbations (Burdett et al., 2001). Walking across a slippery or perturbing surface has been suggested to reduce range of motion and increase ankle and knee stiffness (Fong et al., 2005; Chmielewski et al., 2005). Moreover, Voloshina and Ferris (2015) found increased leg stiffness during running on an IS treadmill. This adaptation was attributed to anticipating the unfamiliar unpredictable perturbations. Despite this, the relative levels of joint stiffness that help maintain stability during uneven shoe-surface perturbations are unknown and warrant investigation.

Hooke’s law states the force required to compress a body is related to the distance displaced and the coefficient of stiffness. This principle only holds true for elastic materials that store and return energy, whereas biological materials are viscoelastic. Biomechanical studies that estimate stiffness are simplistic because they do not account for the numerous components separately and give one value for their combined stiffness. Techniques for analysing stiffness are split into those which measure linear stiffness of the global whole body mechanics and analyses of rotational stiffness about the joints. Detail of the different measurement approaches are given elsewhere (Butler et al., 2003), thus only briefly reported here.

The most common way of calculating vertical (linear) stiffness is to divide the maximum vertical ground reaction force by the change in centre of mass height. Stiffness of the leg uses similar formulae that accounts for the change in leg length (McMahon & Cheung, 1990). Joint stiffness has been quantified by dividing the relative change in joint moment by the change in joint angle, applying a linear regression and taking the coefficient as the value of stiffness. The calculation does not comply with the true mechanical stiffness equation of Hooke's law and has been termed 'quasi-stiffness' (Latash & Zatsiorsky, 1993). However, it does provide information about the resistance of a joint to deform and hence adaptations to shoe-surface perturbations (Baltich et al., 2015).

Alternatively, joint stiffness has been quantified during static conditions by applying external perturbations. This has been demonstrated during static conditions by applying an external perturbation of known force and location. During walking gait, this has been estimated by applying an external perturbation about the ankle joint with a rotating force platform. The angle and moment perturbation response were subtracted from the no perturbation trials and a second order polynomial model applied to calculate impedance (stiffness, damping and inertia) at different portions during stance (Rouse et al., 2014). Applying a perturbation accurately about the joint axis during dynamic movements is not possible in the many research labs, and was a limitation of the Rouse et al. (2014) study. Thus, joint stiffness has been quantified in most unstable or perturbation experiments by calculating quasi-stiffness.

2.5.5. Muscle co-contraction strategies

Simultaneously activating opposing agonist and antagonist muscles about a joint is referred to as muscle co-contraction. This is another strategy of the neuromuscular system to maintain joint stability that is linked to reducing knee and ankle injury (Hirokawa et al., 1991; DeMers et al., 2016). Increased knee and ankle co-contraction has been found after initial walking on an IS (Voloshina et al., 2013). This is in accordance with US studies that found walking in MBT and Reebok Easy Tone shoes increases knee co-contraction, even after a two-week accommodation period (Horsak et al., 2015; Buchecker et al., 2012a). As such, US may train neuromuscular adaptations required to prevent acute knee injuries. On the other hand, elevated co-contraction increases joint loading which has been related with knee osteoarthritis (Hodges et al., 2016).

Mechanical joint stiffness is derived by muscle stiffness, which the neuro-muscular system controls by adjusting muscle activation level (Hogan, 1984; Lee et al., 2006). Muscle co-activation increases ankle and trunk stiffness during static postures (Hunter & Kearney 1982; Lee et al., 2006). It is supposed muscle co-contraction is the underlying mechanism to increasing joint quasi-stiffness. This notion is explored in Chapter 6.

2.5.6. Summary

- Cautious adaptations to gait and lower limb posture improve whole body stability across IS.
- US technologies cause different kinematic adaptations, but all designs appear to affect the ankle to a greater extent than the more proximal knee and hip.
- Joint moments are useful for assessing the contribution of all muscles, including the smaller ankle muscles. This needs to be measured with muscle co-contractions to estimate internal loading. Analysis of ground reaction forces in addition with joint moments will provide further insight into joint stability.
- At the joint level stiffness adaptations are utilised to enhance stability. There is a lack of research investigating these control mechanisms to shoe-surface instability.

2.6. AIMS AND OBJECTIVES

Based on the disadvantages and lack of knowledge of current instability training devices identified in this literature review, the aim of this research was:

To develop and evaluate the biomechanical and neuromuscular adaptations of a novel shoe that provides continuous unpredictable, irregular perturbations during dynamic movements.

Specific objectives were:

1. To develop a surface and a shoe that exhibited random instability (Chapter 3). This surface allowed a condition against which the shoe could be compared.
2. To compare the instability between the developed irregular surface and shoe by examining the movement variability, posture and muscle activation adaptations during walking and running (Chapter 5).
3. To explore the joint stability strategies between the developed shoe and a commercial US during walking and running (Chapter 6).
4. To establish if the shoe also induces instability during functional gym exercises: the forward and lateral lunge (Chapter 7).
5. To evaluate the long-term training effects of the shoe on ankle muscle strength and balance (Chapter 8).

Completing these objectives will increase our understanding of the acute adaptations to unpredictable shoe-surfaces (Objectives 2-4) and the potential long-term training benefits (Objective 5). The knowledge obtained will be used to advise on the potential benefits of incorporating unpredictable instability as a training mechanism in comparison to current commercial training devices that provides predictable (fixed) instability.

CHAPTER 3.

The development of a surface and a shoe with unpredictable instability

3.1. Introduction

In order to achieve the aims and objectives set out in this research, a surface and a shoe with continuous unpredictable perturbations first needed to be developed. This chapter describes the process of design and creation of the irregular treadmill surface (IS), which was used as a research tool to later validate effects of the shoe. Details of the initial prototypes that led to the development of a shoe with random midsole deformations to provide unpredictable instability are then provided.

3.2. Irregular treadmill surface rationale

To develop authentic footwear technologies, in-depth knowledge of human movement must initially be established (Sterzing et al., 2012). To provide proof of concept of unpredictable instability in footwear, the first step was to establish how humans move over IS that exhibit continuous unpredictable perturbations. This fundamental research enhanced our understanding of the biomechanical and neuromuscular IS adaptations.

To date, there is a lack of knowledge about how we traverse natural uneven terrain surfaces. This is partly believed to be due to the difficulties exporting the bulky, heavy, expensive biomechanical equipment (Allen et al., 2015). To investigate IS in laboratories researchers have used de-stabilising rocks (Gates et al., 2012), unexpected drops (Daley et al., 2008; Ferber et al., 2002) or hidden irregular objects underneath grass carpets (Menz et al., 2003a, 2003b; Thies et al., 2005; Marigold & Patla, 2008). These studies are typically limited to between 5 to10 trials and the number of continuous steps to the size of the laboratory, which does not allow time to get into a more regular locomotion rhythm. This is reported to effect the variability of global gait parameters (Barnett et al., 2016). Studying uninterrupted random IS locomotion will increase validity to the type of footwear we wanted to develop and allow movement variability measures to be more accurate.

Voloshina and colleagues have studied the biomechanics and energetics of walking (2013) and running (2015) on an uneven surface by modifying the belt of a treadmill. A secondary canvas belt with loop Velcro was attached with wooden blocks arranged covering the whole surface. The heights of the wooden blocks (1.27, 2.54, and 3.81 cm) were considered too high and unrealistic to incorporate into footwear design. Thomas and Derrick (2003) created an IS treadmill by attaching four boards, of uniform dimension (1.3, 5.1 and 40.6 cm) to the belt of a treadmill at uneven distances to study running adaptations. Details of the material and configuration were not specified. However, the IS was limited to a set height, both studies utilised the same material hardness and the square objects meant there was no difference of surface gradient. This limits the wider application to natural terrain, with infinite ranging surface properties.

To overcome these previous limitations, we wanted to design a treadmill with random instability, with varied surface height, hardness and slope. It is acknowledged the man-made IS treadmill we aimed to develop would not be as variable as natural terrain, which has infinite different surface properties causing both large and subtle perturbations. Additionally, the IS treadmill required materials to be rigidly attached, so the effects of deforming surfaces, such as gravel or sand were not investigated. The type of IS surface sought after was one which caused perceivable perturbations, but would not de-stabilise the locomotion pattern too much, that may require additional harness safety precautions. This was because we later hoped to incorporate this type of instability into footwear.

3.3. Irregular treadmill surface design

The belt of a Woodway treadmill was modified in this research and used in Chapter 5 (Pro XL, Woodway Inc.,WI, USA) and 8 (LOKO S55, Woodway GmbH, Steinackerstraße, Germany). These treadmills have conveyor belt designs with rubber slats (Pro XL: 77

segments, 700mm x 58mm, LOKO: 60 segments, 550mm x 58 mm) orthogonal to the locomotion direction. Each slat was covered with loop material (Velcro) that was cut into strips of the same dimension; this served as the regular surface during testing (Figure 3.1). The IS was created by fixing 4 types of EVA dome shaped inserts (Ø: 140mm) of different height (maximum: 1.0 and 1.5 cm, minimum: both 0.3 cm) and hardness (40 and 70 Asker C) to the treadmill belt by hook (Velcro) material. One insert of each type was ordered randomly into a set, which were arranged non-uniformly across three slats (Figure 3.1). To eliminate visual targeting of foot placements during testing, participants were instructed to focus their gaze on a point on the white board (Figure 3.1). Investigators monitored this ensuring participants could not predict what they would to land on.

Figure 3.1. The irregular treadmill surface with 4 types of inserts attached to the belt by Velcro. This allowed quick changeovers between the regular and irregular surface (bottom right).

This treadmill surface has been examined previously at Li Ning Sports Science Research Center in a basic research study. Results validated that the IS implemented increased instability and during walking (Sterzing et al., 2014a; Apps et al., 2014a) and running (Sterzing et al., 2014b; Apps et al., 2014b). Hence, it was deemed suitable for this research.

3.4. Development of a shoe with unpredictable perturbations

This section gives brief detail about the shoe prototypes that were developed, including the final irregular midsole shoe that was approved for testing in the experimental studies (Chapters 5-7). At the end of the section, a timeline of when the treadmill and shoe prototypes were developed is provided (Figure 3.6).

3.4.1. 'Ball' prototypes

The first prototypes had ball bearings moving freely inside a separate compartment of the midsole. There were 2 rounds of prototypes. Two models were developed: the LN P2-FF had slightly smaller ball bearings in the forefoot and LN P2-RF had slightly larger ball bearings in the rearfoot (Figure 3.2). Shoes were evaluated using subjective ratings of likability, comfort, concentration level, difficulty, stability, variability, safety and energy requirement on 9-point Likert scales. Ten male participants completed an indoor loop (60m x 30m) four times for each shoe condition. After each loop, 2 Likert ratings variables were completed. Walking and running trials were collected at a self-selected pace in each shoe. For comparison to the prototypes a control shoe was tested first to avoid crossover effects from the unstable footwear, followed by mixed order of LN P2-FF and LN P2-RF. Participants commented the prototypes were noisy and heavy compared to a regular shoe. This likely caused the increase in subjective perception of energy requirement and difficulty, but no difference in variability was perceived.

Figure 3.2. The ‘ball’ shoe prototypes: LN P2-FF (top) and LN P2-RF (bottom).

3.4.2. ‘Balls and gel’ prototypes

To make improvements to the ‘ball shoes’, a commercial unstable shoe model, the Bubble Gym shoe (Figure 3.3) was modified. The midfoot pivot was removed and the hollow forefoot and rearfoot protrusions were filled with different numbers of ball bearings and gel (Figure 3.3). The concept of adding gel was to prevent the ball bearings from moving to the same position each step, thus providing a different shoe-surface profile during ground contacts. After the same subjective perception testing being repeated on 10 female participants (prototype shoe was female), it was concluded that the concept did not work because the balls moved to the same areas and the dimension of the sphere were always the same upon landing not providing unpredictable instability.

Figure 3.3. The ‘balls and gel’ prototypes were developed from an unstable shoe model (Bubble gym, Li Ning, top). The midfoot pivot was removed and the forefoot and rearfoot protrusions (left middle) filled with ball bearings (bottom right) and gel material (middle, right).

3.4.3. Hand-made approach

Due to time constraints of the PhD timeline and the slow progress of production of the instability footwear prototypes some ‘hand-made’ prototypes were attempted. The main limitation to achieving unpredictable instability from the previous prototypes was the outsole material between the ball bearings and ground being too stiff. This would have limited the effects of any perturbations. Additionally, the ball bearings were spherical shaped which consequently meant the orientation during ground contact never changed. The Beijing flea markets provided the inspiration of trying irregular shaped buttons inside cotton material. The two rather special creations are illustrated in Figure 3.4. Although not stylish, the unpredictable instability effect was obvious during walking. The button objects moved independently throughout swing, creating a different shoe-surface profile at every ground contact and thus unpredictable perturbations. This crude concept prototype shoe helped explain to the development team at Li-Ning what it was we were trying to achieve.

Figure 3.4. Hand-made prototypes. The first attempt was socks simply attached to a regular training shoe (left) with oddly shaped buttons inside (right). This created the random instability sought after. To demonstrate the desired prototype to the development team, cotton material was stitched and attached to a shoe upper at the rearfoot, midfoot and forefoot by Velcro (middle).

3.4.4. The irregularly deforming midsole prototypes

The tooling for developing highly flexible rubber bags (hardness: 28 Asker C, thickness: 1.5 mm) was opened (Figure 3.5). Bags were separated for 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, the same as typical footwear, but due to the deformable bag material this reduced when wearing the shoe. The bags were originally fixated to the shoe upper (Li Ning Fengchao TD, Li Ning Co, Beijing, size male US 10.0, female US 8.0), and there were flaps that could be opened under the insole into each separate bag to insert objects. Later, we opted to attach the bags to the shoe upper by Velcro. This fixation allowed the objects to move more freely inside and was advantageous because the same shoe upper could be used for the control shoe (section 3.5).

Figure 3.5. The midsole bags with cubes and ball bearing placed inside (left) and attached to the shoe upper (right).

Additionally, the moulds for certain shapes of specific sizes were opened. The shapes tested included: cubes, prisms and ovals made from TPU material (hardness 85A Shore) of small (10 mm^2, oval: 10x20x10 mm) medium and (15 mm^2, oval: 15x25x15 mm) large size (20 mm^2, oval: 20x30x20 mm). We also wanted to test different hardness of objects, but this was not approved due to cost. It was not possible to test every type of object combination in the rubber midsole bags. Instead, a trial and error approach was used to determine what perceived to be the best combination by the footwear researchers at the Sports Science Research Center. After trying many configurations, we determined that the medium cube shapes with the ball bearings (12 mm diameter) from the 'balls and gel' prototypes worked best. This was assumed to be due to the weight of the ball-bearings increasing the vertical space allowing the cubes to move more freely during swing-phase.

Ten male participants completed the same subjective ratings protocol using 4 bag combinations with increasing number of cubes and ball-bearings. This analysis led to a final number of cubes (10) and ball bearings (51) to be tested in total. The number of cube shapes were 4:3:3 and ball bearings were 15:15:21 inside the rearfoot, midfoot and forefoot bags respectively. Originally for the first experimental treadmill study (Chapter 5), males were the target participants because they were tested in the basic research study and the data could be compared. At this time, the product development department were considering developing a commercial instability shoe product based on this research. After initial discussion, they expressed this type of shoe would be targeted at females. Therefore, the participants of the later experimental studies (Chapter 6 and 7) were females. The number of cubes (3, 2, 2) and ball bearings (14, 13, 15) inside the rearfoot, midfoot and forefoot rubber bags was reduced to accommodate the smaller bag size. It is acknowledged that the shapes and number of objects selected may not provide the greatest possible effect of instability. The first aim was to establish if the irregular midsole could provide unpredictable instability before refining specifics with later rounds of prototypes.

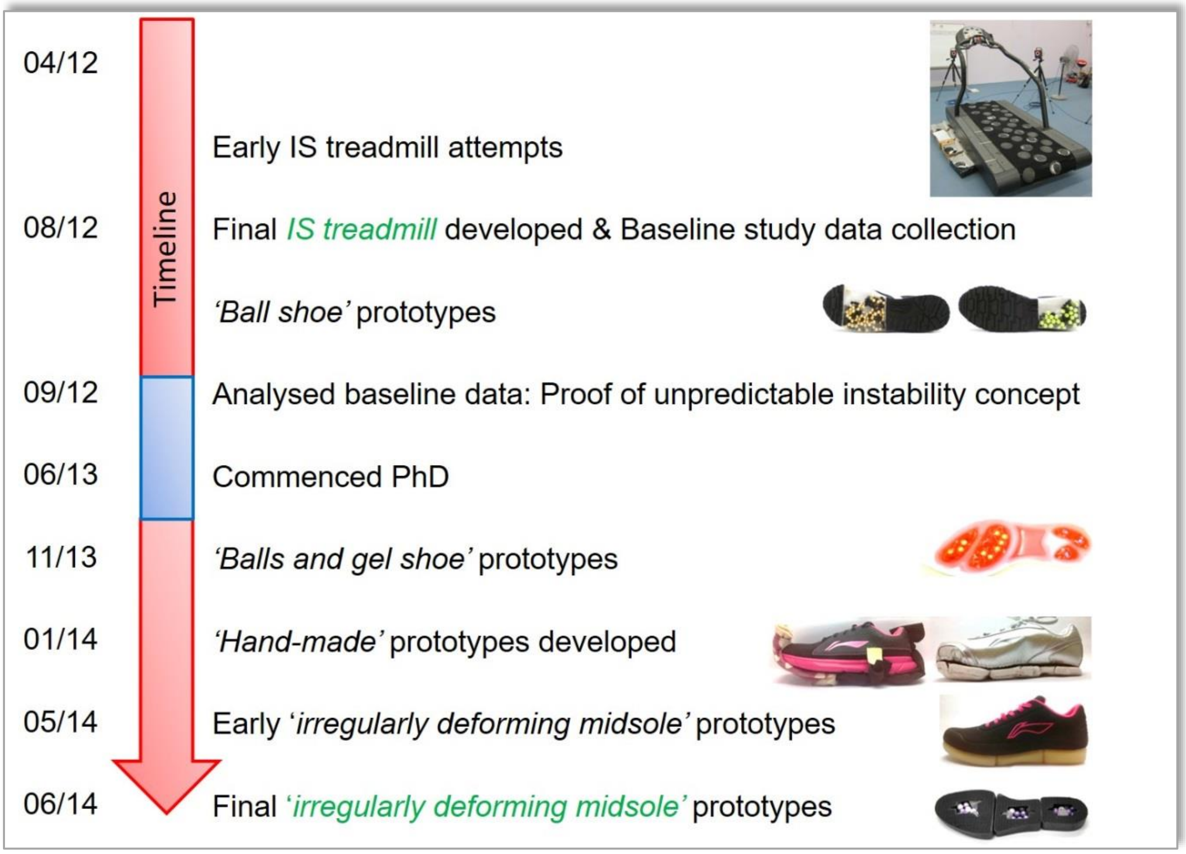

Figure 3.6. Timeline of irregular treadmill surface and irregular shoe midsole prototype developments. Time at Li-Ning Sports Science Research Center (red) and Liverpool John Moores University (blue) is illustrated.

3.5. Control shoe modifications

The literature review identified the reported differences in US studies may be due to the lack of standardised control shoe to compare to (section 2.3.2). Studies have compared to standard stable athletic footwear (Nigg at al., 2006;) or sandals (James et al., 2015; Price at al., 2013), participants' own flat footwear (Stewart et al., 2007; Landry et al., 2010), barefoot (Landry at al., 2010; Schiemann et al., 2014), whilst others have modified regular footwear to be weight-matched to the US (Forghany et al., 2014; Yap et al., 2013). No matter which control shoe is opted for, there will always be movement variability apparent because it is an inherent characteristic of humans (Davids et al., 2006). However, deciding which control shoe to use in this research was important because the level of instability produced by the irregular shoe-surface conditions would be compared to the control, and this varies among different control footwear previously studied. Barefoot conditions, for example, likely increase instability compared to regular footwear and are suggested as a strategy to re-train the ankle muscles which are not required in stable footwear for balance (Nigg, 2009). A barefoot control was considered in this project, but because the footwear and surfaces we studied during tasks of daily life (walking, running and functional movements) that are more commonly done wearing shoes, comparing to a control shoe condition was considered more appropriate.

As mentioned (section 2.5.2), it is beneficial for the control shoe should be identical in features apart from the instability technology of the US. This is common procedure in footwear evaluation to minimise the affects of appearance (Kong et al., 2015; Sterzing et al., 2014c) and influences on biomechanical measures. Typically, US are heavier and have a reduced base of support compared to regular footwear that may influence kinematics and muscle activations. To account for this, the control shoe (CS) utilised the same shoe upper as the irregular midsole shoe (IM). The CS midsoles were developed with the midsoles of the original shoe (Li Ning Fengchao TD, Li Ning Co, Beijing). The medio-lateral midsole shape was cut to identical dimensions of the IM bags and aluminium weights (5g) were glued evenly to replicate the weight of the IM bags (Figure 3.7). The CS midsole weighed

234g and 215g and the irregular midsole shoe weighed 233g and 218g, for the male and female model respectively. Thus, weight and shape midsole differences were minimal. The heel to toe offset of the regular midsole was 10 mm.

Another advantage of the CS design, was that the same shoe upper was used in both IM and CS trials because the midsoles were attached to the shoe uppers by Velcro. This allowed the same reflective marker placement (section 4.1) in all conditions and neutral positions of segments and joint centres were determined from a static trial in the CS only. Utilising a global neutral configuration is advantageous because the absolute angular differences between midsole conditions can be compared. Therefore, kinematic differences between shoe conditions could not be attributed to different marker location, which has been suggested to give different results (James et al, 2015). The deforming material of the IM rubber bags causes the ankle angle is slightly more dorsiflexed in the foot-flat period of stance in walking and running (Figure 5.5). This is because of the virtual kinematic foot segment applied which normalises the ankle zero degrees from the static trial (see section 4.1 for details).

Figure 3.7. The regular midsole removed from the original training shoe upper and cut into same width as IM bags (top right), weights attached (middle right), Velcro attached (bottom right) and used as the control shoe (left).

CHAPTER 4.

General Methods

This chapter provides details of the commonly used equipment and techniques applied in the experimental chapters of this thesis. Specific approaches of each study are provided in the methods section of each chapter.

4.1. Kinematic Model

In the experimental studies (Chapters: 5, 6 and 7), the thigh, shank and foot were modelled as rigid segments in Visual 3D (C-Motion, Rockville, MD, USA). Three-dimensional motion capture systems recorded retro-reflective markers (Ø: 14 mm) which were attached to the lower-limbs and tracked in 6 degrees of freedom, according to the CAST technique (Cappozzo et al., 1995). All joint kinematics were calculated according to the joint co-ordinate system, with cardan sequence of rotations: XYZ (Grood & Suntray, 1983). Positive joint angles about the X-axis represent flexion; and positive ankle Y-axis values reflect eversion.

The kinetic foot segment was defined by anatomical markers attached on the shoe over metatarsal heads 1 and 5, and on the lateral and medial malleoli. Anatomical markers were attached to the malleoli to define the distal end of the shank, and to the medial and lateral femoral epicondyles to define the proximal end. The origin of the segment co-ordinate systems of both the foot and shank were located between the midpoints of the proximal markers (Figure 4.1). Additionally, a virtual kinematic foot segment was created to normalise the ankle joint angle to 0° from the standing static trial, as recommended by C-motion. This was created by precisely aligning the shank co-ordinate system and the virtual foot segment. This foot model does account for the oblique alignment of the subtalar joint axis because the Y-axis (frontal plane) is defined as the cross-product of the Z-axis (transverse plane) and X-axis (sagittal plane) of the shank segment. Therefore, not all of the inversion/eversion rotation of the ankle is captured and is acknowledged as a limitation of this work.

The distal thigh was defined by the knee epicondyle markers and proximally by the hip joint centre. The hip joint centre, and thigh origin, was estimated to be located 8 cm medially from the greater trochanter marker, which is suggested to be representative for healthy individuals (Richards, 2008 pp 121), and accurate for sagittal plane hip and knee kinematics/kinetics (Sinclair et al., 2014). Previous US studies and footwear research in general demonstrates the ankle is more prone to kinematic adaptations rather than more proximal joints (For example: Boyer & Andriacchi., 2009; Sterzing et al., 2015; Chambon et al., 2014). Therefore, a lower limb marker set was used in this thesis that did not include the measurement of hip kinematics. This foot-leg kinematic model was deemed acceptable for examining adaptations of the lower limbs to different conditions at the foot-ground interface during locomotion.

Tracking marker clusters were attached on the lateral side of the thigh (4 markers at LJMU; 5 markers at Li Ning) and shank (4 markers). The foot tracking markers attached to the shoe at the proximal posterior, distal posterior and lateral heel counter. Position and orientation of co-ordinate segment axes relative to tracking markers were determined from a static trial in the anatomical position wearing the control shoe.

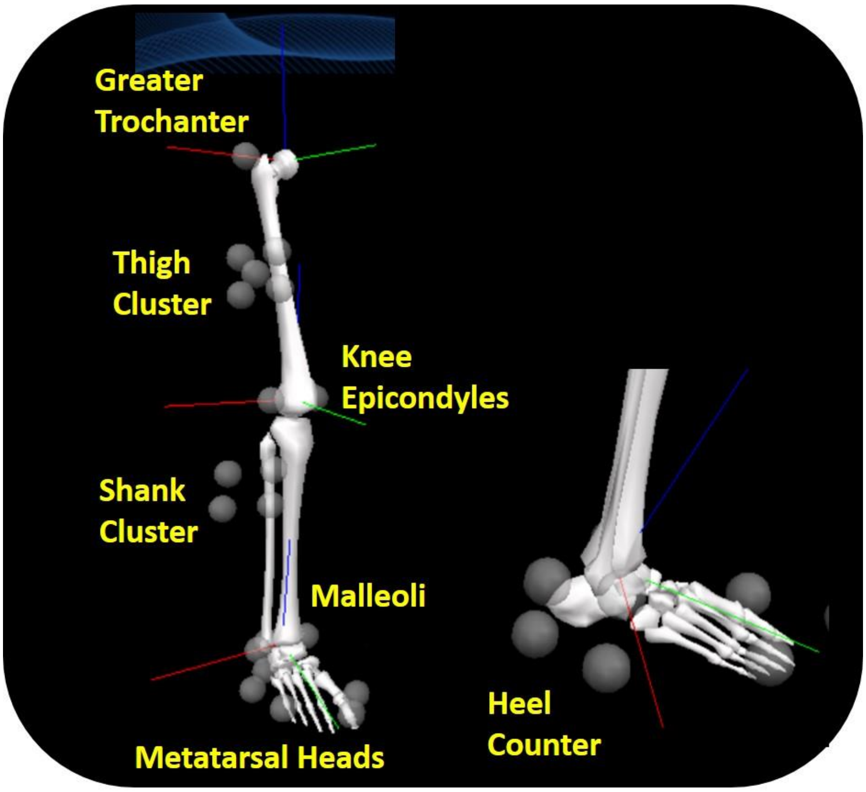

Figure 4.1. Marker set for the right lower-limb. The thigh, shank and foot co-ordinate axes and markers are displayed. The kinematic foot segment has the same co-ordinate axis as the shank.

4.2. Kinematic data processing

To ensure kinematic data had good repeatability in this thesis, the maximum accepted average camera residual from the calibration procedure was 0.5 mm, which is in the lower range suggested by Qualisys (Qualisys AB, Gothenburg, Sweden). Once the marker data was digitised, the marker co-ordinate data were filtered with a 4th order, zero lag, Butterworth digital filter. Pilot work determined the choice of frequency cut-off, by subjectively observing which best removed the high frequency noise (jagged lines), without over smoothing and losing the true peaks and troughs of the signal. To do this joint angular data were overlaid after filtering with different cut-off frequencies: 6, 10, 15, 20 and 30Hz for walking (Figure 4.2), running (Figure 4.3) and the forward lunge (Figure 4.4) data from this research. Based on these the author deemed, 10Hz to be suitable for walking and lunge movements, and 20 Hz for running.

Joint angular velocities were examined in Chapter 7. The same frequency cut-offs are not always adequate for joint angles as velocities (Sinclair et al, 2013). Consequently, the appropriateness of frequency cut-off was also checked by visually observing the joint velocity curves (Figure 4.5). The 10Hz cut-off frequency underestimated joint angular velocity peaks and therefore these were filtered with a 20Hz cut-off frequency.

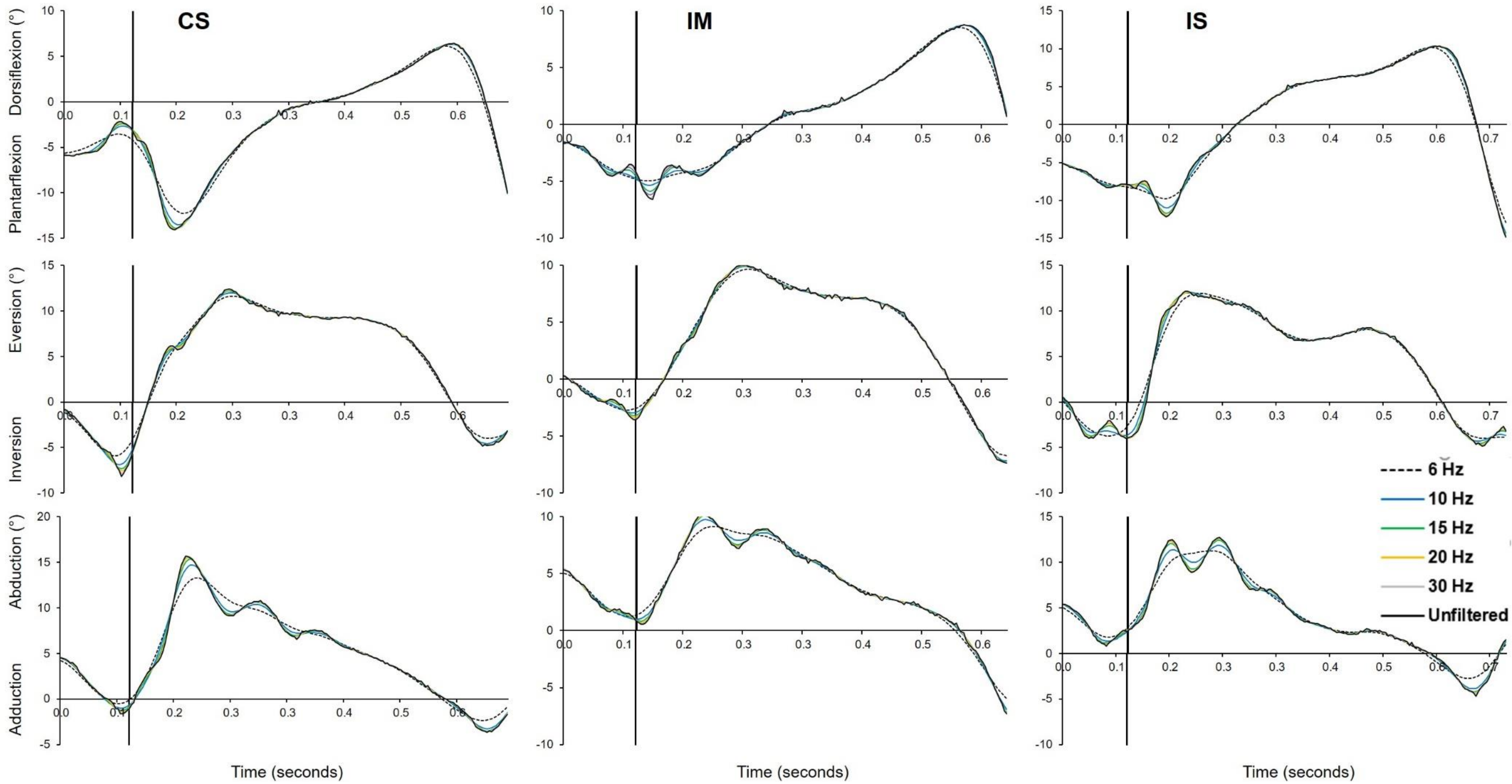

Figure 4.2. Effects of filtering with different frequency cut-offs for a representative ankle angle during **walking**. The sagittal (top), frontal (middle) and transverse plane (bottom) are displayed for the control shoe (left), irregular midsole shoe (middle and irregular surface (right). Time is from 100ms before initial contact (black line) until toe-off.

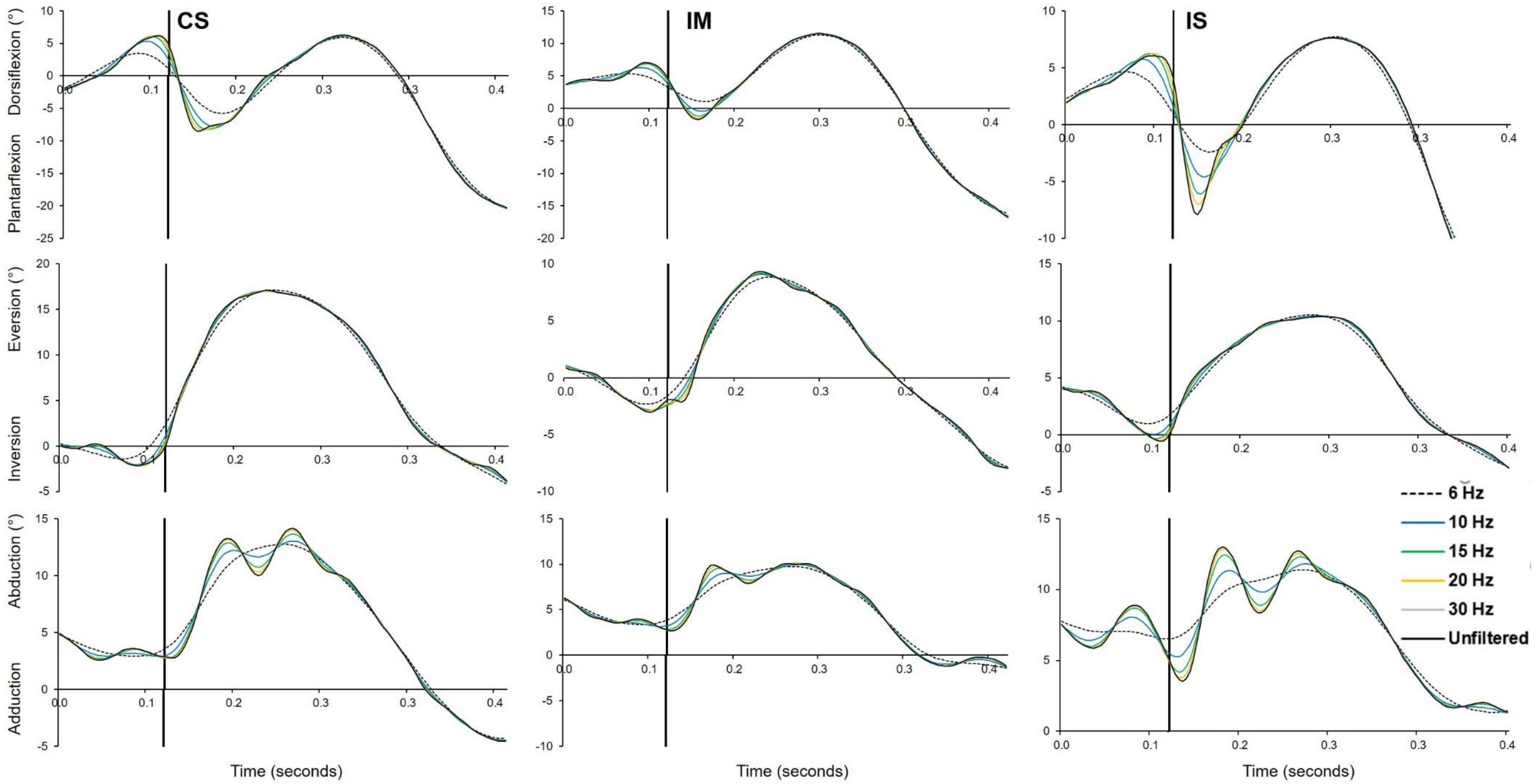

Figure 4.3. Effects of filtering with different frequency cut-offs for a representative ankle angle during **running**. The sagittal (top), frontal (middle) and transverse plane (bottom) are displayed for the control shoe (left), irregular midsole shoe (middle and irregular surface (right). Time is from 100ms before initial contact (black line) until toe-off.

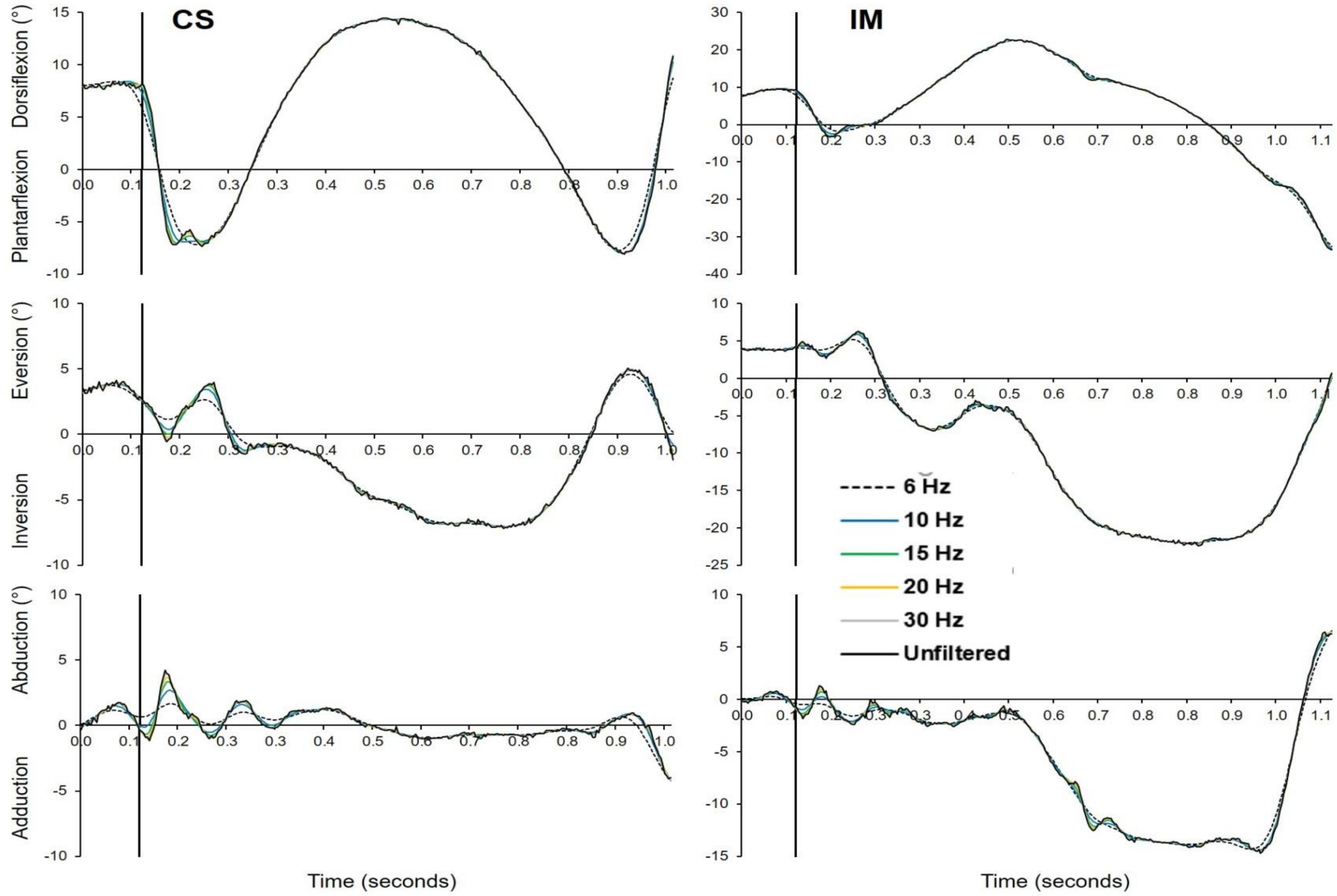

Figure 4.4. Effects of filtering with different frequency cut-offs for a representative ankle angle during a **forward lunge**. The sagittal (top), frontal (middle) and transverse plane (bottom) are displayed for the control shoe (left), irregular midsole shoe (middle and irregular surface (right). Time is from 100ms before initial contact (black line) until toe-off.

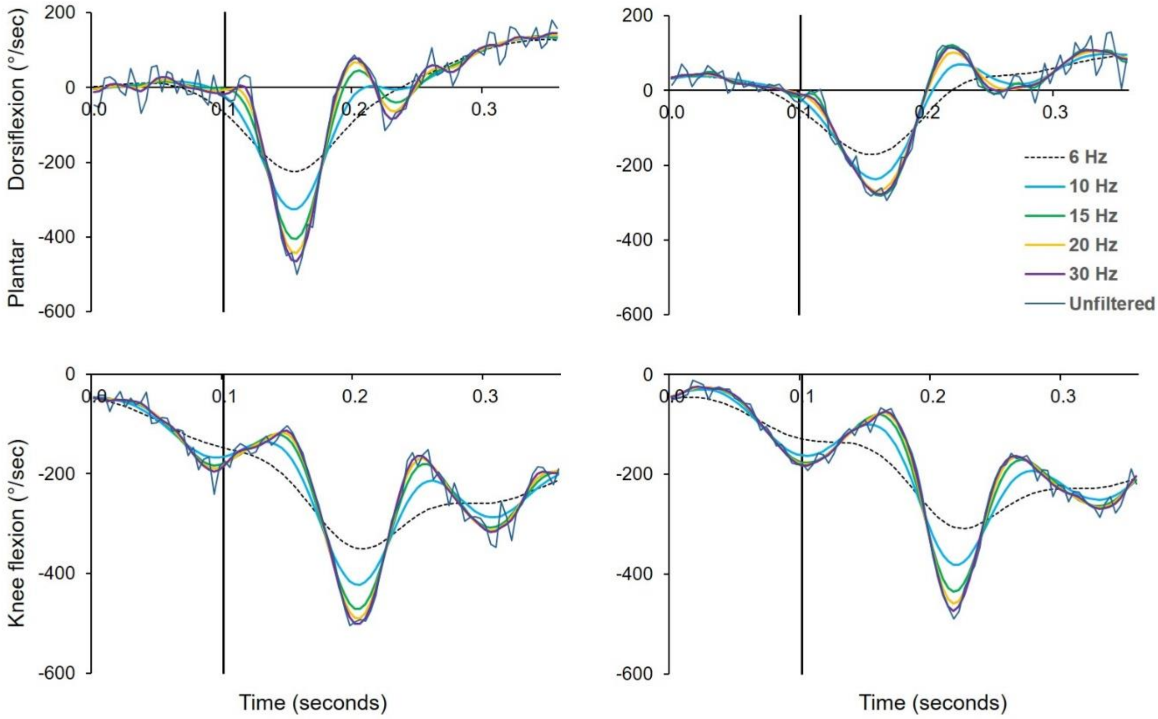

Figure 4.5. A typical sagittal ankle (top) and knee (bottom) angular velocity curves in the control shoe (left) and IM (right) during a **forward lunge**. Time is from 100ms before initial contact (black line) until 0.3 seconds after. Effects of filtering with different frequency cut-offs are displayed. Notice the peak flexion (negative) velocities that were analysed are underestimated with 6, 10 and 15 Hz cut-offs.

4.3. Electromyography system

A wireless telemetric system recorded the myoelectric signal from selected lower limb muscles (TeleMyo DTS, Noraxon Inc., Scottsdale, AZ, USA) at 3 kHz in Chapters: 5, 6 and 7). The sensor electrodes had a differential amplifier attached (specification: input range ± 3.5 mV, input impedance > 100MOhm, common mode rejection > 100 dB, base gain of 500 and baseline noise < 1 uV root mean square) (Figure 4.6). The analogue signal was hardware bandpass filtered (10-500Hz) and converted to a digital signal by the transmitter data acquisition system (16 bit). Before bi-polar electrodes were attached, skin was prepared by shaving, abrading and cleansing with alcohol to reduce impedance. Once all electrodes were attached participants were asked to activate the selected muscles to check for cross talk.

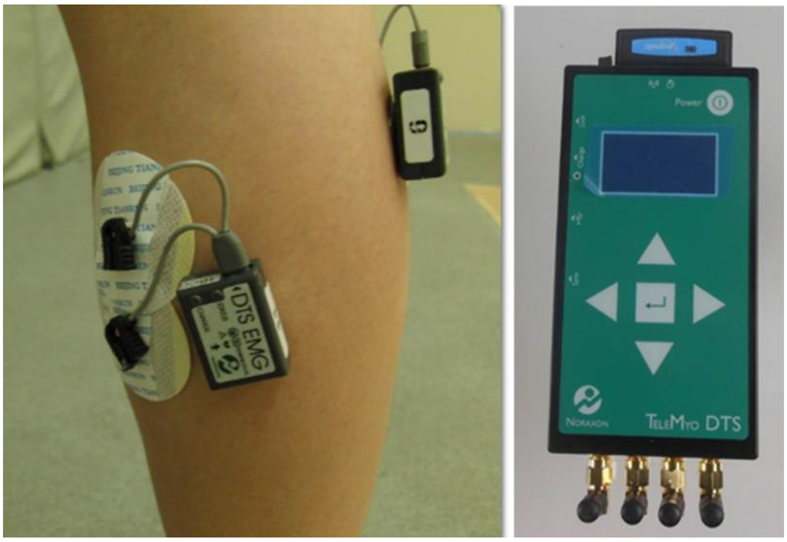

Figure 4.6. A sensor electrodes attached to the gastrocnemius medialis (left) and the DTS belt receiver (right).

4.4. Electromyography data processing

The EMG data were processed in Visual 3D software. The raw EMG signals were digitally band-pass filtered with a bi-directional 4^{th} order Butterworth filter (cut-off frequencies: 10 and 400Hz). After this, data were full wave rectified and smoothed with root-mean square filters (point number specified in chapters) to create linear envelopes. To reduce inter-subject variation, EMG data for each muscle were normalised using the dynamic peak method. To calculate this, linear envelope signals were normalised to the average peak value, across all analysed phases, for each muscle of the control condition of the same locomotion type. This has been applied in previous US studies (Romkes et al., 2006; Buchecker et al., 2012) and has been demonstrated good reliability and sensitivity (Albertus-Kajee et al., 2011).

4.5. Statistics

As no previous data were available in the irregular midsole condition, a priori power analysis was performed on results of the previous fundamental study (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. 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.

Data from the experimental studies in this thesis are reported as the parameter means and standard deviations. To determine significant differences between shoe-surface conditions inferential statistical processing was performed in SPSS (v22, SPSS Inc, Chicago, IL, USA). The study designs were all repeated measures, so dependent on whether data met parametric assumptions, either parametric or non-parametric tests were implemented. Parametric assumptions were met if the data were normally distributed and did not contain any outliers. Normality of data were checked using the Shapiro-Wilk test and visually verified with boxplots, as recommended by Ghasemi and Zahediasl (2012). Data were checked for outliers by the outlier labelling rule of 3 times the inter-quartile range (IQR). Although a multiplier of 2.2 has been recommended, SPSS only displays IQR multipliers of 1.5 and 3 (Hoaglin & Iglewicz, 1987). The more extreme outlier labelling rule was selected because parametric tests were preferred due to their increased statistical power and they were deemed robust enough to cope with outliers just outside the normal distribution. Bonferonni post hoc tests were applied to reduce the likelihood of a type 1 error rate, as recommended (Field, 2013: p547). To indicate the magnitude of any effects, effects size were computed from Chapter 6 onwards to indicate the relevance of findings.

CHAPTER 5.

Biomechanical and neuromuscular adaptations to a randomly deforming shoe midsole compared to an uneven surface during locomotion

5.1. Abstract

A shoe with unsystematic perturbations, similar to natural uneven terrain, may increase instability compared to current unstable footwear technologies. This study compared the instability of a shoe with unpredictably random midsole (IM) deformations, an irregular surface (IS) and a control shoe-surface whilst treadmill walking and running.

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.

Posture alterations at initial contact revealed active adaptations in IM and IS to maintain stability whilst walking and running. Variability of the gait cycle pattern and lower limb kinematics increased in IS compared to the control across locomotion types. Similarly increased variability (coefficient of variation) of 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). Tibilais anterior pre-activation reduced and gastrocnemius activation increased in IM compared to the control whilst walking and running. During running, peroneus longus activation increased in IM and IS. Stability was perceived lowest and injury risk highest in IM, and energy consumption higher in IM and IS compared to the control whilst walking and running (all $p<.001$).

Results indicate IM enhanced instability relative to the control and simulated certain locomotion adaptations of IS, although less pronounced. Thus, a shoe with unpredictable instability revealed potential as a novel instability-training device.

5.2. Introduction

Unstable shoes (US) are specifically designed to create instability with the purpose of providing functional benefits to the user. The concept behind unstable footwear is similar to traditional instability training devices, such as Swiss balls, BOSU balls and wobble boards (Figure 2.1). Such equipment reduces the base of support causing instability, which can be observed through an increase of movement variability (Cimadoro et al., 2013). Regular use promotes coordination improvements through the need to find alternative movement solutions to maintain stability. 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. US in contrast, may allow habitual training during walking, running or aerobic exercises.

As discussed in Section 2.3.2, increased muscle activation is one of the acute responses of wearing US. In MBT increased gastrocnemius medialis activations whilst walking (Price et al., 2013; Romkes et al., 2006) may strengthen the ankle muscles (Kaelin et al., 2011). However, other studies report no significant increases in muscle activation whilst walking (Forghany et al., 2014; Horsak & Baca, 2013; Nigg et al., 2006a; Sacco et al., 2012; Stöggl et al., 2010) or running in unstable shoes (Sobhani et al., 2013).

Improved balance is another suggested training effect of regularly wearing US (Section 2.3.3.). The higher centre of pressure range 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 a dynamic systems perspective has shown this may not always be the case (van Emmerik, & van Wegen, 2000). A better determinant of the postural system's ability could be assessing time to return to a stable posture after an external perturbation (reactive balance). Females older than 50 years old did improve their reactive balance, shown by the time to maintain centre of pressure within a 5x5 mm^2 area after sagittal surface perturbations, after regularly wearing MBT for 8-weeks. Although no significant improvement was found compared to a control group (Ramstrand et al., 2010).

The inconsistent findings in these MBT interventions may due to the measurements being collected while participants are either wearing MBT (Landry et al., 2010) or barefoot (Ramstrand et al., 2010). 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 unstable shoe design, Reflex Control has a thin sole bar along the longitudinal foot axis, compared to the most frequently tested 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 et al., 2015). In addition, reactive balance during one-legged standing improved after a training program in Reflex Control, but not in MBT (Turbanski et al., 2011).

Moreover, although movement variability is initially increased whilst walking in MBT shoes, this variability has been shown to reduce throughout 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. Uneven surfaces may provide a superior training modality by creating a continually changing and unpredictable instability. It is established irregular surfaces (IS) destabilise locomotion, shown by increased muscle activations, a cautious gait pattern and increased movement variability whilst walking (Gates et al., 2012; Marigold & Patla, 2008; McAndrew et al., 2010; Sterzing et al., 2014a; Thies, et al., 2005; Voloshina et al., 2013) and running (Sterzing et al., 2014b; Voloshina & Ferris, 2015). However, IS are often not accessible in urban areas for convenient and frequent use. An alternative and novel solution would be to develop footwear that causes similar irregular and unpredictable instability, during repeated ground contacts. Consequently, we developed a training shoe with random irregular midsole (IM) deformations (Chapter 3). The purpose of this study was to compare the locomotion instability induced by IM and IS during walking and running.

It was firstly hypothesised that IM and IS would cause a similarly increased variability of plantar pressures, and movement of the global gait parameters and the joint kinematics. It was envisaged the differences in variability compared to the control condition would be greater in the plantar pressures and distal joint kinematics rather than global gait parameters. This would indicate if the developed irregular shoe-surface conditions developed do cause unpredictable instability. Secondly, it was hypothesised there will be postural adjustments and increases in muscle activations to maintain balance in IM and IS. These hypothesises were applicable to both walking and running.

5.3. Methods

5.3.1. 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.

5.3.2. Shoe-surface Conditions

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

1. An IM shoe and a regular surface
2. A regular midsole shoe and an IS
3. The regular midsole shoe and a regular surface as a control condition (CC)

Details of these are provided in Chapter 3.

5.3.3. Plantar pressures

In-shoe plantar pressure was measured for the left foot using a Pedar X System insole (Novel, Munich, Germany), containing 99 sensors with spatial resolution between 168 to 178 mm^2, at 200 Hz. Insoles were calibrated directly preceding this research and insoles pressure values set to zero directly before each trial was started. The pressure insoles were 2.2 mm thick and worn in the left and right shoe.

Plantar pressure analysis was performed using Novel Multimask software (Novel GMBH, Munich, Germany). The plantar sole was segregated into six masks by dividing into the rearfoot, midfoot and forefoot at 30 %, 30 %, and 40 % of the insole length respectively (Cavanagh P & Ulbrecht J, 1994) and also medially and laterally at 50 % of the insole width. Peak pressures were calculated for each mask region and for the overall plantar area. Relative loads were calculated as a percentage of the force-time integral for each area to the total force-time integral. To quality check the data, peak pressures of all steps were observed. This revealed extreme peak pressure ‘hot-spots’ occurring in IM trials. Due to these artefacts in the data the plantar pressure results are reported in Appendix A.

5.3.4. 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 (Sterzing et al., 2014a; Sterzing et al., 2014b). This time to accommodate was selected to ensure the immediate adaptations to the shoe-surface conditions were investigated, which is comparable to previous US research (Stöggl et al., 2010; Price et al., 2013; Table 2.2). The 30 seconds data collection time was selected because this enabled enough strides to be collected to ensure the variability results were reliable (Appendix C). Surface EMG and lower limb kinematics were recorded synchronously from the subjects' left leg.

5.3.5. Kinematics

Kinematics were captured by a seven-camera motion analysis system at 300 Hz (Vicon Peak, Oxford, UK). Nineteen reflective markers were attached to define the left thigh, shank and foot segments, as detailed in Section 4.1. Additionally, a marker was placed on the tip and distal posterior heel counter of the right shoe to calculate step length.

Marker data processing details are provided in Section 4.2. 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 (Appendix B). Vertical velocity change of the midpoint between the heel and toe markers identified gait events during walking (O'Connor et al., 2007) and the vertical acceleration of the heel and tip of shoe markers was used during running (Maiwald et al., 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.

Global gait parameters 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. It was expected that IM and IS would have a greater effect during loading in the first half of stance when loading rates are fastest and due to the unpredictable instability. 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.

5.3.6. Electromyography

Surface electromyography (EMG) recorded the left gastrocnemius medialis, peroneus longus, tibialis anterior, bicep femoris, vastus medialis and vastus lateralis muscle activations. Specification of the system, sensor electrodes and details of the analogue to digital conversion and skin preparation are provided in Section 4.3. 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 SENIAM international recommendations (Hermans et al., 2000). Once all electrodes were attached and fixated with tubular bandages, subjects were asked to activate the selected muscles to check for cross talk. 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 and normalised according to the procedures outlined in Section 4.4. The filter to create the linear envelope was an 11-point root mean square moving average filter. 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.

5.3.7. 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, (Figure 5.1) (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).

Figure 5.1. Experimental set-up. 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.

5.3.8. Statistics

All steps (41.0 ± 2.6 for running and 28.6 ± 1.5 for walking) were analysed to compute the mean and variability of all variables for each participant (Appendix C). Variability of gait cycle and kinematic variables were calculated with the coefficient of variation (CV: Mean/Standard Deviation*100), to ensure the ratio of variance was comparable across the different conditions. Most variables followed parametric assumptions (see Section 4.5 for detail) and a one-way repeated measures ANOVA were applied to walking and running data separately. The non-parametric Friedman test were applied to the variables where assumptions were violated. Appropriate Bonferroni adjusted post hoc tests were applied to confirm differences between conditions. An alpha level of 0.05 was set. No corrections were made for the high number of variables tested, due research exploring effects of IM for the first time.

5.4. Results

5.4.1. Kinematics

Differences to the mean global gait parameters were generally small between conditions but consistent across participants whilst walking (Table 5.1) and running (Table 5.2). The gait cycle in IM was characterised by shorter, thus more frequent steps. Variability of the global gait parameters was significantly increased in IS compared to CC. Across variables this increase was greater in running (49 ± 13%) than walking (24 ± 4%) (Figure 5.2). Variability in IM was locomotion dependant. During walking IM had a higher variability than CC (26 ± 14%), similar to IS, whereas during running IM had a lower level of variability similar to CC (3 ± 2%).

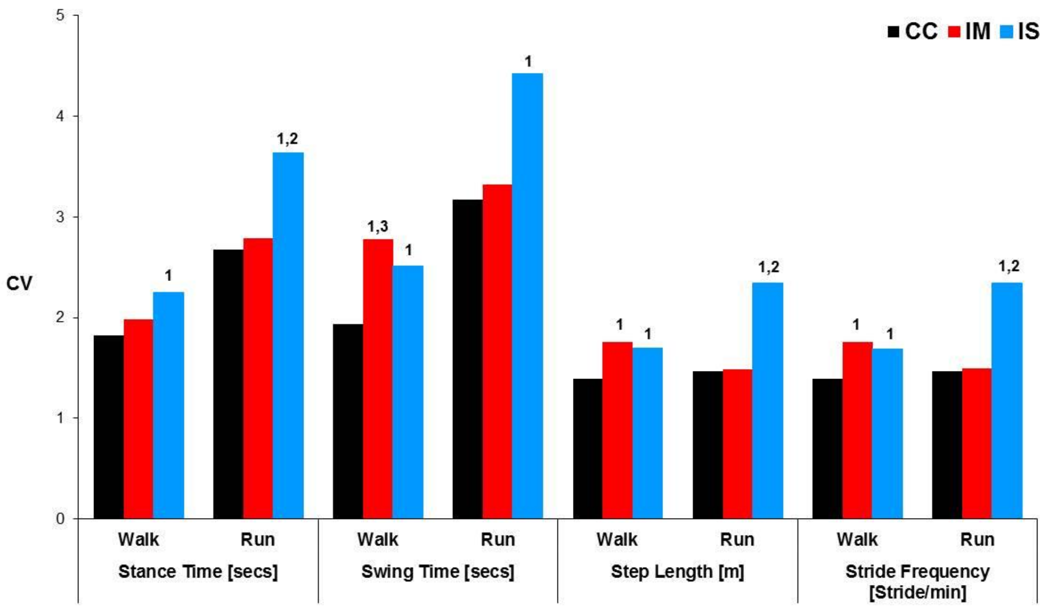

Figure 5.2. Variability (CV) of gait cycle parameters across participants. 1 = significantly greater than CC, 2 = significantly greater than IM, 3 = significantly greater than IS (p<.05). Notice IM has higher values similar to IS during walking and lower values similar to CC during running.

At initial ground contact, knee flexion increased in IM compared to IS and CC whilst walking and running (Table 5.1, Table 5.2). 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 (Figure 5.3). Ankle angles at initial contact ranged around zero degrees, therefore variability could not be computed. Therefore, the standard deviation is reported in Figure 4.4.

During stance, maximum ankle eversion reduced in IM whilst walking and running (Figure 5.4.). 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 most variable in IS, with IM having similar higher variability levels of frontal ankle range of motion (walk: IM = 109%, IS = 94% > CC, run: IM = 143%, IS = 126% > CC) and knee flexion (walk: IM = 60%, IS = 62%> CC, run: IM = 19%, IS = 31% > CC) (Figure 5.3, Figure 5.5) across locomotion. Similarly, to frontal ankle range of motion CV, the standard deviation of maximum frontal ankle angle increased in IS and IM compared to CC (Figure 5.4).

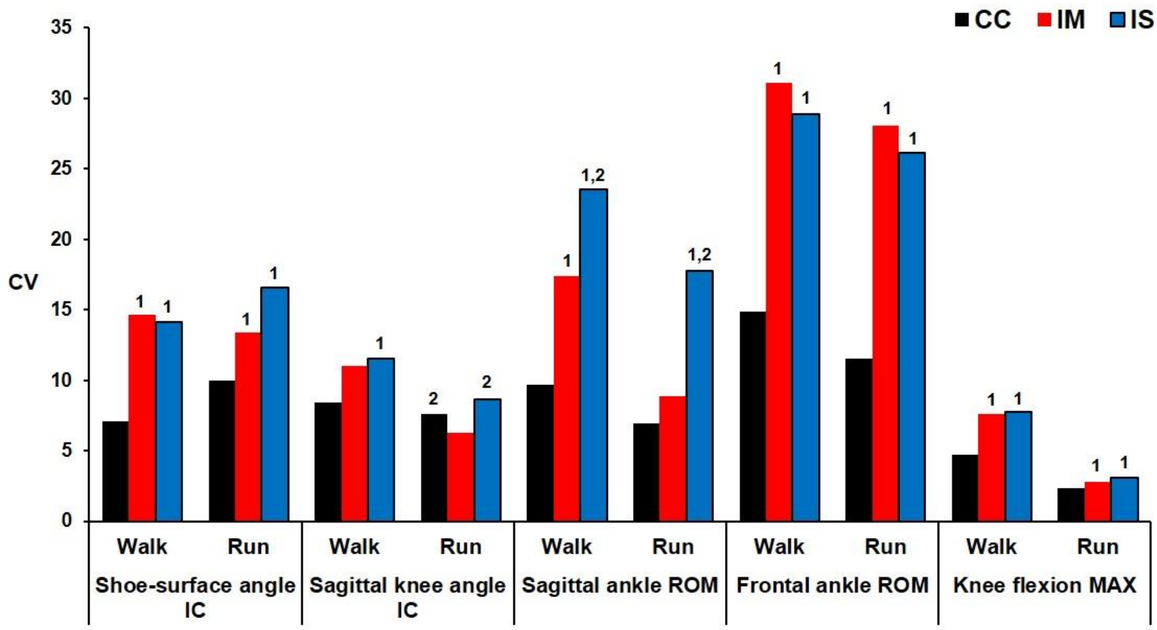

Figure 5.3. Variability (CV) of joint/segment angles at initial contact (IC) and during stance across participants. ROM = range of motion. 1 = significantly greater than CC, 2 = significantly greater than IM (p<.05).

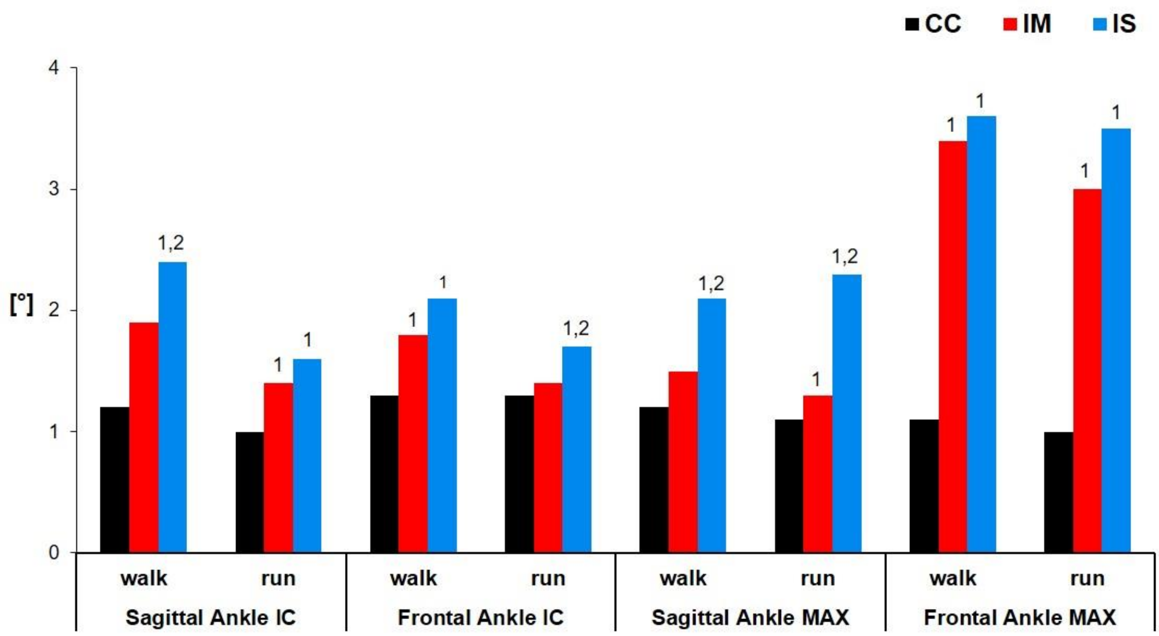

Figure 5.4. Ankle angle variability (SD) at initial ground contact (IC) and maximum (MAX) values during stance across participants. 1 = significantly greater than CC, 2 = significantly greater than IM (p<.05).

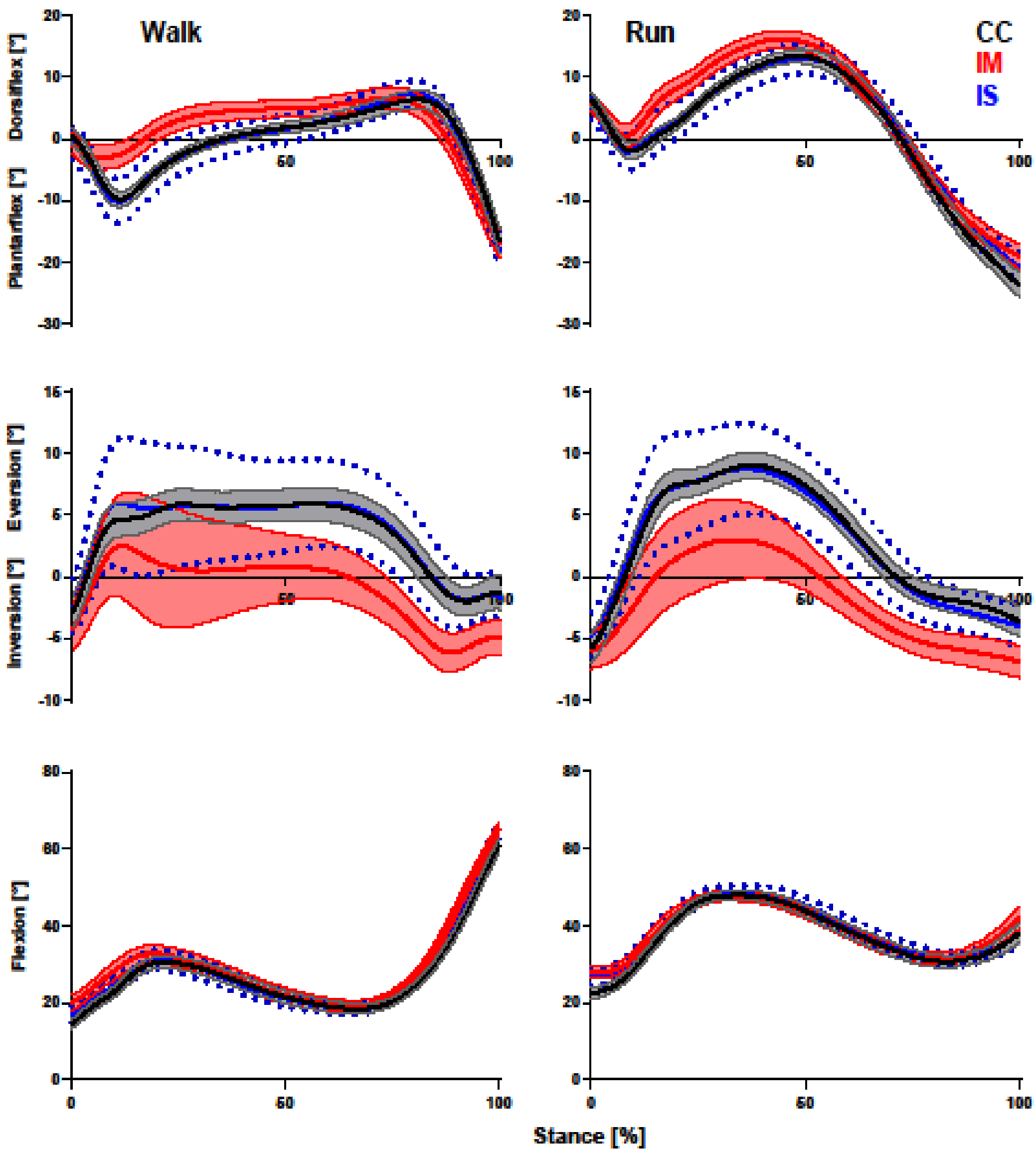

Figure 5.5. Joint angle plotted against stance phase during walking and running across subjects. Solid thick lines represent mean sagittal ankle angle (top), frontal ankle angle (middle) and sagittal knee angle (bottom). CC illustrated by the black line, IM the red line and IS the blue line (mostly overlaid 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.

Table 5.1. Mean (SD) global gait parameters and kinematics during walking across participants

Walking	Variable	CC	IM	IS	ANOVA p-value	Post hoc result
Gait cycle	Stance time [secs]	0.63 (.04)	0.62 (.04)	0.65 (.02)	.010	IS > IM
	Swing time [secs]	0.38 (.02)	0.36 (.02)	0.38 (.02)	<.001	IS, CC > IM
	Step length [m]	0.87 (.05)	0.86 (.05)	0.9 (.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
	*Ankle Inversion MAX [°]	5.9 (3.1)	11.5 (6.1)	10.1 (7.1)	.005	IM, IS > CC
	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. *Maximum value across all steps recorded.

Table 5.2. Mean (SD) global gait parameters and kinematics during running across participants

Running	Variable	CC	IM	IS	ANOVA p-value	Post hoc result
Gait cycle	Stance time [secs]	0.35 (.02)	0.34 (.01)	0.34 (.02)	.014	CC >IS
	Swing time [secs]	0.39 (.04)	0.38 (.04)	0.4 (.04)	.018	IS > IM
	Step length [m]	0.77 (.04)	0.75 (.03)	0.75 (.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 (3.4)	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)	28.0 (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
	*Ankle Inversion MAX [°]	8.7 (3.4)	11.1 (4.8)	9.3 (4.7)	.084	---
	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. *Maximum value across all steps recorded.

5.4.2. Electromyography

Electromyography results showed differences mostly occurred in the shank muscles for both walking (Table 5.3) and running (Table 5.4). Tibialis anterior activation reduced during pre-activation and loading in IM whilst walking compared to CC and IS. During pre-activation whilst running, tibialis anterior activation reduced in IM and IS compared to CC. Peroneus longus activation 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 greater pre-activation in IM than CC during walking and running.

Table 5.3. Normalised mean (SD) electromyography data during pre-activation and loading phases across participants during walking

Muscle	Phase	CC	IM	IS	ANOVA p-value	Significance
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 5.4. Normalised mean (SD) electromyography data during pre-activation and loading phases across participants during running

Muscle	Phase	CC	IM	IS	ANOVA p-value	Significance
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	---

5.4.3. Subjective 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 consumption was perceived greater for IM and IS than CC during walking and running (Table 5.5).

Table 5.5: Subjective perception scores (Mean (SD)) during walking and running across participants

Variable	Locomotion	CC	IM	IS	ANOVA p-value	Significance
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

5.5. 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 variability of global gait parameters 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 instead of 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 (Table 5.1 and 5.2) and more variable frontal ankle motion (Figure 5.3) in IM and IS compared to the control 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 et al., 2005). Also, the maximum ankle inversion angles measured in IM and IS (range: 9.3-11.5^{o}) were within the normal range of frontal ankle motion (17.3^{o}) (Ottaviani et al., 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 variability of global gait parameters in IM and IS during walking, and IS during running is an indicator of reduced stability and has been linked to risk of falling (Sekiya et al., 1997; Hausdorff, 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 (MacLellan & Patla, 2006), 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.

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 et al., 2005). Performance can be achieved consistently through a variety of movement pathways, increasing adaptability to perturbations (Davids et al., 2006; Latash, 2012; Wilson et al., 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 et al., 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 et al., 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 lack of increased peroneus longus activation during walking in IM and IS relates to the perceived risk of injury and energy requirement being lower 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 et al., 2014a). During pre-swing qualitative visual inspection of the peroneus longus EMG activation also appears to increase in IM and IS compared to CC whilst walking and running in some participants. This is likely to stabilise the ankle for push-off, preventing ankle inversion, and aligns with other unstable footwear that creates instability in the medial-lateral direction (Price et al., 2013). However, this requires quantitative evaluation in IM and IS to confirm this effect. The other common finding was a reduced tibialis anterior activation in IM and IS, particularly in IM whilst walking. This result supports previous observations on irregular surfaces (Hettinga et al., 2005; Voloshina et al., 2013), and in unstable shoes (Nigg et al., 2006).

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 et al., 1997) and overground (Wheat et al., 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). The irregular treadmill surface developed did provide continuous unpredictable perturbations, but 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. Also, other characteristics of the IM shoe may have influenced the results other than the varying regional midsole deformation profiles. Specifically the objects inside the IM were perceived by participants and may have caused discomfort that altered their posture at initial contact, for example the flatter foot placement. Furthermore, the lack of cushioning may increase initial loading and cause the higher movement variability during early stance.

5.6. Conclusion

In conclusion, we have created a novel shoe that provides continuously random perturbations and compared the acute adaptations to an irregular surface, which is known to induce instability. The motivation for developing IM was to have a more challenging stimulus than existing US, thus providing greater functional training benefits. The IM successfully increased biomechanical and perceived instability relative to a stable shoe and simulated certain adaptations of an unpredictable IS during walking and running. An additional training benefit of the IM, 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 US, by assessing the adaptability to unpredictable perturbations after regular use.

Chapter 6 investigates whether the unpredictable perturbations of IM required additional joint stability strategies compared to the predictable perturbations of an US. Whether greater training benefits could be acquired is then implicated.

CHAPTER 6.

Lower-limb joint stiffness and muscle co-contractions strategies to instability footwear during locomotion

6.1. Abstract

Unstable shoes (US) continually perturb gait which can train the lower limb musculature, but muscle co-contraction and joint stiffness strategies are not well understood. A shoe with a randomly perturbing midsole (IM) may enhance these adaptations. This study compares ankle and knee joint stiffness, and ankle muscle co-contraction during walking and running in US, IM and a control shoe in 18 healthy females. Ground reaction forces, three-dimensional kinematics and electromyography of the gastrocnemius medialis and tibialis anterior were recorded. Stiffness was calculated during loading and propulsion, derived from the sagittal joint angle-moment curves. Ankle co-contraction was analysed during pre-activation and stiffness phases. Ankle stiffness reduced and knee stiffness increased during loading in IM and US whilst walking (ankle, knee: $p = .008, .005$) and running ($p < .001$; $p = .002$). During propulsion, the opposite joint stiffness re-organisation was found in IM whilst walking (both joints $p < .001$). Ankle co-contraction increased in IM during pre-activation (walking: $p = .001$; running: $p < .001$), and loading whilst walking ($p = .003$), not relating to ankle stiffness. Results identified relative levels of joint stiffness change in response to instability from footwear, providing new evidence of how stability is maintained at the joint level. The unpredictable instability from IM was a more challenging stimulus than the predictable instability of the US, requiring additional joint stability adaptations.

6.2. Introduction

Unstable shoes (US) can be used as functional training devices, as increased muscle activation is required to make postural adjustments and keep the body balanced. This often presents as increased levels of co-contraction whilst walking in US (Horsak et al., 2015; Buchecker et al., 2012a). Co-contraction can distribute internal forces more evenly (Baratta et al., 1988), and may be important for injury prevention (Hirokawa et al., 1991). Mechanical joint stiffness is derived by muscle stiffness, which the neuro-muscular system controls by adjusting muscle activation level (Hogan, 1984; Lee et al., 2006). If this relates to the increased co-contraction at the joint to help stabilise locomotion remains unknown. A stiffer joint will resist displacement from an external perturbation, possibly preventing excessive joint motion and injury (Riemann et al., 2002). On the other hand, moderately reduced range of motion has also been linked with reduced limb stability (Kim & Lockhart, 2012; Salsich & Mueller, 2000).

Previous studies have investigated stiffness adaptations on surfaces with varied hardness. Rotational stiffness about the ankle and knee joint both increase during hopping on softer surfaces (Farley et al., 1998) and whilst running in softer midsole shoes (Baltich et al., 2015), compared to a harder surface and shoe condition. The soft shoe-surface interface may have induced some instability, although in a predictable manner. Similarly to the soft shoe-surface interfaces, joint stiffness has been found to increase in response to upper limb unpredictable perturbations (Burdett et al., 2001). Voloshina and Ferris (2015) found increased leg stiffness during running on an uneven treadmill surface. If this also applies to the lower-limb joints has yet to be investigated.

Joint stiffness has predominantly been based on the angle-moment relationship in previous locomotion studies. The calculation involves dividing the relative change of joint moment by the change in joint angle. Sometimes a linear regression is applied, the coefficient is taken as the value of stiffness, and the r^2 values give an indication of how well the regression fits the data. Although this technique is not a true representation of mechanical stiffness it does reveal how compliant the joint is (Latash & Zatsiorosky, 1993), and thus adaptations to shoe perturbations. Ankle stiffness has been estimated in the foot-flat period during stance whilst walking (Gabriel et al., 2008; Kim & Lockhart, 2012) and running (Hamill et al., 2014; Kuitunen et al., 2001; Baltich et al., 2015). This is a relatively large phase of the gait cycle to take as a single stiffness measure which may obscure subtle changes. Sekiguchi and colleagues (2015) found ankle stiffness increased 2.6 times between the early and late phases of the second ankle rocker period of healthy adults' walking.

As mentioned, US are reported to increase muscle activation (Landry et al., 2010). However, some studies report no increases in muscle activation whilst walking (Nigg et al., 2006a) or running (Sobhani et al., 2013), indicating the US tested were not challenging the neuro-muscular system enough to invoke a response. Plus, effects of US instability are short term, and a less predictable and more challenging training stimulus has been suggested to enhance training effects (Stöggl et al., 2010). A developed irregularly deforming midsole (IM) with unpredictable instability simulates the kinematics of locomotion on uneven surfaces (Chapter 5) and may provide an enhanced training stimulus, requiring heightened joint stiffness and co-contraction strategies. Such a device may be particularly valuable for females because there have been a few reports of gender differences in levels of joint stiffness. Baltich and colleagues (2015) detected females further increased knee joint stiffness levels during running in softer midsole shoes, showing higher sensitivity than the males. Contrarily, Gabriel and colleagues (2008) found reduced ankle stiffness during the third ankle rocker in females during walking. These differences have been linked to females having greater joint laxity and reduced muscle strength; further challenging their joint stability and perhaps making them more susceptible to changes to shoe instability.

Therefore, this study compares ankle and knee joint stiffness and ankle muscle co-contraction of females when walking and running in response to predictable and unpredictable shoe instability. It was hypothesised for walking and running:

1. US and IM would increase ankle and knee joint stiffness compared to a control shoe
2. IM would further increase joint stiffness compared to US in the loading period of stance
3. Increased ankle joint stiffness would be associated with increased muscle co-contraction

6.3. Methods

6.3.1. Participants

Eighteen healthy females were recruited in this study (25.8 ± 2.5 years, 166.6 ± 4.3 cm, 61.8 ± 5.9 kg). Inclusion criteria required participants to be self-reported injury free, heel-toe runners, wear shoe size UK 5.5 ± .5 and have no previous experience with US. All participants took part in recreational exercise (5.5 ± 2.5 hours/week). This study was approved by Liverpool John Moores research ethics committee and participants were informed of the aims of the study and gave written consent prior to testing.

6.3.2. Protocol

Participants walked and ran in three shoe conditions: an US providing predictable instability, IM providing unpredictable instability and a regular shoe as a control (CS) (Figure 6.1). Details of IM and CS are provided in Chapter 3. The US tested was the

Bubble Gym (Li Ning, China), which is characterised with a protruding rocker around the midfoot and smaller protrusions in the rearfoot and forefoot regions. The outsole configuration is aimed to create instability, but its fixed structure means it is predictable. The weight of US, IM and CS was 321g, 218g and 215g respectively.

Figure 6.1. Shoe conditions. Unstable shoe (top), irregular midsole shoe (middle) and control shoe (bottom).

Before overground trials were captured, participants were familiarised to each shoe condition by walking for 90 seconds at 5 km/hr, followed by running at 8 km/hr on a treadmill. During overground trials, a walking speed of 5km/hr (±5%) and running speed of 8km/hr (±5%) were monitored using photocell timing gates (Brower Timing System, Utah, USA) that were placed 1.5 m each side of the force plate. Thus, locomotion speed is consistent with Chapter 5. Before data collection, participants had practice trials to ensure they could land with their right foot on the force plate without targeting. In each shoe condition 20 successful trials were collected where participants walked and ran within the required speed range and their right foot entirely landed on the force plate. The order of experimental locomotion was walk followed by run in the same shoe condition. The CS condition was always first to avoid potential crossover effect from US and IM, whose order was mixed between participants.

6.3.3. Data Collection

Right lower limb kinematics were recorded at 500 Hz by an eight-camera motion analysis system (Qualisys AB, Gothenburg, Sweden). Nineteen reflective markers were attached to define the right thigh, shank and foot segments, as detailed in Section 4.1. Neutral positions of segments and joint centres were determined from a static trial. Kinetic data from a force plate (Kistler 9281B, Amherst, NY, US) were collected in the middle of the runway at 1000Hz.

Surface electromyography (EMG) of the gastrocnemius medialis and tibialis anterior were recorded. Specification of the system, sensor electrodes and details of the analogue to digital conversion and skin preparation are given in Sections 4.3. Pre-gelled bi-polar Ag/AgCl circular electrodes (Noraxon Dual Electrodes, Noraxon, AZ, US) each with a diameter of 10mm an inter-electrode spacing of 17.5mm, were placed parallel to the muscle fibres, on the most prominent part of the muscle belly. All biomechanical measurements were synchronised.

6.3.4. Data Processing and Analyses

Marker data were digitised using Qualisys Track Manager Software (Qualisys, Inc., Gothenburg, Sweden) and further processed together with force plate and EMG data in Visual 3D (C-Motion, Rockville, MD, USA). To reduce the likelihood of artefacts in the joint moment calculation, the same 4^{th} order zero-lag Butterworth filter was applied to kinematic and kinetic data with a cut-off frequency of 10Hz for walking and 20Hz for running (Kristianslund et al., 2012; Section 4.2). Joint moments were calculated using the traditional inverse dynamics approach. The external ground reaction force and centre of pressure location is calculated from the force plate. For each segment the known proximal and distal positions (from marker data) and mass, centre of mass and moment of inertia (based on anthropometric calculations made by Dempster, 1955) are used to calculate the unknown internal joint reaction forces and subsequently the joint moments. Assumptions of this calculation are that the segments are rigid bodies and connect to form a linked model. Full details of this approach, are explained elsewhere (Robertson et al., 2014).

A 20 N threshold determined stance phase of locomotion. Ankle and knee joint angles were calculated, as well as, internal joint moments relative to the co-ordinate system of the proximal segment normalised to body mass (Nm/kg). Sagittal-plane ankle and knee torsional stiffness (k_{joint}) were calculated from the gradient of a least squares linear regression of the ratio change in joint moment (ΔM) to joint angle (Δθ):

$$k_{joint} = \frac{\Delta \mathrm{M}}{\Delta \theta} \tag{1}$$

During walking and running two linear angle-moment phases at each joint were identified to analyse joint stiffness, an earlier period termed the loading phase and a later period denoted as the propulsive phase (Figure 6.2). To improve the fit of the linear regression, the start and end of the stiffness phases were determined by the peak or trough of angle or moment which occurred later and earlier respectively. During walking, ankle stiffness was analysed during the first third (loading phase) and last third (propulsive phase) of the second ankle rocker period. The middle third was excluded from analyses during walking due to non-linearity in US and IM (Figure 6.3). During running, ankle stiffness was similarly analysed in the first half (loading phase) and second half (propulsive phase) of the second ankle rocker. Knee stiffness was analysed in walking and running after a 5% bodyweight threshold was reached until peak knee flexion angle/moment (loading phase).

The 5% bodyweight was selected rather than initial contact to minimise inaccuracies in the position of the ground reaction force vector at lower forces. From peak knee flexion until maximum knee extension, knee stiffness was analysed during walking (propulsive phase) and until the maximum of the anteroposterior ground reaction force during running (propulsive phase). Pilot work determined these linear phases by visual observation of the angle-moment curves and verification with r^2 values; the linear regression was deemed to follow the angle-moment relationship when r^2 values were greater than .90. To determine whether differences in joint stiffness were caused from a relative change in angle or moment the ranges in the loading and propulsive phases were calculated.

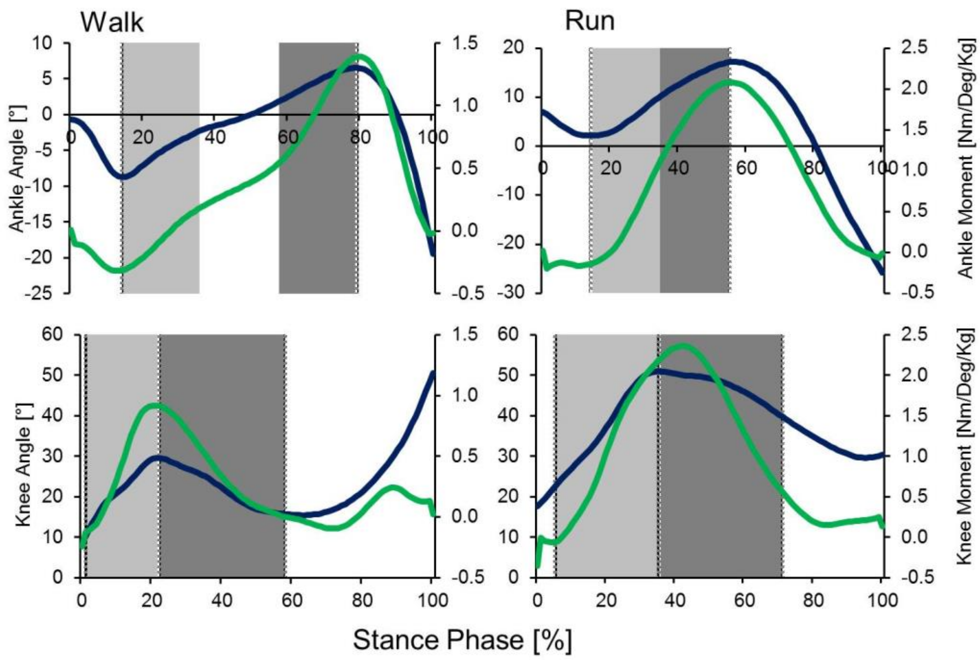

Figure 6.2. Illustration of joint angle and moment phases analysed in stiffness calculations of a typical subject in the control shoe. Angle (blue) and moment (green) for the ankle joint (top) and knee joint (bottom) during walking (left) and running (right). Time is normalised to stance phase. The lighter grey represents the loading phase and the darker grey the propulsive response.

The EMG data were processed and normalised for walking and running separately, according to the procedures outlined in section 4.4. The filter to create the linear envelope was a 51-point root mean square moving average filter. Muscle co-contraction index (CCI) of the gastrocnemius medialis and tibialis anterior (GM-TA) was calculated to estimate ankle co-contraction about the sagittal plane. Based on previous unstable footwear studies (Horsak et al., 2015; Buchecker et al., 2012a) and wanting to quantify ratio and magnitude of muscle activations, the following calculation were applied to calculate muscle co-contraction index (CCI) (Rudolph et al., 2001):

$$CCI = \frac{lowEMGi}{highEMGi} * (highEMGi + lowEMGi) \qquad (2)$$

where lowEMGi is the muscle with the lower amplitude-normalised value at each time point (i). The highEMGi is the corresponding muscle with a higher normalised value. The first part of the equation estimates the ratio and the latter part the magnitude. The CCI can range between 0, where both muscles are not activated, to 200 where both muscles are fully activated (Rudolph et al., 2001). Mean muscle activation and CCI was calculated in the following periods: a pre-activation phase (the 100ms before heel-strike), and during the ankle stiffness loading and propulsive phases.

6.3.5. Statistics

Pilot work confirmed the average joint angle-moment curve, normalised to 101 data points, gave the same joint stiffness results as analysing all 20 trials separately and taking the mean (Appendix D). Therefore, the data from the average angle-moment curve of all trials in each shoe condition per participant were compared statistically. Co-contraction index was calculated in every trial and averaged for each participant across shoe conditions.

Data met parametric assumptions (see Section 4.5). A one-way repeated measure ANOVA with Bonferroni adjusted post hoc tests were conducted for walking and running data separately ($p<.05$). Effect size (η^2) was calculated to indicate the relevance of findings. A strong effect size was defined by $\eta^2 > 0.5$, moderate between 0.5 and 0.3 and low < 0.3 (Field, 2013).

6.4. Results

6.4.1. Joint Kinematics/Kinetics:

Differences in the ranges of joint angles and joint moments were found between shoe conditions for walking (Table 6.1) and running (Table 6.2).

Table 6.1. Mean (SD) ranges of joint angles and moments during **walking** across participants. Positive ranges represent joint flexion, negative joint extension.

Variable	Joint	Phase	CS	US	IM	Effect Size	Significance
joint motion range	ankle	loading	8.42 (1.65)	11.83 (2.40)	8.40 (2.04)	.61	US > CS, IM p < .001
		propulsion	4.98 (1.77)	4.18 (1.83)	2.41 (1.25)	.79	CS > US > IM p < .001
	knee	loading	16.87 (2.40)	18.29 (2.26)	15.18 (2.92)	.56	US > CS > IM p < .001
		propulsion	-12.98 (3.70)	-13.80 (3.76)	-17.05 (5.29)	.54	IM > CS, US p < .001
joint moment range	ankle	loading	-0.57 (0.11)	-0.61 (0.13)	-0.48 (0.10)	.39	CS, US > IM p = .001
		propulsion	0.84 (0.17)	0.84 (0.16)	0.75 (0.18)	.36	CS, US > IM p = .002
	knee	loading	1.05 (0.16)	1.09 (0.26)	1.00 (0.29)	.11	p = .130
		propulsion	-0.91 (0.21)	-0.98 (0.23)	-0.93 (0.24)	.07	p = .276

Table 6.2. Mean (SD) ranges of joint angles and moments during **running** across participants. Positive ranges represent joint flexion, negative joint extension.

Variable	Joint	Phase	CS	US	IM	Effect Size	Significance
joint motion range	ankle	loading	9.77 (4.26)	12.39 (4.19)	12.71 (5.88)	.20	US > CS p = .046
		propulsion	6.41 (1.42)	6.43 (1.31)	5.76 (1.93)	.09	p = .204
	knee	loading	28.97 (2.82)	27.30 (2.60)	25.24 (4.84)	.45	CS > US > IM p < .001
		propulsion	14.35 (4.89)	13.98 (6.42)	13.26 (4.86)	.04	p = .483
joint moment range	ankle	loading	1.27 (0.26)	1.28 (0.32)	1.01 (0.25)	.44	CS, US > IM p < .001
		propulsion	0.91 (0.20)	0.91 (0.15)	0.94 (0.20)	.01	p = .762
	knee	loading	2.51 (0.31)	2.54 (0.39)	2.55 (0.41)	.02	p = .700
		propulsion	2.00 (0.47)	2.02 (0.55)	1.82 (0.57)	.26	US > IM p = .015

6.4.2. Joint Stiffness

The regression showed the stiffness phases analysed followed a very close to linear relationship. Across shoe conditions, joints, phases and locomotion mean (SD) r^2 values ranged between 0.94 (0.04) and 0.99 (0.01). Variations of angle-moment curves between shoe conditions could be observed for individual participants at the ankle joint during walking and running (Figure 6.3.), but were not so distinguishable for the knee joint (Figure 6.4). Significant joint stiffness walking results are presented in Table 6.3 and running results in Table 6.4. In the loading phase, US had reduced ankle stiffness compared to CS, with IM similar to US, but not significantly different to CS during walking. During running, IM had the lowest ankle stiffness and US had reduced levels compared to CS. Knee stiffness increased in IM and US compared to CS in walking and running. During running, IM also had increased stiffness compared to US.

In the propulsive phase, IM had increased ankle stiffness compared to US and CS during walking and compared to US during running. At the knee joint, IM had reduced stiffness compared to CS and US during walking.

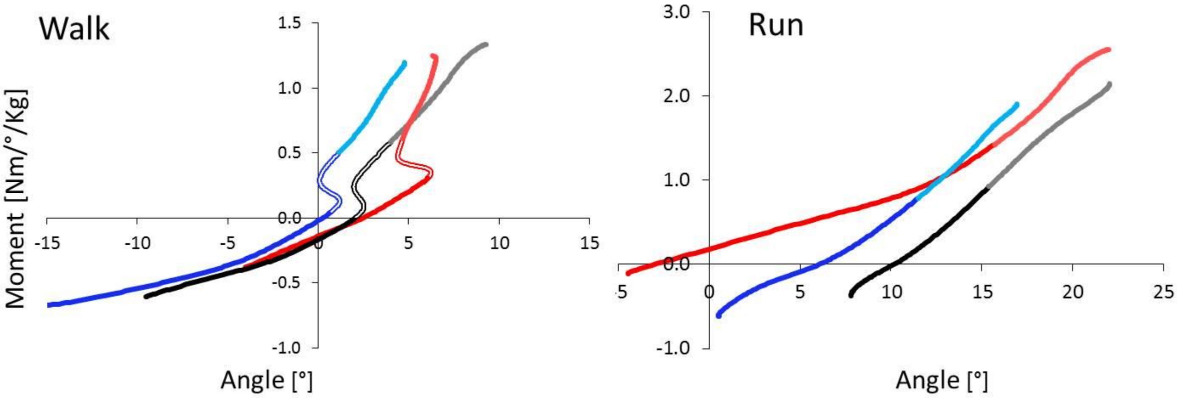

Figure 6.3. Ankle stiffness (angle-moment curve) of a typical subject in the irregular midsole shoe (red), unstable shoe (blue) and control shoe (black) during walking (left) and running (right). The changing colours represent the loading phase (brighter) and propulsive phase (lighter). During walking, notice the lower ankle stiffness (shallower gradient) in US during loading and stiffer ankle (steeper gradient) for IM in the propulsive phase. The hollow line represents the non-linear period which was not analysed. During running, IM had the least stiff and CS most stiff ankle in loading.

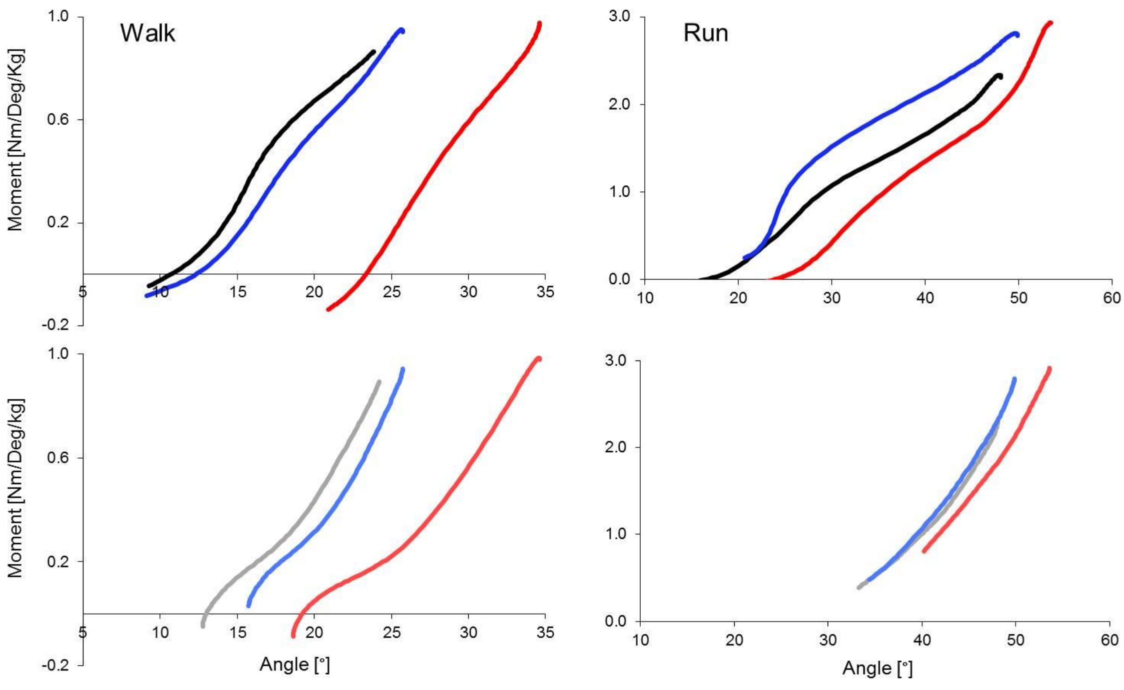

Figure 6.4. Knee stiffness (angle-moment curve) of a typical subject in the irregular midsole shoe (red), unstable shoe (blue) and control shoe (black) during walking (left) and running (right). The loading phase (top, brighter) and propulsive phase (bottom, lighter). During walking in IM, notice the slightly higher knee stiffness (steeper gradient) during loading, and the lower knee stiffness (shallower gradient) during propulsion.

Table 6.3. Mean (SD) ankle and knee joint stiffness across participants in the loading and propulsive phases during **walking.**

Variable	Joint	Phase	CS	US	IM	Effect Size	Significance
joint stiffness	ankle	loading	.065 (.018)	.051 (.013)	.059 (.015)	.29	CS > US p = .008
		propulsion	.191 (.052)	.235 (.104)	.375 (.192)	.51	IM > CS, US p < .001
	knee	loading	.066 (.011)	.071 (.017)	.077 (.021)	.30	IM, US > CS p = .005
		propulsion	.069 (.010)	.068 (.011)	.056 (.009)	.68	CS, US > IM p < .001

Table 6.4. Mean (SD) ankle and knee joint stiffness across participants in the loading and propulsive phases during **running.**

Variable	Joint	Phase	CS	US	IM	Effect size	Significance
joint stiffness	ankle	loading	.126 (.019)	.097 (.024)	.076 (.023)	.73	CS > US > IM p < .001
		propulsion	.141 (.027)	.141 (.024)	.158 (.031)	.24	IM>US p = .020
	knee	loading	.089 (.014)	.093 (.014)	.102 (.023)	.43	IM > US > CS p = .002
		propulsion	.153 (.028)	.157 (.026)	.151 (.022)	.09	p = .228

6.4.3. Electromyography

Significant differences in EMG activation and ankle muscle co-contraction were found between shoe conditions for walking (Table 6.5) and running (Table 6.6). In the pre-activation phase, GM-TA CCI significantly increased in IM compared to US and CS during walking and running. In the loading phase during walking, IM also had significantly higher GM-TA CCI, as well as, in the propulsive phase compared to US only.

Table 6.5. Mean (SD) normalised EMG muscle activation and co-contraction during **walking** across participants.

Variable	Muscle	Phase	CS	US	IM	Effect Size	Significance
muscle activation	GM	pre	9.8 (10.8)	9.3 (6.3)	19.6 (12.2)	.43	IM>CS, US p = .002
		load	14.7 (7.1)	16.1 (7.1)	21.9 (7.4)	.54	IM>CS,US p<.001
		prop	50.0 (6.0)	50.0 (8.7)	45.9 (15.1)	.06	p = .376
	TA	pre	39.1 (11.4)	40.6 (15.4)	23.5 (10.9)	.64	CS, US>IM p < .001
		load	12.1 (4.3)	11.5 (3.9)	12.2 (5.2)	.02	p = .740
		prop	11.7 (5.7)	10.3 (5.0)	11.5 (6.3)	.18	p = .052
muscle co-contraction	GM-TA	pre	8.6 (6.8)	9.5 (4.3)	20.3 (12.7)	.48	IM>CS, US p = .001
		load	9.9 (4.7)	11.0 (4.4)	13.7 (4.9)	.35	IM>CS, US p = .003
		prop	14.2 (6.6)	11.8 (4.5)	13.6 (5.7)	.24	IM>US p = .033

Table 6.6. Mean (SD) normalised EMG muscle activation and co-contraction during **running** across participants.

Variable	Muscle	Phase	CS	US	IM	Effect Size	Significance
muscle activation	GM	pre	10.4 (4.1)	9.1 (4.4)	17.6 (7.5)	.60	IM > CS, US p < .001
		load	46.7 (6.4)	42.2 (11.3)	38.0 (9.6)	.29	CS > IM p = .007
		prop	66.9 (3.1)	56.7 (7.8)	57.1 (13.1)	.32	CS > IM, US p = .006
	TA	pre	57.9 (3.4)	48.4 (10.3)	24.8 (8.2)	.89	CS, US > IM p < .001
		load	20.2 (10.0)	22.9 (11.5)	20.3 (9.7)	.13	p = .116
		prop	21.7 (11.1)	20.6 (10.6)	20.4 (11.0)	.04	p = .519
muscle co-contraction	GM-TA	pre	12.6 (5.6)	10.5 (5.2)	19.2 (7.5)	.54	IM > CS, US p < .001
		load	26.9 (14.6)	28.7 (13.7)	25.9 (12.2)	.05	p = 459
		prop	26.6 (13.1)	25.6 (10.6)	24.0 (3.6)	.05	p = .464

6.5. Discussion

This study compared lower-limb joint motion, moments, stiffness and ankle muscle co-contraction during walking and running in response to predictable (US) and unpredictable (IM) shoe instability. Contrary to our first and second hypotheses there was not a universal increase in joint stiffness due to these types of footwear, but relative levels of ankle and knee stiffness were reorganised. During loading, ankle joint stiffness decreased while knee joint stiffness increased whilst walking and running due to shoe instability (Table 6.4-6.5). This shows that IM and US require a reorganisation of joint stiffness compared to CS. The redistribution from the ankle to the knee was greater in IM than US whilst running. Oppositely, during propulsion in IM ankle joint stiffness increased and knee joint stiffness decreased whilst walking (Fig 6.2). Thus, when comparing IM and US it was shown that unpredictable instability induces greater adaptation requirements than predictable instability. Interestingly, changes to ankle stiffness in IM and US were not associated with a corresponding muscle co-contraction results (Tables 6.5-6.6), rejecting the third hypothesis. In fact, the opposite was found in IM during loading, suggesting joint quasi-stiffness does not align with the previously shown relationship between mechanical stiffness and muscle co-contraction (Hogan, 1984).

The reported knee joint stiffness values during loading are similar to previous studies for walking (Zeni & Higginson, 2009) and running (Baltich et al., 2015). However, due to the novel ankle stiffness phases and lack of literature during the propulsive phases no comparisons can be made. A similar reduction of ankle stiffness and increase of knee stiffness during loading has been found in forefoot strike (FFS) runners compared to rearfoot strike (RFS) runners (Laughton et al., 2003; Hamill et al., 2014). However, in our study participants did not change to a FFS pattern, detected by the shoe-floor angle remaining positive. Thus, the re-organisation of ankle and knee joint stiffness revealed a novel control strategy to deal with shoe instability.

The loading phase ankle and knee stiffness changes in IM and US is related to their position in the kinetic chain. The ankle has least time to respond to perturbations because of the close proximity to the ground. The reduced stiffness allows the ankle flexibility to better absorb any increased loading caused by instability. The knee counteracts the compliant ankle by becoming stiffer to enable a constant locomotion speed and stabilise further up the kinetic chain. The mechanism of altering joint stiffness was mostly through altering the range of motion, as reported previously (Dixon et al., 2010; Chambon et al., 2014). However, an interesting finding was the reduced ankle stiffness in IM during loading was altered through reducing the range of moment (Table 6.1, 6.2). This is proposed to be an active adaptation to increase stability in IM. The flatter foot placement in IM during locomotion (Chapter 5) has been related to a reduced ground reaction force moment arm about the ankle (Braunstein et al., 2010). This would limit the instability effects of IM by keeping the ground reaction force closer to the ankle joint centre. This initial joint stability study was concerned with the global changes to joint stiffness and ankle co-contraction in IM based on similar work on uneven shoe-surfaces rather than the ground reaction force moment arm which was beyond the scope of this research. Future work should clarify if this reduced ankle stiffness is due to changes to ground reaction force moment arm.

When encountering an unexpected slippery surface, increased ankle muscle co-contraction results in less hazardous slips (Chambers & Cham 2007). Therefore, it is likely participants enhanced their ankle stability through GM-TA co-contraction in IM. This contrasts the established consensus that muscle co-contraction increases joint stiffness (Hogan, 1984; Lee et al., 2006). For increased ankle co-contraction was associated with reduced ankle stiffness, derived by angle-moment curves, in IM during loading. Kuitunen and colleagues (2002) similarly found no increase in ankle quasi-stiffness but increased ankle co-contraction whilst running at higher speeds. The authors speculated this could be due to the constant Achilles tendon stiffness having the overriding effect, but this may not be true for the low running speed and walking trials in this study. Contrasting this, Hobara and colleagues (2007) studied hopping where athletes landed on their forefoot. They found the relative force of the triceps surae was greater hopping at higher heights by inhibiting the antagonist tibialis anterior activation, which prevented dorsiflexion excursion and increased leg stiffness. Therefore, the relationship between joint stiffness, calculated from the joint angle-moment curve, and muscle co-contraction is complex and task dependent.

The increased knee stiffness would transmit more load and is a possible risk factor for injury in unstable footwear. A stiffer knee joint has been associated with increased low back pain in running (Hamill et al., 2009) and knee osteoarthritis in walking (Dixon et al., 2010; Zeni & Higginson, 2009), but whether this is the cause or effect of injury/disease is unknown. Furthermore, the increased muscle co-contraction in IM would increase loading at the ankle joint. However, this was the first exposure participants had to the US. Based upon previous instability training interventions, it is envisaged the observed increases of knee joint stiffness and ankle co-contraction may decrease with more time to accommodate. Initial strategies to stiffen joints and increase co-contraction reduce over time as better movement control is learnt and these adaptations are no longer needed for joint stability (Chmielewski et al., 2005; Kim & Lockhart, 2012). Future investigations should therefore monitor adjustments in joint stiffness and muscle co-contraction over time to reveal long term training benefits and risks with unstable footwear.

Limitations in the current study are acknowledged. Accurately quantifying mechanical joint stiffness requires an external perturbation to be applied about the joint axis of rotation (Rouse et al., 2014), which was not part of the current research design. By modelling stiffness as a torsion spring in this study oversimplified the complexities of the various joint components and gave a single value of rotational stiffness and has been termed ‘quasi-stiffness.’ (Latash & Zatsiorsky, 1993). Quasi-stiffness was measured from the joint moment-angle curves, but this is only valid for linear portions of the relationship. Ankle stiffness during the middle third of the foot-flat phase in walking (Fig 2) could therefore not be quantified where the footwear instability may have had an important effect. Only ankle muscle co-contraction was quantified in this study because Chapter 5 showed no thigh muscle activation changes in IM compared to CS. However, US studies found increased knee muscle co-contraction while walking (Horsak et al., 2015; Buchecker et al., 2012a) and it cannot be ruled out the increased knee joint stiffness in US could have been related to muscle co-contraction. Furthermore, the increased knee joint stiffness may have been achieved by the increased gastrocnemius activation in IM because it crosses both ankle and knee joints. If this was the case, the association between quasi-stiffness and muscle co-contraction may hold true for the knee joint. The US had different mass and width dimensions compared to CS and IM that could not be altered without risking reducing any instability effects of the footwear, but we do not expect this to have changed the joint stiffness and muscle activation adaptations substantially. It is acknowledged the haptic sensation of the objects inside the rubber bags under the plantar sole in the IM shoe, which may have caused or contributed to the different joint stability responses rather than the unpredictable perturbations.

6.6. Conclusion

This study identified joint stiffness reorganisations aided stability in unstable footwear during walking and running. The irregular midsole (IM) exhibited unpredictable perturbations and required additional strategies for stability compared to a commercial unstable footwear model: ankle muscle co-contraction and joint stiffness for propulsion. These joint control mechanisms may be applicable to a natural uneven suraface. The greater level of instability caused by the developed IM shoe may induce enhanced neuromuscular training benefits over typical unstable footwear with a fixed outsole. Interestingly, ankle joint quasi-stiffness and muscle co-contraction decreased and increased respectively in the IM shoe during the loading phase of stance. Therefore, they should not be assumed to be always related to each other in dynamic locomotion. However, both measures revealed active adaptation to shoe instability.

Findings from Chapter 5 and 6 provide an overview of the acute biomechanical and neuromuscular adaptations to IM compared to a IS and US whilst walking and running. As recognised in Section 2.3.3, unpredictable instability may be an even more challenging stimulus during dynamic functional exercises. This is investigated in Chapter 7.

CHAPTER 7.

Instability effect of a shoe with random midsole deformations during forward and lateral lunges

7.1. Abstract

Unstable footwear may have positive training effects on the lower limb musculature and sensorimotor system during dynamic gym movements, similar to instability training devices. This study compared the instability of the developed irregular midsole (IM) shoe and the control shoe (CS) during typical gym movements: the forward and lateral lunge.

Seventeen young, female gym class participants completed two sets of ten forward and lateral lunges in CS and IM. Three-dimensional lower limb kinematics, ground reaction forces and thigh and shank muscle activations were analysed. Variables around initial ground contact, during the descending and ascending lunge phases were compared statistically ($p<.05$).

Results suggest responses to IM were similar in forward and lateral lunges in both sets. A flatter foot placement and increased vertical but reduced horizontal velocities at initial ground contact enhanced stability in IM by reducing the horizontal shear forces. Yet, the magnitude and variability of initial loading rates increased. During the descending phase, peak ankle and knee flexion velocities decreased in IM. Variability of frontal ankle motion increased in both phases in IM, by 27% and 39% on average, in the forward and lateral lunges respectively. Sagittal ankle and knee variability were similar between IM and CS. Peroneus longus and gastrocnemius medialis activations increased in the ascending phase in forward (17%, 24%) and lateral (10, 14%) lunges. Prior to initial contact, gastrocnemius medialis activation also increased in IM during forward lunges.

For active female gym goers, IM increased variability during initial loading and frontal ankle motion during lunges, which required postural changes and muscle activations to enhance stability. This demonstrated IM was a more demanding training stimulus compared to a stable shoe. Future work should investigate if regular use can lead to adaptations that benefit sports performance or reduce injury-risk.

7.2. Introduction

Instability training devices frequently have a reduced base of support or are made of soft materials (Figure 2.1) making them unstable, forcing the neuromuscular system to adjust to maintain balance. As mentioned in Section 2.2, this trains the sagittal (Behm & Anderson, 2005) and frontal shank muscles (Strøm et al., 2016) during aerobic exercises, which has been related to reducing the occurrence of ankle injuries (Mcguine & Keene, 2006). Balance sandals are designed like wobble boards and also increase shank muscle activations, as well as, improving balance during functional movements (Blackburn et al., 2003; Michell et al., 2006). US are designed similarly to instability devices and benefit wearers by allowing convenient training during daily life activities. Yet, how they influence functional movements is unknown. The IM required additional joint stability adaptations during locomotion and may also invoke greater instability in gym movements (Chapter 6).

Closed-kinetic chain exercises are advantageous because they are applicable to daily life, sports and involve multiple joint functions (Cordova et al., 1999). As such, the lunge is a functional training exercise that has been used to screen movement control in athletes (Cook, 2003). The proper lunge technique requires flexibility, stability and strength of the lower limbs, and incorrect practice may lead to injury (Kritz et al., 2009). Lunges on unstable support surfaces require higher levels of ability, and may not be suitable for populations with insufficient balance or muscle deficits. The lateral lunge is considered as a progression of the forward lunge, but has been less frequently investigated (Escamilla et al., 2008; Flannagan et al., 2004; Riemann et al., 2013). Novel investigation of the acute biomechanical adaptations of this established gym movement with US would establish if the exercises become even more demanding for the sensorimotor system.

Landry and co-workers found MBT increased muscle activations in the smaller, more distal ankle muscles whilst standing (Landry et al., 2010). However, there are conflicting findings of increases in muscle activations during locomotion (Table 2.1) and balance improvements after regular wear (Ramstrand et al., 2010; Nigg et al., 2009). Moreover, US investigations are limited to standing, walking and running. It is envisaged US may provide greater training effects in active participants during more dynamic functional movements that can be incorporated into gym exercise routines, similarly to instability training devices, but this has not been investigated. Furthermore, the instability induced by IM during walking and running (Chapter 5 and 6) may also be a more difficult training stimulus in gym exercises.

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 goers. Based on Chapter 5 and 6, it was hypothesised IM would create greater instability that would cause increased movement variability and varied ground reaction forces. This would be controlled by increased lower-limb muscle activations and shank and foot adjustments to enhance stability, particularly around initial ground contact and push-off.

7.3. Methods

7.3.1. Participants

Seventeen healthy females who regularly attended gym classes for at least one year participated in this study (21.6 ± 1.6 years, 166.3 ± 4.2 cm, 55.6 ± 3.5 kg, gym class experience 3.3 ± 2.0 years, classes 6.3 ± 1.9 hours/week). The University's research ethics committee approved the study protocol and all participants were informed about procedures prior to signing consent forms. All subjects 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., Liverpool, NY, USA).

7.3.2. Shoe Conditions

Two shoe midsole conditions were tested:

1. The IM
2. The regular midsole shoe as a control (CS)

Details of the irregular midsole shoe and control shoe are provided in Chapter 3.

7.3.3. Protocol

Participants completed two sets of ten forward lunge repetitions and two sets of ten lateral lunge repetitions in IM and CS whilst biomechanical measurements were recorded. The order of lunge movement and midsole condition was randomised to control for any learning or order effect. Between set one and set two of the same lunge movements there was a minimum 45 second rest, and between movements and changing shoe midsoles there was a 5-minute rest to minimise fatigue. Lunge frequency was controlled by a metronome (40 beats per minute) in order to prevent differences occurring between shoe conditions due to different ground contact times, as has been found during gait unstable footwear studies (Romkes et al., 2006; Forghany et al., 2014). The rate (20 lunges per minute) and technique (see below) was deemed suitable after consulting several gym instructors. On the first beat, the right foot should contact the ground before completing the lunge, and on the second beat signalled the right foot should leave the ground to return to the start position. The lunging technique was demonstrated to participants and verbally explained. Before each trial participants practiced the lunge speed and technique until they and the researcher were satisfied to start.

During all lunges participants were asked to place their hands on their iliac crests, look straight ahead and keep their trunk erect, as this is reported to affect lunge biomechanics (Farrokhi et al, 2008). For the forward lunge participants were briefed to start with legs shoulder width apart and take a right step forward. Then, flex the right knee until about 90° with the right thigh approximately parallel to the ground, and the back left leg lowered towards the floor (Figure 7.1). Following knee flexion, extend the right knee, to push back to the starting position. For the lateral lunge participants were informed to start in the same position and take a lateral step to the right. Then, flex their right knee until the right shank was in a vertical position over the right foot. During this instruction was given to keep the trunk straight, not to let the right knee move forward anteriorly and keep the left leg extended (Figure 7.1). After maximal knee flexion, the right knee extended to push back and return to the start position. Participants lunged to their preferred step length and width.

Figure 7.1. Example forward lunge (top) and lateral lunge (bottom) technique.

7.3.4. Kinematics

Motion capture, force plate and EMG data were synchronised and collected for participants' right leg only (the lunge leg). Motion analysis was recorded at 300 Hz by a seven-camera 3D Motion Analysis System (Vicon Peak, Oxford, UK). Nineteen reflective markers were attached to define the right thigh, shank and foot segments, as detailed in Section 4.1. Additionally, markers were placed on the tip of the left shoe and the distal posterior heel counter to calculate step width and length. 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. Marker data processing details are provided in section 4.2.

Qualitative observation of increased peroneus longus muscle activation from Chapter 5 suggested the irregular midsole shoe also created instability during the push-off phase in walking and running. Kinematic variables were therefore selected to determine the preparations for initial ground contact and the instability during the descending (initial contact until maximum knee flexion) and ascending phases (maximum knee flexion until toe-off) of the lunges. Step width and step length were calculated from the distance between the distal heel markers at initial contact. At the instant of initial ground contact sagittal and frontal plane shoe-surface angles and horizontal and vertical foot velocities were computed. Sagittal and frontal plane ankle and sagittal knee kinematics were analysed. During the descending phase, peak ankle and knee flexion velocities were calculated (Figure 4.5). Ankle and knee angles were calculated at initial contact, maximum values and at toe-off. Ranges of motion between initial ground contact to the maximum sagittal ankle/knee angle (ROM1), and from maximum sagittal angle until toe-off were determined (ROM2). Positive sagittal joint angles reflect flexion, and positive frontal ankle angle represents inversion.

7.3.5. Ground reaction forces

Participants' right lunge leg landed on a force plate (90 x 90 cm, AMTI OR6GT, Watertown, MA, USA) flush with the laboratory floor. Lunge ground contact time was determined using a 20 N threshold. Initial contact was offset by 2 frames earlier to ensure the event began at the initial rise of the vertical ground reaction, and was checked visually. The analogue signals were filtered by a 4^{th} order Butterworth filter with frequency cut-off of 50 Hz (Hamill et al., 2014). This frequency cut-off was selected because it removed noise whilst closely following the peaks and troughs of the data. Ground reaction forces were normalised to bodyweight. The initial peak in the vertical ground reaction force, and instantaneous and average loading rate were calculated (Laughton et al., 2003; Figure 7.2). The integral of the horizontal forces during descending and ascending phases were determined.

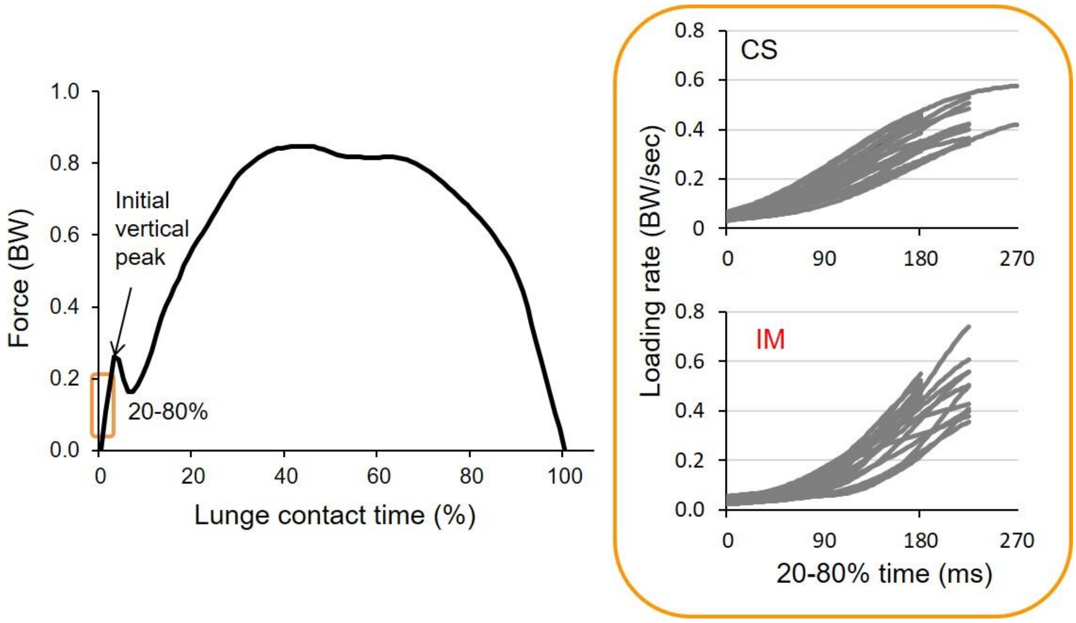

Figure 7.2. Average vertical ground reaction force during 20 forward lunges in the control shoe of a typical subject (left). Instantaneous and average loading rate was calculated between 20-80% period between initial ground contact and the first vertical peak (close up right). The variability and magnitude increased in IM (bottom) compared to CS (top).

7.3.6. Surface Electromyography

Surface EMG activity was recorded of the following right limb muscles: tibialis anterior, peroneus longus, gastrocnemius medialis, vastus medialis and biceps femoris. Specification of the system, sensor electrodes and details of the analogue to digital conversion and skin preparation are given in sections 4.3 and 4.4. The electrodes used were pre-gelled bi-polar Ag/AgCl circular electrodes (Tian run, Beijing, China) of 10mm diameter and were placed with an inter-electrode spacing of 25 mm, parallel to the muscle fibres, on the most prominent part of the muscle belly. Muscles were located according to SENIAM international standards (Hermans et al., 2000) and the electrodes and sensors were fixated with tubular bandages.

EMG data were processed and normalised to the average peak value of each muscle, across all analysed phases in forward and lateral lunges, of CS (section 4.4). The filter to create the linear envelope was a 61-point moving average filter. The mean amplitude was calculated in the following periods to supplement kinematic variables: the 100ms before initial ground contact (pre-activation) to assess feed-forward strategy, and the descending phase and ascending phase. Certain electrode data contained artefacts and were excluded from subsequent analyses. After exclusion the number of subjects (N) per muscle was: gastrocnemius medialis = 15, peroneus longus =15, tibialis anterior = 15, bicep femoris = 14, and vastus medialis = 13.

7.3.8. Statistics

The mean (SD) of all variables were computed, as well as, the variability (coefficient of variation (CV), as detailed in 4.3.7) of the global lunge pattern, joint angles and ground reaction forces. To check for learning effects between the first and second set of lunges within the same shoe conditions paired t-tests were performed. Results showed minimal differences, so the participants' means across set 1 and 2 were collapsed together. Across all variables 95% followed parametric assumptions for both IM and CS (see Section 4.5) and were compared by paired t-tests ($p<.05$). This was advantageous because it allowed the same computation of the effect size (Cohen's d), using the pooled standard deviation (Cohen, 1988). Values of 0.2, 0.5 and 0.8 are considered a small, moderate and large effect size, respectively. No corrections to the alpha level were made, due to the research being exploratory in nature.

7.4. Results

7.4.1. Global lunge characteristics

Few changes to the global lunge parameters were recorded in the forward lunge (Table 7.1) or lateral lunge (Table 7.2). Step width CV was not computed because some participants' mean values ranged around zero. During forward lunges there was an increased stance time and during lateral lunges step length was reduced in IM. Step length was also was more consistent, indicated by the reduced CV, in both lunge directions in IM.

Table 7.1. Mean (SD) global lunge parameters and variability during the **forward lunge** across participants

Forward Lunge		CS	IM	Effect size	Significance
Mean	Stance time [secs]	1.18 (.08)	1.22 (.10)	.769	IM>CS, p = .006
	Step length [m]	.724 (.10)	.718 (.08)	.187	p = .429
	Step width [m]	.100 (.04)	.098 (.04)	.083	p = .738
CV	Stance time	5.3 (1.7)	5.2 (1.2)	.043	p = .713
	Step length	3.2 (1.0)	3.0 (1.0)	.285	CS>IM, p = .039

Table 7.2. Mean (SD) global lunge parameters and variability during the **lateral lunge** across participants

Lateral Lunge		CS	IM	Effect size	Significance
Mean	Stance time [secs]	1.24 (.13)	1.27 (.11)	.355	p = .163
	Step length [m]	.701 (.08)	.681 (.08)	.641	CS>IM, p = .018
	Step width [m]	.072 (.05)	0.075 (.04)	.047	p = .849
CV	Stance time	5.3 (1.7)	5.4 (1.0)	.14	p = .802
	*Step length	3.6 (1.0)	2.9 (0.9)	1.40	CS>IM, p = .023

**Data were not normally distributed*

7.4.2. Kinematics

Kinematic results showed altered movement patterns in IM compared to CS during the forward lunge (Table 7.3) and lateral lunge (Table 7.4). At initial ground contact, participants opted for a flatter shoe surface angle, increased ankle plantarflexion and increased vertical but reduced horizontal foot velocity in IM during the forward and lateral lunges. Knee flexion at initial contact significantly increased in the forward lunge in IM. Sagittal ankle range of motion increased during the descending phase in the forward and lateral lunges in IM compared to CS. Peak dorsiflexion angle increased in IM compared to CS in both lunges. Knee flexion and ankle velocity in the lunge direction (forward lunge: sagittal, lateral lunge: frontal) was reduced in IM. In the forward lunge, knee range of motion decreased in IM compared to CS.

In the ascending phase, sagittal ankle range of motion increased in IM during forward and lateral lunges. During the forward lunge, sagittal ankle angle at toe-off in IM was significantly reduced and negative, indicating participants opted for plantarflexion as opposed to the dorsiflexed foot position in CS (Figure 7.3). No difference was observed in the lateral lunge, where participants utilised plantarflexion at toe-off in IM and CS (Figure 7.4). In the forward lunge, there was less knee flexion at toe-off in IM.

The frontal ankle angle profiles were different between the forward and lateral lunges (Figure 7.3 and Figure 7.4). During the forward lunge, the frontal ankle angles were rather similar to walking and running (Figure 5.4), with the ankle being more inverted in IM compared to CS. During the lateral lunges the ankle inverted after initial contact, followed by an ankle eversion movement until the end of the descending phase, then inverted again, before finally everting before toe-off.

Table 7.3. Mean (SD) kinematics during the **forward lunge** across participants

Forward Lunge	Period	CS	IM	Effect size	Significance
Sagittal Shoe-surface [°]	IC	36.2 (4.8)	30.2 (4.8)	1.337	CS>IM, p < .001
*Vertical Foot Vel [m/s]	IC	0.50 (.14)	0.57 (.13)	1.428	IM>CS, p< .001
Horizontal Foot Vel [m/s]	IC	0.41 (.10)	0.30 (.12)	.930	CS>IM, p = .001
Sagittal Ankle [°]	IC	12.9 (4.9)	9.0 (6.1)	.903	CS>IM, p = .003
	MAX	21.2 (4.6)	23.8 (4.9)	1.181	IM>CS, p< .001
	*TO	3.1 (9.4)	8.0 (14.6)	1.009	IM>CS, p=.001
	ROM1	23.3 (4.7)	24.7 (5.2)	.524	IM>CS, p = .046
	ROM2	30.6 (7.8)	39.4 (11.1)	1.180	IM>CS, p< .001
Peak Plantarflex Vel [°/sec]	ROM1	-289.0 (108.2)	-190.6 (58.4)	1.154	IM>CS, p< .001
Frontal Ankle [°]	IC	6.4 (3.3)	7.8 (3.4)	.820	IM>CS, p = .005
	MAX	-4.7 (4.1)	-1.3 (5.2)	.976	IM>CS, p< .001
	TO	2.5 (2.9)	6.6 (3.8)	1.751	IM>CS, p< .001
	ROM1	10.8 (3.3)	10.7 (2.2)	.066	p = .7889
	ROM2	8.9 (2.8)	10.8 (1.9)	.963	IM>CS, p = .001
Sagittal Knee [°]	IC	35.9 (7.2)	38.5 (6.6)	1.026	IM>CS, p = .001
	MAX	123.3 (8.2)	122.2 (8.0)	.202	p= .417
	TO	39.2 (6.5)	37.0 (6.7)	.824	CS>IM, p = .004
	ROM1	87.4 (5.3)	83.8 (5.5)	.847	CS>IM, p = .003
	ROM2	84.1 (6.1)	85.3 (5.6)	.212	p = .395
Peak Knee Flex Vel [°/sec]	ROM1	-342.2 (67.2)	-313.0 (47.4)	.897	IM>CS, p = .002

IC = initial contact, MAX = maximum flexion/eversion, TO = toe-off, ROM1 = descending range of motion, ROM2 = ascending range of motion. **Data were not normally distributed*.

Table 7.4. Mean (SD) kinematics during the **lateral lunge** across participants

Lateral Lunge	Period	CS	IM	Effect size	Significance
Frontal Shoe-surface [°]	IC	-25.7 (4.6)	-23.3 (3.5)	.840	IM>CS, p = .003
Vertical foot vel [m/s]	IC	0.49 (.14)	0.54 (.14)	.841	IM>CS, p = .003
Horizontal foot vel [m/s]	IC	0.46 (.13)	0.34 (.11)	.987	CS>IM, p = .001
Sagittal Ankle [°]	IC	17.2 (4.4)	14.5 (4.6)	1.530	CS>IM, p< .001
	MAX	29.8 (4.0)	32.3 (4.8)	.818	IM>CS, p = .005
	TO	-21.5 (10.0)	-23.0 (8.4)	.231	p = .354
	ROM1	20.4 (6.8)	23.1 (5.3)	.764	IM>CS, p = .006
	ROM2	53.8 (6.2)	56.4 (6.4)	.490	IM>CS, p = .060
Peak Inversion Vel [°/sec]	ROM1	252.2 (88.6)	171.4 (56.2)	1.323	CS>IM, p< .001
Frontal Ankle [°]	IC	7.0 (4.8)	9.0 (4.7)	.808	IM>CS, p = .006
	MAX	4.0 (4.3)	5.3 (4.6)	.409	p = .111
	TO	10.7 (3.5)	10.0 (4.4)	.167	p = .502
	ROM1	12.3 (2.3)	9.1 (1.9)	1.333	CS>IM, p< .001
	ROM2	10.6 (2.8)	9.7 (2.1)	.353	p = .164
Sagittal Knee [°]	IC	34.2 (4.7)	33.7 (5.2)	.274	p = .276
	MAX	107.3 (10.8)	105.2 (10.0)	.418	p = .104
	TO	31.7 (5.9)	30.7 (5.7)	.234	p = .350
	ROM1	73.1 (9.9)	71.5 (8.2)	.306	p = .226
	ROM2	75.6 (8.0)	74.4 (9.3)	.213	p = .393
Peak Knee Flex Vel [°/sec]	ROM1	-357.8 (96.0)	-313.9 (76.1)	1.101	IM>CS, p< .001

IC = initial contact, MAX = maximum flexion/ eversion, TO = toe-off, ROM1 = descending range of motion, ROM2 = ascending range of motion.

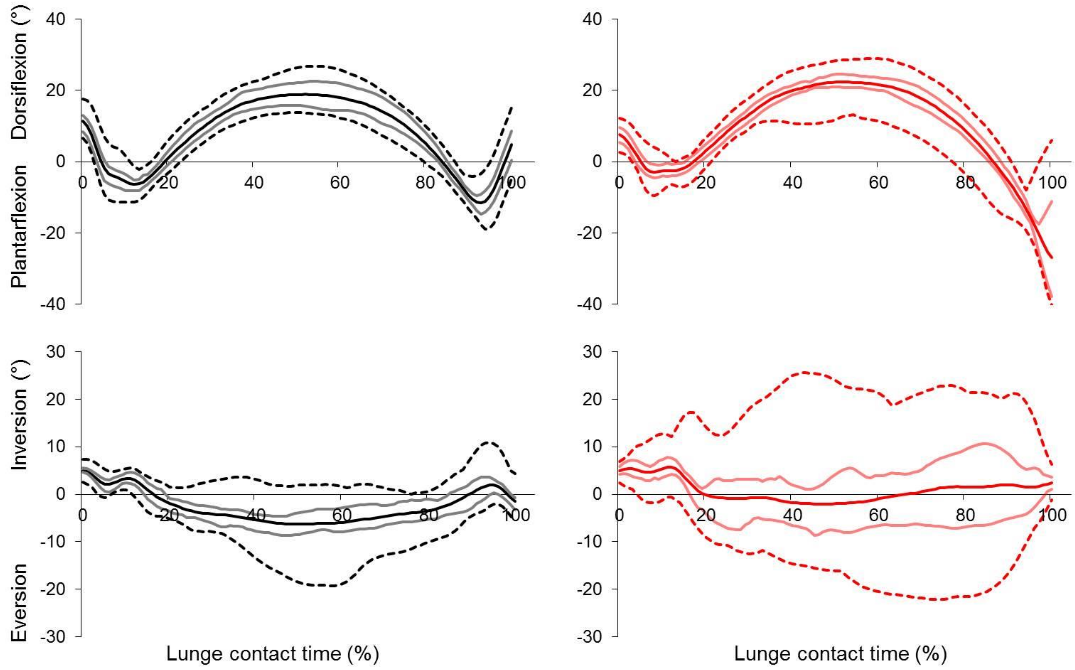

Figure 7.3. Sagittal (top) and frontal ankle angles (bottom) during both sets of **forward lunges** in CS (left) and IM (right) of a typical participant. The median is illustrated as the solid, darker line. To illustrate the variability the 25th and 75th percentiles are displayed as lighter lines, and the minimum and maximum as dashed lines across all 20 lunges for each percentage of contact time. Notice the plantarflexed sagittal ankle at toe-off and the increased frontal ankle variability in the IM.

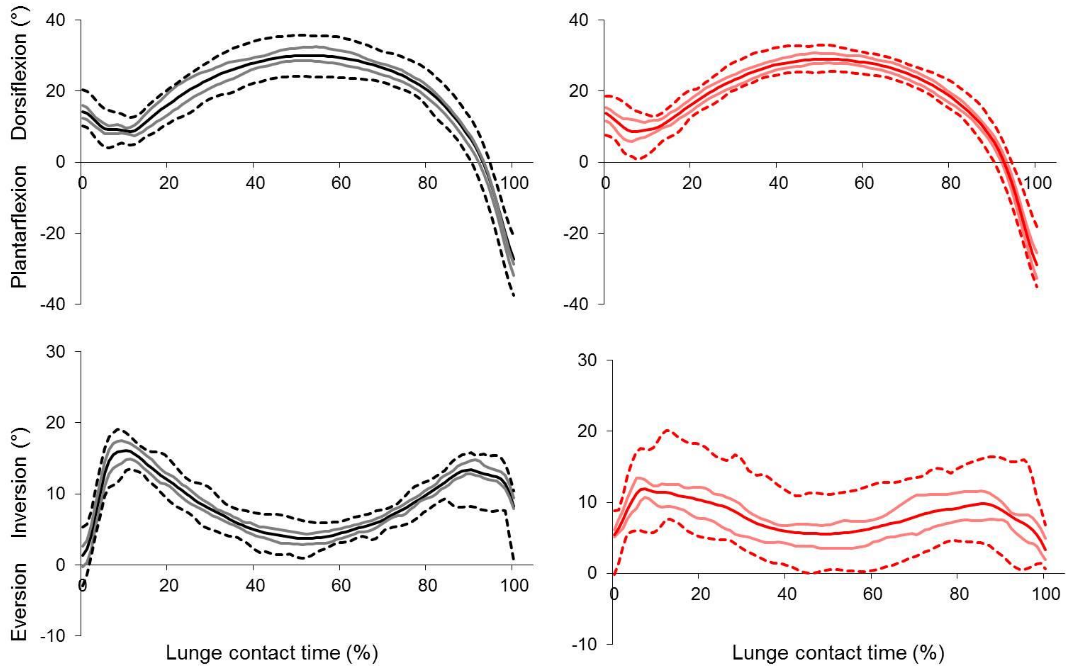

Figure 7.4. Sagittal (top) and frontal ankle angles (bottom) during both sets of **lateral lunges** in CS (left) and IM (right) of a typical participant. The median and variability plotted as in Figure 7.3. Notice the different frontal ankle motion in lateral compared to forward lunges (Figure 7.3).

Few significant differences were found in the variability (CV) of the sagittal ankle and knee joint angles during stance (Tables 7.5, 7.6). Frontal ankle range of motion variability increased in IM compared to CS during the forward lunge (across variables Mean (SD): 27% (6%), Figure 7.3) and lateral lunge (across variables Mean (SD): 39% (11%), Figure 7.4).

Table 7.5. Mean kinematic variability (CV) during the **forward** lunge across participants

Forward Lunge	Period	CS	IM	Effect size	Significane
Sagittal Ankle [°]	ROM1	14.0 (4.5)	17.0 (5.9)	1.001	IM>CS, p = .001
	ROM2	15.3 (5.3)	19.7 (9.1)	.424	p = .100
Frontal Ankle [°]	ROM1	21.4 (7.3)	31.0 (6.7)	.997	IM>CS, p = .001
	ROM2	29.8 (12.3)	37.4 (9.6)	.599	IM>CS, p = .025
Sagittal Knee [°]	ROM1	4.8 (1.6)	5.5 (1.8)	.394	p = .124
	ROM2	5.9 (1.5)	6.0 (0.8)	.073	p = .767

MAX = maximum flexion/ inversion, ROM1 = descending range of motion, ROM2 = ascending range of motion.

Table 7.6. Mean kinematic variability (CV) during the **lateral** lunge across participants

Lateral Lunge	Period	CS	IM	Effect size	Significane
Sagittal Ankle [°]	ROM1	16.6 (6.1)	15.1 (4.6)	.221	p = .376
	*ROM2	13.6 (10.7)	8.3 (7.1)	.469	p = .071
Frontal Ankle [°]	ROM1	15.3 (4.2)	31.3 (10.2)	1.528	IM>CS, p< .001
	*ROM2	20.6 (6.8)	28.8 (8.6)	.839	IM>CS, p = .003
Sagittal Knee [°]	ROM1	4.6 (1.7)	5.0 (1.2)	.273	p = .277
	ROM2	5.2 (1.8)	5.6 (1.6)	.169	p = .497

MAX = maximum flexion/inversion, ROM1 = descending range of motion, ROM2 = ascending range of motion. **Data were not normally distributed*.

7.4.3. Ground reaction forces

The initial vertical ground reaction force was more variable, indicated by the significantly increased CV, in IM compared to CS for the forward lunge (across variables mean (SD) increase: 27 (6) %, Table 7.7) and the lateral lunge (Mean (SD) increase: 31 (3) %, Table 7.8). The magnitude of the initial vertical impact peak and instantaneous and average loading rates also increased in IM across lunge directions. Both the descending and ascending horizontal impulses were smaller in IM compared to CS for both lunges (Figure 7.5).

Table 7.7. Ground reaction force results of the **forward lunge** across participants

Forward Lunge		CS	IM	Effect Size	Significance
Mean	Initial vertical impact peak	.429 (.112)	.470 (.135)	.499	IM>CS, p = .056
	Maximum loading rate	15.6 (6.6)	23.8 (11.1)	1.33	IM>CS, p < .001
	Average loading rate	12.4 (5.6)	15.4 (6.6)	.953	IM>CS, 0.001
	Descending impulse	.076 (.015)	.066 (.009)	.965	CS>IM, 0.001
	Ascending impulse	.082 (.014)	.070 (.010)	1.59	CS>IM, p < .001
CV	Initial vertical impact peak	15.0 (3.4)	19.2 (5.4)	.660	IM>CS, p = .015
	Maximum loading rate	22.7 (5.1)	33.8 (8.2)	1.370	IM>CS, p < .001
	Average loading rate	22.0 (4.0)	30.4 (7.3)	1.294	IM>CS, p < .001
	*Descending impulse	8.9 (4.2)	9.8 (2.5)	.226	p = .366
	Ascending impulse	8.6 (2.8)	9.1 (2.4)	.204	p = .412

Units expressed in body weights; loading rates seconds per bodyweight, impulses in newton/seconds per bodyweight. **Data were not normally distributed*

Table 7.8. Ground reaction force results of the **lateral lunge** across participants

Lateral Lunge		CS	IM	Effect Size	Significane
Mean	Initial vertical impact peak	.736 (.170)	.687 (.162)	.461	p = .076
	Maximum loading rate	27.9 (11.0)	35.0 (15.1)	.804	IM>CS, p = .004
	Average loading rate	20.6 (6.9)	21.3 (7.9)	.140	p = .582
	Descending impulse	.090 (.011)	.080 (.011)	1.03	CS>IM, p = .001
	Ascending impulse	.090 (.011)	.084 (.011)	.699	CS>IM, p = .011
CV	Initial vertical impact peak	12.2 (3.3)	17.9 (3.9)	1.280	IM>CS, p < .001
	Maximum loading rate	21.1 (5.5)	29.4 (6.4)	.869	IM>CS, p = .002
	Average loading rate	19.7 (5.3)	29.5 (7.8)	1.087	IM>CS, p < .001
	*Descending impulse	6.9 (1.4)	7.3 (3.4)	.166	p = .504
	*Ascending impulse	6.9 (1.4)	7.0 (3.2)	.048	p = .847

Units expressed in body weights; loading rates seconds per bodyweight, impulses in newton/seconds per bodyweight. **Data were not normally distributed*.

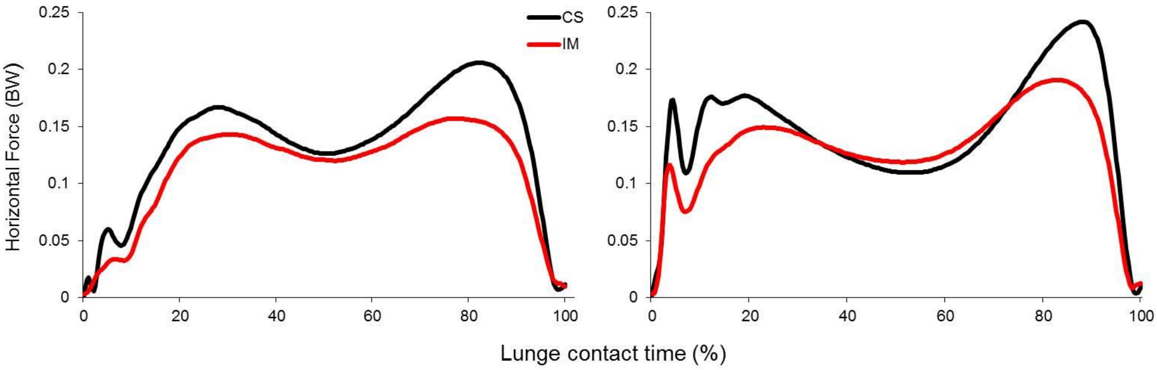

Figure 7.5. Resultant horizontal ground reaction forces during the forward lunge (left) and lateral lunge (right) across participants. Notice the reduced impulse throughout the lunges in IM (red) compared to CS (black).

7.4.4. Electromyography

Relatively few significant muscle activation differences occurred in the forward (Table 7.9) and lateral lunges (Table 7.10), most of which occurred in the shank muscles. During the forward lunges in IM, the mean gastrocnemius medialis and peroneus longus activation increased in the ascending phase (24% and 17%, respectively). The gastrocnemius medialis activation level also increased in the pre-activation phase in IM (29%) during forward lunges.

Similarly in the lateral lunges, the gastrocnemius medialis (tendency only p = .60) and peroneus longus activation increased in the ascending phase in IM compared to CS (14% and 10%, respectively). For both lunges these increased activations could be observed (Figure 6.6). Vastus medialis pre-activation level reduced in IM compared to CS.

Table 7.9. Normalised mean (SD) electromyography results during the **forward lunge**

Muscle	Phase	CS	IM	Effect size	Significance
Gastrocnemius medialis	pre-activation	8.6 (6.4)	11.0 (9.0)	.593	IM>CS, p = .038
	descending	12.0 (4.4)	12.7 (4.8)	.267	p =.318
	ascending	12.0 (6.9)	15.7 (8.8)	.584	IM>CS, p =.040
Tibialis anterior	pre-activation	22.2 (8.8)	21.4 (10.8)	.132	p =.618
	descending	28.7 (6.4)	27.8 (8.5)	.177	p =.505
	*ascending	21.1 (3.8)	20.9 (4.7)	.063	p =.811
Peroneus longus	pre-activation	13.8 (11.0)	14.2 (10.1)	.096	p =.716
	descending	24.4 (7.5)	25.1 (7.4)	.260	p =.332
	ascending	19.7 (13.7)	23.8 (13.7)	.827	IM>CS, p =.006
Bicep femoris	pre-activation	15.3 (10.0)	18.1 (11.1)	.378	p =.181
	descending	27.3 (5.2)	29.6 (6.3)	.413	p =.146
	ascending	28.4 (14.2)	30.8 (15.0)	.356	p =.205
Vastus medialis	pre-activation	15.6 (13.8)	13.8 (10.9)	.301	p =.299
	descending	40.3 (15.4)	38.8 (13.0)	.176	p =.538
	ascending	31.1 (21.8)	32.1 (25.7)	.133	p = .640

**Data were not normally distributed*.

Table 7.10. Normalised mean (SD) electromyography results during the **lateral lunge**

Muscle	Phase	CS	IM	Effect size	Significance
Gastrocnemius medialis	pre-activation	9.4 (7.3)	10.8 (11.1)	.243	p = .362
	descending	13.0 (5.8)	13.2 (5.8)	.061	p = .816
	ascending	21.5 (5.7)	24.9 (9.3)	.528	p = .060
Tibialis anterior	pre-activation	31.4 (12.6)	28.6 (11.5)	.329	p = .223
	descending	33.6 (8.3)	30.4 (8.2)	.602	p = .035
	ascending	21.3 (6.0)	20.1 (5.4)	.336	p = .214
Peroneus longus	*pre-activation	13.8 (8.8)	14.5 (8.5)	.214	p = .421
	*descending	20.2 (10.3)	21.3 (9.5)	.374	p = .169
	ascending	26.1 (8.2)	29.0 (8.1)	.820	IM>CS, p = .007
Bicep femoris	pre-activation	19.5 (12.0)	16.5 (8.2)	.483	p = .094
	descending	29.7 (7.2)	29.2 (5.5)	.084	p = .757
	ascending	32.4 (9.1)	34.4 (10.7)	.215	p = .435
Vastus medialis	pre-activation	20.7 (15.8)	18.1 (15.2)	.632	CS>IM, p = .042
	descending	48.8 (18.8)	46.0 (18.0)	.414	p = .162
	ascending	33.7 (21.8)	32.1 (20.8)	.370	p = .207

**Data were not normally distributed*.

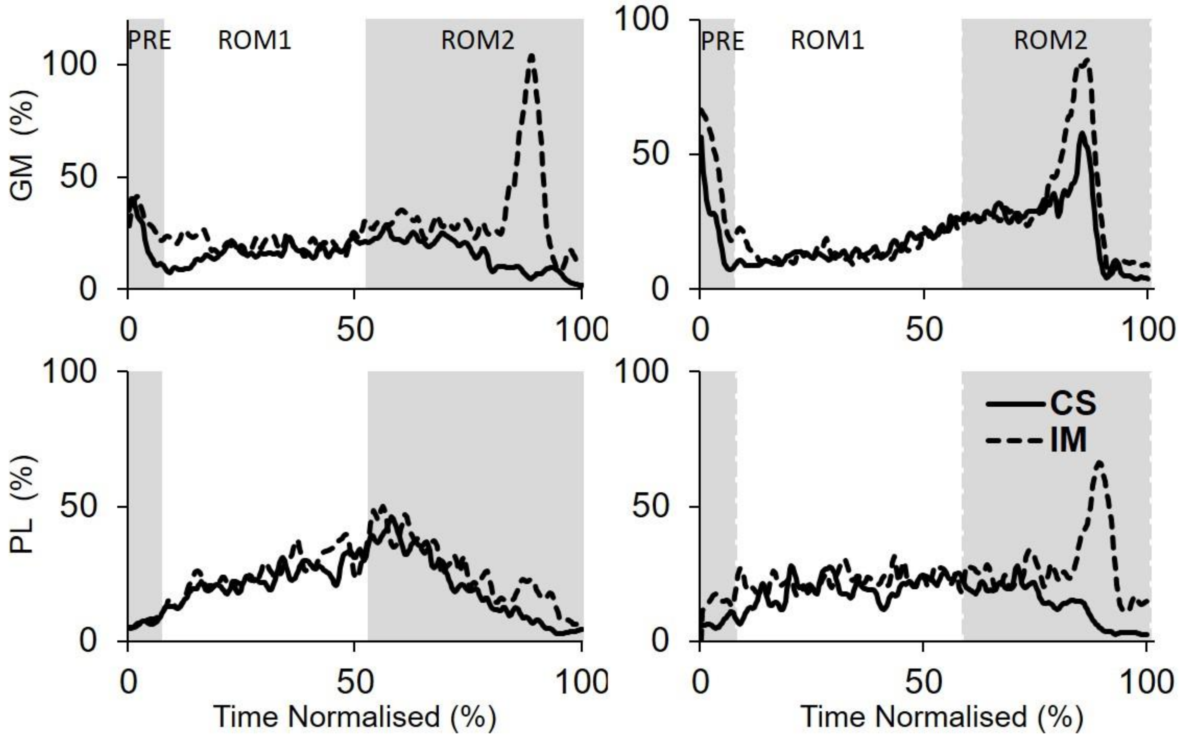

Figure 7.6. Mean normalised gastrocnemius medialis (GM) and peroneus longus (PL) activations during 20 forward (left) and lateral (right) lunges in IM (dotted) and CS (solid) of a participant with increased activations in IM. Time is normalised between pre-activation, ROM1 and ROM2 phases.

7.5. Discussion

This study compared the global lunge characteristics, lower-limb kinematics, muscle activations, and ground reaction forces during forward and lateral lunges IM and CS. The results confirmed the first hypothesis: IM caused greater instability, which was displayed by the increased frontal ankle motion variability, larger and more variable loading rates, and increased shank muscle activations. Contrary to the second hypothesis, the lateral lunge did not create more instability compared to forward lunges in IM. In fact, more adaptations were enhanced in the forward lunge.

The reported lunge stance times were within the rather wide range reported in the literature for healthy adults (Alkjaer et al., 2002; Riemann et al., 2013) and faster than older adults (Flannagan et al., 2004). Despite the metronome to control for lunge contact time, lateral lunge stance times were longer compared to forward lunges, in accordance with previous research (Riemann 2013; Flannagan et al., 2004). Increased stance times were found during forward lunges in IM indicating this was a control strategy to the random midsole perturbations. Alkjaer and colleagues (2002) found ACL deficient patients, who suffer from knee instability, also performed forward lunges slower than healthy controls.

Step lengths for forward and lateral lunges were shorter in comparison to previous studies (Riemann et al., 2013; Escamilla et al., 2008). This relates to the increased maximum sagittal ankle and knee angles achieved in this study. Unlike Escamilla and colleagues (2008), we did not instruct participants to take a stride as long as is comfortable, or impose a specific step length. Instead, participants were free to choose any preferred step length, applicable to exercising at a gym. Step length reduced by 2 cm in IM compared to CS during lateral lunges, signifying this was an active adaptation to reduce the effects of shoe instability. This cautious step length reduction strategy has been reported on IS whilst walking (2.4cm: Voloshina et al., 2013) and running (3.0cm: Voloshina & Ferris, 2015), as well as, in IM (1.3 cm walking, 1.8 cm running; Chapter 5).

Lower-limb kinematic adjustments to IM during forward and lateral lunges were more pronounced than the global lunge characteristics, indicated by the larger effect sizes. At initial ground contact a flatter shoe-surface angle, and increased vertical, but reduced horizontal foot velocities helped reduce the resultant horizontal shear forces in IM. This strategy has been shown to reduce the risk of losing balance by lowering the friction coefficient at the shoe-floor interface (Menant et al., 2008). Knee flexion velocity reduced in the descending lunge phase in IM. This may be an injury prevention strategy to avoid excessive joint motion, as has been suggested in ACL patients (Alkjaer et al., 2002). Although joint stiffness was not analysed, the increased sagittal ankle ranges of motion during both descending and ascending lunge phases would help absorb the increased initial loading rates. Aligned with the results in Chapter 6, knee range of motion reduced in the descending phase in IM during the forward lunges, which would help stabilise the more proximal body segments. However, joint stiffness can only be speculated from reported results and warrants further investigation with the joint kinetics.

Peroneus longus activations increased in the ascending phase in forward and lateral lunges in IM, helping to control the increased frontal ankle motion variability and stabilise the ankle for toe-off. Gait studies have also found US eliciting medio-lateral instability increase peroneus longus activations during pre-swing (Price et al., 2013). This is a positive result because it infers these muscles will be strengthened and conditioned over time, reducing the chance of ankle injury. Salsich and Mueller (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. Thus, based on these findings IM may be suitable for this purpose.

Gastrocnemius medialis activations also increased in IM in the ascending phase, but only significantly during forward lunges. This relates to the plantar flexed ankle at toe-off, which was a stability strategy that prevented the centre of pressure moving posteriorly across the unstable objects in IM (Figure 7.7). Although this may train the triceps surae muscle group, learning to control moving the centre of pressure posteriorly with the IM perturbation would not be. This offers a coach or trainer the chance to decide what technique may be more important to the desired training outcome. 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.

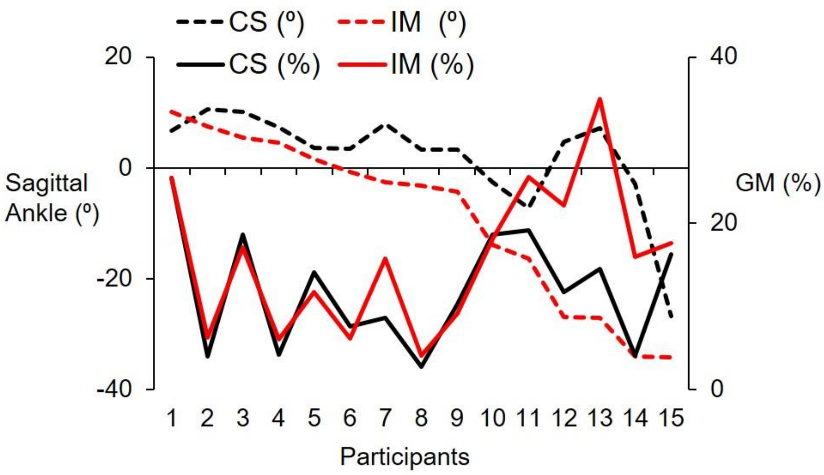

Figure 7.7. Mean normalised gastrocnemius medialis (GM) activations (solid lines) in the ascending phase and sagittal ankle angle at toe-off (dotted lines) for each participant during forward lunges in IM (red) and CS (black). Notice as the ankle angle becomes more plantarflexed (negative) GM activation increases.

Few other differences in lower limb muscles activations were found, suggesting the female gym class participants needed higher levels of instability to have an effect in the descending phase. An instability device that increased muscle activation during the descending phase, where certain muscles act eccentrically, would be desirable for injury prevention as this is when most injuries occur (Jonhagen et al., 2009).

The instability effect of IM on the forward and lateral lunges were relatively consistent. Some results suggested forward lunges in IM may have been more challenging compared to CS. Loading rates increased to a greater extent in forward lunges (27%), opposed to lateral lunges (12%) in IM. Additional adaptations during forward lunges were made in IM to enhance stability: gastrocnemius pre-activation and knee flexion at initial contact increased. The increases of peroneus longus and gastrocnemius activation in the ascending phase compared to CS was also larger in forward lunges (17%, 24%) compared to lateral lunges (10%, 14%). This is because forward lunges present a greater challenge to medio-lateral stability because of the straighter foot progression angle reducing the base of support. During lateral lunges, the length of the foot can be utilised for medio-lateral support (Riemann et al., 2013). Although the foot progression angle was 30º on average at initial contact in lateral lunges which would have reduced the medio-lateral base of support somewhat.

A limitation of this study was the set lunge speed (20 bpm), which was imposed to reduce biomechanical differences due to speed variations which have been demonstrated previously during gait (Romkes et al., 2006; Sekiya et al., 1997). However, this may have been faster or slower than participants' preferred pace and not as applicable to gym training situations. Longer contact times in the IM footwear may present greater challenges to stability, as found in walking compared to running (Chapter 5). Some discomfort from the objects inside the IM bags may also have altered the biomechanical response in the lunge movements, it is suggested this haptic sensation be removed from future prototypes.

7.6. Conclusion

Biomechanical research of US is limited to standing, walking and running. Higher frontal ankle motion and loading variability showed IM was a more demanding training stimulus, which induced a cautious foot placement at initial contact and increased shank muscle activations. If these adaptations were maintained during a long-term training intervention, sensorimotor balance ability and ankle muscle strength would very likely improve. Thus, IM may provide additional training benefits during functional gym exercises over typical gym footwear, although this requires investigation. Future studies should determine if regularly doing functional exercises wearing IM trains the sensorimotor system and ankle musculature, and how this might enhance performance or reduce injury-risk, such as ankle sprains. However, the IM prototype was not durable enough to use for a long-term intervention, so this was investigated on the IS treadmill which provided similar unpredictable instability at the shoe-surface interface (Chapter 8). This limited the training intervention to locomotion. Walking was selected because this is most applicable to daily activities where a shoe would offer a convenient training modality.

CHAPTER 8.

The effect of 6-weeks irregular surface walking on ankle muscle strength and balance

8.1. Abstract

Positive training benefits of US from recent studies are inconsistent. Greater instability from an IS providing continuous unpredictable perturbations may have a more profound effect. This study investigated the effects of a 6-week IS walking intervention on ankle muscle strength and postural balance. Healthy participants (n=8) were tested before (pre) and after (post) a 6-week training program whilst walking on an IS treadmill and were compared to a control group (n=7) by a mixed ANOVA ($p<.05$). Ankle strength for plantarflexion, dorsiflexion, eversion and inversion was determined from isokinetic concentric peak torques. Postural balance was assessed by the sway area and variability of centre of pressure velocity (SD) during static 1-legged standing. Results showed no significant differences between the training group and control. Both groups showed significantly increased dorsiflexion ($p= .017$) and eversion ($p = .005$) peak torques post-test. Sway area, medio-lateral centre of pressure range and velocity variability also reduced post-test. Although not significantly, the training group displayed greater post-test strength increases and postural sway decreases than the control on average for: dorsiflexion peak torque (training group: +42%, control: +8%), eversion peak torque (+24%, +13%) sway area (-36%, -27%) medio-lateral range (-19%, -12%) and variability of velocity (-19%, -14%). Findings indicate regular IS walking may improve aspects of ankle strength and balance, but more research is needed before IS training can be recommended as a training tool for in healthy individuals. Future studies should incorporate a longer training period with measurements made midway through the intervention and include a familiarisation session.

8.2. Introduction

Unstable shoes (US) are endorsed as a therapeutic device that can improve health and wellbeing. With regular wear US are proposed to strengthen the lower limb muscles, enhance balance and reduce joint pain (Nigg et al., 2012). Kaelin and colleagues (2011) reported ankle eversion strength gains after 3 months of regularly wearing US in participants with unstable ankles. However, not all US studies report increased muscle activations for healthy participants (Forghany et al., 2014; Horsak & Baca, 2013; Nigg et al., 2006a; Sacco et al., 2012; Stöggl et al., 2010; Table 2.1), suggesting ankle strength gains would be inconsistent. Moreover, balance studies report conflicting results after regularly wearing US, with some studies reporting improvements (Landry et al., 2010) and others not (Ramstrand et al., 2010; Nigg et al., 2009).

Therefore, we developed a novel US prototype that creates greater instability, which would require further neuro-muscular and postural adjustments, provoking more consistent training effects. A shoe with random midsole deformations (IM) was developed, with the idea that the unpredictable perturbations would be a greater challenge to the sensorimotor system than current commercial US with predictable instability. In Chapter 6 this concept was validated: additional ankle muscle activations and co-contractions, as well as altered levels of ankle and knee stiffness were found in IM compared to a typical US during walking and running. Thus, it was theorised regular wear of IM would offer greater ankle muscle strength and balance improvements than regular US. Ankle muscle imbalances are a risk factor for sprains (Baumhauer et al., 1995). Although it is unlikely healthy participants will have muscle imbalances, it is important to monitor any changes IM has on strength ratios of opposing muscles groups.

However, the IM shoes developed were designed for testing purposes only and are not suitable for regular wear. The prototypes were not robust enough for a training intervention study to be undertaken, due to the velcro material breaking down and frailties of the rubber bags. In order to establish the long-term benefits of regularly wearing IM, a training program involving unpredictable instability is warranted. In Chapter 5, there were similar adaptations whilst walking and running in IM compared to an irregular surface (IS). Highest instability was observed on IS by the increased kinematic variability, which induced postural adjustments and shank muscle activations. Hence, training effects of regularly walking on IS would be even more apparent than IM. The purpose of this study was to investigate the influence of regular IS treadmill walking for 6-weeks on ankle muscle strength and postural balance ability in healthy adults. It was firstly hypothesised ankle strength, particularly eversion, would be improved after the IS training group compared to a control group, but there would be no changes in muscle strength ratios. Secondly, it was hypothesised static single-leg balance would improve in the IS training group compared to the control group. This would be indicated by a reduced centre of pressure sway area and velocity variability.

8.3. Methods

8.3.1. Participants

Fifteen healthy, sports science students were recruited in this study and randomly assigned into two groups. The training group (TG) included 8 participants (2 males, 6 females, Mean (SD) 26.1 (2.9) years, 170.0 (6.3) cm, 71.0 (13.6) kg) and the control group (CG) included 7 participants (2 males, 5 females, Mean (SD) 24.3 (2.6) years, 168.9 (6.0) cm, 65.8 (8.7) kg). Both groups were tested twice at the biomechanics lab at Liverpool John Moores University with a minimum of 6-weeks between pre and post-testing sessions. Inclusion criteria required participants to be self-reported injury free, and have no known neuromuscular or balance disorders. The study was approved by the University's research ethics committee and participants were informed of the aims of the study and gave written consent prior to testing.

8.3.2. Training intervention

During the pre-test session, the TG selected their normal preferred walking speed on a treadmill (Pro XL, Woodway Inc., USA) with a regular surface. After the pre-tests, the CG were asked to continue their normal amount of routine exercise and not start a new training activity. Within one week of the pre-test session, the TG commenced a 6-week training program consisting of 3, 1-hour sessions per week. The training program consisted of walking at their preferred speed (Mean (SD): 3.8 (0.5) km/hour) on the same treadmill which was modified to have an irregular surface (IS). Details of the IS are provided in Chapter 3. During training sessions, participants were asked to look ahead, so they could not predict where on the insert surface they would land.

8.3.3. Muscle strength assessment and analysis

Ankle muscle strength of the right leg was determined from a Biodex System 3 isokinetic dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA) at 100 Hz. The seatback was 85° upright, and the chair, dynamometer and limb support under the thigh positioned to ensure the knee angle approximated 20-30°, according to Biodex guidelines. The seat position was adjusted as necessary to ensure the ankle movements were perpendicular to the long-axis of the dynamometer. The exact equipment positions were recorded for each participant to ensure identical set-up in the post-tests. During plantarflexion-dorsiflexion measurements the dynamometer was rotated by 90° and the ankle joint aligned laterally with the dynamometer axis (Figure 8.1). During inversion-eversion measurements the dynamometer was tilted 50° towards participants and the ankle joint aligned with the dynamometer axis vertically (Figure 8.1). Once participants were positioned, belts and straps were securely fastened around the trunk, waist and right thigh to ensure the torque applied was by the ankle muscles only (Baltzopoulos, 2007). Participants wore their own socks during testing. To secure the foot from sliding during the eversion and inversion measurements velcro was attached to the footplate and the sock (Figure 8.1, left). Additionally, two straps wrapped around both the forefoot and ankle which were tied to the footplate.

Isokinetic concentric tests for plantarflexion, dorsiflexion, eversion and inversion were performed maximally for 5 repetitions at 60^0/sec at the full range of motion. The order was randomised between plantarflexion-dorsiflexion and inversion-eversion measurements, but consistent across pre and post-test sessions. Prior to data collection participants were familiarised with the set-up, procedure and had a 3-repetition warm-up where they were asked to give at least 50% of their maximum effort. Before trials participants were encouraged to give their maximum effort and were given visual feedback from the joint moment data during tests. The highest 3 peak torques from the 5 repetitions of each test were averaged and normalised to the body weight (Nm/kg) (Figure 8.2). Muscle strength ratios were also calculated for dorsiflexion-plantarflexion and eversion-inversion joint torques.

Figure 8.1. Biodex 3 dynamometer set-up for plantarflexion/dorsiflexion (top) and inversion/eversion (bottom) measurements. The dynamometer was tilted so the ankle rotates about the sagittal (left) and frontal (right) axis.

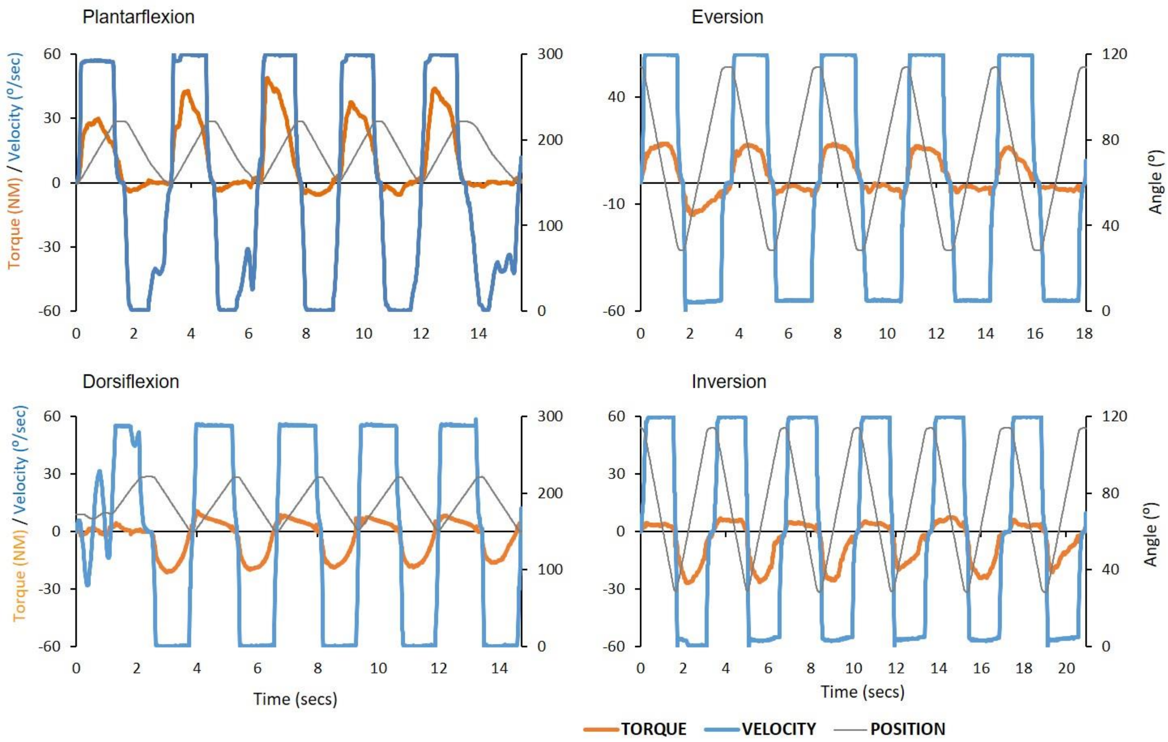

Figure 8.2. Example isokinetic concentric torque, velocity and position profiles during a pre-test session of a typical participant for plantarflexion (top, left), dorsiflextion (bottom, left), eversion (top, right) and inversion (bottom, right). Peak torques occurred during the isokinetic period. Often 1-2 peaks were slightly lower per trial so the three highest peaks were averaged for analysis.

8.3.4. Postural balance assessment and data analysis

Postural balance was assessed with the flamingo test, which has shown good reliability in healthy particiapnts (Ekdahl et al., 1988). Participants were instructed to stand still, with their right barefoot on a force plate (Kistler 9281B, Amherst, NY, US) whilst looking at a target 2.5 m in front of them at a height of 1.65 m. Ground reaction forces were recorded for two trials of 30 seconds, with a minimum rest interval of 30 seconds between trials. A trial was deemed successful if participants maintained the left leg 5 cm above the ground and kept hands on the iliac crests (Jakobsen et al., 2011).

Centre of pressure (COP) co-ordinate data were filtered with a 4th order low-pass Butterworth filter with frequency cut off of 10 Hz (Salavati et al., 2009) and exported for analysis in Excel (Microsoft Corp., Redmond, Washington). Sway area from a confidence ellipse was calculated by multiplying the SD of the medio-lateral and antero-posterior COP co-ordinate data by π (Jakobsen et al., 2011; Gerbino et al., 2007; Tropp et al., 1983). It is a measure of static postural ability to keep still and has been shown to differentiate balance ability between dancers and soccer players (Gerbino et al., 2007). The trial with the smaller elliptical area was used for additional variables and statistical analysis (Jakobsen et al., 2011). Postural control was assessed by calculating the COP range (Landry et al., 2010) and SD of the velocity in the antero-posterior and medio-lateral directions (Paillard & Noé, 2006).

8.3.5. Statistics

Paired t-tests verified there were no significant differences in height ($p = .848$), weight ($p=.891$) and age ($p=.219$) between the TG and CG. The mean of all variables for each participant during pre and post-tests were computed and checked for parametric assumptions (see section 4.5). A two-way mixed design ANOVA assessed the difference between groups (TG and CG) and within participants across testing sessions (pre-tests and post-tests). The alpha level was set at 0.05. To indicate the relevance of findings, effect sizes (η^2) were calculated for the main effects of group and testing session. A strong effect size was defined by $\eta^2 > 0.5$, moderate between 0.5 and 0.3 and low < 0.3 (Field, 2013).

8.4. Results

Across results, there were no significant interactions or differences between the TG and CG, but significant differences between pre and post-test sessions.

8.4.1. Muscle strength

There were significant main effects for testing session, with increased dorsiflexion and eversion peak torques, and eversion-inversion ratio during the post-test session (Table 8.1).. The TG had larger percentage increases of ankle muscle strength across all muscles groups compared to the CG, although no result reached significance. Mean percentage increases for the TG and CG respectively were: 7% versus 0% for plantarflexion, 42% versus 8% for dorsiflexion, 24% versus 13% for eversion and 16% versus -5% for inversion.

Table 8.1. Mean (SD) joint torques and muscle strength ratios across groups and testing sessions.

	TG		CG		Training group:	Test session:
	pre-test	post-test	pre-test	post-test	η2	η2
Plantarflexion [Nm/Kg]	.69 (.23)	.73 (.17)	0.83 (.37)	0.83 (.37)	.05	.05
Dorsiflexion [Nm/Kg]	.28 (.07)	.40 (.12)	.38 (.12)	.41 (.11)	.12	.37*
Eversion [Nm/Kg]	.27 (.06)	.34 (.07)	.32 (.13)	.36 (.13)	.04	.46**
Inversion [Nm/Kg]	.34 (.11)	.40 (.10)	.41 (.15)	.39 (.13)	.02	.05
PF-DF ratio [%]	43.0 (13.5)	57.3 (27.6)	51.4 (19.9)	55.9 (21.5)	.01	.16
EV-INV ratio [%]	84.8 (22.1)	90.1 (27.8)	83.4 (27.3)	101.4 (43.4)	.01	.28*

PF plantarflexion, DF Dorsiflexion, EV Eversion, INV Inversion. Significant differences indicated by: **p<.01, *p<.05.

8.4.2 Postural balance

Both groups showed significantly improved postural balance in the post-test session for 3 out of 5 variables (Table 8.2). No significant results were found between the TG and CG, but sway area showed a greater reduction in the TG (-36%) than the CG group (-27%) at post-test. Medio-lateral range and variability of velocity (SD) also reduced to a greater extent post-test in the TG (-19, -19%, respectively) than the CG (-12, -14%). No changes were found in the anteroposterior direction.

Table 8.2. Mean (SD) centre of pressure results across groups and testing sessions

	TG		CG		Training group:	Test session:
	pre-test	post-test	pre-test	post-test	η2	η2
Sway area [mm^2]	97.7 (45.0)	62.2 (12.8)	107.0 (41.0)	78.2 (26.1)	.06	.41*
AP Range [mm]	32.2 (9.1)	26.9 (4.0)	33.7 (5.4)	31.9 (10.0)	.04	.13
ML Range [mm]	23.8 (4.5)	19.3 (3.5)	24.4 (4.3)	21.4 (3.8)	.05	.40*
AP Velocity SD [mm/s]	22.2 (5.2)	18.7 (4.9)	25.8 (8.9)	24.7 (6.3)	.20	.27*
ML Velocity SD [mm/s]	26.0 (6.0)	21.1 (4.5)	30.2 (10.0)	26.0 (3.2)	.18	.13

AP Antero-posterior, ML medio-lateral. *significant difference (p<.05).

8.5. Discussion

This study investigated the effects of 6-weeks irregular surface (IS) walking training on ankle muscle strength and postural balance. The results showed no significant interactions, not confirming our first hypothesis that ankle muscle strength would increase in the training group (TG) compared to a control group (CG). The second hypothesis that the TG would significantly improve postural balance compared to the CG was also rejected. Instead, there were significant improvements for both the TG and CG at post-test. Ankle eversion and dorsiflexion concentric peak torques increased, plus, sway area and medio-lateral centre of pressure range and velocity variability reduced. Although not statically significant, the TG had greater improvements than the CG. Additional research is needed to determine if these greater improvements relate solely to the irregular IS training.

There are several potential reasons why there were no significant differences between the TG and the CG. The 6-week IS training period was selected because it has been applied to US balance interventions (Landry et al., 2010; Ramstrand et al., 2010: Nigg et al., 2009). Heitkamp et al. (2001) found improved static single-leg balance and muscular strength after just 12 balance training sessions in healthy adults. The balance training involved a variety of exercises on unstable devices including a Swiss ball, roller-board, trampoline and pedalo stepper. Sessions were completed 2-3 times a week for 25 minutes. The IS surface in this study was likely less demanding for the sensorimotor system which may require a longer training period for balance improvements. Likewise, ankle muscle strength gains were only reported after 3 months after performing exercise in US in participants with unstable ankles (Kaelin et al., 2011). A longer period for IS training may enable ankle muscles to strengthen.

The healthy, young sports science students tested is another possibility for lack of differences between groups. The IS treadmill modification was developed with the idea of unpredictable instability would promote greater training effects than US. Yet, the IS instability still may not have been a challenging enough stimuli. Similarly, figure skaters did not improve static single leg balance after a 4-week neuromuscular training program involving unstable support surfaces (Kovacs et al., 2004). It may be there is a limit on the amount of improvement that can be made for those who would already have had good postural balance ability and no ankle muscle strength deficits. This effect was observed for sway area in the TG, where participants that had values less than 100 mm^2 at pre-test made smaller reductions at post-test than those with greater pre-test sway area (Figure 8.3). However, Kovacs and colleagues (2004) did find balance improved on the skaters' ice skate, indicating more balance improvements can occur during more challenging tasks.

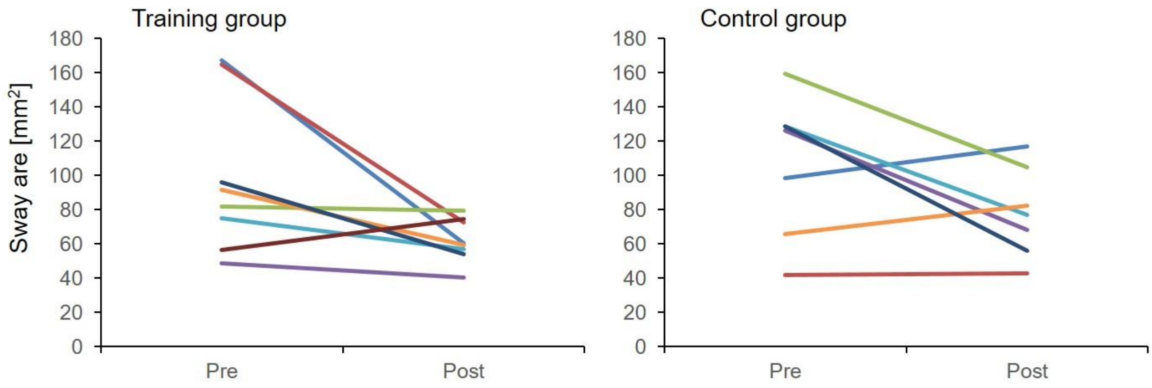

Figure 8.3. Sway area results for each participant displayed by a line at pre and post-test for the training group (left) and control group (right). Notice the large reduction post-test participants with highest sway area at pre-test.

Alternatively, fast adaptations may have occurred during the first IS training session that reduced the need for participants to increase ankle muscle activations or make postural adjustments. Such biomechanical adaptations and energy expenditure reductions has been shown to occur in the first hour of walking in US (Blair et al., 2013; Yap et al., 2013). This possibility was explored in 2 participants from the TG, by measuring how they adapted to the IS during their first training session (Appendix E). Results showed the initially increased knee joint variability and shank muscle activations reduced during the hour training session to a similar level as the regular control surface. This provides some evidence to suggest acute adaptations occur and the neuro-muscular system quickly learns how to enhance stability on IS. Therefore, future research on IS interventions may wish to focus on populations with balance or strength deficits. Older adults are reported to improve 2-legged static balance after a US intervention (Landry et al., 2010) and those with ankle instability improve functionality after an ankle strength training intervention (Sekir et al., 2007).

Ankle strength and balance improvements occurred in the CG during the post-test session, despite participants not doing any additional exercises than their usual routines.

This is the same as the previous research on static balance performance after an US intervention (Ramstrand et al., 2010) and dynamic balance after an instability training device intervention (Eisen et al., 2010). Possibly, a learning effect occurred from pre to post-testing sessions, where participants where familiarised to the measurement procedures. This makes it difficult to determine if the training group improvements occurred due to the IS training or due to a learning effect. Therefore, a familiarisation session on a separate day before pre-test measurements are made, or more comparisons to the CG midway through the training intervention are recommended.

Limitations in this study are acknowledged. Firstly, the sample size was small (n = 15) and the number of participants in the TG (n = 8) and the CG (n=7) was not equal. This yielded low statistical power for the variables analysed and increased the chance of a type 2 error (Field, 2013). The CG did not complete the same exercise as the TG, therefore differences in the results may not be from the irregular surface but from the 3 hours per week treadmill walking. Although it is not expected this would have had a large influence because participants were healthy, future uneven shoe-surface training studies should include the same training routines apart from the shoe-surface instability. Static postural balance was assessed as this has been demonstrated to improve after US and balance training interventions (Landry et al., 2010; Michell et al., 2006). However, as the intervention involved walking with random surface perturbations, this may have a larger effect on dynamic balance ability. Moreover, isokinetic concentric torques were measured at 60°/second only. Measuring eccentric contractions and a range speeds would reveal a more complete understanding of training effects on ankle strength. Maximum ankle plantarflexion and eversion velocities on IS from the data from Chapter 5 were 399±24º/sec and 264±102º/sec respectively, indicating faster speeds may have increased validity to the training intervention. However, at faster speeds the peak torque is more likely not to be in the isokinetic period (Figure 8.2). Despite these limitations, this initial training study with unpredictable instability demonstrates good potential to conduct a longer, more robust study to explore training adaptations.

8.6. Conclusion

Regular walking on an irregular surface improved postural balance and ankle muscle strength, but not significantly from the control group. A longer training period, with more control group comparisons is needed before IS training can be recommended as a training tool for in healthy individuals.

CHAPTER 9.

Synthesis

9.1. Introduction

Current evidence suggests unstable footwear (US) may not provide a challenging enough stimulus to require consistent biomechanical alterations to maintain stability. This may partly be due to healthy participants, who have good sensorimotor control, being the focus of previous research. Moreover, the outsoles of US provide fixed instability, which is predictable, allowing adaptations to be learnt quickly reducing their de-stabilising effect. It was foreseen that continual unpredictable instability would provide the more difficult stimulus needed to force the neuromuscular system into making postural adjustments that would result in training benefits with regular use. This instability could be provided by an irregular surface (IS) or a shoe, the advantage of a shoe being that training effects can occur during activities of daily life.

Therefore, the aim of this research was to develop and evaluate the biomechanical and neuromuscular adaptations to wearing novel shoes that provide continuous, unpredictable perturbations during dynamic movements (the irregular midsole (IM)). Specific objectives in experimental studies were:

1) Compare the instability induced by IM and IS, and a control shoe-surface condition during walking and running (Chapter 5).
2) Explore the joint stability strategies between IM and an US during walking and running (Chapter 6).
3) To investigate the adaptations to IM compared to a control shoe during aerobic exercises: the forward and lateral lunge (Chapter 7).
4) To evaluate the short-term training effects on ankle muscle strength and balance in IM. As this was not feasible, training effects of unpredictable instability were investigated after 6-weeks of regular walking on the IS (Chapter 8).

This final chapter synthesises the findings of studies to discuss the acute (Chapter 5, 6 and 7) and short-term adaptations (Chapter 8) to continuous unpredictable shoe-surface instability. The implications are used to provide guidance on the effectiveness of unpredictable instability training and offer recommendations for future research.

9.2. Acute adaptations to shoe-surface instability

The studies that evaluated the acute effects (Chapter 5, 6 and 7) of IM and the IS suggest they increase instability compared to a control shoe-surface. This was shown by changes to the variability of movement (Chapter 5, 7), loading rates (Chapter 7) and plantar pressures (Appendix A). Adaptations to the global gait parameters, lower-limb kinematics, joint stiffness and muscle activations were made to enhance stability. Chapter 6 showed additional joint stability responses were required in IM compared to an unstable shoe.

9.2.1. Variability caused by unpredictable shoe-surfaces

Movement variability of both the ankle and knee joint, and the global gait parameters were assessed in this research. The implications of joint variability are discussed in Section 9.3. Increased variability of the global gait parameters have been used to quantify stability (Table 2.3) and are linked to the risk of falling (Hausdorff et al., 2001; Thies et al., 2005). Based on this principle, the findings of increased variability of step length, stride frequency and swing time in IM compared to the control whilst walking suggests greater risk of falling. Companies wishing to develop a training device with unpredictable instability should not overlook this potential injury risk (see 9.4.1. for further detail on injury).

However, the relationship between global gait variability and stability is limited. Firstly, the changes to step length and step time (Chapters 5 and 7) have revealed responsiveness to recover from perturbations during gait (Cordero et al., 2003), where no change to variability would represent an abnormal response (McAndrew et al., 2010). Secondly, with increasing task difficulty there was a reduction in the variability global gait parameters compared to the control shoe. Increased step length variability was found whilst walking in IM compared to the control shoe, but there was no difference in running and a reduction during lunges. This interesting observation is explained by the larger boundaries of stride dynamics possible to maintain stability whilst walking compared to running or lunge movements. This is referred to as the attractor state (Davids et al., 2003), where variability within this region is considered the ‘good variability’ (Latash, 2012), allowing exploration of the degrees of freedom and acquisition of flexible motor patterns (Van Emmerick & van Wegan, 2002). Thus, the increased variability of global gait parameters whilst running in IS (Chapter 5) and in the control shoe during lunges (Chapter 7) compared to IM may indicate participants were challenged by the unpredictable perturbations and were restricting their motion within the attractor. Yet, this can only be speculated from the analyses applied in this thesis.

The problem of quantifying this ‘good’ from ‘bad’ variability, which are outside of the attractor and would be de-stabilising, with parameters, such as step length and step width has been recognised (McAndrew et al., 2010). As determining loss of stability control from global gait variability is not possible, it is recommended better end-point or global variables are selected in future studies, such as trunk acceleration (Moe-Nilssen & Helbostad, 2005; Marigold & Patla, 2006). Another possible solution is the uncontrolled manifold approach, which has recently been applied to walking gait. In this analysis, the maintenance of the COM is considered the task goal (attractor) and the lower-limb joint combinations are the elements (degrees of freedom) available to achieve this (Papi et al., 2015; see Latash, 2012 for detail of this analysis). This analysis may be valuable in future research that wants to relate the global and joint level variability, and distinguish whether or not it is detrimental to performance.

9.2.2. Biomechanical responses to unpredictable shoe-surface instability

Chapters 5 and 7 revealed a cautious locomotion strategy was adopted to enhance stability to IM and IS compared to a control shoe-surface. The IM reduced the step length during walking, running and lateral lunges. Lower-limb kinematics at initial contact were also altered, such as, the flatter shoe-surface angle whilst walking and running and performing forward lunges. These adaptations reduce the risk of slipping by decreasing the horizontal ground reaction forces in the initial loading and push-off phases (Figure 7.5) (Menant et al., 2008). Furthermore, knee flexion increased at initial contact, would increase stability by lowering the centre of mass (MacLellan & Patla, 2006).

Chapter 6 investigated joint control strategies in IM (unpredictable instability) and an US (fixed instability) whilst walking and running. Results showed enhanced adaptations in IM compared to the US, indicated by the re-organisation of ankle and knee joint stiffness in the propulsive phase during walking and the greater re-organisation during the loading phase during running. This important finding demonstrated IM is a more challenging stimulus than the US (Figure 6.1) tested. The reduced ankle stiffness in IM was caused by reducing the dorsiflexion moment; this mechanism was observed on the video data to be caused by moving the ground reaction force vector closer to the joint axis. This is proposed to be a protective response to limit the ankle range of motion caused by the unpredictable perturbations of IM.

9.2.3. Neuromuscular responses to unpredictable shoe-surface instability

Results revealed IM altered the activations of certain shank muscles in phases of the movements tested compared to the control and US. Notably, gastrocnemius pre-activation increased in IM compared to the control shoe (Chapter 5 and 6), and the US whilst walking and running (Chapter 6). Given one of the claimed benefits of wearing US is to increase muscle activations, the unpredicted perturbations provided by IM may provide a stimulus that can do this more reliably.

Previous unstable footwear and uneven surface research during gait has not specifically focussed on muscle pre-activation levels, but have generally focused on the stance phase and its sub-periods. Although some previous MBT studies found increased gastrocnemius or soleus activation during the loading phase of walking (Romkes et al., 2006; Price et al., 2014; Forghany et al., 2014), similar to IM (Chapter 6). Due to the anteroposterior rocker in MBT, it is likely this footwear may increase gastrocnemius pre-activation level, but this warrants further investigation.

Tibialis anterior pre-activation significantly reduced whilst walking and running in IM (Chapter 5 and 6), which is in accordance with other US (Romkes et al., 2006; Forghany et al., 2014; Sacco et al., 2012) and IS research (Hettinga et al., 2005; Voloshina et al., 2013). This is associated with flatter shoe-surface angle at initial contact (Figure 9.1). Gastrocnemius medialis pre-activation has been correlated with ankle position at initial contact whilst running over a step of different heights (Muller et al., 2010). Therefore, ankle muscle pre-activations represent a feedforward strategy to position the ankle and foot in a stable position at initial contact (Section 9.2.2). During forward lunges in IM, increased gastrocnemius medialis activations also related to maintaining a plantar-flexed ankle positon, but in the ascending phase (Figure 7.7).

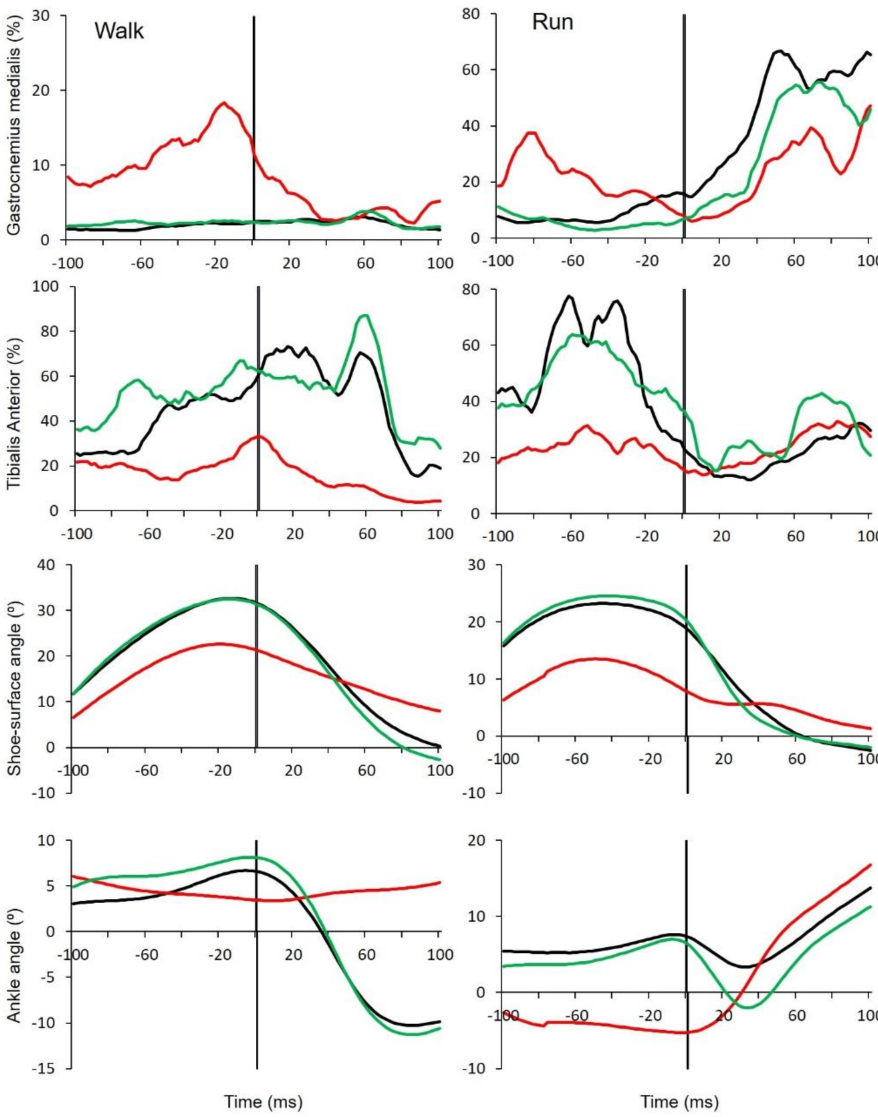

Figure 9.1. Mean gastrocnemius medialis and tibilais anterior activations (top), as well as, shoe-surface angle and sagittal ankle angle (bottom) during walking (left) and running (right) in an example participant. Shoe conditions IM (red), US (Bubble Gym, green) and CS (black) are displayed. Time is between 100ms before and after initial contact (black line). These changes in ankle muscle activations positioned the ankle and foot to increase stability at initial contact.

A promising result was the significantly increased peroneus longus activations in the loading phase whilst treadmill running and walking (trend only $p = .062$) in IM and on IS during the ascending phase of the forward and lateral lunges (trend only $p = .060$) in IM. The reason for the non-significant increase whilst walking may relate to the smaller sample size ($n = 12$), reducing the statistical power. However, the small effect size ($\eta^2 = .245$) suggested the magnitude of effect was limited. Increased peroneus longus activations in the ascending phase of the forward and lateral lunges suggests IM provides a greater stimulus in more demanding tasks.

Largely no changes were observed in thigh muscles activations across movements (Chapter 5 and 7). This aligns with the majority of previous US research during locomotion (Table 2.1). In contrast, increased vastus medialis, vastus lateralis and medial hamstring activations during walking (Voloshina et al., 2013) and vastus medialis and medial hamstring activations during running (Voloshina & Ferris, 2015) have been reported on an irregular treadmill surface. This likely relates to the more challenging treadmill surface used in these studies. The wooden blocks used had increased heights (1.27, 2.54 and 3.81cm) compared to the inserts attached to the IS treadmill of this research (1.0 and 1.5 cm) and objects inside the IM bags (cubes: 1.5 cm, ball bearings: 1.2 cm).

9.3. Short-term training effects of shoe-surface instability

Due to the IM materials not being robust, only the acute adaptations could be explored. Therefore, a training intervention on the IS that provided a similar level of unpredictable instability (Chapter 5) was used to assess the short-term training effects (Chapter 8). We opted to maintain the same IS treadmill so results would be more applicable to the IM. No increased ankle strength or static balance improvements were found relative to a control group. Thus, no support is provided for these short-term training benefits with the unpredictable instability provided by the IS used for healthy participants. Yet, this provides practical implications for studying different populations (Section 9.5). Alternatively, the IS needs to provide greater perturbations required to de-stabilise healthy participants.

The increased gastrocnemius medialis and peroneus longus activations in the studies that investigated acute responses (Chapter 5, 6 and 7) gave some reasoning that unpredictable instability may increase ankle muscle strength after regular exposure. However, the activation response was greater during the more demanding tasks: running and lunges compared to walking. The 6-week IS walking intervention was selected because it is most applicable to activities of daily life where IM could be used as a convenient training modality. Perhaps, unpredictable instability during demanding tasks would provide a greater training stimulus for ankle strength and balance improvements in healthy participants.

Perturbation training causes stiffening of the lower-limb joints initially, but this reduces with more exposure time (Chmielewski et al., 2005). Advantageous functional movement patterns are learnt to offset perturbations, such as increasing knee flexion that reduces loading to the joint (Hurd et al., 2004). Some evidence of this was observed by the sagittal ankle ranges of motion increasing between 1 and 20 minutes on the IS in the pilot study (Figure D.5). This suggests participants were relaxing, allowing their degrees of freedom to be unlocked after initially being exposed to IS (Vereijken et al., 1992). Results of the pilot implied fast adaptations occurred to improve stability, but more evidence is needed before this discerns the lack of training benefits in the training intervention. Investigating kinematic adaptations learnt to improve dynamic stability warrants further investigation to fully explore the short-term training benefits of unpredictable instability.

Improving the functional level of joint variability is a proposed benefit of training with unpredictable shoe-surface instability (Chapter 5). 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 et al., 2005; section 2.4). Performance can be achieved consistently through a variety of movement pathways, increasing adaptability to perturbations (Davids et al., 2006; Latash, 2012; Wilson et al., 2008). There is some evidence to suggest this also reduces the risk of chronic overuse injuries (Hamill et al., 1999; Heiderscheit et al., 2002; section 2.4.4). Only recently, has the movement variability forced onto the system by instability training (Strøm et al., 2016; Cimadoro et al., 2013) been associated with the positive functional benefits from the dynamics systems perspective in US research (Stöggl et al., 2010; Buchecker et al., 2017). Thus, altering environmental constraints, such as shoe-surface instability, could train this optimal level of variability.

Movement variability of the ankle and knee were analysed in this thesis by comparing the dispersion of data from the coefficient of variation and the maximum ankle inversion angle across stance phases of gait cycles (Chapter 5). Across walking and running, IS displayed the greatest variability, with IM having similar levels of frontal ankle motion and sagittal knee variability (Chapter 5 and Appendix E). Throughout forward lunges, IM displayed increased frontal ankle motion compared to CS, and during the descending phase in lateral lunges (Chapter 7). The pilot study (Appendix E) provided some evidence that within 1-hour of walking on the IS treadmill acute adaptations occurred suggesting the sensorimotor system had learnt to stabilise the level of knee variability (Figure D.4), but ankle variability remained high (Figure D.2, D.3). As the foot segment is closest to the perturbation of the IS, and based on these findings, it is likely shoe-surface instability will have a greater training effect on the ankle opposed to the more proximal knee.

9.4 Practical implications

The content of this thesis has implications for the design of future unstable footwear, as well as, sporting and clinical applications.

9.4.1. Unstable footwear recommendations

Fitting in exercise to maintain a healthy lifestyle in the ever-busy schedules of modern day life is considered one of the major barriers (Mailey et al., 2014). A key advantage of US is that they allow training to occur during tasks of daily life. Unpredictable instability, rather than fixed instability provided by US, requires greater joint stability adaptations and deemed a more demanding stimulus (Chapter 6). Yet, the lack of training effects in the intervention (Chapter 8) suggests more dynamic functional movements or larger shoe-surface perturbations are required, but this might not be applicable to daily routines. One suggestion to this conundrum would be to market this type of functional footwear with different difficulty or instability level. This would allow the consumer to choose the level of instability most appropriate for their needs. For example, grade 1 could represent low instability training, grade 2 moderate instability training and grade 3 high level instability training. This would need biomechanical testing of prototypes with different size of objects inside the IM bags to evaluate the level of instability they are achieving during different movement activities.

However, a major concern for manufacturers is safety, as they are liable for any personal injuries to consumers caused by their products (Health and Safety act, 1987). Instability training devices may reduce injury risk through the associated neuromuscular training benefits (McGuine & Keane, 2006; Hrysomallis, 2007). Yet, performing exercises using such devices poses a heightened risk because they purposefully aim to destabilise the sensorimotor system, which increases the risk of falling or causing injury. US with similarly designed reduced bases of support (rocker soles) also may increase the risk of falls (Albright & Woodhall-Smith, 2009). Despite this risk, more than 25 unstable shoe brands have been developed (Nigg et al., 2012).

Products with unpredictable shoe-surface instability have not been developed for commercial use, only in clinical settings overseen by a physician, such as perturbation training. Although this research favours the idea of wearing a shoe incorporating unpredictable instability during activities of daily life, the reason it may not have been already adopted could be that they are not considered safe. Across all testing sessions with IM and IS within the experimental studies of this thesis no injury or fall incidents occurred, and the investigators felt the risk was low due to the perturbations being small. In Chapter 5, maximum ankle inversion across all steps during walking and running in IM and IS was within a normal range (Ottaviani et al., 1995). Measuring how close the maximum joint angles are to the end of their range of motion is proposed as a first step to address potential injury risk in US designs. Yet, it remains to be seen if a company will develop a commercial product with similar instability to IM and IS used in this research and how they will manage the safety risk.

Li Ning did part fund this PhD and provide the footwear and materials, but no commercial footwear was manufactured for the research. Rather the experimental studies involved basic research questions to explore if the biomechanical and neuromuscular effects of a shoe with random instability might be beneficial. Therefore, future publications associated with this work will state the following conflict of interest statement:

"While Li Ning did fund this study and provide the footwear/materials there is no conflict of interest to declare. The footwear/materials are not commercial products but developed to answer a fundamental research question about unexpected perturbations. The primary author wrote the paper with no influence from the funder. Dr Thorsten Sterzing was affiliated with the funder and helped design, interpret and proof read the manuscript but has no obligations to Li Ning regarding publications of basic research questions."

9.4.2. Shoe-surface instability in sports settings

Across sports, ankle sprains are among the most common (Fong et al., 2007). This research identified the unpredictable instability shoe-surface conditions, IM and IS, had greatest effect on the ankle:

- The only changes to lower-limb muscle activation levels occurred in the shank.
- Frontal ankle joint movement variability increased to the greatest extent, where lateral ankle sprains occur.
- Reduced shoe-surface and ankle plantarflexion angles occurred at initial contact.

Therefore, it is proposed IM and/or IS could be useful in the prevention and rehabilitation of ankle sprains. It has been reported regular training on balance boards reduces ankle injury risk (McGuine and Keene, 2006), and wobble boards reduces the risk of sports related injuries in general (Emery et al., 2005). Yet not all prospective instability training device studies report reduced injury risk (Söderman et al., 2000). The unpredictable perturbations from IM and IS may better represent the scenarios when ankle inversion injuries occur, especially during closed kinetic chain functional movements (Blackburn et al., 2003).

It has been speculated feedforward muscle pre-activation is more important to maintaining ankle stability rather than feedback muscle responses (Gutierrez et al., 2009). Particularly, reduced peroneus longus pre-activation has been found in those with functional ankle instability compared to healthy participants (Suda and Sacco; 2011; Caufield et al., 2004). Suda and Sacco (2011) also found gastrocnemius lateralis activated earlier in healthy participants. Thus, the observation of earlier activation of the gastrocnemius medialis in IM during forward and lateral lunges may be suitable to train feedforward ankle muscle activations to reduce injury risk. Future research is needed to support this assumption.

Regular walking on IS did not significantly increase measures of ankle muscle strength and static single-legged balance in the healthy participants. More dynamic movements and challenging instability are suggested (Chapter 8). However, those with functional ankle instability from lateral sprains have reduced strength and balance deficits. Regular training with IS or IM may enhance rehabilitation by helping to restore ankle strength and balance, as found using MBT (Kaelin et al., 2011).

9.4.3. Shoe-surface instability in clinical settings

Falls are a major public health problem, costing the NHS £2 billion per year and being the most common form of death in those over 65 years (Fenton, 2014). Various assessments are available to clinicians to identify fall risk, such as questionnaires, various functional tests including biomechanical tests using inertial sensors, EMG and accelerometers (Howcroft et al., 2013). However, these tests are carried out on level, stable ground when 24% of falls occur walking across uneven surfaces (Berg et al., 1997). Evaluating how well individuals adapt to uneven surfaces could be implemented by conducting a gait analysis on the IS and/or IM conditions developed in this research. This may identify those persons who are at increased risk of falls on uneven terrain better than current assessment measures. Moreover, as traversing uneven terrain is more demanding to the sensorimotor system, this may enable potential fallers to be detected at an earlier stage.

As well as a preventative measure, shoe-surface instability could be implemented during the advanced stages of gait re-training in rehabilitation. A similar intervention of stepping over obstacles in a virtual environment whilst treadmill walking showed improvements in gait speed, step length and clearance height over obstacles in stroke patients over a 2 week training period (Jaffe et al., 2004). Re-learning strategies to cope with IS in a safe environment may increase the chances of maintaining stability in the real world and reduce fall risk.

9.5. Future research directions

Directions for future research are recommended based on the findings and limitations of this thesis to advance our understanding of shoe-surface instability. Across experiments conducted, the only comparison to another unstable footwear brand was to the Bubble Gym shoe (Chapter 6). Conducting similar studies with a range of US brands would enhance the external validity of this work. Specifically, comparing to MBT and to Reflex Control Schuh (although not commercially available) would help discern any enhanced effects of the IM technology from increasing levels of predictable instability from a predictable fixed outsole (Schiemann et al., 2015; Turbianski et al., 2011).

Short-term training benefits of IM could not be explored because a durable prototype was not developed. A training intervention with IM will allow the true training effects to be revealed and better guidance given to the design of US. The findings from Chapter 8 can be used to design a more comprehensive training intervention using the recommendations provided. It was also noted participants could perceive the objects inside the rubber bags under the plantar sole in the IM shoe, which may have caused the kinematic adaptations, rather than the instability. It is suggested future prototypes should aim to reduce this haptic sensation.

These studies were limited to young healthy participants, but as recognised IM may have a more profound effect on those with ankle muscle strength or balance deficits. Future research should assess these different populations. As identified, falls in the elderly (Section 9.4.3) and ankle sprains in sports players (Section 9.4.2) are major issues that may be preventable by sensorimotor training (Kim & Lockhart, 2012; Granacher et al., 2006;

McGuine & Keene, 2006; Hrysomallis, 2007). US interventions allow convenient training throughout activities of daily life and would likely improve adherence to training programs.

Future studies should include more comprehensive measurement and analysis approaches. EMG analysis was limited to the magnitude of muscle activations, but more in-depth analysis will allow greater insight into muscle function and adaptations to shoe-surface instability. Muscle onset latency improves after sensorimotor training with unstable support surfaces, such as wobble boards and may be an important mechanism to injury prevention (Granacher et al., 2006; Clarke & Burden, 2005). Wavelet analysis is a powerful tool that would enable the timing, intensity and frequency to be evaluated simultaneously, with the latter also giving indications of the muscle conduction velocity (von Tscharner, 2000). This analysis has revealed differences whilst standing in MBT (Landry et al., 2010), but is limited to just 6 participants whilst walking (Nigg et al., 2006).

Surface EMG is limited to measuring the large superficial muscles, whereas it has been suggested US has a greater training effect on the smaller ankle muscles (Nigg et al., 2012). Using specially designed circumferential electrode arrays enables measuring activations of these small muscles (Landry et al., 2010). Assessing muscle strength by scanning muscle volume after a training intervention would increase the validity of this work, assuming instability training does have the largest influence on the smaller muscles.

This research suggests the IM shoe may improve the level of functional joint variability. To investigate this, more complex analysis approaches are required such as the uncontrolled manifold approach or co-ordination variability analyses to provide detail about how different joints work together to maintain stability. Besides, joint stiffness analysis from the angle-moment curves revealed differences in sagittal ankle and knee adaptations in IM compared to the Bubble Gym shoe. Although limitations with this technique were acknowledged (Chapter 6; Latash & Zatsiorsky, 1993), future stiffness analyses should develop frontal plane joint stiffness methods and evaluate joint impedance by applying a known perturbation about the joint (Rouse et al., 2014).

9.6. Thesis Conclusion

The biomechanical and neuromuscular responses to a shoe that provided continual unpredictable perturbations was evaluated for the first time. The developed IM increased instability to a similar level as an IS and more than a control shoe-surface, indicated by increased variability of frontal ankle and sagittal knee motion. Kinematic adaptations were made to enhance stability, which were associated with increased ankle muscle activations to position the foot and stabilise the ankle. Additional joint stability strategies were required in IM compared to an US, with fixed instability, indicating unpredictable perturbations are more demanding. Short-term training with the IS did not confer increased ankle muscle strength or postural balance enhancement. A more challenging training programme that examines kinematic and variability responses during unpredictable perturbations warrants investigation before IM can be validated as having benefits to consumers. Instead, the unpredictable instability provided by IM and IS may aid those with ankle muscle strength and balance deficits. These findings could contribute to the development of more effective clinical interventions for ankle injury prevention and rehabilitation programs in sporting applications, as well as, falls prevention and gait re-training.

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APPENDICES

Appendix A: Plantar pressure in the irregular midsole

Plantar pressure measurements were recorded in IM whilst walking and running (Chapter 5). This was to determine whether IM provides a more varied shoe instability than a control shoe-surface condition (CC). It was hypothesised that IM and IS would cause a similarly increased variability of plantar pressures compared to CC.

Specification of the plantar pressure measurement system and details of the data collection and analysis are provided in section 5.3.3. As stated, artefacts in the form of pressure 'hot-spots' were observed in the results (Figure A1). This was likely caused from landing on a corner of one of the cubes inside the IM bags in certain steps. Therefore, the results are reported in this Appendix because despite this limitation in the data it provides further evidence the IM does indeed provide continually different instability, although they should be interpreted with caution. To help limit the effects of these artefacts on the results, any peak pressure values in steps that were greater than 1.5 times the mean across all steps within the trial were removed from subsequent analyses. To assess variability of pressure distributions the insole pressure profiles were visually observed and compared statistically using the tests outlined in Section 5.3.8 on the coefficient of variation (CV) of the peak pressures and relative loads in the different pressure mask regions (5.3.3.).

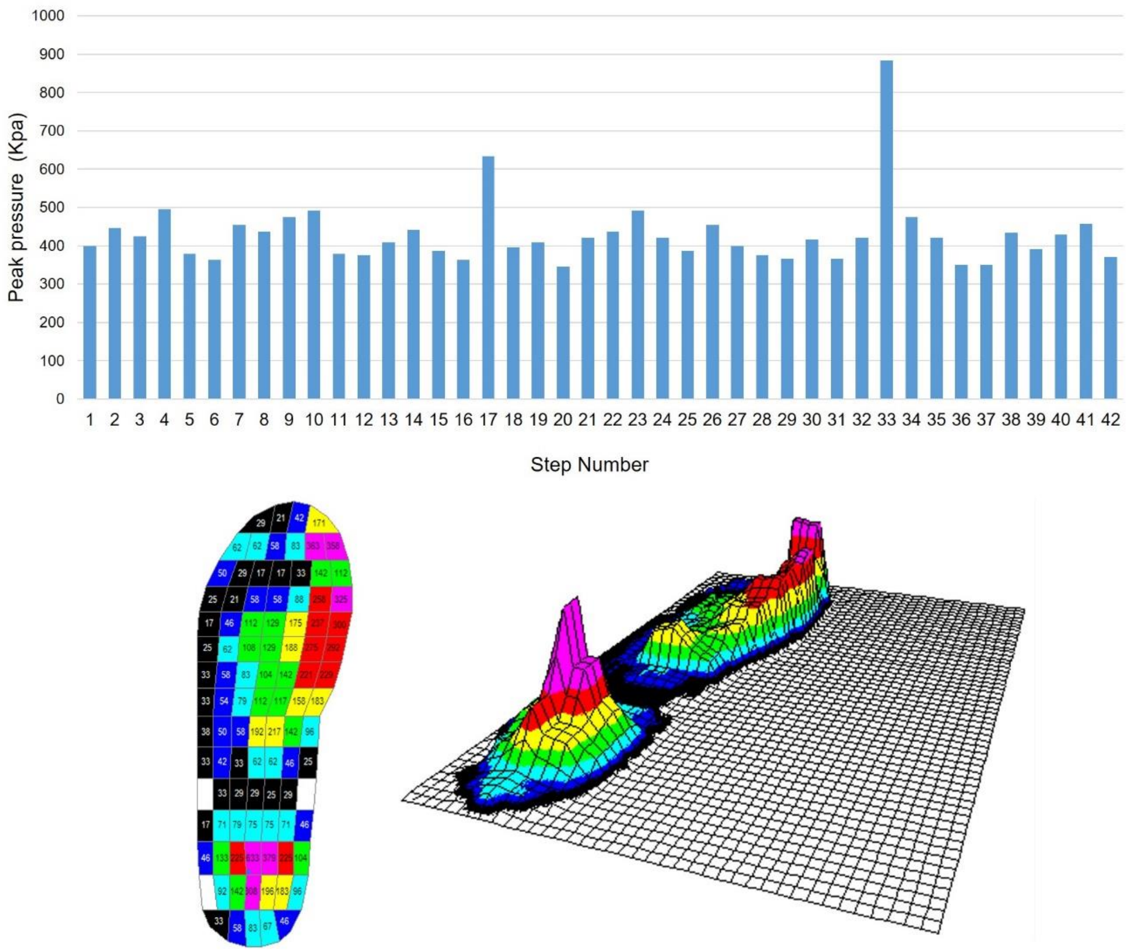

Figure A1: The maximum peak pressure recorded across all sensors for each step in a running trial (top). Two pressure spikes occurred on step 17 and 33 which were greater than 1.5 times the mean peak pressure across all steps. The pressure 'hot-spot' for step 17 is visually displayed in 2D (bottom left) and 3D (bottom right).

Results:

Compared to CC, IM had increased variability of peak pressure distributions that could be qualitatively observed from the insole pressure profiles (Figure A2-A3). Variability (CV) of peak pressures was greater in IM and IS compared to CC for the overall plantar surface and all mask regions whilst walking (Table A1) and running (Table A2). In the medial and lateral rearfoot, IM was more variable than IS, whilst in the lateral midfoot IS was more variable than IM for walking and running. Distribution of the relative loads was more variable in IS than CC in all mask regions for walking (Table A3) and running (Table A4).

In the rearfoot, IM had increased variability of relative loads compared to CC whilst walking and running. In the midfoot IM was also more variable than CC during running.

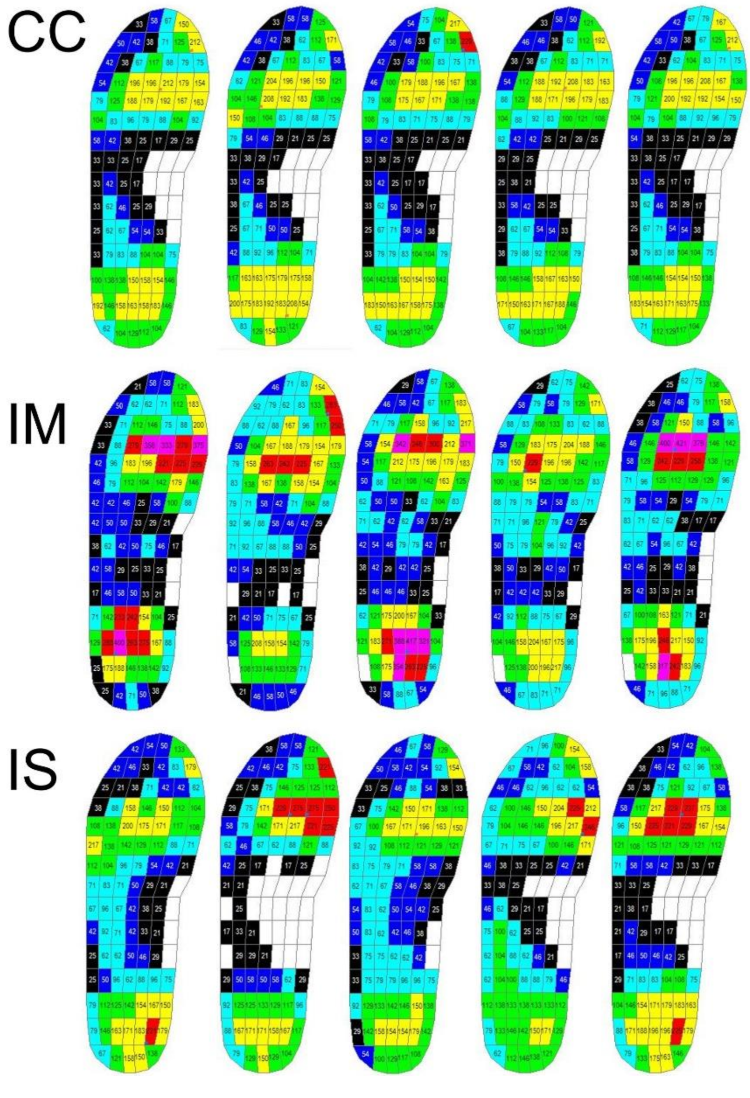

Figure A2: Regional peak plantar pressure distributions during 5 consecutive steps in the control shoe condition (top), IM (middle) and the irregular surface (bottom) during **walking** in a typical participant. The different colours represent the level of peak pressure (KPa) recorded under each sensor. Lowest to highest pressures are illustrated by: white, black, blue, cyan, green, yellow, red and pink. Notice the more varied array of pressure in IM compared to CC.

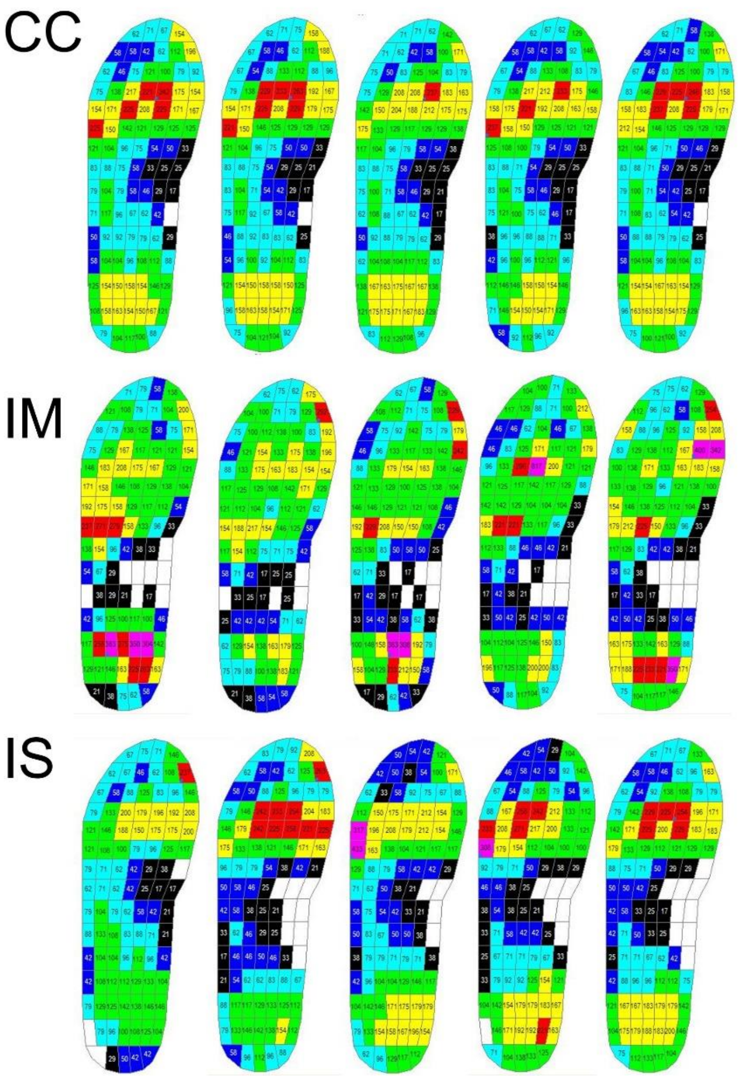

Figure A3: Regional peak plantar pressure distributions during 5 consecutive steps in the control shoe condition (top), IM (middle) and the irregular surface (bottom) during **running** in a typical participant. The different colours represent the level of peak pressure (KPa) recorded under each sensor. Lowest to highest pressures are illustrated by: white, black, blue, cyan, green, yellow, red and pink. Notice the more varied array of pressure in IM compared to CC.

Table A1: Mean (SD) variability (CV) of peak pressures during **walking** across participants

Region	CC	IM	IS	p-value	Post hoc result
Overall	11.2 (2.9)	16.1 (4.4)	16.9 (3.0)	<.001	IM & IS > CC
Lateral Rearfoot	8.1 (3.2)	22.9 (6.0)	11.5 (2.8)	<.001	IM > IS > CC
Medial Rearfoot	6.7 (2.4)	22.7 (5.5)	12.6 (3.0)	<.001	IM > IS > CC
Lateral Midfoot	13.1 (5.3)	24.5 (6.2)	35.8 (5.7)	<.001	IS > IM > CC
Medial Midfoot	9.3 (8.7)	25.3 (11.1)	33.0 (7.0)	<.001	IM & IS > CC
Lateral Forefoot	12.4 (3.9)	21.1 (7.4)	22.0 (4.6)	<.001	IM & IS > CC
Medial Forefoot	10.9 (3.7)	18.1 (5.2)	18.2 (2.7)	<.001	IM & IS > CC

Table A2: Mean (SD) variability (CV) of peak pressures during **running** across participants

Region	CC	IM	IS	p-value	Post hoc result
Overall	8.1 (1.8)	14.8 (3.0)	16.9 (2.8)	<.001	IM & IS > CC
Lateral Rearfoot	13.6 (3.9)	28.2 (7.5)	18.0 (3.0)	<.001	IM > IS > CC
Medial Rearfoot	11.7 (3.3)	25.6 (7.4)	17.0 (2.9)	<.001	IM > IS > CC
Lateral Midfoot	8.8 (3.4)	22.5 (6.9)	30.3 (7.5)	<.001	IS > IM > CC
Medial Midfoot	8.5 (4.3)	24.0 (8.6)	26.4 (4.3)	<.001	IM & IS > CC
Lateral Forefoot	8.1 (2.4)	24.8 (8.6)	23.4 (3.0)	<.001	IM & IS > CC
Medial Forefoot	8.6 (1.9)	16.9 (4.4)	17.6 (3.4)	<.001	IM & IS > CC

Table A3: Mean (SD) variability (CV) of relative during **walking** across participants

Region	CC	IM	IS	p-value	Post hoc result
Lateral Rearfoot	10.0 (3.9)	18.6 (5.4)	15.2 (2.7)	<.001	IM & IS > CC
Medial Rearfoot	9.6 (3.1)	17.0 (4.8)	15.0 (3.2)	<.001	IM & IS > CC
Lateral Midfoot	18.1 (7.2)	18.5 (6.0)	48.4 (11.1)	<.001	IS > IM & CC
Medial Midfoot	21.7 (7.1)	22.6 (11.2)	88.7 (28.7)	<.001	IS > IM & CC
Lateral Forefoot	13.5 (2.5)	15.1 (4.7)	23.5 (5.5)	<.001	IS > IM & CC
Medial Forefoot	14.4 (3.2)	11.7 (3.2)	20.9 (3.5)	<.001	IS > CC > IM

Table A4: Mean (SD) variability (CV) of relative during **running** across participants

Region	CC	IM	IS	p-value	Post hoc result
Lateral Rearfoot	14.3 (5.2)	19.7 (5.1)	20.9 (5.8)	<.001	IM & IS > CC
Medial Rearfoot	12.4 (4.5)	16.2 (4.7)	19.3 (5.8)	<.001	IS > IM > CC
Lateral Midfoot	7.3 (1.8)	13.6 (4.1)	34.0 (9.5)	<.001	IS > IM > CC
Medial Midfoot	9.0 (3.0)	18.8 (6.3)	52.6 (12.6)	<.001	IS > IM > CC
Lateral Forefoot	6.5 (1.6)	9.7 (2.5)	15.4 (2.6)	<.001	IS > IM & CC
Medial Forefoot	7.0 (1.6)	7.3 (1.8)	12.1 (2.5)	<.001	IS > IM & CC

Interpretation:

In summary, the plantar pressure results support the evidence that IM does provide unpredictable instability from the more varied distribution on plantar loads compared to a control shoe. The localised peak pressure variability for the individual sensors was greater in IM than the general distribution of loads among the different foot regions. This suggests the objects inside the IM rubber bags were moving around, but not causing a more global change in the locomotion pattern.

Appendix B: Kinematic ground contact algorithm selection

The purpose of this pilot test was to determine which ground contact algorithm from the literature worked best for the walking, running and shoe conditions of this research.

Two participants completed 5 walking and running trials overground across a force plate and 90 seconds on a treadmill in CC and IM shoe conditions, according to the protocols in Chapter 5 and 6 respectively. A footswitch (DTS FootSwitch, Noraxon, Inc., Scottsdale, AZ, USA) was placed inside the right shoe during all trials to give a better indication of contact on the treadmill. Right lower limb kinematics were recorded at 500 Hz by an eight camera motion analysis system (Qualisys AB, Gothenburg, Sweden), and processed as detailed in Section 4.2.

Three ground contact algorithms from literature were identified that were appropriate for this research:

- Hreljac and Marshall (2000)
    - Initial contact (IC) defined by the vertical acceleration peak of the heel marker and toe-off (TO) by the horizontal acceleration peak of the 5th metatarsal marker.
- O'Connor et al. (2007) (Foot velocity algorithm)
    - IC and TO defined by the vertical velocity peaks and troughs of the midpoint between the heel and toe markers
- Maiwald et al. (2009) (Foot Contact Algorithm)
    - IC defined by the vertical acceleration peak of the heel marker and TO by the vertical acceleration peak of the toe marker

The events of IC and TO were qualitatively compared between the ground contact algorithms, footswitch and force platform.

Results showed the footswitch IC and TO events aligned well with the onset and offset of the vertical ground reaction force. The ground contact algorithm IC and TO events were more difficult to interpret, with certain algorithms aligning better with the footswitch events for walking or running and others better in the stable shoe condition. This may be due to high data capture frequency (500Hz), different filters applied and the IM shoe condition used compared to previous studies (60Hz: Hreljac and Marshall, 2000; O'Connor et al., 2007 and 240Hz: Maiwald et al., 2009). The foot velocity algorithm (O'Connor et al, 2007) was deemed most suitable for the walking trials (Figure A1) and the foot contact algorithm (Maiwald et al, 2009) for running trials (Figure A2), which were applied in Chapter 5. There was a small delay for IC when using the foot velocity algorithm (Figure A1). This was observed in the video and the reduced first sagittal ankle rocker. Where this occurred the IC event was moved 5 frames earlier. For some subjects in some shoe conditions the algorithm did not select the correct peak in velocity or acceleration. In these cases the event was moved to the correct velocity/acceleration peak.

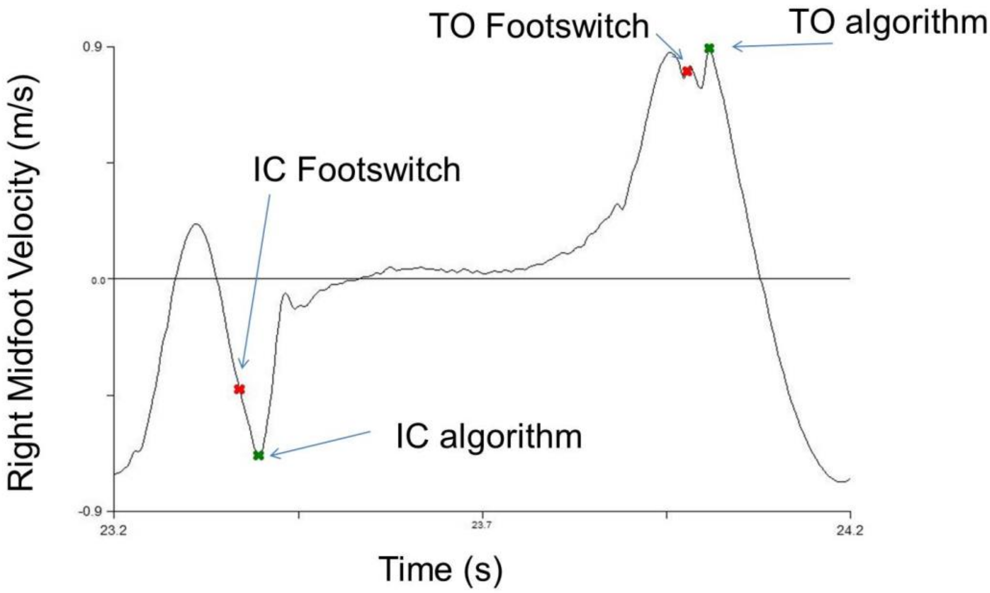

Figure A1. Example IM treadmill data of the vertical velocity of the midpoint between the heel and toe markers. The IC and TO events foot velocity algorithm (green) and footswitch (red) are displayed.

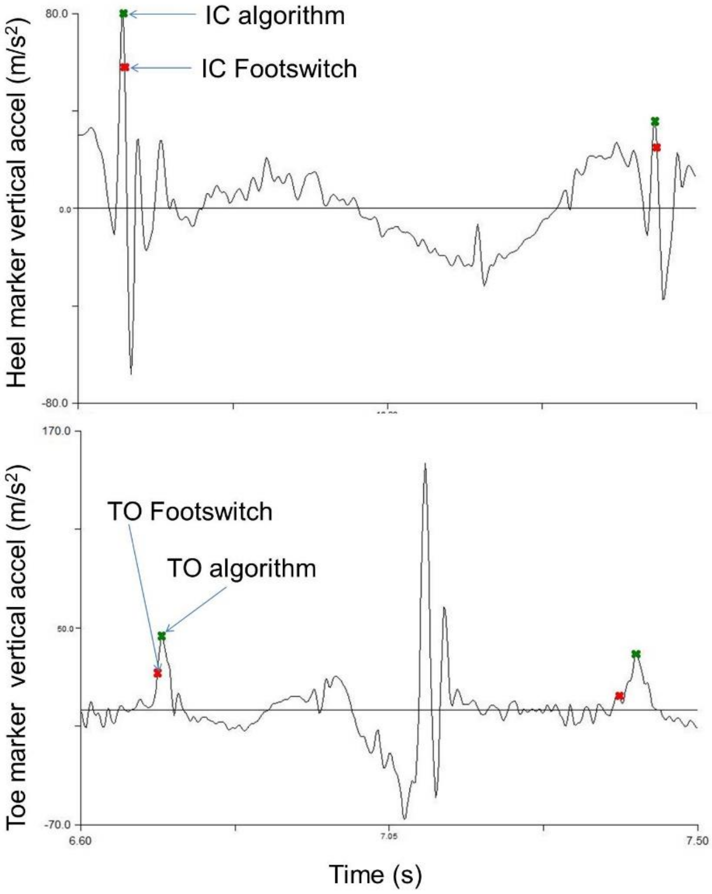

Figure A2. Example IM treadmill run data of the vertical acceleration of the heel (top) and toe (bottom) markers. The IC and TO events foot contact algorithm (green) and footswitch (red) are displayed.

Appendix C: Determining how many steps to analyse

In Chapter 5 all steps (mean 41+/- 2.6 for running and 28.6 +/-1.5 for walking) were analysed. This was based on pre-analysis showing reduced fluctuations in standard deviation with more than 30 steps (Figure B.1).

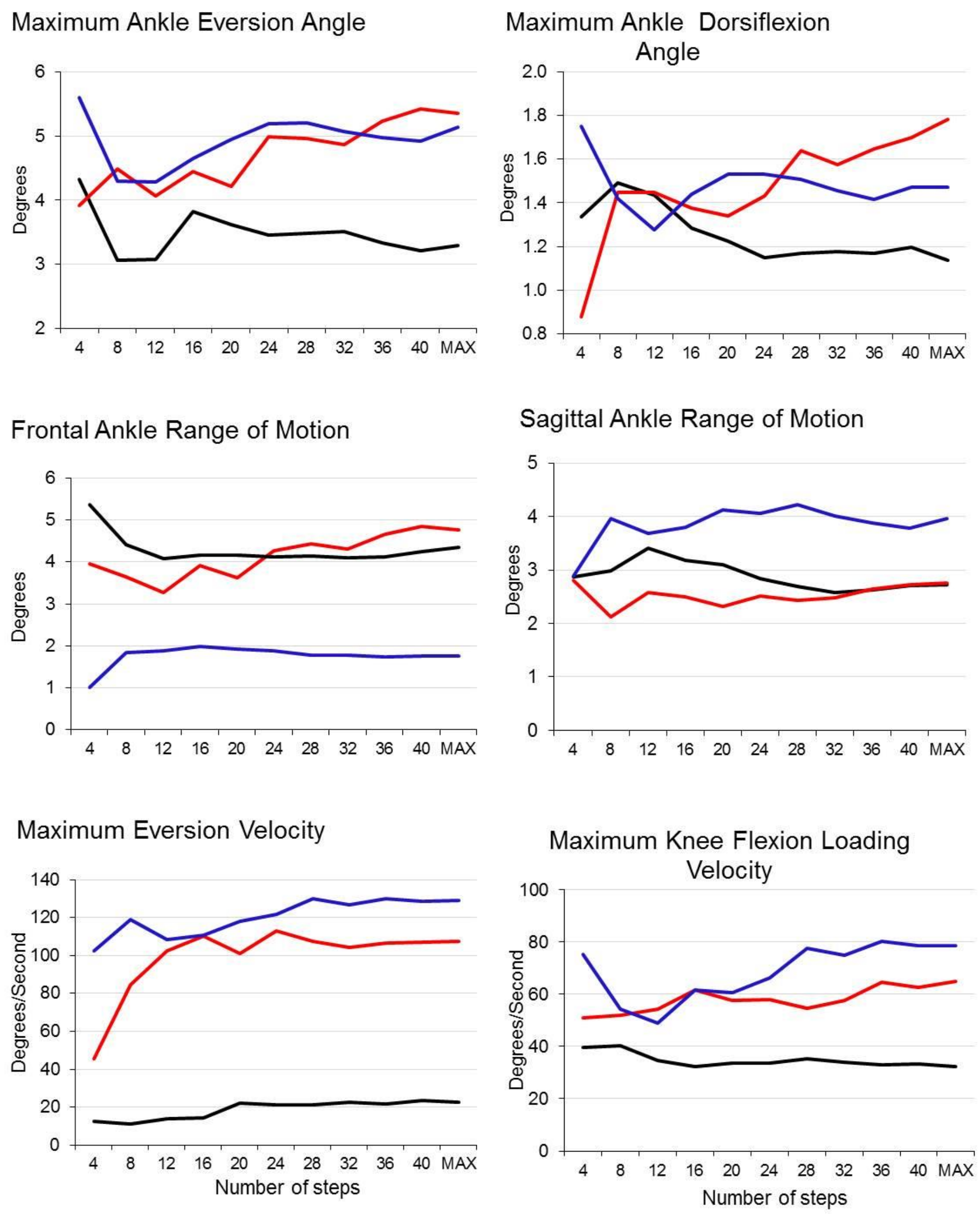

Figure B.1. Standard deviation calculated with increasing number of steps for a typical subject **running** in CC (black), IM (red) and IS (blue). Note standard deviation was increased for CC and decreased for IS on certain graphs to display the fluctuations of SD more clearly.

Appendix D: Determining linear joint-moment phases to analyse quasi-stiffness

The aim of this appendix is to determine if joint stiffness, calculated by a linear regression of joint angle-moment curves, could be computed using the average regression of all 20 trials. Mean knee joint stiffness was calculated using 2 different methods for 1 participant from Chapter 6. Firstly, the linear regression was applied to the angle-moment curve of every trial and the average of these calculated. Secondly, the linear regression was measured using the average joint angle-moment curve for each shoe condition normalised to 101 data points.

Results showed there was no difference between either method in the knee stiffness values or how well the regression fit the data (r^2 values) (Table C1). Therefore, for convenience when analysing the average stiffness values from the normalised angle-moment plots were compared statistically in Chapter 6.

Table C.1. Knee stiffness results (Nm/Kg/deg) and r^2 values during the propulsive phase of a typical subject for all 20 trials in each shoe condition. The average of these and of the mean angle-moment curve normalised to 101 data points were the same to 2 decimal places.

Trial number	**CS**		**US**		**IM**	
	Stiffness	r^2	**Stiffness**	r^2	**Stiffness**	r^2
1	.065	.97	.070	.94	.052	.95
2	.068	.97	.067	.93	.054	.93
3	.079	.98	.068	.95	.054	.95
4	.077	.96	.072	.94	.058	.95
5	.070	.96	.066	.94	.047	.93
6	.065	.96	.069	.94	.053	.93
7	.068	.96	.067	.94	.058	.93
8	.065	.94	.063	.95	.049	.95
9	.066	.96	.070	.94	.047	.93
10	.060	.97	.067	.93	.047	.93
11	.068	.95	.071	.93	.050	.93
13	.065	.94	.071	.93	.046	.90
14	.066	.98	.069	.92	.056	.92
15	.068	.94	.067	.90	.052	.95
16	.069	.95	.065	.93	.049	.94
18	.067	.96	.068	.93	.053	.96
19	.073	.94	.068	.95	.053	.90
20	.067	.94	.068	.92	.043	.96
Average of 20 trials	.068	.96	.068	.93	.051	.94
Mean of normalised curve	.068	.96	.068	.93	.051	.94

Appendix E: The acute kinematic and muscular adaptations of the lower extremity to uneven surface walking

The purpose of this pilot study was to explore adaptations learnt within the first hour of walking on an uneven treadmill surface. It was hypothesised gait cycle and kinematic variability, and muscle activations would be initially increased but reduce over the hour to a similar level as a regular surface. This was based on the lack of improved static balance and ankle muscle strength from Chapter 8.

Methods:

Two female participants from the main training group underwent additional biomechanical measurements during the pre-test and first irregular surface (IS) training session. Their age (participant 1: 26, participant 2: 25 years), height (171, 173 cm), weight (69.8, 66.3 kg) and shoe size (7, 6 UK) were reflective of the group. Participants were familiarised to the control surface and IS during the pre-test session, where they walked on each surface at their selected normal walking speed for 2 minutes each. Their gait speed was slightly faster than the group average (participant 1: 4.5 km/hr, participant 2: 4.0 km/hr). During this pre-test familiarisation, biomechanical measurements were recorded for 60 seconds after 1 minute of walking on the control surface and IS. On the day of the first training session, the regular one hour IS walking procedure was followed (see section 8.3.2), except participants wore the lab shoes and biomechanical measurements were recorded after walking for 1, 20, 40 and 60 minutes. All steps were analysed during each 60 second period recorded (Mean (SD): 49.4 (1.6), range: 47-52).

Three-dimensional lower-limb kinematics were recorded using the same set-up and procedure as Chapter 6 (Section 6.3.3). Additionally, markers were placed on the shoe tips and on the distal heel counters. Marker data were processed and ground contact events as outlined section 5.3.4. Characteristics of the gait cycle were derived from ground contact times. Mean and variability of the global gait parameters, maximum ankle and knee angles and velocities, and initial contact angles and ranges of motion were computed.

Surface electromyography (EMG) of the gastrocnemius medialis, gastrocnemius lateralis, tibialis anterior and peroneus longus were recorded and processed as detailed in Section 6.3.3. Unfortunately, the Noraxon system crashed before the initial measurement during the training session (minute 1) for participant 1 and results are for participant 2 only. Muscle pre-activations and stance phase activations were overlaid and qualitatively assessed.

Results:

Pre-test kinematic results on IS were mostly comparable to minute 1 of the training session. The gait cycle results are illustrated in Figure D.1. Participant 1 took slightly faster, shorter steps at the beginning of the training session (minute 1 mean: 1.14 seconds), similar to IS and control at pre-test results, that gradually got slower and longer during the hour (minute 60 mean =1.19 seconds). Participant 2 took shorter, faster steps during the pre-test than the training session, but the difference was negligible.

Gait cycle variability (SD and CV) also differed between participants. Participant 1 tended to reduce variability during the training session, but this was not apparent for participant 2. During the pre-test participant 2 had increased variability in IS compared to the control, alike the previous results in Chapter 5, but participant 1 had similar pre-test levels of variability between both surface conditions. These results imply gait cycle adaptations to IS are specific to the individual. Possibly, participant 1 gained greater whole body control during the hour training session on IS because increased gait cycle variability has been linked with risk of falling (Hausdorff et al., 2001; Thies et al., 2005). Although, this could be an adaptation to treadmill locomotion in general, and not the IS.

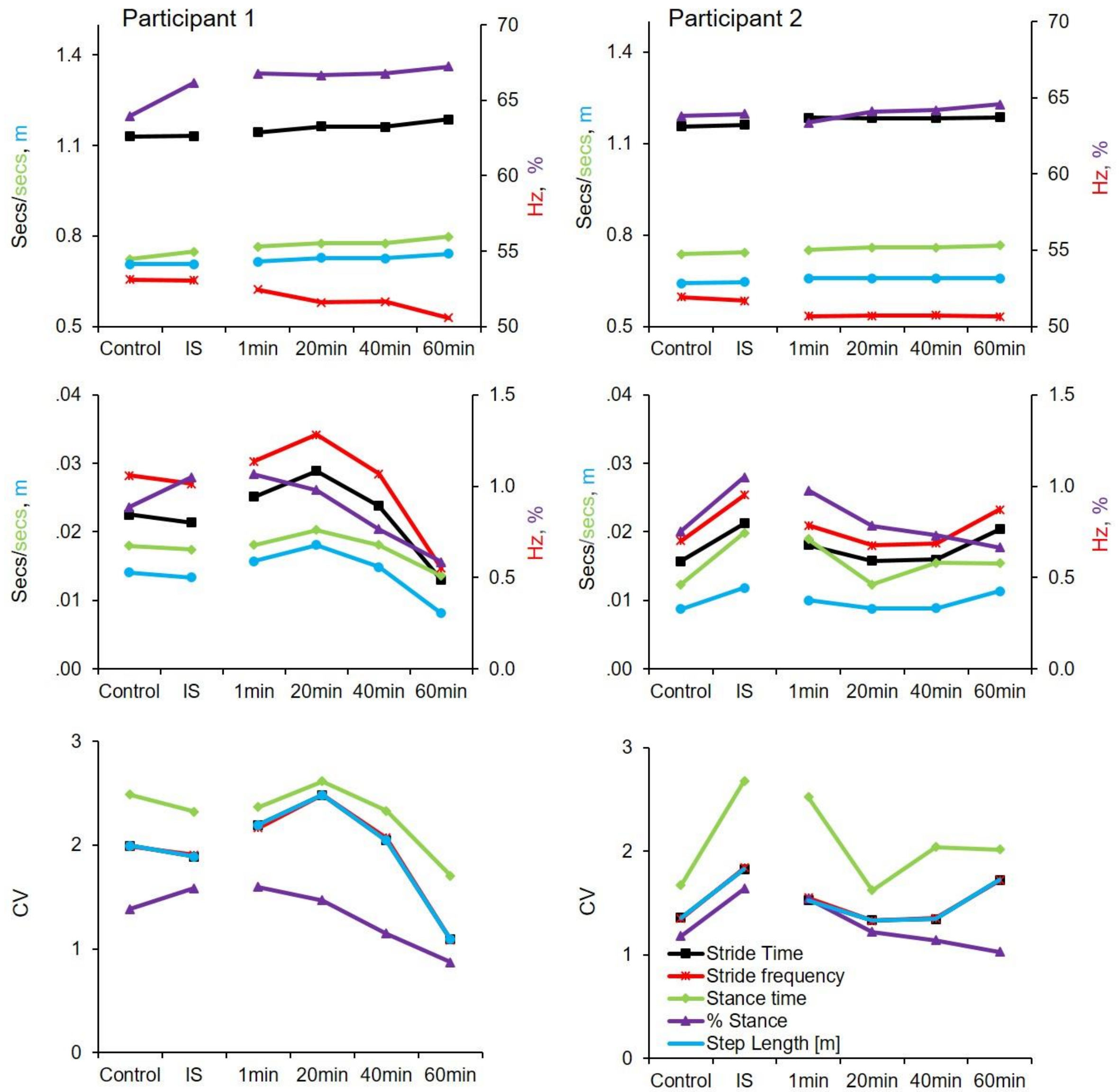

Figure D.1. Gait cycle results for participant 1(left) and participant 2 (right). The mean (top), SD (middle) and CV (bottom) are displayed for stride time (black), stance time (green), percentage of time in stance (purple), step length (blue) and stride frequency (red). Pre-test results are separated from the hour training session.

The sagittal ankle, frontal ankle and sagittal knee kinematic results are illustrated in Figures D.2 to D.4 respectively. Variability of sagittal and frontal ankle kinematics during stance remained higher on all analysed time points on IS than on the control at pre-test.

Due to the ankles close proximity to the ground, and surface perturbation, it is likely no adaptation (acute or long-term) can be made to reduce ankle variability on IS. Mean ankle eversion velocity were also higher in IS than the control in all time points analysed.

Nevertheless, a common strategy was found at the sagittal ankle to suggest participants were more comfortable walking on IS by the end of the training session. Mean sagittal ankle ranges of motion increased after minute 1 of the training session (Figure D.5).

Mean sagittal knee flexion angles and velocities were increased in participant 1 compared to participant 2. Adaptation strategies also differed. In participant 1 the maximum angle, range of motion and flexion velocity tended to increase after minute 1 of the training session, whereas they gradually reduced even further throughout the hour period in participant 2. Regardless of these individual strategies, knee variability reduced during the training session to a similar level as the pre-test control in both participants. This was more apparent for participant 2, where variability stabilised by minute 40 of the training session. The SD of participant 1 reduced by minute 20 for knee range of motion, and possibly flexion velocity as well. This appeared to be a greater effect for the CV due to the increased mean values. Therefore, participants increased knee control during the training session, indicated by the lower variability that reflected the control at pre-test. This learning effect would have enabled greater stability further up the kinetic chain.

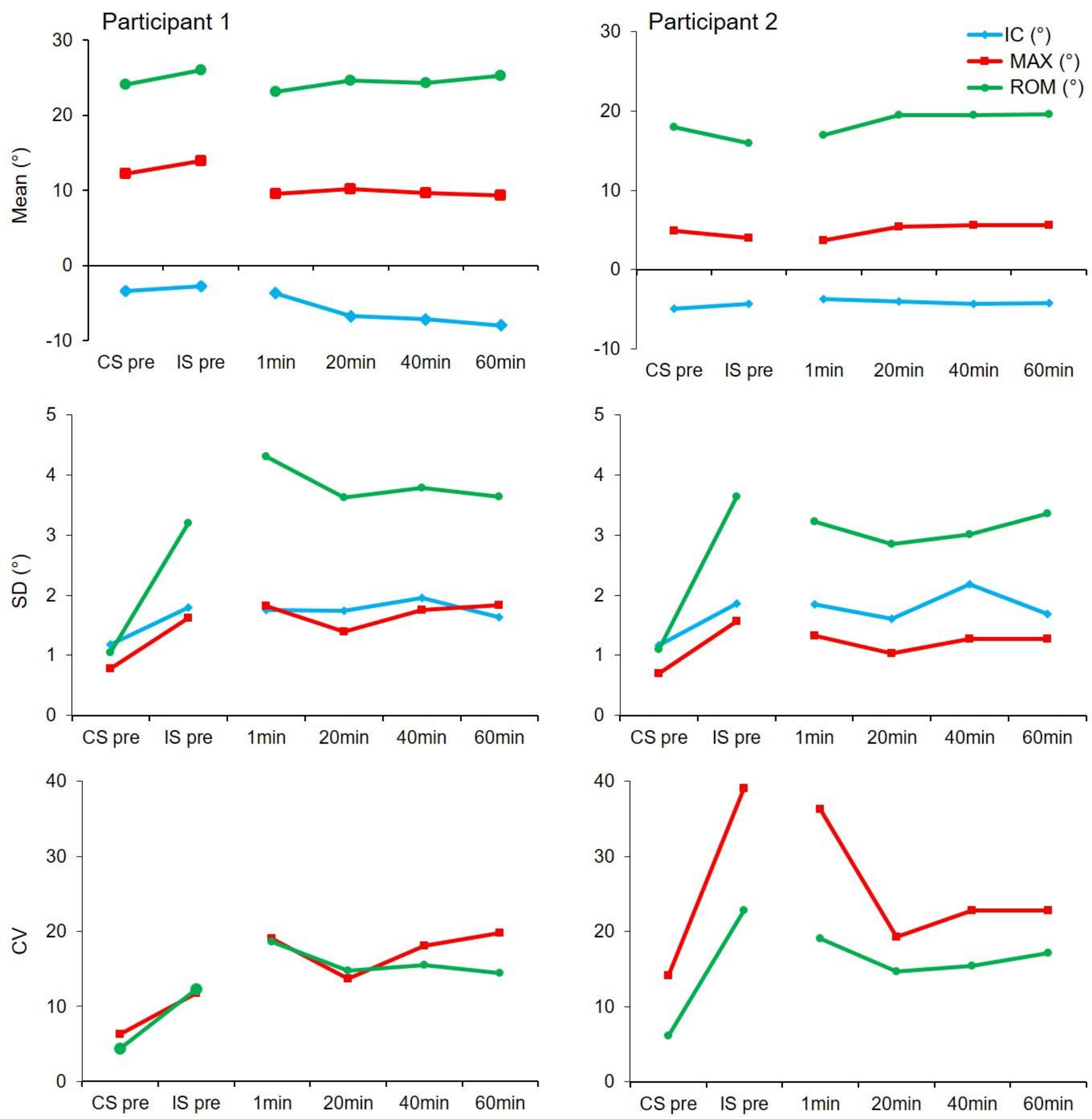

Figure D.2. Sagittal ankle kinematic results for participant 1(left) and participant 2 (right). The mean (top), SD (middle) and CV (bottom) are displayed for initial contact (blue), maximum angle (red) and range of motion (green). Pre-test results are separated from the hour training session.

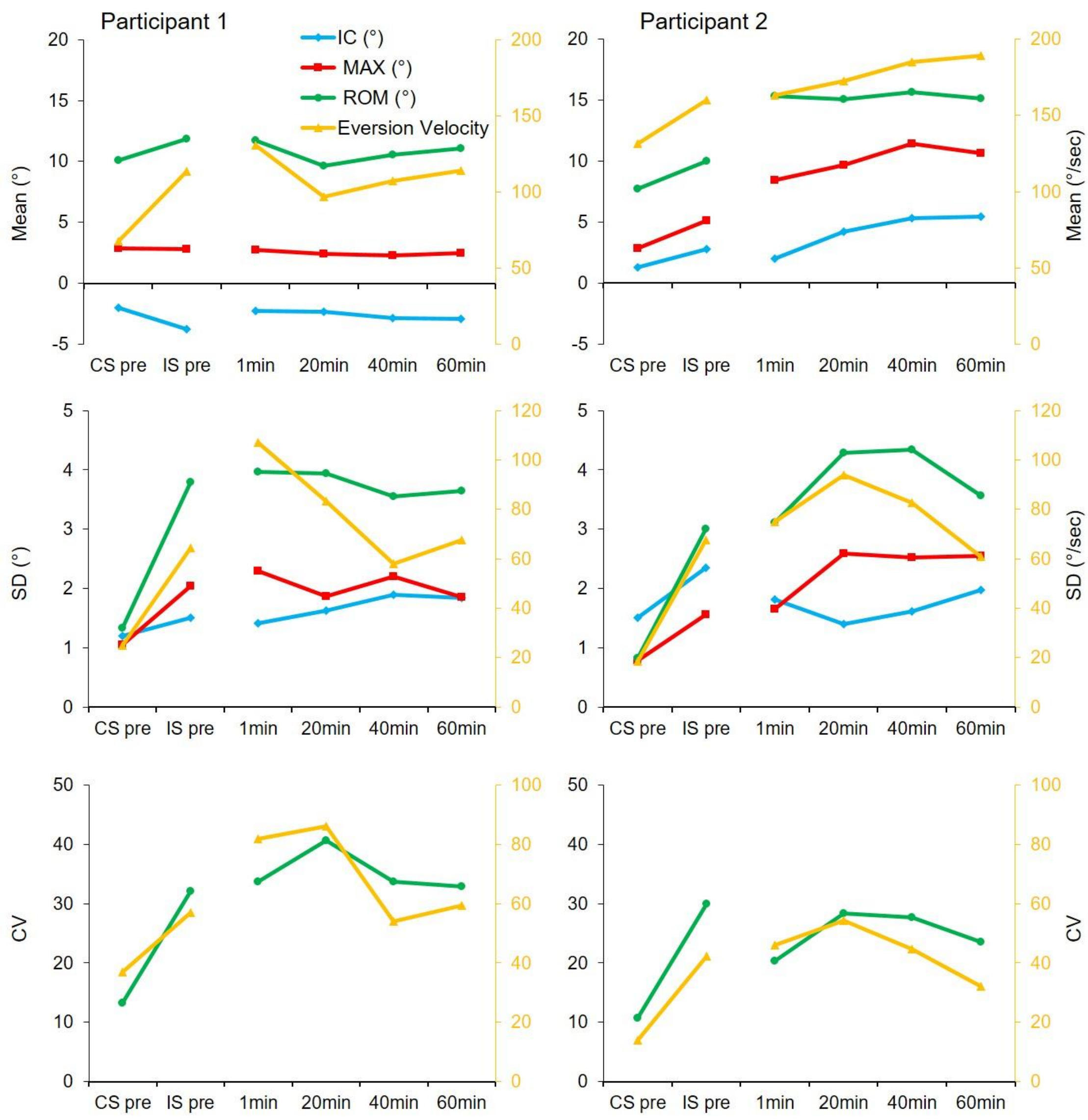

Figure D.3. Frontal ankle kinematic results for participant 1(left) and participant 2 (right). The mean (top), SD (middle) and CV (bottom) are displayed for initial contact (blue), maximum angle (red), range of motion (green) and eversion velocity (yellow).

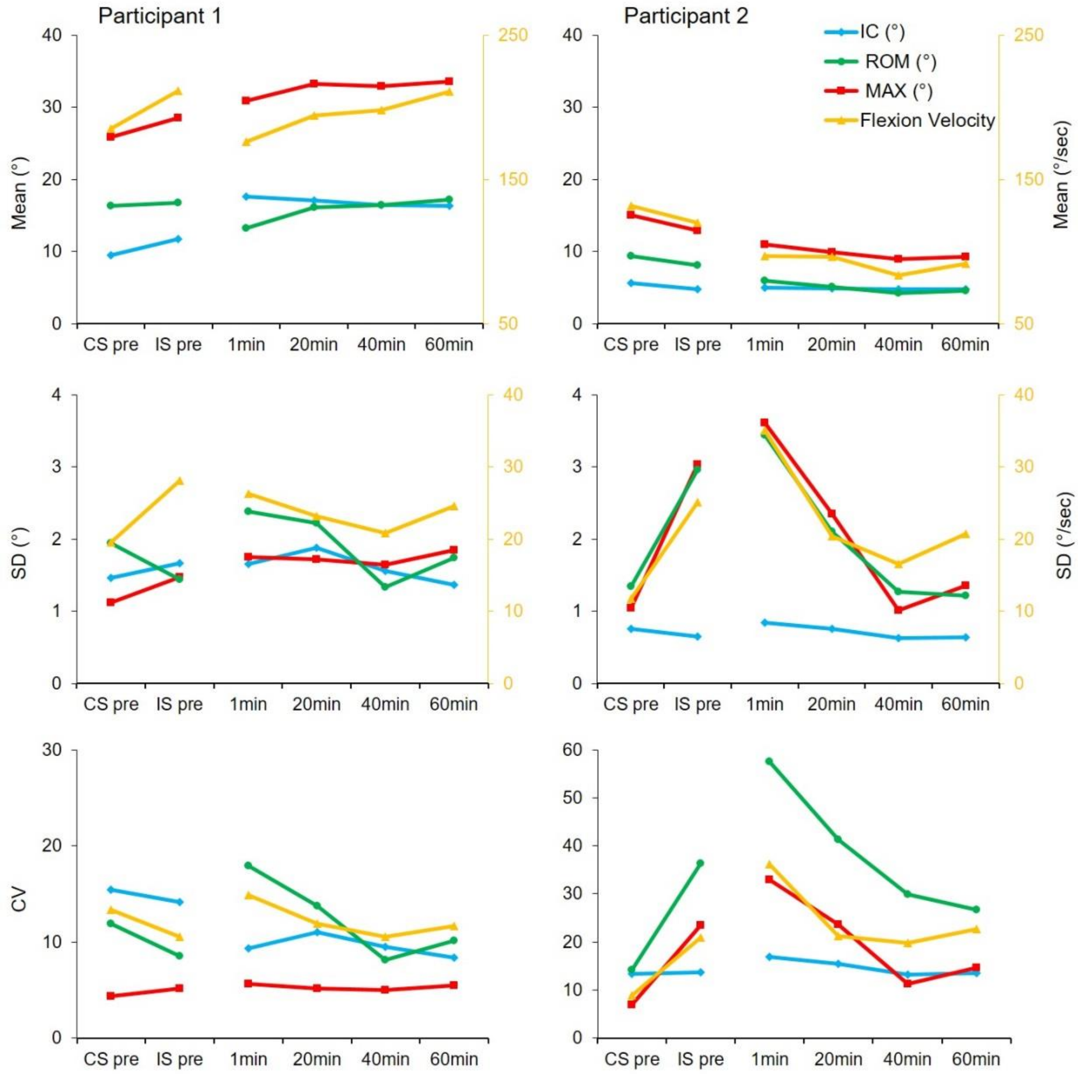

Figure D.4. Sagittal knee kinematic results for participant 1(left) and participant 2 (right). The mean (top), SD (middle) and CV (bottom) are displayed for initial contact (blue), maximum angle (red), range of motion (green) and flexion velocity (yellow).

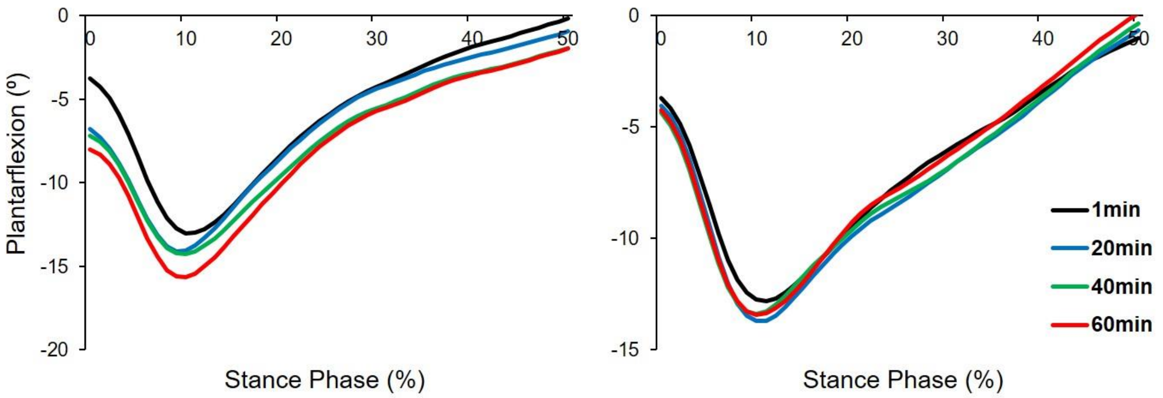

Figure D.5. Mean sagittal ankle angle during the first half of stance for participant 1 (left) and participant 2 (right). Notice the increased peak plantarflexion after 1 minute suggesting the degrees of freedom are being relaxed.

Electromyography

The only EMG differences observed during the training session occurred between minute 1 and the other time periods recorded (Figure D.6). Most notably, mean activations were increased in the gastrocnemius medialis, gastrocnemius lateralis and peroneus longus during the second half of stance phase at minute 1. Interestingly, peroneus longus activation level during the 100ms preceding initial contact were reduced during minute 1.

In comparison to the pre-test, IS had similarly increased shank muscle activations during the second half of stance phase compared to the control surface (Figure D.7). This implies that within the first 20 minutes walking on IS muscle activations followed a more similar pattern to the control. It also suggests regular walking on IS would have similar effects on ankle muscle strength to walking on the regular control surface.

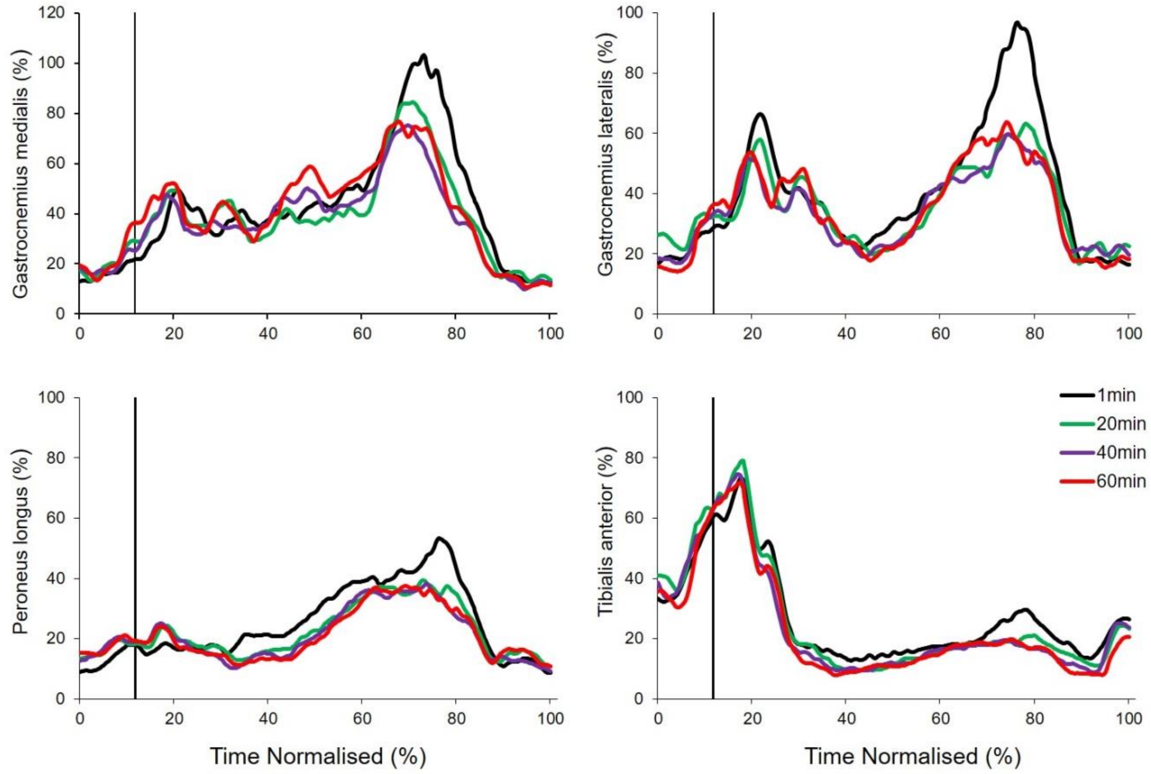

Figure D.6. Mean normalised shank EMG activations at different time points during the training session on IS for participant 2. Time is normalised between the 100ms before initial ground contact and toe-off, average initial contact illustrated by the black line. Notice small differences between 1 minute (black line) and the later time points.

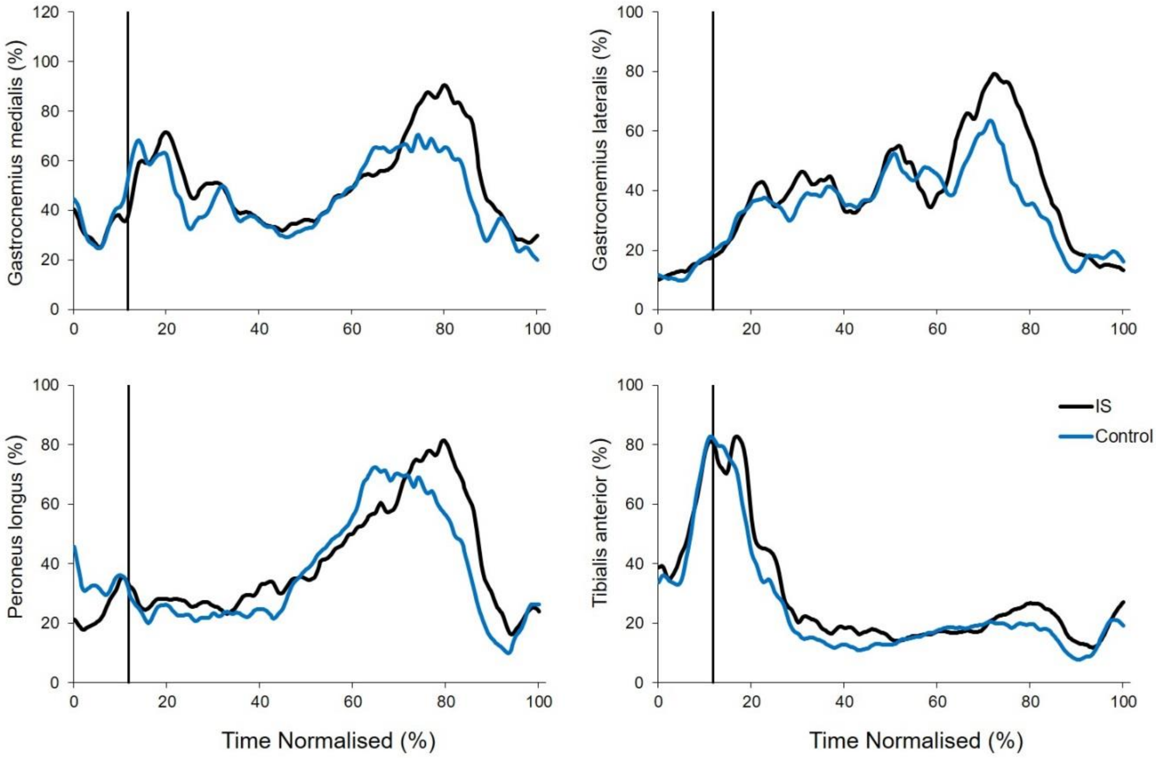

Figure D.7. Mean normalised shank EMG activations during the pre-test for IS (black) and the control (blue) for participant 2. Time is normalised between the 100ms before initial ground contact and toe-off, average initial contact illustrated by the black line.

Conclusion:

The initially increased knee joint variability and shank muscle activations reduced during the hour training session to a similar level as the regular control surface, partially confirming the hypothesis. Thus, some reasoning for the absence of balance and ankle strength gains from the 6-week IS training intervention is provided. These acute adaptations show the neuro-muscular system quickly learns how to enhance stability on IS. As only 2 participants were tested (1 for EMG) the results may not be applicable to the larger training group. However, this pilot highlights the importance of doing individual, rather than group analyses that may mask different strategies to increase stability, such as the mean knee kinematics.