The effects of foot structure, footwear and technique on knee joint loads in over ground running

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The effects of forefoot structure, footwear and technique on knee joint loads in over ground running

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Abstract
Research is yet to consider the influence of forefoot structure on forefoot pronation, its interaction with footwear and the ramifications for knee joint loading when performing endurance running (ER). This thesis investigated the relationships between forefoot structure, forefoot pronation, footwear choice, running technique and knee joint loading.

Chapter four examined the measurement error of ER kinematics and kinetics within a test session, between sessions on the same day and between two days. Absolute measurement error for all kinematic and kinetic comparisons were ≤7.62° and ≤0.59 $Nm \cdot Kg^{-1}$ respectively. Results were used to assess habituation to novel footwear conditions and calculate sample size in subsequent studies.

Using data from chapter four, chapter five investigated time to habituate in novel running conditions (barefoot, minimal and maximally-cushioned footwear) in a sample of recreational runners. Twenty-one minutes was sufficient to establish consistent hip, knee and ankle sagittal plane kinematics, where variability was less than or equal to previously established within-session data.

Post habituation, chapter six investigated associations between foot structure, forefoot pronation and peak-knee adduction moment, and the effect of running condition on forefoot pronation. Hallux angle and phalange width accounted for 35% of variance in forefoot pronation ($P = 0.029$). Results also showed forefoot pronation was significantly associated ($P < 0.05$) with peak-knee adduction moment (r = -0.57, r = -0.77, r = -0.61, for barefoot, minimal and structured-cushioned shoes, respectively). A medial translation in the centre of pressure was not associated with increased forefoot pronation. Footwear also influenced forefoot pronation. Minimal footwear had greater forefoot pronation compared to barefoot ($P = 0.042$) and the structured-cushioned condition ($P = 0.001$).

Chapter seven examined the effects of footwear on lower-limb kinematics and kinetics. Compared to barefoot and minimal shoes, a more extended knee and dorsiflexed ankle at initial contact, increased peak-knee flexion moment, and reduced the peak-dorsiflexion moment were observed in maximally-cushioned shoes. An extended lower limb follows previous work that suggests insulation of mechanoreceptors would encourage a running technique that projects the foot more anteriorly to reduce ground contacts for a given distance. These kinematic changes also suggested overstride would increase as participants change from barefoot to maximally-cushioned footwear.

Subsequently, chapter eight investigated the effects of footwear on overstride and its association with peak-knee adduction moment. Changing from maximally cushioned, to minimal shoes or barefoot, reduced overstride relative to the hip, whereas overstride relative to the knee decreased from maximally cushioned to barefoot only. Results also showed moderate to strong positive correlations between overstride and peak-knee adduction moment in all running conditions. Findings suggest footwear influences overstride, overstride was associated with peak-knee adduction moment, and reducing overstride might reduce peak-knee adduction moment, a variable associated with injury.

Following observed relationships in chapter eight, chapter nine attempted to reduce peak-knee adduction moment. Twelve recreational endurance runners performed either a 30-minute run, or 30 minutes of gait retraining. The intervention had a primary focus on reducing overstride following the reported relationship between overstride and peak-knee adduction moment. Controlling for baseline measures, there was no significant difference between overstride, trunk lean and subsequently peak-knee adduction moment. The lack of difference was attributed to the short duration and the acute nature of the coaching session. Similar investigations over a longer period of time are warranted.

Collectively, phalange width and hallux angle contributed to forefoot pronation, and forefoot pronation was associated with peak-knee adduction moment when running, a measure associated with injury. This suggests those with compromised forefoot structure, might be at risk of injury, particularly when attempting to run barefoot or in minimal shoes that lack support. As participants changed from barefoot to minimal to maximally-cushioned footwear overstride increased, with medium to strong positive correlations for overstride and peak-knee adduction moment. This suggests runners with a large overstride are likely to be exposed to increased peak-knee adduction moment and potentially injury, and reduced overstride might present a means to reduce injury.

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Work arising from this thesis

Full text publications

Wilkinson, M., **Stoneham, R.,** Saxby, L. (2018) Feet and Footwear: Applying Biological Design and Mismatch Theory to Running Injuries. International Journal of Sports and Exercise Medicine.

Conference proceedings

Stoneham, R., Barry, G., Wilkinson, M. The effect of footwear on stride length and sagittal-knee-flexion moments in overground running in a recreational-endurance-running population. World Congress of Biomechanics, Dublin Ireland, 8-12th July 2018.

Stoneham, R., Barry, G., Wilkinson, M. Relationships between foot structure, function and peak-knee adduction moments in different footwear conditions during overground running. 16th Staffordshire conference on clinical biomechanics, Staffordshire University, 27th April, 2018.

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In preparation

Stoneham, R., Barry, G., Wilkinson , M. Time to habituation when running barefoot, in a minimal and maximally-cushioned shoe conditions for 30 minutes. *Projected: Footwear Science*.

Stoneham, R., Barry, G., Wilkinson, M. The effects of footwear on overstride and its relationship with peak-knee adduction moment. Projected: *European Journal of Sport Science.*

Declaration

I declare that the work contained in this thesis has not been submitted for any other award and that it is all my own work. I also confirm that this work fully acknowledges opinions, ideas and contributions from the work of others.

Any ethical clearance for the research presented in this thesis has been approved. Approval has been sought and granted by the Faculty of Health and Life Sciences Ethics Committee at Northumbria University.

I declare that the Word Count of this Thesis is **55,254** words.

Name: Richard C Stoneham

Signature:

Date:

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List of abbreviations

BF	Barefoot
COP	Centre of pressure
COM	Centre of mass
CCRS	Conventional cushioned running shoe/s
t_c	Contact time
ER	Endurance running
FFS	Forefoot strike
IC	Initial contact
LR	Loading rate
MS	Minimal shoe
MCS	Maximally-cushioned shoes
MAX	Maximum
MIN	Minimum
MFS	Mid-foot strike
OSH	Overstride relative to the hip
OSK	Overstride relative to the knee
RFS	Rear foot strike
RCT	Randomised control trial
ROM	Range of motion
STJ	Subtalar joint
TFJ	Tibiofemoral joint
TTJ	Transverse-tarsal joints
TO	Toe off
VPP	Virtual pivot point

1.0 Introduction

1.1 Endurance running

Humans primarily locomote by walking and running, with the presence of a flight phase distinguishing the latter from the former (Bramble & Lieberman, 2004; Novacheck, 1998). Sprinting and endurance running (ER) represent two forms of running, with ER characterised by slower speeds, and larger ground-contact times (Bramble & Lieberman, 2004; Dugan & Bhat, 2005; Novacheck, 1998). The focus of this thesis will be the biomechanics of ER.

Endurance running is an enormously popular form of exercise in western culture with over 54 million people making the choice to run recreationally each year in North America (Running USA, 2014). It has been suggested that such popularity has an evolutionary basis, with archaeological and anthropological evidence suggesting ER shaped the structure of *Homo sapiens* and offered biological advantages for persistent hunting success (Bramble & Lieberman, 2004; Perl, Daoud, & Lieberman, 2012; Rolian, Lieberman, Hamill, Scott, & Werbel, 2009).

1.2 Evolutionary adaptations

The lower limbs present several evolutionary adaptations that suggest humans have evolved to be ER specialists (Bramble & Lieberman, 2004). Examples are increased articular joint surfaces, the anatomical structure of the gluteus maximus, increased leg length, lower-limb mass redistribution and tendon length (Bramble, & Cutright-Smith, 2006; Bramble & Lieberman, 2004; Jungers, 1988; Lieberman, Raichlen, Pontzer, Myers & Steudel, 1985; Thorpe, Crompton, Guenther, Ker, & McNeill Alexander, 1999). When examining the foot, it can be argued this structure is especially adapted for ER following dramatic structural changes in response to the demands of bipedal locomotion.

The architecture of the human foot is one inherited from primates with the original primary purpose of climbing trees, to one that now acts as the singular interface with the ground during bipedal locomotion (Harcourt-Smith, O'Higgins, & Aiello, 2002; Morton, 1935). The modern human foot serves a range of important duties that facilitate safe locomotion, with primary

functions being to adapt to terrain, support body weight, absorb and return energy, as well control the progression of bodyweight during stance (Chou et al., 2009; Dugan & Bhat, 2005; Ker, Bennett, Bibby, Kester, & Alexander, 1987; Morton, 1935). Despite strong arguments that the human structure has evolved through natural selection to be an ER specialist, injury rates are high (van Gent et al., 2007).

1.3 Injury rates

Injury rates in running are reported anywhere between 20 and 79% (van Gent et al., 2007). The knee joint is often reported as the most common site of injury (Taunton et al., 2003; van Gent et al., 2007), with research reporting, patella tendinopathy, patellofemoral pain and medial tibial stress syndrome as the most common complaints (Lopes, Hespanhol Jr, Yeung, & Costa, 2012; Taunton et al., 2002).

Several explanations have been put forth to explain such high injury rates, such as age, gender, training load and biomechanical loading of the lower limbs (Lopes et al., 2012; Satterthwaite, Norton, Larmer, & Robinson, 1999; Taunton et al., 2002). Whilst some previous research has investigated foot structure measures such as arch index, foot posture (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014; Miller, Whitcome, Lieberman, Norton, & Dyer, 2014; Neal, Griffiths et al. 2014) and others measures of foot motion such as rear foot eversion (Messier & Pittala, 1988), the relationship between forefoot structure and forefoot kinematics has not been investigated in ER. Specifically, forefoot pronation, defined as the inward rotation of the forefoot about its longitudinal axis (figure 1.1) is a measure theoretically linked to forefoot structure (hallux alignment) (Morton, !935). How forefoot struture might relate to forefoot pronation and how this might relate to injury have not been fully explored in ER. Despite a longstanding interest in the effect of shoe design, there has also been little attention on how footwear choice might influence the forefoot structure and forefoot pronation relationship, and subsequently the loading of the kinetic chain above the foot. Morton (1935) and Chou et al. (2009) both present compelling evidence that forefoot structure influences static balance and the control of the centre of pressure (COP) in static tasks. However, research investigating the relationship between forefoot structure and forefoot pronation in dynamic

tasks such as ER, and how this might relate to lower-limb joint loading, in particular the knee, remains to be explored.

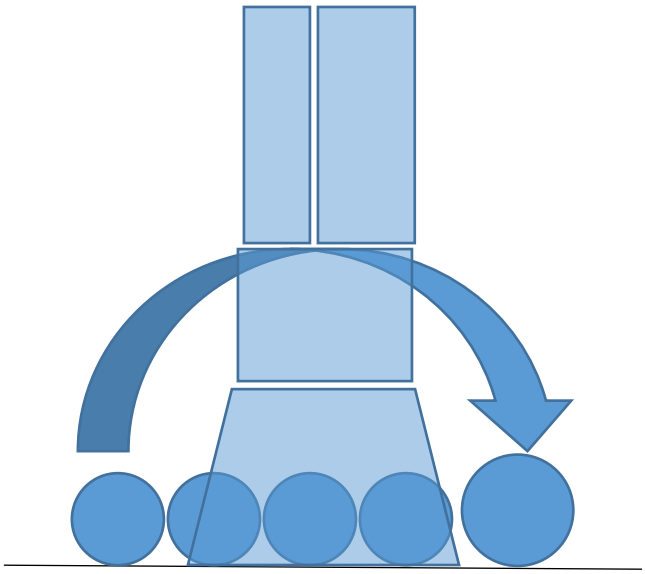

Figure 1. 1 to illustrate the term forefoot pronation in the frontal view of the right foot. Circles indicate toes and the arrow indicates the rotation implied by the term forefoot pronation. This, a lift of the lateral border and clockwise rotation as indicated by the arrow about the long axis of the foot.

1.4 Foot structure

Foot structure varies between populations depending on footwear choice, or of lack of, with those who are habitually shod characterised by a narrower foot and increased plantar pressures underfoot compared to barefoot populations (D'AoUt, Pataky, De Clercq, & Aerts, 2009; Shu et al., 2015) (figure 1.2). A clear differentiating factor between populations is the alignment of the hallux. Shu et al. (2015) reported a significantly more adducted hallux angle for shod participants. Hallux alignment is important as it has been suggested that hallux alignment has evolved through natural selection to control the progression of body weight in locomotion (Morton, 1935). This suggestion was supported in practice by Chou et al. (2009) who reported that when the hallux was restrained, the ability to control the COP was compromised. Moreover, a narrow foot width, and subsequently reduced medio-lateral functional axis might also compromise the control of foot motion during stance. This suggestion is supported by Hoogvliet, van Duyl, de Bakker, Mulder, and Stam (1997) who reported that as functional-

foot breath was reduced, larger compensatory frontal plane kinematic motion occurred. Applying these findings to ER, a compromised foot structure might compromise the control of the ground-reaction force (GRF) when running, and therefore the loading of the joints more proximal to the foot.

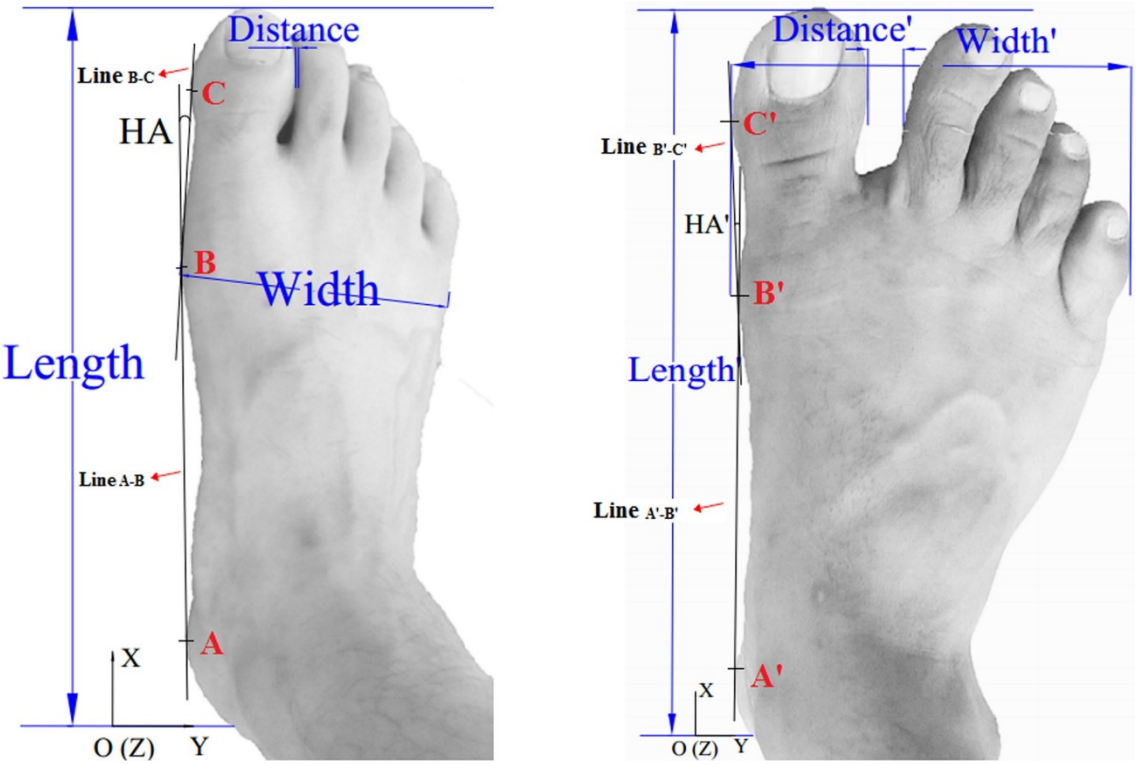

Figure 1. 2 Dorsal view of foot structure data in a sample of Chinese shod runners (left) and a sample of habitually unshod Indians (right) (Shu et al., 2015).

Modern footwear design has the potential to compromise several aspects of foot structure (Hoffmann, 1905). For example, a narrow toe box that compresses the toes towards a central point, and a toe spring that raises the toes from the ground might theoretically compromise the ability of the Hallux to oppose excessive forefoot pronation and therefore compromise the control of the GRF during stance. How forefoot structure and footwear choice are associated with forefoot pronation, and how forefoot pronation might relate to the loading of lower-limbs are primary aims of this thesis.

1.5 Footwear

Modern running footwear is often advertised to reduce a runner's likelihood of injury. However, a systematic review by Richards, Magin, and Callister (2009) concluded that the prescription of this type of footwear to reduce injury is not evidence based. Furthermore, there is an extensive body of knowledge that reports that modern running footwear has the potential to increase loading of lower-limb joints compared to barefoot (Bonacci, Vicenzino, Spratford, & Collins, 2014; Kerrigan et al., 2009). This combined with the argument that humans have evolved to be ER specialists has seen a movement in running research to investigate barefoot running and subsequently barefoot-inspired footwear termed 'minimal' shoes.

Barefoot running has been suggested to offer several biomechanical advantages compared to a conventional cushioned running shoe (CCRS) for reducing the likelihood of injury such as reduced stride length, increased stride frequency, reduced resultant knee joint loads, improved proprioceptive feedback and a trend for runners to a adopt a non-rear foot strike (RFS) strategy (Daoud et al., 2012; Dudley, Pamukoff, Lynn, Kersey, & Noffal, 2017; Kerrigan et al., 2009; Robbins, Waked, Gouw, & McClaran, 1994). These barefoot induced changes are particularly important following that a 10% reduction in stride length has been associated with a 3-6% reduction in the likelihood of stress fractures (Edwards, Taylor, Rudolphi, Gillette, & Derrick, 2009). Transitioning to barefoot running has been shown to reduce peak-knee adduction moment, a measure prospective work has identified to be associated with injury rates (Dudley et al., 2017; Kerrigan et al., 2009). The change to a non-RFS footfall strategy has been suggested to be of importance for injury potential. Daoud et al. (2012) reported a RFS doubled the occurrence of repetitive stress injuries compared to other foot strike strategies. However, it is important to note that the foot might be at risk when running barefoot in a modern environment, and as such the minimal shoe movement was born.

A consensus on minimal shoe design was provided by Esculier, Dubois, Dionne, Leblond, and Roy (2015) who defined minimal footwear as a shoe that provides minimal interference with foot movement as a result of its high flexibility, low heel to toe drop, weight and stack height, and the absence of motion control and stability devices. However, while this theoretically

suggests such footwear would elicit similar biomechanics to barefoot running, findings have been equivocal with research suggesting minimal footwear is similar to barefoot biomechanics (Squadrone & Gallozzi, 2009) and others suggesting minimal shoe biomechanics are not the same (Bonacci et al., 2013). Potential explanations for such inconsistencies in findings could be underpinned by differences in methods. For example, Squadrone and Gallozzi (2009) recruited habitual barefoot runners, whereas Bonacci et al. (2013) recruited runners who were new to barefoot running and did not assess if data recorded were representative of stable biomechanics. Such inconsistencies in findings between footwear conditions need to be addressed to help establish a consensus as to the benefits or potential risks of minimal shoes for endurance runners. To assist current and future research, this thesis will assess time to habituation in a variety of different footwear conditions.

In contrast to the concept of barefoot and minimal shoe design, maximally-cushioned shoes are a type of shoe design recently introduced to the running-shoe market. This footwear is characterised by extreme cushioning in the midsole with manufacturers reporting such a design to benefit comfort and impact characteristics (Sinclair, Richards, Selfe, Fau-Goodwin, & Shore, 2016). Research investigating this type of footwear is in its infancy with the effects of such a design on ER biomechanics largely unknown and worthy of further investigation. The effects of maximally-cushioned shoes on ER biomechanics, how this compares to other types of running conditions and how these changes relate to factors associated with injury are yet to be explored. This line of investigation is essential for runners, practitioners and researchers alike.

1.6 Running technique

There is no one correct way to run, but rather a series of optimal locomotive solutions based on internal and external factors. For example, terrain, speed and the feedback provided from the foot-shoe-terrain interaction are examples of internal and external factors that inform locomotion solutions (Gruber, Silvernail, Brueggemann, Rohr, & Hamill, 2012; Nilsson & Thorstensson, 1989; Robbins et al., 1994). Previous work has shown that footwear has the potential to alter these inputs resulting in varying locomotion solutions that are specific to

footwear design. For example, when running barefoot or in minimal shoes there is a trend to reduce stride length (Kerrigan et al., 2009; Squadrone, Rodano, Hamill, & Preatoni, 2015). Changes in stride length as a function of footwear are important following that research reports a 10% reduction in stride length reduced the likelihood of stress fractures (Edwards et al., 2009), reduced sagittal and frontal peak knee moments (Firminger & Edwards, 2016) and reduced energy absorbed at the hip, knee and ankle (Schubert, Kempf, & Heiderscheit, 2014). However, there are a variety of lower limb configurations that can produce an identical stride length (figure 1.3), this warrants a more in depth analysis of 'overstride', a measure used to quantify the infinite geometric solutions possible for any given stride length.

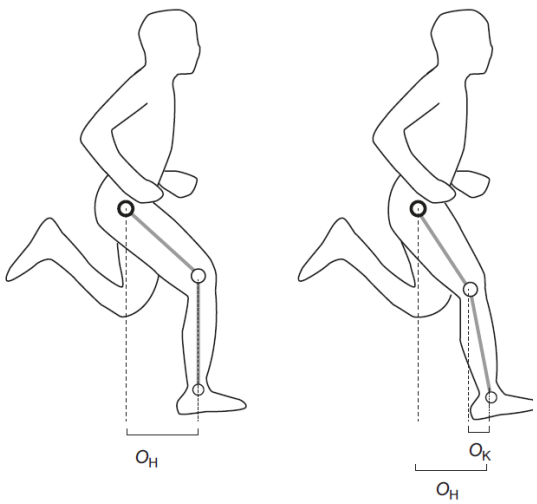

Figure 1. 3 Image of two endurance runners reporting an identical stride length but with differing overstride configurations. Left: overstride hip only. Right: a combination of overstride hip and knee (Lieberman, Warrener, Wang, & Castillo, 2015).

Lieberman, Warrener, et al. (2015) quantify overstride as the anterior projection of the ankle joint relative to the hip and knee at initial contact. These measures are of interest following recent work that reported an increase in overstride was associated with an increase in the vertical component of the impact peak and braking impulse (Heiderscheit, Chumanov, Michalski, Wille, & Ryan, 2011; Lieberman, Warrener, et al., 2015). An increase in the vertical component of the impact peak is of particular interest following the GRF is used to calculate external joint moments that act on the human body when performing ER. An external

joint moment is the turning moment created by the GRF that acts about respective joint centres (Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2013). An internal joint moment represents the net turning moments created by the respective internal structures such as muscles and ligaments that act about the same joint centres in equal magnitude but in the opposing direction (Robertson et al., 2013). Following recent reports that the external peak-knee adduction moment was a prospective characteristic of runners who are at risk of injury (Dudley et al., 2017), clinical work showing increased external peak-knee adduction moment is indicative of knee joint health (Sharma et al., 1998) and the association between GRF components (a measure used to calculate external joint loads) and overstride (Lieberman, Warrener, et al., 2015), this thesis will give specific attention to the external peak-knee adduction moment and its potential association with overstride. Research is yet to investigate a relationship between overstride and the loading of the knee joint.

In addition, the effects of different types of footwear on measures of overstride are yet to be investigated. These types of investigations are important following the theoretical association between increased overstride and knee joint loads, and previous work that suggests that running barefoot or in minimal shoes can reduce stride length. Collectively, investigations into the association between overstride and knee joint loads and how overstride might change as a function of footwear choice is yet to be explored, but is important given that runners with increased peak-knee adduction moment are at greater risk of injury.

1.7 Intervention

Unlike anatomical structures, running technique can be manipulated within a short time period. For example, participants in Heiderscheit et al. (2011) ran at a variety of stride lengths ranging from -10 to +10% of their prefered stride length within a single test session. Stride length has received much attention with consistent evidence suggesting a reduced stride length reduces biomechanical measures associated with injury etiology (Edwards et al., 2009; Napier et al., 2015; Schubert et al., 2014). Specifically, a reduced stride length has been shown to decrease peak GRF and peak-knee adduction moment (Kerrigan et al., 2009; Schubert et al., 2014), the latter a charactistic of injured runners (Dudley et al., 2017). As

previously disscussed, overstride underpins stride length and provides a more specific means to describe lower-limb geometry. However, only Heiderscheit et al. (2011) and Lieberman, et al. (2015) have used overstride as an outcome meansure. Results suggest that as overstride increases, loading rate, the vertical component of the GRF impact peak (Lieberman, et al., 2015), and braking impulse (Heiderscheit et al., 2011) increase. Together, this work suggests that stride length, and potentially overstride can be manipulated within a single session and that a reduced overstride migth reduce lower-limb loading patterns associated with injury.

Additionally, it is proposed that trunk lean shares an inherent relationhsip with overstride. Following logic that as trunk lean increases the centre of mass (COM) is projected more anteriorly, work by Horak and Nashner (1986) suggested that to increase dynamic stability while running, and to prevent falling, the step reflex strategy would increase overstride to maintain dynamic balance. Following, an improved running posture (reduced trunk lean) might reduce the step reflex, this in turn might reduce overstride and therefore the peak-knee adduction moment, a variable when increased was characteristic of injured endurance runners and compromised joint health (Dudley et al., 2017; Sharma et al.,1998).

To date, a coach-led gait retraining intervention to reduce peak-knee adduction moment as a function of reduced overstride relative to the hip and knee, and trunk lean in a sample of recreational runners has not been undertaken. This thesis will undertake such an investigation in a sample of recreational endurance runners and if successful might provide a means to reduce the likelihood of injury for endurance runners.

1.8 Aims of the thesis

The aim of the thesis is to investigate how forefoot structure, forefoot pronation, footwear and running technique effect the external-peak-knee adduction moment following previous work that reports the knee joint as the most common site of injury, prospective work indicating an increased external peak-knee adduction moment was characteristic of injured runners, and this measure being attenuated/ augmented by running condition and running technique. To investigate how these measures effect the external-peak-knee adduction

moment and the potential relationships that exist between them, a series of experiments will be carried out.

- Chapter four: To describe the measurement error of the sample of interest in addition to the systematic error of measurement equipment, a reliability study will be undertaken to assist with the estimation of sample sizes in subsequent chapters. Within-session data will also be used to quantify stability in a subsequent chapter investigating the effects of habituation time in a variety of novel running conditions.
- Chapter 5: Based on the variability observed in chapter four and due to lack of consensus on the effects of different running shoe designs on lower-limb kinematics and kinetics, chapter five will assess the time taken for novice runners with no experience running barefoot or in either minimal or maximal-cushioned footwear to achieve consistency in measures equal to that observed in the within-session reliability data.
- Chapter 6: Following habituation in chapter five so that running characteristics are representative of habituated biomechanics, chapter six will investigate the impact of forefoot structure on forefoot pronation, how forefoot pronation might change as a function of footwear and importantly whether forefoot pronation is related to knee-joint loading given an increased load is characteristic of injured runners. This thesis will investigate such relationships and shed light on forefoot structure, footwear and forefoot pronation interactions.
- Chapter 7: Following habituation and forefoot structure investigations, 3-D lower-limb running biomechanics will be compared between conditions (barefoot, minimal and maximally-cushioned) in an attempt to help establish a consensus on the effects of footwear conditions on habituated individuals.
- Chapter 8: In addition to kinematics and kinetics of chapter seven, chapter eight will investigate how footwear impacts novel measures of running technique such as overstride and whether these measures might relate to the peak-knee adduction moment. This is yet to be investigated and of interest given changes in stride length

have been reported to change knee joint loads and knee joint loads are related to injury.

- Chapter 9: Following findings of chapter eight, chapter nine will aim to reduce knee-joint measures associated with injury by improving posture and subsequently overstride given the theoretical relationship between overstride and knee joint loads.

It is hoped that these investigations will contribute to the body of research aiming to reduce injury rates in recreational endurance runners and shed light on foot structure, forefoot pronation, footwear, and knee joint load interactions.

2.0 Literature review

2.1 Introduction

Upright bipeds locomote primarily by walking and running (Bramble & Lieberman, 2004; Mann & Hagy, 1980; Novacheck, 1998). Walking and running are biomechanically-distinct gaits characterised by the absence and presence of a flight phase respectively (Bramble & Lieberman, 2004; Dugan & Bhat, 2005). Running can be further separated into two distinct gates; sprinting and ER based on shorter ground-contact times and the absence of heel contact in the former (Kivi, Maraj, & Gervais, 2002; Mann & Hagy, 1980; Mann, Moran, & Dougherty, 1986). The focus of this literature review will be the biomechanics of ER.

Reber, Perry, and Pink (1993) estimated that 40 million North Americans participate in ER for fitness, recreation or competition. The popularity of ER is supported by a recent Sports and Fitness Industry Association (SFIA) report showing that 54 million people made the choice to participate in running in 2014 (running at least 6-plus days/yr) (Running USA, 2014). Of those 54 million classified by the SFIA as running participants, approximately 30 million ran at least once a week. This would seem to suggest that ER is a popular means of exercise in western culture.

The lack of specialist equipment needed to perform ER, the low cost of participation and accessibility could explain the high participation rates observed. Endurance running offers psychological benefits for participants including reduced anxiety, depression and anger (Bodin & Hartig, 2003) and physiological benefits, such as weight loss and increased aerobic capacity (Achten, Venables, & Jeukendrup, 2003; Jones & Carter, 2000). Moreover, high participation rates are not a modern phenomenon. The popularity of ER dates back to 776 b.c. with the origin of the Olympic games (Chalkley & Essex, 1999).

The original Olympic games comprised events such as best trumpeter, horse/chariot racing and a variety of running races (with and without light armour) (Chalkley & Essex, 1999; Hilbe, 2009). Most events of the original games are no more, but the popularity of running as an organised-competitive sport has remained. The modern games have a variety of ER races

ranging over distances from 3 km to marathon performed by men and women on track and road. It has been argued that the sustained popularity of ER has an evolutionary basis in light of archaeological and anthropological evidence that shaped the anatomy of *Homo sapiens*, and offered a survival advantage in the form of the persistence hunting strategy (Bramble & Lieberman, 2004; Liebenberg, 2006).

2.2 Evolutionary adaptations to running

Running has not always been recreational. It has been suggested that the genus *Homo* evolved to perform ER (Bramble & Lieberman, 2004; Carrier et al., 1984; Rolian et al., 2009). Both biological and archaeological records suggest that the shape and structure of *Homo sapiens* has been adapted to facilitate ER (Bramble & Lieberman, 2004; Carrier et al., 1984; Lieberman et al., 2006). Based on this evidence, Bramble and Lieberman (2004) propose that among the primate family, a distinctive set of specialist adaptations make the activity of ER unique to humans.

2.2.1 Upper body adaptations

Independent rotation of the trunk relative to the lower body is thought to be key to maintaining dynamic stability when running (Bramble & Lieberman, 2004). Because the skeleton is not affected by the GRF during the flight phase, additional mechanisms are needed to act against internal rotation of the trunk (Pontzer, Holloway, Raichlen, & Lieberman, 2009). Independent thorax rotation and arm swing provide such a mechanism (Hinrichs, 1987; Pontzer et al., 2009). Pontzer et al. (2009) argues that arm swing is a passive response to the forces exerted by the swinging of the legs. Based upon this theory, rotational moments at the pelvis are transferred superiorly to the shoulder girdle and subsequently the arms. This is thought to allow the stabilising muscles of the trunk and shoulder girdle and other connective tissues to act as ‘springs’, with the magnitude of elastic recoil proportional to the forces placed upon them (Pontzer et al., 2009). The proportional force provided by arm swing has the effect of facilitating counter-rotational moments around the vertical axis of the thorax, subsequently

cancelling out the rotational-angular momentum of the lower limbs to near zero (Bramble & Lieberman, 2004; Hinrichs, 1987).

It is the lengthened trunk vertically displacing the thorax from the pelvis that permits counter rotation. This adaptation was first seen to be partially developed in *Australopithecus*, however, a slender-elongated structure is only fully observed in *Homo erectus* (Bramble & Lieberman, 2004). Archaeological research also reports a structural difference between *Homo* and *Pan,* with *Homo* possessing a distinct independence between the pectoral girdle and the head-and-neck complex (Bramble & Lieberman, 2004; Diogo & Wood, 2011). The independence of the *Homo* shoulder girdle allows limited head yaw, a factor known to improve visual stability during running (Pontzer et al., 2009). This presents a possible advantage for persistence hunting, and therefore a reason that the adaptation would be retained by natural selection.

Another derived-structural adaptation in *Homo* is the broadening of the shoulders which provides a means to increase the turning moments around the vertical axis. Broader shoulders facilitate increased turning moments via a greater moment arm, while simultaneously allowing for a reduced forearm mass (Bramble & Lieberman, 2004). Putting this mass distribution in context, when compared to chimpanzees, human forearms have 50% less mass relative to total body mass (Bramble & Lieberman, 2004). Francis and Hoobler (1986) demonstrated the consequences of increased forearm mass when they observed that increasing the mass of the forearm increased the oxygen cost in running, but not walking. This suggests that forearm-mass reduction is a running-specific adaptation in *Homo* that would reduce the energy cost of ER, improving the success of persistence-hunting (Liebenberg, 2006).

Based on these findings, it can be argued that an elongated trunk, independence of the head-and-neck complex, broadening of the shoulders and a reduction in forearm mass all represent upper-body evolutionary adaptations that could improve ER performance and play a key role in the survival of the early Hominin.

2.3 Lower body adaptations

2.3.1 Increased articular joint surfaces

An important adaptation to consider in the evolution of *Homo* is the skeletal response to the increased forces associated with ER. Endurance running exposes the body to greater force compared to walking (Bramble & Lieberman, 2004; Keller et al., 1996). Keller et al. (1996) reports peak vertical GRF ranging from 1.1-1.5 times bodyweight (BW) in walking compared to around 2.5 BW in running. This force is subsequently transmitted up through the body to the skull (Mercer, Bates, Dufek, & Hreljac, 2003; Mercer, Devita, Derrick, & Bates, 2003). It is possible to attenuate some of this force via joint excursion and particular foot strike strategies, but other than this, the remaining force is attenuated by the bones and joints (Bramble & Lieberman, 2004; Derrick, Hamill, & Caldwell, 1998; Lieberman, 2012b). It has been suggested that expanding of joint-articular surfaces presents a mechanism to effectively share the contact force over a greater area (Bramble & Lieberman, 2004). The archaeological evidence is clear that compared to ancestors of *Pan* and *Australopithecus,* the genus *Homo* has an increased articular-joint contact area in the sacroiliac joint and most of the lower-limb joints (relative to body mass) (Bramble & Lieberman, 2004; Jungers, 1988, 1991). This evolutionary adaptation, absent in the upper limbs (Jungers, 1988), might be for walking due to the preclusion of the forelimbs in locomotion. However, it is thought to play a more meaningful role in attenuating the forces associated with ER, and theoretically allowed early *Homo* to run with reduced joint loads, and thus reduced potential for injury (Bramble & Lieberman, 2004; Jungers, 1988).

2.3.2 Gluteus Maximus

During the first half of stance the human trunk is flexed relative to the vertical. This, in combination with the increased forces associated with running, magnifies forward pitching and instability of the trunk observed at heel strike (Bramble & Lieberman, 2004; Lieberman et al., 2006; Maus, Lipfert, Gross, Rummel, & Seyfarth, 2010). Several structural features in humans have been proposed as evolutionary adaptations to promote stability during running

(Bramble & Lieberman, 2004). Most notable is the enlargement of the gluteus maximus which is one of the most distinguishing features of humans compared to other primates (Bramble & Lieberman, 2004; Lieberman et al., 2006). For example, compared to our closest relatives, apes possess neither the pelvic structure or muscle architecture, and it is argued the change is a response to the demands of bipedal locomotion (Lieberman et al., 2006). Furthermore Lieberman et al. (2006) demonstrated that the gluteus maximus is quietest during level and uphill walking, but increases in activity, and alters its timing with increases in running speed. This suggests gluteus maximus enlargement was an important evolutionary adaptation to help control the pitch of the trunk at foot strike when running.

Unfortunately, based on origins and insertions of fossil evidence alone, reconstruction of the relative size and configuration of any muscle is not feasible (Zumwalt, 2006). However, a structure of a human-like pelvis is thought to be evident by around 1.9 million years ago in *Homo erectus*, with attachment sights for the gluteus maximus that are widened and rough along the superior iliac crest (Lieberman et al., 2006). Although this pelvis anatomy is demonstrated clearly in *Homo erectus*, the argument remains as to when a human-like organisation was first developed (Haeusler, 2002; Lieberman et al., 2006). Moreover, Lieberman et al. (2006) also reports that as well as trunk stabilisation, the human gluteus maximus acts to decelerate the swing leg, and might also help control hip flexion and extend the thigh. This evidence plausibly suggests that gluteus maximus enlargement was first seen in *Homo erectus*, a time when upright locomotion required trunk stabilisation.

2.3.3 Leg length

Every step taken during ER consumes approximately 100 J per stance phase in the form of muscle action supporting the body weight (Ker et al., 1987). Following this, it is energetically favourable if a distance is covered in as few steps as possible, thus supporting the body weight fewer times (Kram & Taylor, 1990). A common trait of a cursorial-specialists which exploits this relationship is elongated-lower limbs. Humans are no exception, with their elongation of the lower limbs reducing the energetic cost of locomotion (Bramble & Lieberman, 2004; Kram & Taylor, 1990). Humans typically increase running velocity by increasing stride length, as

opposed to stride frequency (Cavanagh & Kram, 1989). Average stride length of the adult human population is approximately 2.5 meters (Mercer, Devita, et al., 2003). This stride length is thought to be a product of both the elastic properties of human tendons and the length of the lower limbs (Bramble & Lieberman, 2004). In locomotion, a long stride theoretically decreases the energy cost of ER by increasing ground-contact time (Kram & Taylor, 1990). Increased contact time (t_c) has been shown to be inversely related to the energy cost of running in a cross-species-comparative study, with the relationship explained by the reduced rate of force development during support of the body weight as t_c increases (Kram & Taylor, 1990). However, while increased stride length and t_c are theoretically linked across species, there comes a point where both begin to increase energy cost when running. This is because increased t_c as a function of increased stride length increases the amount of stored elastic energy dissipated as heat (Cardinale, Newton, & Nosaka, 2011). It follows that the ideal combination is long-lower limbs with an optimal t_c to maximise the potential of the stretch-shortening cycle. Based on this evidence, ancestors with longer limbs could have performed ER with a reduced energetic cost. Natural selection generally favours adaptations that reduce energy cost as saved energy can be used for reproduction and feeding of offspring (Lieberman, 2013).

When looking at the fossil records of human evolution, it is agreed that the femur length (a proxy of lower-limb length) was short in *A.afarensis*, (McHenry & Coffing, 2000). Conversely, there is uncertainty about the femoral length in other hominoid species due to the nature of fossil quality. That said, there is sufficient evidence to suggest that *A.Africanus* also had a short femoral length relative to humeral length (McHenry & Coffing, 2000). It is only around 1.8 million years ago in *Homo erectus* when longer femoral length is agreed to have been fully established (Bramble & Lieberman, 2004). However, it is important to note an increase in limb size, and ultimately mass, increases the metabolic demand of limb movement (Myers & Steudel, 1985).

2.3.4 Lower-limb-mass distribution

The elongation of the lower limbs and subsequent increased mass increases the cost of locomotion as a result of a greater mass-moment of inertia (Bramble & Lieberman, 2004; Hogberg, 1952). However, basic engineering principles dictate, and research has shown, that the proximal redistribution of mass represents an adaptation whereby the cost of locomotion could be reduced (McLester & Pierre, 2007; Myers & Steudel, 1985). Myers and Steudel (1985) demonstrated that moving two 1.8 kg weights fixed to the upper lower leg to the more distal position of the ankle increased the metabolic cost of ambulation by around 12%. Drawing on this research, it seems that the redistribution of distal-limb mass presented an important adaptation to overcome the metabolic cost of longer-lower limbs.

The relative mass of the distal-lower limb in fossils of hominoids are not known. However, bone structure of the foot is known, with humans presenting shorter and more cursorial-like bone structure than their ancestors (McHenry & Coffing, 2000). This evidence tentatively suggests that human-lower limbs have evolved similar to other cursorial specialists, reducing their lower-limb bone mass. This presents a means to reduce the metabolic demands of ER through a reduced mass-moment of inertia that theoretically could have enhanced persistence-hunting performance by allowing hunters to run at greater speeds for a given energetic cost, or for longer at a given running velocity (Myers & Steudel, 1985).

2.3.5 Tendon adaptations

Humans possess a series of muscle-tendon adaptations that contribute to the task of ER, examples include short muscle fascicles and elongated tendons compared to Chimpanzees (Thorpe et al., 1999). Possessing a compliant limb while running allows humans to store and release elastic energy in their tendons. Ker et al. (1987) calculated that if a 70 kg man running at 4.5 $m \cdot s^{-1}$ expends 100 J per stance phase around 35% of the energy is returned via the elastic properties of the Achilles tendon alone. This provides evidence for the argument that the adaptation of elongated tendons, in particular the Achilles tendon, reduces metabolic cost and can improve ER performance.

Currently there are no early and well-preserved *Homo* calcanei available, making it difficult to estimate a timeline for the evolution of an increased Achilles tendon length. However, the transverse groove of the anatomical insertion of the Achilles tendon on the calcaneus of a chimpanzee is similar to that in early australopithecine specimens (Bramble & Lieberman, 2004). This contrasts with the wider and taller anatomy of the *Homo sapiens* attachment site (Bramble & Lieberman, 2004). This led Bramble and Lieberman (2004) to hypothesise that a developed Achilles tendon was absent in *Australopithecus*, most likely originating in *Homo* as a function of selective pressure for improved ER.

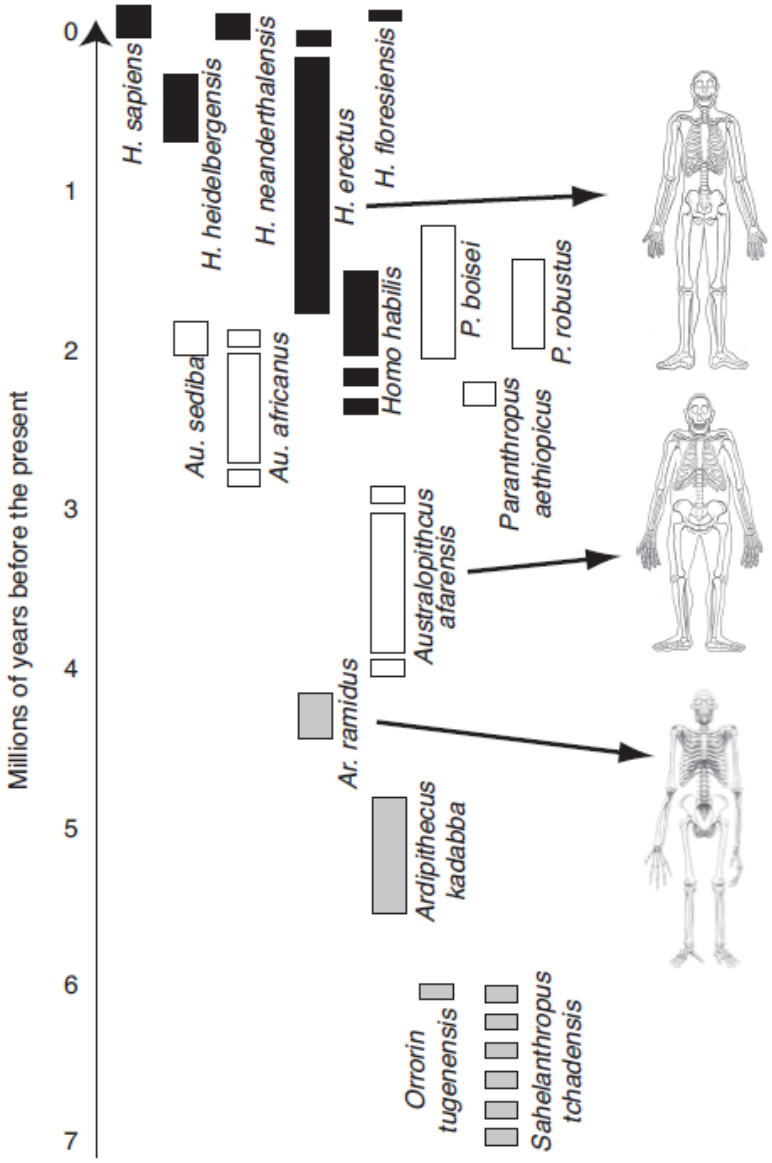

Figure 2. 1 Dates of species of hominins with respective skeletons from each major stage of human evolution. *Homo* genus species are black, those from the genus Australopithecus are white and early hominin species and genera are grey (Lieberman, 2011).

Table 2. 1: General structural adaptations of the human skeleton and their functional purpose in bipedalism.

Author (Year)	Anatomical location	Feature	Functional purpose	Walking or running
Diogo and Wood (2011)	Upper body	Independence between the pectoral girdle and the head and neck complex	Counter rotation of the head and neck complex for visual stability	Running
Bramble and Lieberman (2004)		Narrow and lengthened trunk	Counter rotations of the thorax and pelvis	Running
Bramble and Lieberman (2004)		Broadening of the shoulders	Counter rotation of the trunk and pelvis	Running
Francis and Hoobler (1986) Bramble and Lieberman (2004)		Reduced forearm mass	Improved economy	Running
Jungers (1988, 1991)	Lower body	Expanding of joint articular surfaces	Reduced skeletal strain	Running

Lieberman et al. (2006)	Expansion of the gluteus maximus	Trunk stabilisation	Running
Bramble and Lieberman (2004)	Elongation of the lower limbs	Improved economy	Running and Walking
Myers and Steudel (1985)	Lower-limb-mass redistribution	Improved economy	Running and Walking
Thorpe et al. (1999)	Elongated tendon length	Energy storage Improved economy	Running
Liebenberg (2006) Carrier et al. (1984)	Reduced body hair and increased eccrine sweat glands	Thermoregulation	Running
Carrier et al. (1984)	Mouth breathing	Thermoregulation	Running

Table 2. 2: Structural adaptations of the human foot and their functional purpose in bipedalism

Author (Year)	Feature	Functional purpose	Walking or running

Dugan and Bhat (2005)	Structural architecture of the transverse-tarsal joints	Forefoot adaptability to running surface	Walking and running
Chou et al. (2009)		Shock absorption	Walking and Running
Duerinck, Hagman, Jonkers, Van Roy, and Vaes (2014)		Increased base of support	Walking and running
		Improved directional stability	Walking and running
		Rigidity during push-off	Walking and running
Ker et al. (1987)	Medial-longitudinal arch of the foot	Shock absorption	Walking and Running
Stearne et al. (2016)		Restitution of potential elastic energy	Walking and running
		Improved economy	Walking and running
Hicks (1954)	Windlass mechanism	Rigidity during push-off	Walking and running
Noe, Voto, Hoffmann, Askew, and Gradisar (1993)	Enlarged Calcaneus	Reduced skeletal strain	Walking and running
D'AoUt et al. (2009)	Hind-foot to forefoot ratio	Improved plantar-pressure distribution	Walking and running
Chou et al. (2009)		Improved directional stability	Walking and running
Rolian et al. (2009)	Phalange length	Reduced kinetic demands at the metatarsal phalangeal joint	Running
		Improved economy	Running

Chou et al. (2009)	Abducted hallux	Directional stability	Walking and running
Hicks (1954)		Facilitate the Windlass mechanism	Walking and running
Plank (1995)		Excessive pronation prevention	Walking and running
Mei, Fernandez, Fu, Feng, and Gu (2015)		Attenuate peak plantar pressure	Running
D'AoUt et al. (2009)		Increased medio-lateral axis	Walking and running
Hoogvliet et al. (1997)			

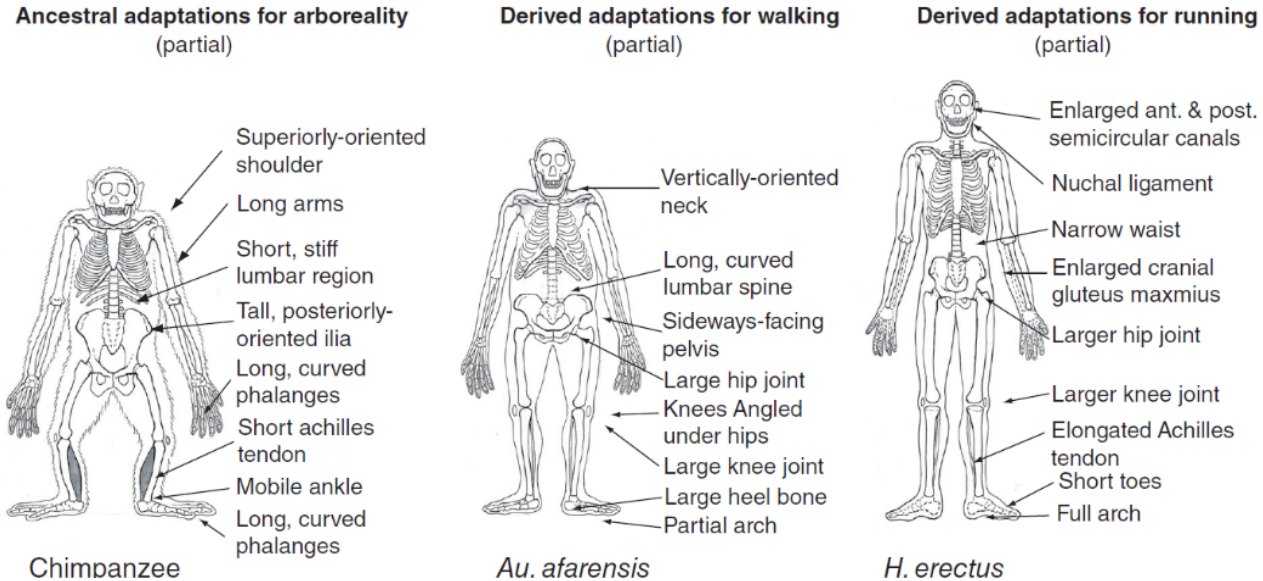

Figure 2. 2 Comparison of major locomotor adaptations in Chimpanzees, Australopiths and humans. Anatomical features highlighted are structural adaptations for arboreal locomotion in Chimpanzees, derived features for walking in Australopithecus, and derived features for running in homo (Lieberman, 2011).

2.4 Running injuries

Whilst Lieberman's hypothesis that ER drove aforementioned anatomical and structural adaptations, Lieberman and colleges rely on assumptions about frequency of endurance running and the quality of the running environment; however, it is difficult to be certain of such assumptions. In light of this critique, Pickering and Bunn (2007) have challenged the certainty of the ER hypothesis suggesting it is unclear that ER was the only driving mechanism behind aforementioned adaptations in absence of knowledge of running frequency and environment. Rather, it is suggested that whilst the ER hypothesis is logical, its certainty remains unclear. Therefore, if adaptations were not solely driven by endurance running, this might in part explain why modern humans are injured so often when performing ER.

Beyond the certainty of the ER hypothesis, the current human form (*Homo erectus*) appears to be structurally adapted to the demands of endurance running yet remains susceptible to running-related injuries. Published injury rates range anywhere from 20-79% (Daoud et al., 2012; Taunton et al., 2003; van Gent et al., 2007) or alternatively 2.5 to 62 injuries per 1000 hours of exposure to running (Buist et al., 2010; Jakobsen, Kroner, Schmidt, & Kjeldsen, 1994; Lun, Meeuwisse, Stergiou, & Stefanyshyn, 2004; Lysholm & Wiklander, 1987). Whilst it could be suggested that that a sedentary lifestyle potentially facilitates a lack of stimulation in structures responsible for dealing with the high forces associated with exercise, and therefore a lack of conditioning might contribute to injury rates, it is important to note high injury rates are also observed in highly active running groups that run upwards of 40 miles per week (Walter, Hart, McIntosh, & Sutton, 1989). This suggests that although lifestyle might contribute to injury this is not the only contributing factor. To date, there has been little attention to the effects of forefoot structure on injury mechanism.

A variety of injury sites are associated with ER (Lopes et al., 2012; Taunton et al., 2003). The most common are the knee (Fredericson & Misra, 2007; Jacobs & Berson, 1986; Taunton et al., 2003) and lower leg (Jakobsen et al., 1994; Lopes et al., 2012). Of injuries seen at the knee and lower leg the most prevalent are patellofemoral pain and medial tibial stress syndrome. The etiology of injuries is debated.

There is some evidence to suggest that age (Satterthwaite et al., 1999; Taunton et al., 2003), gender (Satterthwaite et al., 1999) and anatomy (Taunton et al., 2002) can predispose an individual to injury. However, in a systematic review, van Gent et al. (2007) reported that the evidence for these variables is limited, with the influence of 'training factors' being more widely accepted as fundamental to ER injury (Hreljac, 2004; Lopes et al., 2012; van Gent et al., 2007). van Gent et al. (2007) identified strong evidence that male runners were predisposed to running injuries if they ran more than 64 $km \cdot week^{-1}$.

Based on the suggested link between overtraining and increased injury rates, it seems logical that an increase in speed, and (secondary to increased GRF) increased demands on the musculoskeletal system would also be associated with increased injury rates (Edwards, Taylor, Rudolphi, Gillette, & Derrick, 2010; Lopes et al., 2012; van Gent et al., 2007). Using computer modelling, Edwards et al. (2010) reported that a reduction of 1 $m \cdot s^{-1}$ could reduce loading of bone material, and thus the likelihood of stress fractures by 10%. Another training factor reported to contribute towards running-related injuries was a history of previous injuries (Marti, Vader, Minder, & Abelin, 1988; Saragiotto et al., 2014; van Gent et al., 2007). A possible explanation for this link is provided by Taunton et al. (2003) who reports that of those who were injured and went on to re-injure themselves, 42% were not fully rehabilitated from a previous injury to the same anatomical location. This suggests that the lack of patience to fulfil a rehabilitation programme might have led to the subsequent injuries. Notably, Daoud et al. (2012) demonstrated that injury rates could also be influenced by running technique, specifically by footfall pattern. In a three-year-prospective study on collegiate track runners, those with a RFS had twice the likelihood of repetitive-strain injury as those with a forefoot strike (FFS). Daoud et al. (2012) suggested that the observed increase in injury rates for RFS runners could be attributed to the impact peak and increased rates of loading seen with a RFS (Barr & Barbe, 2002; Lieberman, 2012b). Collectively, this evidence for injury causality is in line with the conclusions of Lopes et al. (2012) who suggests that the majority of injuries in long distance running are related to overloading of the musculoskeletal structures of runners' lower limbs.

Large and cyclical loading of muscles and bones presents a mechanism for overuse injuries (Barr & Barbe, 2002; Daoud et al., 2012; Edwards et al., 2010; Hreljac, Marshall, & Hume, 2000; Lieberman, 2012b; Lopes et al., 2012; Schaffler, Radin, & Burr, 1989). This type of loading causes a reduction in the mechanical properties of musculoskeletal structures and in turn causes damage in the bone such as micro cracks (Barr & Barbe, 2002; Burr et al., 1998; Lieberman, 2012b; Schaffler et al., 1989). However, if remodelling rates exceed damage caused by overloading, potential for injury is avoided (Hreljac, 2004; Lopes et al., 2012). If overload exceeds remodelling rates, increased mechanical hysteresis and micro cracks will result (Craig, 2008; Edwards et al., 2010; Hreljac, 2004). These micro cracks can propagate into macro cracks ('shin splints') and cause the musculoskeletal structures to become brittle (Burr et al., 1998; Lieberman, 2012b). Repeated high force in combination with increased hysteresis presents a possible injury mechanism for both bones and other soft tissues such as tendons and ligaments (Lieberman, 2012b). This is because increased mechanical hysteresis will convert a small amount of energy to friction and heat, but a greater amount into structural damage (Burr et al., 1998; Lieberman, 2012b).

Increased rates of loading are also associated with increased mechanical hysteresis and the propensity for injury (Ferber, Davis, Hamill, Pollard, & McKeown, 2002; Hreljac et al., 2000; Lieberman, 2012b). Ferber, Davis, Hamill, et al. (2002) investigated the kinetic differences between ten females with a history of stress fractures and ten gender matched controls. Ferber, Davis, Hamill, et al. (2002) reported that those with a history of stress injuries exhibited a 32% increase in their initial loading rate (LR) (158.61 BW^{-1} vs 108.89 BW^{-1}; $P = 0.03$) and a 34% increase in the average LR (117.93 BW^{-1} vs 77.52 BW^{-1}; $P = 0.03$). In a similar study, Hreljac et al. (2000) also investigated kinetic difference between runners with a history of repetitive strain injuries and controls who were injury free their entire running career. Their results support that of Ferber, Davis, Hamill, et al. (2002), with an increased vertical LR in those with a history of overuse injuries (Injured: 93.1 ± 23.8; Non-injured:76.6 ± 19.5). Furthermore, Zadpoor and Nikooyan (2011) conducted a meta-analysis investigating differences between magnitude and LR of the vertical GRF in runners with and without a history of lower-limb-

stress fractures. They concluded LR was significantly different between groups, with the previously injured group demonstrating significantly higher average and instantaneous loading rates. This suggests those who employ a running technique that produces greater loading rates are at an increased risk of injury to musculoskeletal tissue than runners with lower rates of loading (Hreljac et al., 2000; Lieberman, 2012b).

To date, previous work has largely focused on global measures derived from GRF data such as peak, average and instantaneous LR and their association with injury rates (Phan et al., 2017; van der Worp, Vrielink, & Bredeweg, 2016). However, these measures are not specific to a joint that is commonly injured. Currently there is a paucity of prospective work investigating overground 3-D lower-limb biomechanics in endurance running and how these measures might share a relationship with injury rates. Dudley et al. (2017) provided one such study, prospectively recording the lower-limb ER kinematics and kinetics of 32 overground endurance runners who were free from injury six months prior to the start of a 14-week long season. Dudley et al. (2017) reported of the 31 runners who completed the study 12 of 13 injuries occurred on the dominant limb and increased peak-knee adduction moment was associated with injured runners (Injured: 1.32; Uninjured: 0.93 $Nm \cdot kg^{-1}$). Given that the most common site of injury is the knee (Taunton et al., 2002; van Gent et al., 2007), it was logical that a measure reflecting loading at the knee was associated with injury rates. Furthermore, it has been reported by Willy, Manal, Witvrouw, and Davis (2012) that increased peak-knee adduction moment differentiates males who suffer from patellofemoral pain from healthy controls. This is particularly important given patellofemoral pain is one of the most common types of injury reported at the knee (Taunton et al., 2002). In other work investigating walking gait, Sharma et al. (1998) retrospectively reported patients with advanced medial-tibiofemoral osteoarthritis reported a significant correlation between peak-knee adduction moment in the left and right knee and K-L grade (a measure of osteoarthritis severity) (r = 0.68 and 0.6, respectively) and joint-space width (a surrogate measure of joint health) (r = 0.45 and 0.4, for left and right knee respectively). The relationship between peak-knee adduction moment and K-L grade persisted after controlling for age, sex and severity of pain (r = 0.71 and 0.61, in

left and right knees respectively). For every one-unit increase in peak-knee adduction moment, there was a 0.63mm decrease in joint space width. This highlights the effects of increased peak-knee adduction moment on joint health and injury, and underlines the importance of investigating peak-knee adduction moment.

Additionally, if the knee has not evolved to deal with increased loads in the frontal and transverse plane, then an increased knee-adduction moment might induce secondary adaptive running mechanics that place excessive load on other lower-limb joints and this might further increase a runner's propensity for injury. Collectively, the knee is the most common site of injury, is not well adapted to deal with increased loads in the frontal plane and increased adduction loads are associated with injured runners and clinical populations. Future work should investigate the loading of the knee joint, with a particular focus on peak-knee adduction moment in light of previous work showing this loading pattern to be associated with injured endurance runners and knee joint health.

If evolution has adapted the human structure to perform ER with minimum energy expenditure and risk of injury, it is possible that the high injury rates observed in modern-shod-western runners might be due to maladapted structure (i.e. loss of natural function), incorrect use of that structure (i.e. poor running technique) and subsequently aberrant lower limb loading. This is an unconventional view of running injury, but is rooted in both basic physics and evolutionary biology. It also provides a possible explanation for the high running-injury rates in mammals supposedly adapted for running. The aforementioned evolutionary evidence presented in this review provides the basis for the assumption that natural selection has adapted various structures of the human body specifically for the demands of ER. Physics provides the context for the suggestion that, used appropriately, these structures are adapted to the magnitudes and rates of loading associated with running, however used incorrectly or poorly maintained, these structures might succumb to the high forces and loading rates that running can produce. This is the theory on which this thesis is based. To examine the current evidence in light of the theory, some key definitions are required. For the remainder of the thesis, structure and function consistent with the evolutionary perspective shall be described

as ‘natural’ form and function. Patterns of form and function in published literature inconsistent with the evolutionary perspective shall be described as ‘normal’ form and function. To begin the argument for the theory that a mismatch between what is natural and what is normal might account for the high injury prevalence in adapted running specialists, it seems logical to begin with an examination of the structure that first makes contact with the ground.

2.5 Foot anatomy and function

The architecture of the human foot has evolved from a structure inherited from primate ancestors, and originally adapted for climbing and quadrupedal locomotion (Harcourt-Smith et al., 2002; Morton, 1935), to one which now acts as the connection between the ground and the rest of the body, performing a variety of important tasks that facilitate bipedal locomotion (Dugan & Bhat, 2005; Harcourt-Smith & Aiello, 2004; Hicks, 1954; Rolian et al., 2009). Primarily, these functions are to adapt to the terrain, support the body weight or multiples of it, absorb and return energy, and act as a lever to control the progression of body weight in the intended direction (Chou et al., 2009; Dugan & Bhat, 2005; Ker et al., 1987; Morton, 1935). The structure and evolutionary adaptations from the precursor primate foot that facilitate these functions are complex and warrant in-depth discussion. The complexity and elegance of nature’s solution to functional problems dealt with by the foot are said to have moved DaVincci to say that “The human foot is both a masterpiece of engineering and a work of art”.

2.5.1 Shock absorption (pronation)

For shock absorption in running, it is essential that the foot is able to adapt to the landing surface (Dugan & Bhat, 2005; Ker et al., 1987). The foot therefore must be pliable when making contact with the terrain. Assuming the common RFS gait observed in western populations (Lieberman et al., 2010; Rodgers, 1988), it follows that the foot is unlocked via a combination of ankle dorsiflexion, subtalar eversion and forefoot abduction, (commonly referred to as pronation) during the first 20% of stance (Dugan & Bhat, 2005; Rodgers, 1988). In addition, simultaneous rear-foot eversion and internal-tibial rotation also act to pronate at

the subtalar joint (STJ) (Dugan & Bhat, 2005). Pronation at the STJ is important as it facilitates a solid contact of the plantar surface with the landing surface (Dugan & Bhat, 2005), however this is not the only joint to pronate in the foot.

Secondary to pronation of the STJ, pronation at the transverse-tarsal joints (TTJ) causes the respective axes to become parallel and further contribute to foot mobility (Rodgers, 1988). This mobility has been reported to increase the adaptability of the plantar surface to the landing surface (Dugan & Bhat, 2005). Improved mobility also enables shock absorption (Dugan & Bhat, 2005), as well as a greater base of support through the spreading of the metatarsals. Duerinck et al. (2014) demonstrated this spreading most clearly when reporting the forefoot width (the distance between the metatarsal heads one and five) increases by 9% from heel strike to midstance when walking. Following these observations, it seems logical that an adaptable forefoot that spreads upon loading would also allow for greater directional stability. This is because a wider and compliant forefoot would be able to mould to a greater proportion of the terrain, providing a wider base of support and subsequently a greater axis of leverage to minimise unnecessary movement in the frontal plane (Chou et al., 2009; Pollock, Durward, Rowe, & Paul, 2000; San Tsung, Zhang, Fan, & Boone, 2003).

In clinical practice, the foot posture index (FPI) is often used to provide a description of foot type. This provides clinicians with an indication as to whether static foot posture is supinated, neutral or pronated. The measure has been validated (Redmond, Crosbie, & Ouvrier, 2006) and following has been used in research. For example, when categorised as either pronated, neutral or supinated Dahle, Mueller, Delitto, and Diamond (1991) and Yates and White (2004) both reported pronated feet predispose runners to knee pain and medial tibial stress syndrome compared to control groups. However, whilst this index provides an indication of static foot type, the foot is not rigid and moves under load in dynamic tasks such as ER. Whilst the FPI shows promise practicality for clinicians, other approaches such as 3D analysis that quantify pronation throughout stance present a continuous and dynamic alternative during ER. Together this highlights the importance of investigating pronation in ER for injury, as well as utilising 3D analysis equipment that can quantify pronation dynamically.

2.5.2 Energy storage and return

Post pronation the unlocked foot provides an opportunity for mechanical loading of the elastic tissues and the arch of the foot (Ker et al., 1987; Miller et al., 2014; Stearne et al., 2016; Wager & Challis, 2016). In particular, it is thought that the compression of the medial-longitudinal arch contributes most to the restitution of potential elastic energy in foot structure (Kelly et al., 2014; Ker et al., 1987; Lieberman, 2012b; Rodgers, 1988; Stearne et al., 2016). Stearne et al. (2016) demonstrated this *in vivo* by restraining the compression of the foot's medial longitudinal arch and reporting the subsequent effect on the metabolic cost of running. When running with a 'full arch insole' (reducing arch compression by 80%) they estimated an 8.8% reduction in the potential elastic energy stored, with a similar increase of 6% in the cost of travel. This is less than the predicted 17% of Ker et al. (1987), however speeds used by Ker et al. (1987) were greater (4.5 $m \cdot s^{-1}$) than that of Stearne et al. (2016) (2.7 $m \cdot s^{-1}$). This supports suggestions of Ker et al. (1987) that the contribution of the longitudinal arch of the foot is somewhat related to running speed. In support of this suggestion Lai, Schache, Lin, and Pandy (2014) investigated the effects of running speed on lower-limb elastic-structure contribution to the total positive work generated by the gastrocnemius and soleus muscle-tendon units. They reported the contribution of the tendon to elastic-strain energy increased from 53 to 74% and 62 to 75% for the soleus and gastrocnemius, respectively, as running speed increased. This supports the idea that elastic structure contribution is speed dependant. It therefore seems that higher running speeds present the potential to relatively reduce the cost of ER. For this reason, the medial longitudinal arch of the foot represents one of most important evolutionary lower-limb adaptations for ER in humans (Bramble & Lieberman, 2004; Lieberman, 2012a; Stearne et al., 2016).

2.5.3 Supination

The point of maximum pronation represents the end of the shock absorption period of stance and marks the point where heel lift and supination at the STJ begin (Chan & Rudins, 1994). Supination is the opposite of pronation. Characterised by plantarflexion, subtalar inversion

and forefoot adduction, it plays a key role in creating a stable-rigid lever for push off (Chan & Rudins, 1994; Dugan & Bhat, 2005).

Supination at the STJ commences as the heel rises, the stance limb begins to externally rotate and concentric action of the gastrocnemius causes the hind foot to invert (Dugan & Bhat, 2005). This hind-foot inversion also causes the TTJ axis to converge and lock to form a rigid structure (Bruckner, 1987; Dugan & Bhat, 2005). Simultaneously, the Windlass mechanism increases foot stability, with the flexion of fixed phalanges initiating an important sequence of bone articulations as the heel rises (Hicks, 1954). As the phalanges rotate around the metatarsal heads, plantar fascia tension is increased (Dugan & Bhat, 2005; Hicks, 1954). This increase in tension causes the plantar fascia to pull upon the plantar aponeurosis connected to the medial tubercle of the calcaneus (Dugan & Bhat, 2005). As a result of this tension, Hicks (1954) *in vitro* investigation reported that the metatarsal phalangeal joint shifts approximately 1 cm towards the calcaneus, this raises the medial longitudinal arch of the foot and further forces the TTJ into a rigid, packed and flexed position (Dugan & Bhat, 2005; Hicks, 1954; Mann & Hagy, 1979). It is this close-packed position that facilitates a solid-rigid lever for push off, a construct essential for transfer of both elastic recoil and active-muscle force to the running surface.

Intrinsic foot muscles also contribute to the transformation of the foot from a shock absorber to a rigid lever for push off. It is thought that when intrinsic foot muscles produce force they assist joint stability in a similar fashion to the plantar fascia (Dugan & Bhat, 2005). Kelly et al. (2014) investigated the contribution of intrinsic-foot muscles to participant's foot-joint stability while in a seated position with the knee flexed to 90°. From this position, participants had vertical loads applied to the knee joint to stress the longitudinal arch. This load was incrementally increased up to and beyond body weight. Results suggested that as the vertical load increased, the longitudinal arch deformed and stretched, and the electrical activity of the intrinsic foot muscles increased, as would be expected based on muscle spindle reflexes (Burke, Hagbarth, & Löfstedt, 1978). Kelly et al. (2014) went on to demonstrate that artificial stimulation of intrinsic foot muscles beyond natural activity levels allowed muscles to act with

a buttressing effect to the deformation of the longitudinal arch. These findings led Kelly et al. (2014) to conclude that intrinsic foot muscles have the potential to act in a similar manner to the longitudinal arch of the foot, controlling deformation of the arch, storing and releasing elastic energy, and possibly reducing mechanical demands, while increasing stiffness of the longitudinal arch towards final push-off.

Taking these factors together, energy stored in the longitudinal arch during pronation and load bearing is returned during supination to reduce the metabolic cost of running by between 8 and 17% depending on running speed (Ker et al., 1987; Stearne et al., 2016). With this in mind, restraining this mechanism would logically decrease elastic energy return, increase metabolic cost, muscle activity and injury potential.

2.5.4 Hind-foot to forefoot ratio

The hind foot is primarily made of up of the calcaneus, a large-solid structure with a fatty pad on the inferior aspect to help attenuate high impacts at initial contact (Lieberman, 2012a; Noe et al., 1993). The rear foot region has received much attention with research primarily investigating eversion excursions and velocities and their relation to injury etiology and injury rates. For example, Stacoff, Denoth, Kaelin, and Stuessi (1988) suggested the amplitude of rearfoot eversion might underpin the development of running injuries by imposing high loads on medial structures at the ankle joint. Retrospectively Vtasalo and Kvist (1983) reported the shin splint group had significantly ($P< 0.01$) greater angular displacement between initial contact and the maximum everted ankle angle. Additionally, Hreljac et al. (2000) suggested runners who had never sustained an injury reported significantly greater pronation velocity and a larger rearfoot supination angle at initial contact. This study could have been furthered by assigning differentiating variables to specific injury mechanisms. Conversely, when investigating differences between rearfoot motion in runners who suffer from patellofemoral pain and controls, Messier, Davis, Curl, Lowery, and Pack (1991) report rearfoot kinematics were not good discriminators between groups. This suggests specific measures of rearfoot motion might explicitly relate to specific injury mechanisms. In contrast to the array of

research conducted on the rearfoot the forefoot lacks the depth of attention assigned to rearfoot eversion measures.

The forefoot lacks the solidity seen in the hind foot and is generally broader, more mobile (Dugan & Bhat, 2005; Hicks, 1954). The relationship between a broad-mobile forefoot with a slim-rigid hind foot and the potential for injury is best demonstrated by D'AoUt et al. (2009) who compared the effects of habitual footwear use on foot structure and foot function. D'AoUt et al. (2009) demonstrated that a habitually-unshod population had a broader forefoot compared to a western-habitually-shod population. D'AoUt et al. (2009) then reported that those with a broader forefoot had a more uniform pressure distribution throughout the plantar surface when walking across a pressure plate, arguing similar to others, that a more evenly-distributed pressure could reduce the propensity of overuse injuries (Nagel, Fernholz, Kibele, & Rosenbaum, 2008; Weist, Eils, & Rosenbaum, 2004).

A wide forefoot also provides functional benefits for directional stability. Investigating the effects of hallux amputation, Chou et al. (2009) restrained asymptomatic participants' hallux to 30° of dorsiflexion using a splint, effectively reducing the phalangeal width. They reported that 'directional stability', quantified as a % of COP movement in the intended direction, was significantly worse post restraint. The results of Chou et al. (2009) and other static unipedal investigations suggest that a reduced-forefoot width results in a smaller mechanical lever to control the foot and therefore the COP (Chou et al., 2009; Hoogvliet et al., 1997).Together, this evidence suggests foot structure and any footwear that constrains forefoot width might increase plantar pressure and diminish foot control during static and walking trials; a factor associated with overuse injuries when running (D'AoUt et al., 2009; Hrysomallis, 2007), however, this is yet to be investigated in ER.

2.5.5 Hallux

As the foot is the only part of the anatomy to contact the ground, it makes sense that the magnitude and direction of the GRF, as well as plantar pressures, will in part be influenced by its anatomical structure (Chou et al., 2009; Morton, 1935; Yavuz et al., 2009). In particular,

evolutionary modifications of the hallux, such as increased thickness and abduction, represent important functional adaptations to provide directional stability and prevent injury (Chou et al., 2009; Mei et al., 2015; Morton, 1935). From a biomechanical perspective, these adaptations are important considering that the magnitude and direction of forces and peak-plantar pressures contribute to overload-related injuries (Edwards et al., 2010; Lieberman, 2012b; Lopes et al., 2012; Mei et al., 2015).

The main functional role of the hallux is to correctly direct the body weight during stance, while facilitating the windlass mechanism and creating a rigid lever for push-off (Chou et al., 2009; Lieberman, 2012a; Morton, 1935; Yavuz et al., 2009). The contribution of the hallux in directing the progression of the body weight was first discovered by Morton (1935), and recently demonstrated by Chou et al. (2009). Chou et al. (2009) removed the functional influence of the hallux by splinting it in 30 degrees of dorsiflexion, and reported that the 'directional control score', (a measure of the ability to direct the COP) significantly regressed in the forward-left, forward, and forward-right direction, when performing dynamic tasks. Morton (1935) and Plank (1995) also demonstrated a compromised functional capacity of the foot when the hallux was abducted (hallux Valgus) in walking trials. Both authors reported excessive pronation in feet with hallux valgus, as an adducted hallux can no longer produce forces to oppose the inward role of the foot due to the mechanically-compromised position. Consequently, this presents the potential for compromised loading of joints proximal to the foot as a result of compromised control of the GRF, and might explain the high injury rates observed at the knee (Taunton et al., 2003).

In contrast to the compromised function reported with hallux Valgus, Mei et al. (2015) demonstrated that the naturally-abducted hallux seen in habitually-barefoot populations (D'AoUt et al., 2009) demonstrated a unique ability to attenuate peak-plantar pressure by sharing the pressure at the forefoot during running. Based on previous discussions of injury mechanisms, this reduction in forefoot pressure might reduce hysteresis in the metatarsal heads and other tissue such as the plantar fascia, therefore reducing injury risk (Barr & Barbe, 2002; Mei et al., 2015).

In a neurophysiological study investigating the origins of manual dexterity, Hashimoto et al. (2013) mapped the neural and somatotopic representation of the fingers and toes in both living humans and monkeys. Results revealed both monkeys and humans represent fingers separately in the primary-sensorimotor cortex, similar to how they are separated in the hand. However, when investigating the toes, monkey toe function was fused. In contrast, humans had independent cortical representation of the hallux, with the representation of the lesser toes overlapping. Supporting this observation, Aiello, Dean and Cameron (1990) report that unlike the anatomy of chimpanzees and orang-utan, the human flexor hallucis longus inserts only to the hallux. Based on these observations, authors hypothesize that independent representation and anatomy of the hallux results from bipedal locomotion and underlines the importance of the hallux in bipedal locomotion.

In summary, it can be suggested that selective pressure for success in ER has adapted cortical organisation of the brain and the anatomical structure of the hallux to deal with the demand and high force associated with ER. Additionally, such pressure facilitated an anatomical position that contributes to pronation control. This evidence also confirms that malalignment of the hallux compromises the ability to direct body weight through the longitudinal axis of the foot, leading to compromised control of the GRF. Such loading patterns could potentially increase proximal loading mechanics at the knee joint. These proposed links are yet to be investigated.

2.5.6 Proprioception

Minimal footwear is not a new concept and dates back approximately 9,000 years, with its use attributed to a variety of tasks such as crossing coral reefs (Stewart, 1972). Notably, both new and old minimal-shoe designs facilitate essential sensory feedback via their thin and flexible soles. Also, as modern foot structure is a compromise inherited from primate ancestors, humans have not evolved the specialist protective-plantar surface tissues of other running-specialist (E.g. hooves of horses and pads of dogs). Together, it seems that humans require a form of protection for daily living, but one which does not compromise the sensory feedback mechanisms; a mechanism essential for injury avoidance.

During running, foot, footwear and terrain interactions stimulate mechanoreceptors providing important somatosensory information about impacts (Patel, Fransson, Johansson, & Magnusson, 2011). Both slow and rapid-adapting mechanoreceptors can be found in the foot. Slow-adapting-plantar mechanoreceptors provide information about how pressure is spatially distributed on the skin; whereas rapid-adapting mechanoreceptors provide information about the magnitude and change in magnitude of pressure exerted on the skin (Kavounoudias, Roll, & Roll, 1998; Patel et al., 2011). It is this information that is used by runners to modify technique accordingly (Gruber et al., 2012; Hsu, 2012; Lieberman, 2012b; Robbins & Waked, 1998; Robbins, Waked, Allard, McClaran, & Krouglicof, 1997).

Using the information provided by mechanoreceptors, runners adopt a variety of kinematic-running strategies to reduce running loads associated with shod running (Derrick, 2004; Gruber et al., 2012; Lieberman, 2012b). Gruber et al. (2012) investigated whether the RFS footfall pattern, a pattern some associate with injury (Daoud et al., 2012), and commonly seen in western populations was related to the lack of proprioceptive feedback imposed by CCRS. When running on a soft-cushioned material, similar to a running shoe, 80% of participants ran with a RFS. Conversely, when the material was removed, only 35% ran with a RFS and 27.5% and 37.5% changed to a mid-foot strike (MFS) and FFS, respectively. Gruber et al. (2012) suggested that running barefoot with a MFS/FFS was a kinematic response to the change in surface compliance and thus the likely increase in impact force when running on a hard surface. However, other variables besides surface compliance could have explained the change in foot strike angle, such as, stride frequency, stride length and overstride. To address this issue Lieberman, Castillo, et al. (2015) used a general linear mixed model approach to control for their influence. Authors reported that with the influence of internal, external, and acquired variables controlled, participants demonstrated a significant trend to RFS when running on a compliant surface ($P = 0.01$). Moreover, of the participants observed, those who habitually ran barefoot demonstrated a greater likelihood to MFS/FFS (1.88° ±0.85) when running on a hard surface. This underlines the importance of proprioceptive feedback to inform kinematic strategies when running and further highlights the detrimental effects of

modern footwear has on running technique and injury rates given discussions establishing the link between increased injury rates and foot strike strategy (Daoud et al., 2012).

McNitt-Gray and Yokoi (1990) investigated the effects of drop landing surfaces on vertical GRF when reporting the kinetic consequences of proprioception attenuation. McNitt-Gray and Yokoi (1990) found that when gymnasts were instructed to drop from identical heights (0.69m), gymnasts landed with greater impact forces on a soft-compliant surface compared to hard surface as a function of reduced hip and knee flexion. Although not in the context of running, this study lends support to the argument that proprioception of peak impacts is essential to inform force-attenuating-landing strategies and avoid increased joint loading. Furthermore, Robbins and Gouw (1990) propose this observation of increased force when landing upon more compliant surfaces could be an attempt to compress compliant surfaces and gain a secure support base in response to the sense of unstable equilibrium underfoot. Together this suggests that proprioceptive feedback informs kinematic strategies, such as a FFS gait to reduce impact forces underfoot, but when impact is masked, and the surface underfoot is unstable, humans employ increased stiffness to compress compliant and unstable surfaces.

Another strategy reported to reduce joint loading when running is increased knee-joint flexion at initial contact. Derrick (2004) reported that this strategy attenuates the propensity for injury by reducing effective mass. Derrick (2004) suggests that such strategies mostly occur when proprioceptive feedback suggests that the likelihood for injury is greater, such as running on irregular surfaces. However, increased knee flexion at initial contact has the potential to increase metabolic demands through the positive relationship it shares with maximum knee flexion and increased metabolic cost (Derrick, 2004). This demonstrates that proprioceptive feedback reduces the likelihood of injury and has the potential to initiate strategies that sacrifice metabolic efficiency to reduce injury risk.

The proprioceptive feedback provided by the plantar surface also provides important information for dynamic stability (Nurse & Nigg, 2001; Robbins et al., 1997; Robbins et al., 1994). In classic investigations by Robbins, Waked, and McClaran (1995) and Robbins et al.

(1994), the effects of footwear on the ability to estimate foot slope and effectively walk across a thin beam were investigated. Robbins et al. (1995) reported that wearing footwear significantly increased error estimation, and Robbins et al. (1994) reported that an increase in the thickness and reduction in hardness of the midsole increased balance failure when walking across a beam by 54 and 77% respectively. From these results, authors suggest that any material attenuating plantar-surface feedback reduces positional awareness of the foot and the ability to adapt and execute a safe movement strategy.

Based on the available evidence, it seems that the plantar surface of the foot has evolved to be essential for avoidance of injury. The evidence suggests that interference with sensory feedback might increase the propensity for injury. This underlining the importance of proprioceptive feedback in reducing injury rates. However, the foot still needs to be protected from the urban environment, something which is often littered with sharp and potentially dangerous objects. This provides a strong argument for minimal shoes to reduce injury rates while maintaining some proprioceptive feedback.

2.5.7 Ontological development of the foot

Shoe choice is not a decision to be taken lightly, with research reporting the detrimental effects that specific types of footwear have on the structural development of the foot dating back well over 100 years (Hoffmann, 1905; Hsu, 2012; Kadambande, Khurana, Debnath, Bansal, & Hariharan, 2006; Lieberman, 2012b; Miller et al., 2014; Walther, Herold, Sinderhauf, & Morrison, 2008). In particular, anthropometric research provides an in-depth insight into the effects of footwear on the previously discussed evolutionary adaptations of the foot. For example, the flexibility of the forefoot, the muscular strength of intrinsic foot muscles and the development of the longitudinal arch have all been shown to be compromised by non-anatomically-shaped footwear (Bramble & Lieberman, 2004; Kadambande et al., 2006; Rao & Joseph, 1992; Stearne et al., 2016; Walther et al., 2008). These general findings suggest that wearing ill-fitting shoes reduces the capacity of the foot to develop and function naturally.

Julius Wolff (1836–1902) was one of the first to report that when environmental loads placed on bones are changed by habitual activity (such as wearing restrictive footwear), a bone will remodel to adapt to the new demands (Frost, 1990). This principle is fundamental to the relationship between footwear and foot structure and demonstrates that the mediating variable in the ontological development of foot shape is the force exerted on the bones. This principle is particularly important when considering the development of children's feet (Hoffmann, 1905; Walther et al., 2008). Hoffman's (1905) classic investigation of the morphological differences between shod and un-shod populations discussed this issue. Hoffmann (1905) reported that the foot is especially plastic at the young age associated with the transition from barefoot to conventional-modern footwear, and given continued restraint, the foot can be easily mal-adapted. Walther et al. (2008) also reports that in young children (up to the age of four) the bones are particularly malleable as they are yet to ossify and are easily influenced by unnecessary-restrictive forces (D'AoUt et al., 2009; Hubbard, Meyer, Davidson, Mahboubi, & Harty, 1993; Kadambande et al., 2006; Whitaker, Rousseau, Williams, Rowan, & Hartwig, 2002).

Based on this evidence, it seems logical that if footwear restrains foot movement, or attenuates the natural forces experienced by the foot bones during walking or running, the consequences would most likely be a compromised foot structure and associated-compromised function.

Following the principles of Wolff and its applications to anatomical-foot development (Frost, 1990) it seems logical that an altered loading pattern would also compromise intrinsic-foot-muscle function. Work by Miller et al. (2014) reported a reduction in the anatomical cross sectional area was observed and subsequently intrinsic-foot muscles became weak, following the excessive support provided by footwear, and subsequent lack of stimulation (secondary to a compromised loading pattern). However, neither muscle activity, strength nor function was examined by Miller et al. (2014). In contrast, Kelly, Lichtwark, Farris, and Cresswell (2016) reported an increase in peak activity (flexor digitorum brevis +60%) and total stance muscle activation (flexor digitorum brevis +70%; abductor hallucis +53%) when running shod

compared to barefoot. Based on the latter observation, the potential for injury could increase in response to the overloading of intrinsic-foot muscles when shod.

If modern-shoe design compresses the foot into a compromised shape, then it seems logical that this compressive force will also promote a compromised foot shape, intrinsic-muscle function and subsequently poorer control of the foot (Chou et al., 2009; Kadambande et al., 2006; Miller et al., 2014; Shine, 1965). Moreover, physics dictates a body is better balanced over a larger base of support (Pollock et al., 2000) and conventional footwear limits natural-anatomical development, so it can be hypothesised that metrics of foot structure such as "ball-of-foot" or "phalange" width will be related to dynamic control of bodyweight and subsequently injury risk. This is yet to be explored.

2.6 Elements of shoe design that could change foot development

As previously discussed, footwear dates back around nine thousand years and has served several roles such as protecting feet from sharp coral (Stewart, 1972). Recently, there has been a boom in footwear targeted specifically towards endurance runners, claiming to enhance performance and reduce injury risk (Lieberman et al., 2010). Notably, a review examining scientific investigations in support of these claims reported that no evidence exists to suggest that current running shoe design can accomplish these aims (Richards et al., 2009). Taking Richards et al. (2009) findings further and remembering that this thesis is in part focused on foot function secondary to foot structure, an in-depth discussion of the effects that individual elements of conventional-cushioned-running shoes have on foot function is warranted.

2.6.1 Narrow toe box

Investigations of habitually-barefoot populations consistently report that the widest part of the foot is the forefoot as a result of the abduction of the hallux and flailing of the phalanges (D'AoUt et al., 2009; Mei et al., 2015; Shu et al., 2015). Conversely, when examining the design of a CCRS, the narrowing of the toe box towards a central point represents a poor fit for the natural-anatomical foot structure (Hoffmann, 1905; Kadambande et al., 2006). This

contrast between natural foot anatomy and the CCRS results in the most anterior aspect of the forefoot being forced centrally producing a cascade of compromising effects on foot function.

A consequence of fitting a naturally-wide forefoot into a narrow toe box is compression of the metatarsals (Hoffmann, 1905; Morton, 1935). This compression can cause the first metatarsal to both elevate and supinate, resulting in the commonly cited Morton's foot syndrome (MTFS) (Morton, 1935; Rodgers & Cavanagh, 1989; Sauer, Biancalana, & Filner, 2010). This translated position of the first metatarsal represents the origin for a series of foot-function issues. The elevation of the first metatarsal head shifts load bearing medially to the second metatarsal head, increasing plantar pressure under the second metatarsal head compared to those without MTFS (Rodgers & Cavanagh, 1989). This is important in the context of injury, with those who suffer from increased plantar pressure often developing the painful condition of plantar keratosis under the head of the second metatarsal and Morton's Neuroma, a mechanical entrapment neuropathy of the interdigital nerve (Hockenbury, 1999; Hunt, McCormick, & Anderson, 2010).

Secondary to compression caused by narrowing of the toe-box, a reduction in forefoot width will also follow (D'AoUt et al., 2009; Shine, 1965; Wolf et al., 2008). In theory, this will compromise the directional control of bodyweight, given that wider base of support and uncompromised hallux have been shown to improve dynamic stability (Chou et al., 2009; Hoogvliet et al., 1997). Moreover, time spent in a narrow toe box also increases the likelihood of hallux valgus, and thus a narrowing of the phalanges. Shine (1965) reported a significant linear and positive association in men and women ($P<0.001$) for years of shoe wear, hallux-valgus angle and the percentage of the population suffering from hallux valgus. Furthermore, prospective work by Munteanu et al. (2017) reports this observation is not hereditary. Munteanu et al. (2017) examined the development of hallux valgus over the life span in 74 and 54 pairs of identical twins and non-identical twins, respectively, and reported no evidence for genetics to predict hallux valgus. Instead, regularly wearing a compressive toe-box design predicted the likelihood of the condition developing. These findings suggest that not only does

the wearing of shoes impact the structure of the foot, but time spent in conventional-non-anatomically-shaped shoes increases the likelihood of hallux valgus.

In the context of running-related injuries, excessive pronation resulting from misalignment of the hallux is of particular importance (Hunt et al., 2010; Plank, 1995). A narrow toe box forces the hallux into adduction, and promotes hallux valgus (Shine, 1965). As previously discussed, this position of adduction prevents the hallux counteracting the inward roll of the forefoot causing excessive pronation (Morton, 1935; Plank, 1995). Excessive pronation shifts the weight to the medial aspect of the first MTP joint. This overloading of the MTP joint is important, given that the literature cites excess pronation as a key component in the development of disorders such as bunions, hammer-toe (a crossover deformity) (Hockenbury, 1999; Hunt et al., 2010) and running-related injuries (Messier & Pittala, 1988). Furthermore, if the function of the hallux is compromised and the foot structure can no longer oppose the inward roll of the foot then this might also compromise loading of joints more proximal to the foot.

Collectively, forefoot width and the natural alignment of the hallux play an important role in maintaining static and dynamic stability, directing body weight through the longitudinal axis and preventing excessive pronation. However, if a narrow toe box compresses these structures, it follows that both hallux valgus and a reduced phalange width will compromise a runner's ability to control foot motion and the associated GRF. The link between foot structure, footwear and the ability to control foot motion in dynamic tasks like running can be deduced from first principles (Wilkinson & Saxby, 2016), but is yet to be explored in practice.

2.6.2 Toe-spring

Conventional running shoes are typically designed with an upward-curve towards to most anterior point of the shoe under the forefoot, termed a 'toe-spring' (Willwacher, König, Potthast, & Brüggemann, 2013). The toe-spring is thought to facilitate the passage of body weight forwards towards toe-off late in stance, as well as to attenuate forefoot pressure (Willwacher et al., 2013). However, a toe-spring raises the toes into a dorsi-flexed position

such that they are not in contact with the ground at midstance, this reduces the active base of support and the contact of the hallux whereas previously discussed it acts to direct the body weight in the transverse plane (Chou et al., 2009; Morton, 1935). Furthermore, toe flexion tightens the plantar-arch tissues via the Windlass mechanism (Ker et al., 1987). This naturally occurs immediately before toe-off locking the fore-and hind-foot creating a rigid lever for effective transfer of force into the ground. It is however, inappropriate prior to this when the foot should be compliant for stability, shock-absorption and energy storage (Ker et al., 1987; Rodgers, 1988; Stearne et al., 2016).

Though the upward curve of the toe-spring was originally used as a symbol of status (Stewart, 1972), and more recently has been argued to allow for the forward progression of the weight in shoes that were unable to bend (Willwacher et al., 2013), modern research suggests a toe spring will compromise stability and reduce shock absorption. This increases the likelihood of overuse and impact-related joint injuries at the ankle and knee, (Barr & Barbe, 2002; Hrysomallis, 2007; Lieberman, 2012b; Lopes et al., 2012; Schaffler et al., 1989; Willems et al., 2005).

2.6.3 Arch support

Arch height has been reported by research as an important anatomical indicator of foot structure. To date research has investigated the effects of arch height and its association with injury in endurance running populations (Kaufman, Brodine, Shaffer, Johnson, & Cullison, 1999; Pohl, Hamill, & Davis, 2009). Evidence from both authors suggest that pes cavus and pes planus have potential for injury in endurance running population. Specifically, participants with pes cavus had nearly twice the incidence of stress fractures compared with those of average arch height (Kaufman et al., 1999). Also, it has also been reported that runners that suffer from plantar fasciitis report significantly lower arch height (pes planus) as compared to controls (Pohl et al., 2009). Whilst this evidence demonstrates foot anatomy has been investigated for some time with results highlighting its importance in predicting injury, how footwear choice might impact arch height in dynamic tasks such as ER has received little attention.

Footwear with in-built arch support is thought to have been designed with the intention of altering foot and lower-limb mechanics associated with injury such as excessive pronation (Franz et al., 2008; Kerrigan et al., 2009; Messier & Pittala, 1988; Nigg, 2001). However, as previously mentioned, Richards et al. (2009), argue that no evidence exists to suggest that CCRS design improves performance or reduces injury risk in endurance runners. Furthermore, medial arch support is suggested to interfere with the natural collapse of the medial-longitudinal arch, decreasing shock absorption and energy return (Hsu, 2012; Ker et al., 1987; Lieberman, 2012b; Perl et al., 2012; Stearne et al., 2016).

If a solid medial arch opposes the natural collapse of the longitudinal arch during ER, then it seems logical that the potential for energy restitution will also be compromised. Stearne et al. (2016) reported that blocking the arch (with a full-arch insole) reduced arch compression by approximately 80%, increased mechanical work by 8.8% and subsequently increased metabolic cost (+6%) when running. Authors also reported that the difference in metabolic cost between a low and high arch (compared to no support) was not significantly different suggesting that arch strain increases in a non-linear fashion, and that even minimal support is detrimental. These findings support earlier work by Ker et al. (1987) who was first to report that the longitudinal arch of the foot provides a restitution of energy of approximately 17%. The magnitude of the energy return differs markedly between the two studies. This can be partially explained by the difference in speeds that each investigation used (Stearne et al. (2016): 2.7; Ker et al. (1987): 4.5 $m \cdot s^{-1}$). This suggests that elastic contribution might be velocity dependant. Velocity-dependant elastic contribution would have been a logical musculoskeletal adaptation for natural selection to retain, as if the cost of locomotion increases with speed (in response to increased forces), a mechanism whereby energy restitution increases in a similar trend would be beneficial for energy conservation in the early *hominin*. Furthermore, it is also noteworthy to mention that Ker et al. (1987) used cadavers from individuals who suffered from a disease that necessitated their amputation, thus a possibility exists that the mechanical properties of the tissue might react differently to compression compared to the *in vivo* testing of Stearne et al. (2016). Based on this evidence, it seems that

blocking the medial arch can increase the metabolic cost of running, as well as the likelihood of injury through increased mechanical demands that will be placed on the lower limbs (Lopes et al., 2012).

Secondary to this arch compression, if the mid foot does not compress when bearing weight, the foot will be unable to fully pronate and conform to the support surface (Dugan & Bhat, 2005; Wolf et al., 2008). This has consequences for running injuries, given that previous research not only demonstrates that forefoot pronation is essential in facilitating forefoot pliability (Dugan & Bhat, 2005), but also because pliability of the forefoot facilities a wider the base of support. It is this wider base of support that provides improved dynamic stability and therefore reduced likelihood of injury (Chou et al., 2009; Hoogvliet et al., 1997), a mechanism inhibited by conventional-cushioned-running shoes.

Intrinsic muscles of the foot, like other muscles in the body are subject to atrophy with lack of use. Constrained by arch supports, intrinsic-foot muscles crossing the medial-longitudinal arch are likely to weaken (Bruggemann, Potthast, Braunstein, & Niehoff, 2005; Hsu, 2012; Lieberman, 2012b; Lieberman et al., 2010; Miller et al., 2014). In practice, intrinsic-foot muscles have been shown to influence the elevation and functional capacity of the medial-longitudinal arch (Bruggemann et al., 2005; Kelly et al., 2014). This evidence suggests that arch support should be absent from shoe design, however little research has been conducted in this area. Only recently, Miller et al. (2014) investigated the effects of minimal shoes on the anatomical cross sectional area of specific foot muscles, reporting an increase in anatomical cross sectional area of key intrinsic-foot muscles after 12 weeks wearing a minimal shoe devoid of arch support, reduced cushioning, and a forefoot-heel offset of < 4mm. This demonstrates that the current CCRS design induces atrophy of intrinsic foot muscles that support the longitudinal arch (Kelly et al., 2014), and contribute to a rigid-foot structure in supination. However, although foot strength was inferred, neither foot muscle strength nor activity was directly measured. In a study examining the effects of shod footwear on longitudinal arch kinematics and intrinsic foot muscle activity while running barefoot and shod, Kelly et al. (2016) reported an increase in peak activity (flexor digitorum brevis +60%)

and total stance muscle activation (flexor digitorum brevis +70%; abductor hallucis +53%) when shod. Furthermore, increased-intrinsic-muscle activation was observed alongside a reduction in longitudinal arch compression (25%). It was suggested that the cushioning of the CCRS acted in series with the foot to reduce system stiffness. Following this, intrinsic-foot muscles increase activity to maintain system stiffness by increasing longitudinal arch stiffness. Most recently, Holowka, Wallace, and Lieberman (2018) tested the hypothesis that the regular use of shoes that restricts foot motion (arch support) are associated weaker foot muscles and reduced stiffness. Results showed north-western Mexican habitually shod men from urban areas reported a significantly reduced cross sectional area for abductor hallucis and abductor digiti minimi muscles, and reduced stiffness in the longitudinal arches compared to minimally shod individuals. Additionally, adductor hallucis cross sectional area was associated with longitudinal arch stiffness. These results suggest the use of CCRS subject intrinsic foot muscles to atrophy and reduce longitudinal arch stiffness.

In summary and in agreement with Richards et al. (2009), arch support in a CCRS presents no benefit to ER. Employing footwear with an arch support can increase the mechanical demands of lower-limb muscles and reduce their cross-sectional area. In addition, excessive activity of intrinsic-foot muscles presents a possible mechanism that could predispose runners to injury via increased and prolonged demands on small-specialised muscles.

2.6.4 Elevated Heel

The addition of elevated and cushioned heels for performance dates back to the early 1970's, but before this, elevated and cushioned heels can be seen in 1832 where rubber soles were added to the soles of shoes for greater durability (Davis, 2014). The next major progression came in 1964, when the first pair of cushioned-athletic-Japanese running shoes were manufactured (creating with it the company ASICS) (Vanderbilt, 2008). These shoes soon found their way to the USA, were Arthur Lydiard, a world-class running coach promoted their use to large groups of heart-attack rehabilitation patients. Bowerman, a student of Lydiard wrote a best-selling book entitled 'Jogging' detailing his realisation of the benefits of jogging (Bowerman & Harris, 1967). However, Bowerman had the unproven idea that increasing

overstride as a function of a heel-to-toe gait, would be the least tiring over long distances (McDougall, 2010). In response to this ill-informed idea Bowerman and Knight developed the Cortez, a running shoe with an elevated cushioned heel attenuating the discomfort associated with the RFS resulting from the overstriding pattern. This shoe was hugely successful, and laid the foundations of the company now known as Nike.

More recently, elevated heels have been marketed to protect the foot from large impacts forces and associated high rates of loading when employing a RFS strategy (Kerrigan et al., 2009; Lieberman, 2012b). This marketing is largely based on machine testing protocols that do not account for proprioceptive feedback and human behavioural responses to compliant materials (Robbins & Waked, 1997). Based on advertising and mechanical testing alone, a runner could be forgiven for assuming an elevated and cushioned heel might protect them from injury, however, this elevation has been argued by some to encourage running mechanics associated with injury (Daoud et al., 2012; Kerrigan et al., 2009).

From a biomechanical perspective, a cushioned heel under the posterior aspect of a CCRS elevates the heel from its natural position (Gruber et al., 2012; Lieberman, 2012b; Lieberman et al., 2010). This elevation of the hind foot means that when a runner attempts to employ a MFS, the additional material under the hind foot provides a dorsiflexion off-set and encourages the commonly seen RFS pattern, a pattern associated with factors such as such as increased effective mass, impact transients and injury rates (Daoud et al., 2012; Lieberman, 2012b).

Also, when employing a FFS, a dorsiflexion off-set will attenuate joint excursion at the ankle and loading of the Achilles tendon will be limited (Lieberman, 2012b). This decreased joint excursion will reduce the amount of eccentric work performed by the triceps-surae complex, a factor thought to be essential in controlled lowering of the hind foot in a FFS gait, reducing contact forces in this strike pattern (Lieberman et al., 2010; Perl et al., 2012).

Collectively, this evidence suggests that elevating the heel not only encourages a runner to RFS which some associate with increased impact forces, but also reduces the range of motion

(ROM) over which the Achilles tendon has influence. Following this, and in the context of energy utilisation, it can also be argued that a cushioned heel will reduce the potential for energy restitution at the Achilles tendon complex. However, it should be noted that substrate, speed and overstride also play a role in the loading mechanics of running (Gruber et al., 2012; Lieberman, 2014; Lieberman, Warrener, et al., 2015) and elevated heels are but only a single factor in the multifactorial nature of overuse injuries. In addition, it is important to note that sensory insulation provided by excessive cushioning from CCRS also encourage overloaded running mechanics (Robbins et al., 1994).

2.6.5 Sensory insulation

Proprioceptive feedback provides key information about the position and magnitude of forces under the foot (Kavounoudias et al., 1998; Patel et al., 2011). However, the cushioning in modern footwear, in particular under the heel, impairs sensory input when performing static and dynamic tasks (Gruber et al., 2012; Robbins et al., 1994; Rose et al., 2011). Gruber et al. (2012) investigated the effects of sensory insulation and foot-strike strategy (a factor associated with injury) while running on a cushioned surface (similar to CCRS). Gruber et al. (2012) demonstrated that when sensory feedback is attenuated (running barefoot on a cushioned surface), runners typically adopted a RFS; however, when fully sensate (running barefoot on a hard surface), a strong relationship between change in foot-strike strategy (RFS to FFS) and change in surface conditions (compliant to hard) was evident ($P = 0.0008$). This demonstrates the extent that CCRS masks sensory feedback, and lends support to the argument that CCRS encourage the overloading of running mechanics given the association between RFS and injury (Daoud et al., 2012). These findings are in agreement with the biological imperative, that is, a task will be performed with minimum energy expenditure while simultaneously minimising injury risk (Alexander, 1989; Sparrow, 2000). However, with afferent feedback masked by the soft-cushioned running surface, a runner will stride out (De Wit, De Clercq, & Aerts, 2000; Kerrigan et al., 2009), a strategy known to minimise energy expenditure as a function of increased t_c and reduced frequency of times the bodyweight is supported per distance (Kram & Taylor, 1990). However, while economical, this strategy has

been associated with increased lower-limb joint moments (Kerrigan et al., 2009) and the potential for overuse injuries.

When hypothesising the functional implications of sensory insulation on injury potential, Robbins et al. (1994) theorised that the plantar surface would not deform under high loads while wearing athletic footwear. This led Robbins et al. (1991; 1994) to suggest that sensory insulation will lead to an underestimation of forces, and following this, overuse injuries as a function of this 'perceptual illusion' (Robbins & Gouw, 1991). Nurse and Nigg (2001) support this argument with their work on the effects of plantar desensitisation on a dynamic balance task. They demonstrated that when specific areas of the foot (rear-foot and forefoot) were desensitised, peak pressure moved to an area that was still sensate; and when the whole foot was desensitised, pressure increased under metatarsal heads. The change in distribution and increases in pressure when fully desensitised were suggested to protect lower areas of sensitivity from risk of injury when sensory feedback was lost, and to use areas with greater feedback to improve balance. The observations of overall increased pressure when fully desensitised support the theory of Robbins and Gouw (1990) that participants' increase lower-limb stiffness, and subsequently plantar-pressure in an attempt to compress compliant proprioceptive-attenuating materials to improve sensory feedback.

When the effects of increasing hardness and therefore increasing sensory feedback were investigated, Nigg (1986) reported increased shoe stiffness decreased impact. In context, changing from a shore of 35 to 25 increased impact from 2170 to 2300N. This supports the argument that increased shoe stiffness has the potential to decrease joint loading. Although not argued by the author, this reduced impact could be the product or combination of improved proprioceptive feedback provided by the stiffer material, or the reduced need to compress an unstable-compliant surface. These findings are particularly pertinent when running in CCRS when the foot is placed upon an unstable-compliant surface and proprioceptive feedback is reduced. Based on these findings it can be suggested that wearing stiffer shoes decreases impact forces, potentially increases proprioceptive feedback and subsequently decreases the potential for injury. Furthermore, this evidence also supports the argument that a minimal shoe

design could reduce the likelihood of injury and maximally-cushioned footwear might increase the likelihood of injury.

Sensory insulation also has the potential to compromise running-related postural control. Rose et al. (2011) investigated the effects of sensory insulation on the ability to perform a running-related postural-control task. Eighteen participants employed three different conditions in a randomised order when performing a running-like action (barefoot, minimal shoe and CCRS). Participants were instructed to jump to a force platform from an elevated platform (70 cm away and 10 cm elevated) from their dominant limb to their non-dominant limb. The findings of Rose et al. (2011) support the argument of this review, suggesting that the excessive sensory insulation, provided in this case by the CCRS filters essential sensory input. This resulted in a significant increase in the anterior-posterior, medio-lateral, vertical and dynamic-postural stability index (resultant index), in CCRS compared to the barefoot condition. Results also suggested that minimal shoes (Vibram 5-fingers) elicit non-significant differences between the barefoot condition in the anterior-posterior, vertical and dynamic-postural-stability index. This suggests that a CCRS mask important sensory feedback for postural-control tasks, and that minimal footwear with thin soles potentially improve postural control compared to CCRS.

In summary, a variety of stability measures as well as running-related postural control appear to be compromised by footwear that inhibits the sensory feedback provided from the running surface. This loss of normal loading mechanics on the plantar surface, and the need to compress complaint materials upon impact present a strong argument that sensory insulation increases an individual's potential for running related injury.

2.7 Running technique

The technique of running is not a set pre-programmed kinematic pattern which we have evolved in response to the need to survive, but rather an ever varying skill driven by the biological imperative, that is, the drive to perform a task with an optimum balance between minimum energy expenditure and injury potential (Gruber et al., 2012; Kram & Taylor, 1990; Sparrow, 2000). It can be argued that a natural style of running is one that uses sensory

feedback mechanisms and natural anatomical structures to minimise energy expenditure and reduce the potential for injury. Following this, if the internal or external factors that influence the potential for injury or energy expenditure change, the skill of running will adapt accordingly to best serve the biological imperative.

Generally, there is no incorrect way to locomote, but rather less or more appropriate solutions based upon external and internal factors. Substrate, speed and proprioceptive feedback are examples of factors that influence choice of locomotive solution (Cavanagh & Kram, 1989; Gruber et al., 2012; Hatala, Dingwall, Wunderlich, & Richmond, 2013; Nilsson & Thorstensson, 1989; Robbins & Gouw, 1991). Research suggests there are two primary strategies for humans to locomote (Srinivasan & Ruina, 2006). Using computer modelling, Srinivasan and Ruina (2006) reported two gait strategies when the outcome measure was set to minimse energy expenditure and the independent variable was velocity. The classic inverted pendulum was optimal at low velocities, but a bouncing-running gait was optimal at high velocites. The finding of a bouncing gate is of particular interest, as the model did not account for elastic-like structures which would have further exaggerated this bouncing locomotion strategy. However, the findings of this study could be argued as rudimental as the spring like tendons of the human body were not accounted for (Blickhan, 1989), the knee joint was absent, and the energetic cost of leg swing was neglected. That said, the finding of a bouncing like gait is in aggreement with previous observations using models that account for these factors (Geyer, Seyfarth, & Blickhan, 2006). In addition, a spring-like system that flexes upon impact would attenuate the high impact forces assosiated with running by converting translational energy into rotational energy and storing elastic energy for the latter half of stance (Derrick, 2004; Lieberman et al., 2010). These findings suggest that a spring-like bouncing gait is optimal for economy and safety at ER speeds, and by employing these mechanics, the potential for injury is decreased, yet paradoxially, injury rates remain high in normal running populations.

Normal running can often lead to injury because the running technique is more pendular like, with an extended lower limb at initial contact (De Wit et al., 2000) rather than the more spring

like and flexed technique seen in natural running. At low speeds a pendular technique is advantageous, as the force is lower and the energy cost is lower compared to that of the spring like model. This is because a near extended leg in stance efficently exchanges potential-kinetic energy and kinetic energy with each step (Bramble & Lieberman, 2004). However, at higher speeds, a spring like technique with joint excursions is needed to deal with higher forces and the increased risk of injury. Though the energy cost of a bounding stratergy is higher, some energy can be retured through elastic structures in the lower limbs (Bramble & Lieberman, 2004). The transition between walking and running strategies was investigated by Sasaki and Neptune (2006) who modelled the muscle fibre and series elastic components of healthy adults while walking and running both below and above their preferred walk-to-run transition speed. Simulations revealed that running above their preferred-transition speed produced a greater elastic contribution compared to below. This supports the notion that mechanical demands placed on muscles via increased GRF (secondary to increased speed) play an influential role in walk-to-run transition speed. However, if sensory-input mechanisms such as mechanoreceptors are insulated runners will not be able to detect increased GRF's and accurately inform alterations to their running gait, and a normal running gait ensues, bringing with the potential for increased kinetic loading.

2.7.1 Kinetic consequences of common/ 'normal' running technique

Differences between natural and normal running technique have been reported to cause changes in the forces that act on the body (Kerrigan et al., 2009; Lieberman, 2012b; Lieberman et al., 2010). The most common being the impact transient observed during the loading response in normal running (Divert, Mornieux, Baur, & Mayer, 2005; Lieberman, 2012b; Lieberman et al., 2010). Lieberman et al. (2010) highlighted the presence of this kinetic variable as unnatural in their landmark investigation on the effects of foot strike strategies on GRF's in Kalenjin and American runners. Lieberman et al. (2010) reported peak vertical GRF during the impact period was approximately three times lower in habitual barefoot runners who FFS compared to habitual shod runners who RFS (habitually barefoot and FFS: 0.58 ± 0.21 BW; habitually shod and RFS: 1.74 ± 0.45 BW). This suggested those who employ a

‘normal’ RFS could be predisposed to injury as a result of an increased musculoskeletal load. Lieberman et al. (2010) also reported that the LR of habitual barefoot runners who FFS was similar to the habitual shod runners who RFS (habitually barefoot and FFS: 64.6 ± 70 $BW \cdot s^{-1}$; habitually shod and RFS: 69.7 ± 28.7 $BW \cdot s^{-1}$), but several times lower than habitually shod runners who RFS when barefoot (463.1 ± 141 $BW \cdot s^{-1}$). This suggested that wearing cushioned running shoes helps attenuate loading rates of habitually RFS runners, from which it could be argued that a cushioned running shoe represents a safer means to locomote based on LR alone. However, these findings do not provide insight into individual joint loading, and do not correlate well to the findings of Daoud et al. (2012) who observed that RFS runners had twice the likelihood of suffering repetitive stress injuries compared to FFS runners over a two-year period of collegiate-track running.

Research suggests that a RFS gait, typical of a normal running strategy, and secondary to sensory insulation, is associated with increased injury rates (Daoud et al., 2012). Lieberman et al. (2010) findings suggest a mechanism, as habitually shod runners with a RFS gait demonstrated increased peak vertical loads compared to the habitually barefoot and forefoot-striking Kalenjin’s. Indeed, this would seem to suggest that foot-strike strategy might account for overuse injuries, and this argument is further supported by the relationship that exists between musculoskeletal damage and increased in peak loading (Barr & Barbe, 2002). However, Hatala et al. (2013) observed Daasanach barefoot tribe members from Kenya to also RFS, like the Americans of Lieberman et al. (2010) who recorded greater peak impact loads and loading rates compared to their habitually barefoot and forefoot striking counterparts. This observation was unexpected, as a RFS gait has been associated with injury (Daoud et al., 2012), therefore, it is suggested that other variables also influence a individuals kinetics and potential injury risk. Hatala et al. (2013) offered suggestions to explain the contrast in findings. Compared to the work of Lieberman et al. (2010) the substrate the Daasanach tribe ran on was soft and compliant, which might have acted to attenuate vertical GRF’s, and play a role in foot-strike-strategy selection (Gruber et al., 2012). Additionally, the reduced average speed seen in Hatala et al. (2013) (3.3 $m \cdot s^{-1}$), compared to Lieberman et al. (2010) (5.1 – 5.9 $m \cdot s^{-1}$)

represents a means to reduce the load under the hind foot of Daasanach tribe members to that which would not require a runner to adopt a FFS strategy (Nigg, Bahlsen, Luethi, & Stokes, 1987). Lastly, the Kalenjins' studied in Lieberman et al. (2010) were noted to run in excess of 20 Km per week, in contrast to the short distances reported for the Daasanach tribe. This would suggest that Daasanach tribe members might not be conditioned to the sustained increase in eccentric work at the Achilles-tendon complex when employing a FFS, and a RFS prevailed. These possible explanations support this review in that 'natural running' is not a set skill, but rather a skill that is adaptive and dependant on sensory feedback, something that is attenuated in the normal running technique and might provide a possible explanation to the high injury rates in normal runners. In light of these conflicting findings, research studies have begun to suggest that foot-strike strategies do not fully account for injury mechanisms (Lieberman, 2012b), but rather that injury mechanisms are a consequence of overall running technique, a skill influenced by footwear choice, the employment of which often differentiates the normal and natural running technique (Kerrigan et al., 2009).

2.7.2 Effects of footwear

Modern athletic footwear is purported to play a beneficial role in long distance running and prevention of injury. In particular, modern shoes designed with arch supports, toe springs, elevated heels and pronation control systems are thought to reduce injury (Franz et al., 2008; Kerrigan et al., 2009; Lieberman, 2012b; Willwacher et al., 2013). However, as previously discussed, Richards et al. (2009) failed to identify any research meeting their criteria and concluded that the prescription of this shoe type to distance runners is not evidence based. In spite of this, this type of footwear is used by millions of runners worldwide on a weekly basis.

These types of footwear not only have little effect on injury prevention, but alter the mechanics of ER. Kerrigan et al. (2009) reported the consequences of this type of footwear design on running technique when comparing CCRS to barefoot and reported a CCRS increased external-joint moments, increasing the stress placed on biological tissues. This demonstrated that CCRS increase lower-limb joint loads, however the CCRS shoe is not the only type of footwear available to runners, with the recent advent of minimal and maximally-cushioned

footwear. Minimal and maximally-cushioned footwear represent two types of footwear at the opposite end of the shoe-market spectrum, with the former permitting the natural movement of the foot, minimal support and improved sensory feedback, and the latter designed with structural support and extensive midsole cushioning compared to CCRS (Esculier et al., 2015; Sinclair et al., 2016).

2.7.3 Effect of minimal shoes on running kinematics and kinetics

Until recently the term minimal shoe was used without standardisation (Ryan, Elashi, Newsham-West, & Taunton, 2013), however, to develop and validate a rating scale that could be used to determine the degree of minimalism for a shoe, a standardised definition was required. Esculier et al. (2015) sought the opinions from forty two experts from eleven countries around the world, and with 95% agreement defined a minimal shoe as, "Footwear providing minimal interference with the natural movement of the foot due to its high flexibility, low heel to toe drop, weight and stack height, and the absence of motion control and stability devices". From this definition and previous discussions, a minimal shoe theoretically allows the foot to take advantage of several anatomical structures while simultaneously promoting an improved running technique.

A commonly reported kinematic adaptation to running in a minimal shoe is increased plantarflexion at initial contact (relative to conventional footwear) (Gruber et al., 2012; Squadrone et al., 2015). This change is proposed to be a response to the removal of the proprioceptive insulation provided by thick mid-soled footwear, and the subsequent detection of high impact forces underfoot when running with a normal running technique (De Wit et al., 2000; Gruber et al., 2012; Robbins et al., 1994). As previously discussed, the implications of such a change in running technique were highlighted by Daoud et al. (2012) who reported RFS runners incur approximately twice as many repetitive strain injuries as those who FFS. Furthermore, with factors such as shoes mass, strike type, stride frequency and running experience controlled Perl et al. (2012) were able to report runners using minimal footwear significantly improved their cost of transport compared to CCRS, citing the loading of

anatomical structures such as the longitudinal arch as the biomechanical mechanism for such improvements.

Conversely, the effects of minimal footwear on sagittal plane knee angles are not consistent, with some reporting a more flexed knee joint at initial contact (Squadrone et al., 2015; Willy & Davis, 2013) and others reporting no difference between minimal and CCRS (Sinclair, Greenhalgh, Brooks, Edmundson, & Hobbs, 2013; Squadrone & Gallozzi, 2009). Increased knee flexion at initial contact is relative to injury potential following this measure is a precursor for increased knee flexion at midstance, reduced effective mass, and potentially the length of the external moment arm when the GRF is greatest (Derrick, 2004). Additionally, Gerritsen, van den Bogert, and Nigg (1995) reported incereased knee flexion at initial contact reduced the GRF impact peak by 68N per degree of flexion, concluding alterations in initial contact kinematics dictate impact forces. This suggests minimal shoes can improve kinematics at the knee that are assosiated with joint loads, however, others suggest minimal footwear does not effect knee kinematics. A potential explanation for such inconsistencies in findings in minimal footwear research is habituation time. If participants have not been provided with sufficient time to habituate to minimal footwear, it is difficult to conclude whether a kinematic change in running technique, or alternatively lack of, is product of habituated running mechanics or a participant's initial response to a novel-footwear condition. For example, Moore and Dixon (2014) report that 20 minutes of treadmill running was necessary to habituate treadmill runners with no experience of barefoot running to barefoot treadmill running. Putting this in context, it is likely that the results of Sinclair, Greenhalgh, Brooks, et al. (2013) who provided five minutes to habituate to minimal footwear drew conclusions from running biomechanics that were an initial response to a novel-footwear condition. Investigations of the time necessary to produce stable and consistent ER biomechanics in novel footwear conditions while performing overground running do not exist and are necessary for future research using novel-footwear conditions. Time to habituation while running overground barefoot, in minimal shoes, and maximally-cushioned shoes is yet to be reported in overground running.

Minimal shoes have been reported to improve joint-specific loading patterns associated with injury. Sinclair (2014) investigated the effects of minimal footwear on knee and ankle loads, and reported minimal footwear was associated with reductions in patellofemoral kinetic parameters and sagittal knee joint moment, concluding this type of footwear might serve to reduce the incidence of injuries at the knee. Furthermore, in a comparison between minimal, conventional and maximally-cushioned footwear on patellofemoral joint kinetics, Sinclair et al. (2016) reported a significant reduction in patellofemoral force and pressure as well as patellofemoral force per mile when wearing minimal footwear. Specific to the knee joint moments, Firminger and Edwards (2016) investigate the individual contribution and combined effects of stride length and minimal footwear. Firminger and Edwards (2016) reported that wearing minimal footwear significantly reduced the sagittal knee joint load (P = 0.003), but not the peak-knee adduction moment (P = 0.544). Conversely, Bonacci et al. (2013) reported no significant difference between conventional footwear and minimal footwear for either sagittal or adduction peak moments. However, participants in Firminger and Edwards (2016) and Bonacci et al. (2013) were new to minimal footwear and likely did not provide a habituated representation of overground minimal-shoe ER biomechanics.

While there has been much focus on the effects of footwear choice on surrogate measures of running injury, there is a lack of consensus in research findings as to the effects of minimal footwear on running injury rates. Based on previous findings that running in a minimal shoe increases plantarflexion and knee flexion, and reduces kinetic measures associated with injury, or in other cases shows no significant difference (Sinclair, 2014; Squadrone & Gallozzi, 2009; Squadrone et al., 2015), it is logical that minimal shoes might reduce injury rates. Ryan et al. (2013) examined the effect of progressive minimal-footwear exposure on injury propensity. Participants were novel to the minimal design and were assigned to one of three groups: neutral, partial minimal and full minimal whilst wearing Nike Pegasus 28, Nike Free 3.0 or Vibram 5-Finger Bikila, respectively. After a 12 week progressive training plan incrementally increasing from 20%-58% of experimental shoe exposure, 23 injuries were reported. Results showed the partial-minimal condition experienced three times the number of injuries as the

neutral group and there were five more injuries in the partially-minimal group than the fully-minimal group. This suggests minimal footwear exposes a runner to a greater risk of injury than neutral footwear and partially-minimal were more dangerous than fully-minimal shoes. This supports the opinion of this thesis that sensory insulation might lead to overloading of anatomical structures and when combined with a lack of structural support (partially-minimal) this might increase injury prevalence. However, this also suggests running in minimal footwear increases injury risk relative to neutral footwear. A potential explanation might be that exposure time (12 weeks) was too acute for the necessary structural adaptations to occur that can accommodate subsequent changes in anatomical loads. A training plan spread over a greater amount of time might yield different results. Furthermore, whilst injury rates were of interest, biomechanical differences between injured and uninjured groups would have helped explain findings following increased joint loads are often related to injury mechanisms.

An investigation comparing the effects of minimal shoes to other available footwear, in a sample habituated to minimal shoe runners investigating lower-limb 3-D kinematics and kinetics has not been studied. Such an investigation is important following previous work that identifies increased knee-joint moments to be associated with injured runners (Dudley et al., 2017).

2.7.4 Effect of maximally-cushioned footwear on running kinematics and kinetics

In opposition to the barefoot/minimal concept, maximally-cushioned footwear has recently entered the running-shoe market. Unlike minimal shoes, there is no established consensus as to what is a maximally-cushioned shoe. This is likely because research into maximally-cushioned footwear is in its infancy, and as such, a definition is yet to be determined. Of the available research, features consistently associated with a maximally-cushioned shoe are extensive cushioning (>20mm) in the midsole and a minimal heel-toe drop (Agresta et al., 2018). Currently, maximally-cushioned footwear manufacturers promote this footwear to increase comfort and attenuate impact (Sinclair et al., 2016), despite a lack of research to support these claims. The effects of maximally-cushioned shoes on running gait compared to other available footwear options warrant investigation.

As previously discussed, barefoot running and minimal shoe running are reported to increase plantarflexion compared to CCRS, with the sensation of high forces underfoot suggested as a biomechanical explanation for the change from RFS to FFS (De Wit et al., 2000; Gruber et al., 2012). Previous work has reported midsole cushioning insulates somatosensory feedback and the detection of peak forces underfoot influences foot-strike strategy (Gruber et al., 2012; Robbins et al., 1994). Following, it would be expected that extensive cushioning of maximally-cushioned footwear would further insulate somatosensory feedback, and in line with previous discussions addressing the biological imperative, dorsiflexion at initial contact would increase. However, Sinclair et al. (2016) reported no significant difference between conventional footwear (4.83°) and maximally-cushioned footwear (4.78°). A possible explanation being a lack of time provided for habituation to a novel footwear condition, underlining the need for research investigating habituation times. Additionally, Agresta et al. (2018) reported increased plantarflexion at initial contact for minimal footwear (-7.7° ± 5.7) and maximally-cushioned footwear (-4.2° ± 5.1) compared to participants' native footwear during an initial 10 minute run in each footwear condition. Moreover, there were no differences observed post four weeks of exclusively running in this footwear, suggesting participants' running technique was stable. Authors suggest that trends to increase plantarflexion might be explained by shoe design, citing work by Horvais and Samozino (2013) that reports foot strike angle was associated with heel height and heel-to-toe drop across a range of speeds (r = 0.542 – 0.695). However, with no reported characteristics for native footwear, explanations for this trend are difficult. Furthermore, this study was conducted on a treadmill, a testing condition known to change lower-limb biomechanics, thus these findings should be interpreted with caution (Nigg, De Boer, & Fisher, 1995). Based on previous discussions on sensory insulation induced by extensive cushioning it seems logical that increased dorsiflexion would be observed when wearing maximally-cushioned footwear, however, current reports are conflicting suggest a more plantarflexed foot or no difference. This contrast between theory and previous work that is limited by study methods underline the need for research using habituated participants and knowledge of all footwear conditions to establish a consensus in the emerging field of maximally-cushioned footwear biomechanics.

The effects of maximally-cushioned-footwear on knee kinematics have not been explored in detail. Sinclair et al. (2016) compared minimal, CCRS, and maximally-cushioned footwear, and reported after five minutes of running in minimal and maximally-cushioned footwear, a sample new to footwear conditions reported main effects for peak flexion and ROM. Authors reported peak knee flexion and ROM were significantly larger in maximally cushioned and CCRS compared to minimal, concluding differences in peak-knee flexion and therefore ROM were likely a result of additional energy being absorbed at the knee in a RFS gait. However, this conclusion is limited to biomechanical results that were likely a representation of novel responses to novel footwear conditions. If the most distinguishing feature of maximally-cushioned footwear is the excessive cushioning in the midsole then the findings of studies investigating the effect of conventionally cushioned footwear compared to minimal footwear could tentatively be extrapolated to suggest the effects of maximally-cushioned footwear. It has been reported that as midsole thickness increases from a minimal shoe to a CCRS knee extension increases at initial contact (Squadrone et al., 2015; Willy & Davis, 2013). These kinematics adaptations are noted as important given that increased knee extension at initial contact is associated with increased extension at midstance and increased effective mass (Derrick, 2004). Investigations into the effects of maximally-cushioned footwear on 3-D kinematics have not been fully explored in recreational runners habituated to maximally-cushioned footwear. Investigations of this nature providing a comparison to a variety of other footwear conditions are necessary to understand the effects of maximally-cushioned footwear on lower-limb joint kinematics.

Maximally-cushioned footwear has been reported to worsen lower-limb joint kinetics that are associated with injury etiology. To date, maximally-cushioned footwear research has focused on patellofemoral joint kinetics (Sinclair et al., 2016), impact shock attenuation (Sinclair, 2017), Achilles tendon load (Sinclair, Richards, & Shore, 2015) and GRF data (Sinclair, Greenhalgh, Brooks, et al., 2013). Sinclair et al. (2016) investigated the influence of maximally-cushioned footwear on patellofemoral kinetics when running overground and reported patellofemoral joint peak contact force and peak pressure were significantly greater

in the maximally-cushioned and CCRS than minimal footwear. Sinclair et al. (2016) concluded as patellofemoral peak force and pressure are associated with patellofemoral injury etiology, a minimal shoe might reduce a runner's propensity for patellofemoral injuries. Sinclair (2017) also investigated the effects of minimal, maximally-cushioned and CCRS on impact shock attenuation during running by measuring tibial and sacral accelerations. Sinclair (2017) reported peak-tibial accelerations and shock attenuation were significantly less in minimal footwear compared to CCRS and maximally-cushioned footwear. This suggests that wearing maximally-cushioned footwear placed increased demands on biological tissues by increasing the magnitude of impact therefore increasing the likelihood of stress related lower-limb injuries. There is also a trend for maximally-cushioned footwear to shift loading from the ankle to the knee. Sinclair et al. (2015) reported the effects of minimal, conventional and maximally-cushioned footwear on Achilles tendon load and reported running in minimal footwear increased Achilles tendon force per step compared to maximally-cushioned footwear, and per mile compared to both maximally-cushioned and CCRS. However, this study did not report the patella tendon force. A reduction in patella tendon force would be expected following previous work by Sinclair (2014) reporting increased midsole cushioning (barefoot compared to minimal), increased Achilles tendon force, but decreased patella tendon force and knee flexion moment. Following, it can be argued that maximally-cushioned footwear would increase knee joint loads and therefore the potential for injury at the knee. Whilst this investigation investigated sagittal plane joint moments, research is yet to investigate the effects of this footwear on non-sagittal knee joint moments. This is important given that increased peak-knee adduction moment is characteristic of injured runners (Dudley et al., 2017).

2.7.5 The effects of footwear on spatiotemporal variables

Following ER changes in kinematics and kinetics, footwear also changes spatiotemporal variables. Reduced stride length is associated with improved loading patterns associated with injury (Edwards et al., 2009; Firminger & Edwards, 2016; Heiderscheit et al., 2011; Schubert et al., 2014). In a recent systematic review, Schubert et al. (2014) investigated the effects of

stride frequency manipulations and therefore stride length manipulation when speed was held constant. Authors reported reduced GRF, reduced energy absorbed at the hip, knee, and ankle joints and improved shock attenuation when stride length was reduced. Corroborating these findings, Heiderscheit et al. (2011) reported a 5 and 10% percent increase in stride frequency, when running at a constant speed reduced mechanical energy absorbed at the knee, while the hip was only significant at 10%. Following these reported benefits of increased stride frequency and reduced stride length, barefoot running and minimal footwear are often used as a mean to improve running technique and joint loading patterns.

Kerrigan et al. (2009) investigated the effects of barefoot running in sample of recreational runners, but did not report barefoot running experience. Results showed the sample significantly reduced their stride length ($P = 0.001$) when transitioning from shod (2.29 m) to barefoot (2.15 m), concluding change in stride length was likely driven by shoe characteristics that promote comfort. Additionally, Bonacci et al. (2014) reported running barefoot, in a minimal shoe or a racing flat significantly reduced stride length compared to a CCRS, and when investigating a range of different minimal footwear designs (compared to a CCRS) Squadrone et al. (2015) showed a trend for increasing stride length as recreational runners transitioned from barefoot to minimal to conventional. The latter finding suggests that as cushioning increased so did stride length. This is particularly important with the recent advent of maximally-cushioned shoes. Following increased cushioning and somatosensory insulation increasing stride length, it could be hypothesised that maximally-cushioned footwear might increase stride length and increase lower-limb loading patterns associated with injury.

Stride length can be altered by manipulating flight time or overstride. The latter defined as the anterior projection of the lower-limb relative to the COM or the hip joint centre at initial contact (Heiderscheit et al., 2011; Lieberman, Warrener, et al., 2015). However, Lieberman, Warrener, et al. (2015) argue the COM is dependent on trunk lean, and measuring from the hip joint centre, a measure close to the COM, provides an improved and consistent representation of overstride. There are a variety of ways a participant can overstride yet have an identical stride length. For example, a runner could flex both their hip and knee joint in one

lower-limb configuration, yet reduce their hip flexion and increase their knee extension in the next trial and produce an identical stride length. This highlights the importance of quantifying the position of both the hip and knee relative to the ankle joint at initial contact.

The biological imperative provides a possible explanation for the potential underlining effects that footwear have on the alterations on overstride and therefore observed stride length when comparing footwear when running. The subconscious drive to minimise energy cost of movement is well documented (Alexander, 1989; Sparrow, 2000). In the context of locomotion, Kram and Taylor (1990) have shown that the cost of running is inversely related to ground-contact time. It therefore follows that it is energetically favourable to cover a given distance with a longer ground-contact time, which is facilitated by an increased overstride, increasing stride length. However, while least energetically costly, this movement strategy might not be the least injurious. This is because excessive midsole cushioning and the associated sensory insulation encourage increased overstride. The cushioned heel insulates the mechanoreceptors at the calcaneus from the true forces acting upon the foot, and therefore allow runners to perceive an increased overstride as safe (Robbins & Gouw, 1991; Robbins et al., 1994). Furthermore, this sensory insulation and subsequent increase in overstride is associated with a RFS, as the foot is now not falling below the knee, but is extended anterior to the knee. It is this over projection which will also likely cause an increase in stiffness of joints on contact (Lieberman, Warrener, et al., 2015). This theoretically suggests that shoe design influences overstride and that overstride could plausibly increase the potential for injury.

Investigations into the effects of overstride are in their infancy and currently have only been investigated in the sagittal plane. Previous work by Heiderscheit et al. (2011) was first to investigate overstride, reporting increased braking impulse as overstride increased. However, overstride was measured relative to the COM, a measure easily influenced by trunk lean. More recently, Lieberman, Warrener, et al. (2015) demonstrated that overstride relative to the hip and knee was associated with increased posteriorly directed braking force and the vertical component of the GRF impact peak, respectively, and that overstride and the resulting forces

are reduced with increased stride frequency. However, these forces and their relationship with overstride are yet to be investigated in three dimensions for a runner's lower joints. The influence of other elements of technique (such as posture) on overstride, are also yet to be determined. Similarly, the influence of footwear on the consequences of overstride is also yet to be investigated.

2.7.6 The effects of trunk lean on running technique

The virtual pivot point (VPP) model is a proposed control mechanism that uses limb position to provide dynamic stability in human locomotion (Maus et al., 2010; Maus, Rummel, & Seyfarth, 2008). This is achieved by consistently directing the GRF through a single-virtual point in the body referred to as the VPP, and by doing so, applying external moments to the lower-limb joints to maintain dynamic stability (Maus et al., 2010). This effectively allows the VVP to act as a simulated-sagittal hinge from which the torso rotates. This converts the conceptually difficult task of maintaining dynamic stability of an inverted pendulum to a less complex one, in which the human system rotates around a point that is intrinsically stable (Maus et al., 2010; Van Bommel, 2011).

It is this observation that the GRF acts through a VPP located superior to the COM, and the understanding that external moments are calculated as the product of the magnitude of the GRF, and the perpendicular distance between the GRF vector and the joint centre that suggests trunk lean as a mechanism to increase joint loading (Maus et al., 2010; Simonsen, Dyhre-Poulsen, Voigt, Aagaard, & Fallentins, 1997). Because an increased trunk lean projects the COM more anteriorly, in an attempt to maintain dynamic stability while running, and to prevent falling, the step reflex strategy would increase overstride in an attempt to increase the dynamic base of support (Horak & Nashner, 1986). As a result of this increase in overstride, and the observation that the GRF acts through a VPP superior to the COM, the VPP model predicts that this strategy would increase the perpendicular distance between the hip joint centre and the GRF vector, and therefore increase the internal-hip extension moment required during weight acceptance. However, despite this understanding, some research still advocates trunk lean as means to run safely (Teng & Powers, 2014).

Teng and Powers (2014) prescribe athletes to run with a flexed trunk, utilising trunk lean as a means to reduce knee-joint loading without shifting demands to the ankle joint. However, Teng and Powers (2014) report a simultaneous and significant increase in the hip-joint power when running with increased trunk lean. The latter observation makes sense and supports the VPP model, in that an increased trunk lean will increase hip extension demands during ER by inducing a greater step reflex strategy in the form of an overstride (Horak & Nashner, 1986). Preece, Mason and Bramah Preece, Mason, and Bramah (2016) investigated the relationship between trunk lean and running velocity in both recreational and elite runners. Results demonstrated recreational runners increase trunk inclination by approximately one degree for every increase of $1 m \cdot s^{-1}$; whereas elite runners demonstrated no trend. Authors argued a possible consequence of trunk lean is that runners might place their foot in front of their body (overstride) in an attempt to achieve equal anterior-posterior distance between COM and COP. However, it is more likely that an increase in stride length will be a step reflex, given the well establish relationship between perturbed-dynamic balance (in this case trunk lean) and the step reflex (Horak & Nashner, 1986). It can also be implied that a flexed running posture is associated with poorer running performance, given that recreational runners with poorer 10km run times (recreational: 43 ± 3; elite: 32 ± 2 minutes) employed the flexed posture. This observation is supported by Lieberman, Warrener, et al. (2015) who demonstrated that overstride, a variable theoretically linked to trunk lean increased braking forces and thus cost of locomotion. This time difference provides further support to the VPP model, as a flexed posture will increase overstride, and thus the distance between the hip-joint centre and the GRF increasing joint loading. This is an important observation given that increased loading at joints can increase metabolic cost of travel and increases injury risk. An increase in the external hip joint moment is suggested to be associated with injury, as increased external peak moments and loading rates are linked to micro trauma in bones and other biological tissue (Barr & Barbe, 2002; Burr et al., 1998; Lieberman, 2012b). This theoretically suggests that trunk lean and overstride could increase injury risk. These variables and their influence on joint loading, and in conjunction with different types of footwear conditions are yet to be fully examined.

2.8 Gait retraining to improve joint loading

Gait retraining strategies designed to improve running performance or reduce the likelihood of injury are not novel in sports research, with studies primarily focusing on lower-limb spatiotemporal variables (Dallam, Wilber, Jadelis, Fletcher, & Romanov, 2005; Edwards et al., 2009; Fletcher, Bartlett, & Romanov, 2010). A reduced stride length, increased stride frequency and subsequently reduced t_c alter both kinematic and kinetic variables associated with performance and injury (Bowersock, Willy, DeVita, & Willson, 2016; Edwards et al., 2010; Heiderscheit et al., 2011). Bowersock et al. (2016) investigated the effects of varying step length (and subsequently overstride), as well as the effect of foot strike patterns on the kinetics of the tibiofemoral joint (TFJ) and the medial compartment of the TFJ, an area where degenerative disease is commonly observed. While running at a constant speed, Bowersock et al. (2016) and colleagues reported, regardless of foot-strike pattern, a reduced step length (-10%) equated to a reduced peak contact force, force impulse per step/kilometre and a reduced LR over the entire TFJ and medial compartment. Conversely, when the step length was increased (+10%) all kinetic variables increased ($P < 0.05$). This demonstrating the potential of stride length and theoretically overstride to decrease the likelihood of injury.

Warne et al. (2016) also investigated the influence of gait retraining, with the primary instructions to reduce stride length and increase stride frequency, while simultaneously running 'quietly as possible'. Warne et al. (2016), allocated participants into one of two intervention groups (n = 12), both received gait retraining, but only one used minimal footwear. Participants undertook six weeks of progressive gait-retraining, reporting a reduced LR. Post-hoc analysis revealed a non-significant reduction (18%) in CCRS group, and a significant (33%) reduction in LR when employing minimal shoes; however, they did not conduct a priori power calculation therefore the non-significant difference in CCRS might be a product of sample size. This finding suggests that reduced stride length reduces LR, and minimal shoes have an additive effect. The effects of reducing stride length are further supported by Edwards et al. (2009) who used musculoskeletal and finite element modelling techniques to show that reducing stride length by 10% reduced the likelihood of probabilistic

stress fracture between 3 to 6%. A reported decrease in biological stress is important as *in vitro* tests show cartilage loaded with large force promotes chondrocyte cell death, as well as deleterious-structural changes, such as, wide and long cartilage lesions (Bowersock et al., 2016). Collectively, these findings highlight minimal shoes and reduced stride length can potentially reduce the likelihood of injury in gait retraining

Heiderscheit et al. (2011) investigated the influence of manipulating lower-limb-gait mechanics, but examined the influence of stride frequency on energy-absorption demands in the lower-limb joints. They recruited 45 participants and recorded their running mechanics across a range of stride frequencies at their preferred-running speed. Stride frequencies ranged from -10% to +10% of preferred stride frequency and increased in 5% increments. Corroborating the results of Bowersock et al. (2016), Heiderscheit et al. (2011) demonstrated that increasing the stride frequency (subsequently reducing the stride length) resulted in improved-joint kinetics. Furthermore, Heiderscheit et al. (2011) also reported that increased overstride was associated with increased braking impulse, suggesting a greater demand would be placed on muscles in the latter half of stance to maintain a constant speed. However, any further analysis in relation to overstride was not reported. Importantly, this demonstrates the potential for overstride manipulation to reduce the kinetic demands placed on lower-limb joints, and thus reduce the strain applied to biological tissue. To date little is known about the effects of overstride on lower-limb joint loads and warrants further investigation. Remembering overstride underpins stride length, previous work suggests footwear influences stride length, and increased lower-limb moments, in particular increased peak-knee adduction moment are associated with injured runners, future coaching interventions should investigate overstride.

While spatiotemporal factors provide insight into injury mechanisms, it is important to note that few studies address the global technique of running. The control of the upper body has received little attention, and investigations of this nature are sparse. A global approach to gait retraining addressing variables such as trunk lean as well as lower-limb mechanics seems

logical given the strong relationship which exists between both trunk lean and the step reflex in the form of an overstride (Horak & Nashner, 1986).

Trunk lean shares an inherent relationship with stride length because of the reflex to increase overstride and improve dynamic stability in response to a anteriorly projected COM with a flexed trunk (Horak & Nashner, 1986). Following the principles of the VPP model, a flexed trunk will change how the GRF acts upon the body by changing the perpendicular distance from joint centres to the GRF, and thus changing lower-limb joint loading. Based on this theoretical understanding, increased trunk lean augments internal-joint moments, and could increase metabolic costs. Conversely, it can also be argued that a reduced trunk lean could improve performance due to the attenuated metabolic demands. This suggestion is supported by data presented by Preece et al. (2016) who observed that elite runners with faster 10Km run times do not increase trunk lean as a function of speed; however, recreational runners with slower 10 km time trials increase trunk lean by about one degree for every 1 $m \cdot s^{-1}$ increase in speed. This finding gives strength to the argument that an upright posture improves performance and reduces injury potential as a result of reduced metabolic cost and joint loading, respectively. Importantly, this demonstrates that trunk lean can differentiate between elite and recreation runners, and that the influence and manipulation of trunk lean warrants further investigation. To date little is known about trunk lean, posture and its effect on individual joints in three dimensions, therefore future research should investigate the effects of trunk lean on individual-joint loads, especially given the reflexive relationship it shares with overstride and the established effects this has on injury potential (Edwards et al., 2009).

Coaching strategies that encourage runners to reduce stride length, increase stride frequency and subsequently reduce t_c have the potential to reduce injury by reducing the magnitude loading patterns associated with injury. However, there is an apparent gap in the literature which lacks examination of the role of trunk lean and overstride in running. Of the few studies available, the role of trunk lean is equivocal with some authors prescribing a flexed trunk, and others not. Based on the strong link between overstride and trunk lean and basic principles of engineering and physics, it appears a reduced trunk lean and overstride have the potential to

reduce joint loading patterns associated with injury. The influence of a coaching intervention on trunk lean and overstride is yet to be fully explored.

2.9 Main findings and conclusions

This review suggests that humans are ER specialists who possess a series of traits retained through selective pressure for ER success. However, injury rates range anywhere between 20-79%. This review proposes that when used appropriately, the foot and lower-limb structure is well adapted to the natural forces that act upon it, therefore injury rates should be low; however, if used incorrectly, or functionally constrained, biological tissue and structures can succumb to the high forces experienced during ER. The relationship between foot structure, forefoot kinematics, footwear, running technique is not simple and it is important to remember that the context of the running environment is also important with protection from the urban environment a necessity for the human foot.

Human foot structure is inherited from our primate ancestors and is not designed to deal with high plantar surface loads. Following this, footwear has long been worn by our ancestors to protect their feet from the variety of harsh environments. In contrast to the simple footwear used by our ancestors that allowed the foot to function naturally, recent times have seen an influx of footwear that inhibit the natural development of the foot from a young age. Design features such as constrained toe-box, toe-spring, and arch support have all been reported to inhibit natural foot development. These restrictive and compressive forces applied to the foot during stance are reported to compromise natural-foot development, foot function and therefore the capacity of structures such as the hallux to control the GRF in static balance tasks. Minimal shoes might provide a solution to this problem if the design is devoid of these compressive and restrictive forces. The consequences of foot structure and footwear choice on forefoot kinematics and joint loading patterns associated with injury is yet to be fully explored in ER.

The review also discussed the potential effects of footwear on running technique. A thick-cushioned sole provides sensory insulation that masks feedback to mechanoreceptors. This

feedback was reported as essential to inform impact-avoidance strategies. Furthermore, with research reporting humans will employ a running strategy that both minimises energy expenditure and injury risk, it follows that participants will overstride in response to increasing midsole cushioning. This is particularly important with the introduction of maximally-cushioned footwear to the running shoe market, characterised by excessive cushioning. However, research to date is complicated by a lack of investigations addressing habituation times in novel footwear conditions, and drawing conclusions from runners potentially unhabituated to novel shoe types. This highlights the need for research in this area. Increased overstride is associated with decreased stride frequency and increased trunk lean and increased GRF components. An increased GRF might increase peak-knee adduction moment following moments are calculated as the function of the GRF and the perpendicular distance from the respective joint centre, a loading pattern associated with injured overground runners. This provides a rationale for minimal footwear that allows some sensory feedback, yet protects the foot from the modern environment. Evidence presented in this review suggests that excessively-cushioned footwear has the potential to increase trunk lean and overstride. The consequences of which are yet to be investigated with 3-D analysis.

Future research should focus on studies that establish the following:

1. Time to habituation in novel footwear conditions
2. The association between foot structure and forefoot kinematics.
3. The effect of footwear on forefoot kinematics.
4. The association between forefoot kinematics and peak-knee adduction moment.
5. The effects of footwear on overstride and how this relates to the loading of the knee joint.
6. Interventions that aim to reduce the peak-knee adduction moment.

2.9.1 General aim

The aim of this thesis is to investigate select relationships between the following variables: foot structure, forefoot kinematics, running technique and peak-knee adduction moment. The

specific relationships of interest are outlined in section 2.9.2 'Specific study aims'. Additionally, time to habituation in novel footwear conditions will be investigated. Within these studies the effects of different types of footwear condition (barefoot, minimal shoe and maximally-cushioned footwear) on habituation, forefoot kinematics, running technique and lower-limb joint loads will also be investigated.

2.9.2 Specific study aims

1. To examine the reliability of gait laboratory measures both within-day and between-days to inform sample size and natural variability for proceeding chapters in the thesis.
2. To report the time to habituation in a sample of recreational overground runners' novel to barefoot running, minimal and maximally-cushioned-shoe running.
3. To investigate the association between forefoot structure and forefoot pronation, and how this relates to peak-knee adduction moment during overground running when barefoot, in minimal and maximally-cushioned shoes.
4. To explore the effects of barefoot running, minimal and maximally-cushioned shoe running on lower-limb kinematics and kinetics in a sample of recreational overground runners post a 30-minute habituation run.
5. To investigate the effects of barefoot, minimal and maximally-cushioned running shoes on overstride and its relationship with peak-knee adduction moment during overground endurance running in recreational runners.
6. To examine the effects of an acute posture focused intervention on peak-knee adduction moment in a sample of recreational overground endurance runners.

3.0 General Methods

This chapter provides details of the methods used for system set-up, test protocols, and data processing used to generate outcome measures in the thesis. Chapter specific statistical approaches and methods are discussed in detail in the relevant chapters that follow. The term "clear difference" defines a comparison between variables where a 90% confidence interval does not contain 0.

This thesis derived findings from three dimensional analyses with floor embedded force plates to calculate spatiotemporal, kinematic and kinetic variables. This system was used in all experimental chapters.

3.1 Ethical approval and location

Approval for studies described in this thesis was sought from and granted by the Faculty of Health and Life Sciences Ethics Committee at Northumbria University (see appendix A, B and C).

3.2 Participant enrolment to studies

The flow diagram below illustrates the 37 participants enrolled into the three experimental investigations that encompass this thesis.

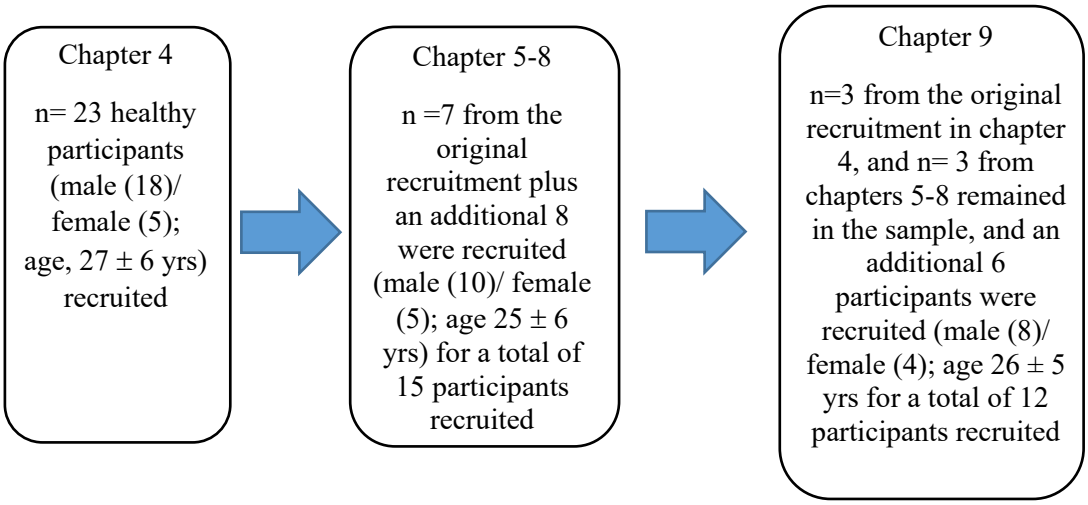

Figure 3. 1 Diagram to visualise participant cross over from chapters four through to nine.

3.3 Three-dimensional analysis methods

3.3.1 Three-dimensional gait laboratory calibration

The biomechanical gait analysis suite used in this thesis comprised fourteen 3-D infra-red motion analysis cameras (12 x T20 cameras and 2 x T40 cameras) (Vicon MX, Oxford, UK). These cameras were calibrated following a standardised protocol, with a five-marker calibration wand (Vicon, Oxford, UK). The calibration was deemed to be accurate when all cameras (Vicon T20/40, Oxford, UK) produced an image error ≤ 0.2mm. The origin of the cameras was set once the wand had been placed in a predetermined position representing the centre of the volume. This allowed cameras to deduce their orientation for that particular session. Kinematic data was set to record directly into Vicon Nexus software (version 1.7) at 200Hz.

A total of four floor embedded force plates (OR6-7, AMTI, Watertown MA, USA) (width = 464mm; length =508mm; depth = 82.6mm) located within a 2.7 m by 0.93 m floor space were used for kinetic data collection. Force plates were connected to an amplifier (MiniAmp MSA-6, AMTI, Watertown MA, USA), which amplified force with a gain of 1000. Amplified signals from the force plates were connected to one of two available Vicon MX Giganet core processing units (Vicon, Oxford, UK) by way of a patch box. According to manufacturers, the force plates had a linearity of ±0.2% and a stated hysteresis of ±0.2%. Kinetic data were captured at 1000Hz.

3.3.2 Participant preparation for three dimensional analyses

Participants were instructed to arrive to the testing session well rested. Upon arrival participants were asked to provide informed consent, and provided with an opportunity to ask questions pertaining to the described procedures. Unless stated otherwise, participants were provided with footwear that corresponded to a specific testing session (barefoot/ minimal shoe/ maximally-cushioned shoe). Participants were then provided with compression clothing to improve biomechanical representation when running. For repeated measures on separate days, participants were requested to arrive well rested and at a similar time of day.

The ‘Plug-In Gait’ and ‘Oxford-Foot Model’ were used to derive spatiotemporal, kinematic and kinetic data of a participant’s dynamic trials. For the models to be representative of a participant, a series of anthropometric measurements were taken and associated with the model. These measures were in accordance with the recommendations of the ‘Plug-in Gait’ foundation notes (Vicon, 2010) and are as follows:

Body Mass

The mass of all participants was measured to the nearest 0.1 kg before the commencement of a testing session using a balance-beam scale (SECA, Vogel & Halke GmbH, Hamburg, Germany; precision of 0.1 kg). Before each testing session the scales were set to absolute 0 and a turn screw was finely adjusted until the scales were as near perfectly balanced as possible. For a participant’s weight to be recorded accurately and without fluctuation, they were instructed to remove their footwear and stand as still as possible on the weighing platform while maintaining a neutral stance. The scales were then adjusted to bring them into balance and a reading was recorded.

Stature

Stretch stature was assessed using a Holtain stadiometer (Holtain Ltd., Crymych, Wales) following the guidelines outlined by the International Society for the Advancement of Kinanthropometry (Marfell-Jones, Stewart, & de Ridder, 2001). Prior to assessing a participant’s stature, the accuracy of the stadiometer was checked using a one-meter rule and was altered where necessary. Participants were instructed to remove shoes and stand with their feet and heels together, while having the upper part of their back and buttocks touch the scale. The head was positioned so it conformed to the Frankfort position (when the orbitale is in the same horizontal plane as the tragion). Once in this position, participants were asked to inhale while the headboard was adjusted to their height. The board was adjusted (compressing the hair) so that it made contact with the vertex of the skull. From this position a participant’s height was recorded.

Leg length

Leg length represents the line from the anterior superior iliac spine (ASIS) to the medial malleolus, via the knee joint. In line with the recommendations of the Vicon system (Vicon, 2010), this measurement was taken while standing and bearing weight in the anatomical position. Leg length was recorded to the nearest millimetre using a measuring tape (SECA 201, Birmingham, UK).

Knee width

Knee width was measured in mm and defined as the medio-lateral width of the knee across the line of the knee axis. Similarly to leg length, this measure was recorded while standing and bearing weight in the anatomical position. Measurements were recorded using manual-anatomical callipers (Bicondylar Caliper, Holtain, Crymych, UK).

Ankle width

Ankle width was measured in mm and defined as the medio-lateral width across the malleoli. Similar to previous measures, this was performed while bearing weight in the anatomical position. Measurements were recorded using manual-anatomical callipers (Bicondylar Caliper, Holtain, Crymych, UK).

Depending on the model used, a series of retroflective markers (Ø=14mm), including four wand markers with lateral stems of (80mm) for the thigh and lower leg were attached over predetermined anatomical landmarks according to manufacturer recommendations (Vicon, 2010, 2012). When wearing footwear markers were applied to a point on the shoe that best approximated the underlying anatomical location of the foot. The locations names and definitions are as follows:

Upper body

- The seventh cervical vertebrae (C7) was identified by its distinctively long and spinous process. When not directly observable a participant was asked to flex their neck to exaggerate its prominence.

- The tenth thoracic vertebrae (T10) was identified by first identifying T12 that was identified as the most inferior spinous process attached to the ribs and is typically smaller than L1. From here, two spinous process were counted superiorly to identify T10.
- The clavicle marker (CLAV) was defined as the jugular notch where the clavicles meet the sternum.
- The sternum marker (STRN) was located upon the Xiphoid process of the sternum.
- To identify marker placement for the shoulder (L/RSHO), the joint was palpated to identify the acromio-clavicular joint. Markers were attached bilaterally.
- A single marker was then placed over the right shoulder blade (RSHOLD) to assist auto-label identification of both right and left sides.

Lower Body

- Two markers were placed upon the anterior superior iliac spines (L/RASI).
- Two markers were placed upon the posterior superior iliac spines. These were inferior to the position of the sacro-iliac joints (L/RPSI).
- The left thigh wand marker was placed upon the left-distal-lateral aspect of the lower third of the thigh so that it would not impede arm swing. The placement of the wand marker was in line with the hip and knee joint centres as well as the knee flexion/extension axis and helped to define rotation (LTHI). Similarly, the right thigh wand marker was also placed in line with the hip and the knee joint centres and flexion extension axis, but was placed at the proximal-lateral aspect of the lower third of the thigh (RTHI).
- To accurately identify the position for the lateral-epicondyle knee markers (L/RKNE), participants were asked to flex and extend their knee whilst sitting. The movement of the skin over the lateral aspect of the knee was examined to identify an area of minimal skin movement. This position represented flexion/extension axis and the location for marker attachment.

- The second set of wand markers were placed upon the lower leg. The left lower leg wand marker was placed upon the lateral-distal aspect of the left lower leg (LTIB). The right lower leg wand marker was placed over the proximal-lateral aspect of the lower third of the lower leg (RTIB). The placement of the marker was in line with the ankle and knee joint centres and the ankle plantarflexion/dorsiflexion axis. The anterior/posterior positioning of each of the lower leg wand markers reflected the external rotation of the lower leg with respect to the knee flexion axes when standing in a neutral-anatomical position.
- Markers were placed upon the lateral aspect of each malleoli, which corresponded to an imaginary line acting through the transmalleolar axis (L/RANK).
- Markers were then placed over the left and right second metatarsal heads (L/RTOE), on the mid-foot side of the equinus break between the forefoot and midfoot.
- Markers were placed upon the posterior aspect of the calcaneus (L/RHEE).

Oxford-Foot Model (OFM) location

- To identify the location of the Sacrum (SACR), a single marker was placed mid-way between both the LPSI and RPSI.
- To identify the lateral head of the fibula (HFB), skin inferior to the location of the lateral aspect of the knee joint was palpated both inferiorly and superiorly on the dominant limb.
- To identify the tibial tuberosity (TUB) participants were asked to slowly flex and extend their knee while the patella tendon was palpated to identify the tibial tuberosity.
- Approximately mid lower leg, a marker was placed over the anterior aspect of the anterior crest of the shin (SHN).
- When using the OFM, the heel marker (HEE) of the dominant limb was placed as inferior as possible on a line that bisected the calcaneus, but not so that it would be dislodged during ER.

- A calcaneus peg-marker (CPG) was then attached superior to the heel marker and on a line bisecting the calcaneus. The orientation of this peg-marker reflected the varus/valgus orientation of the heel.
- For static trials a marker was placed proximal to the heel and in line with the peg marker bisecting the calcaneus (PCA). This could be removed post-static trial.
- A marker was placed over the medial malleoli (MMA) that corresponded to an imaginary line that acts through the transmalleolar axis.
- A line was traced inferior from the medial malleoli until a small ridge was palpated that identified the anatomical location for the sustaniculum Tali (STL).
- Following the identification of the STL, a marker was placed equidistant on the lateral boarder of the hind foot representing the lateral calcaneus (LCA).
- Palpating the extensor hallucis longus (EHL) tendon while participants dorsiflexed their toes identified the proximal-dorsal aspect of the first metatarsal (P1M). The marker was placed medial to the EHL.
- After the identification of P1M, distally palpating the first metatarsal identified the distal and medial aspect of the first metatarsal (D1M).
- On the lateral border of the foot, the lateral-distal fifth metatarsal head was located (D5M).
- Tracing the lateral border of the fifth metatarsal posteriorly, the lateral-proximal metatarsal head was identified (P5M).
- The hallux marker (HLX) was defined as the distal medial aspect of the first medial phalanx.

Once anthropometric measures had been associated with a participant's model in Vicon Nexus and marker attachment was complete, a static trial with a minimum of 600 frames was captured. Figures 3.1 and 3.2 depict the described anatomical locations for marker placement.

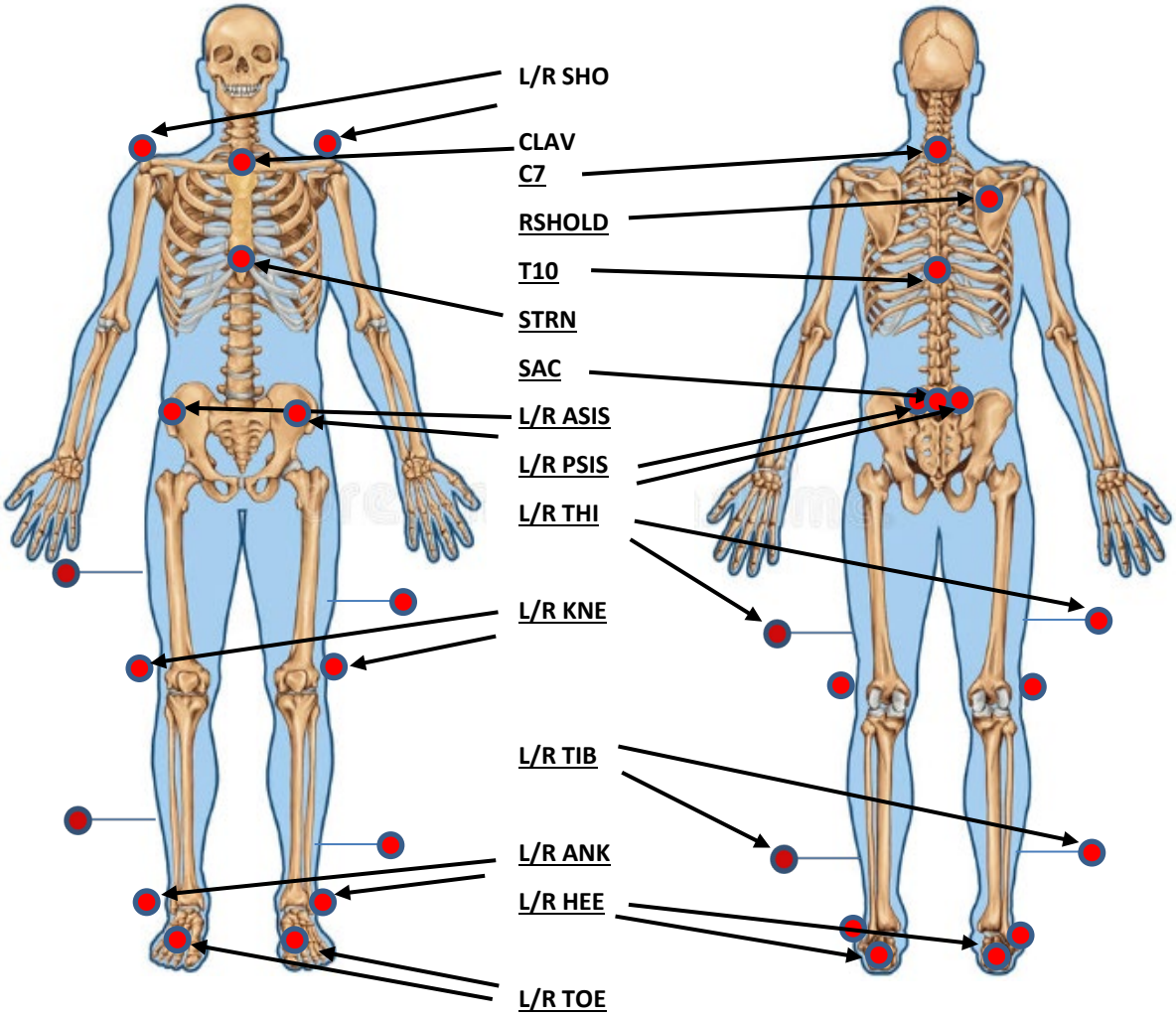

Figure 3. 2 Marker placement for the 'Plug-In Gait' model.

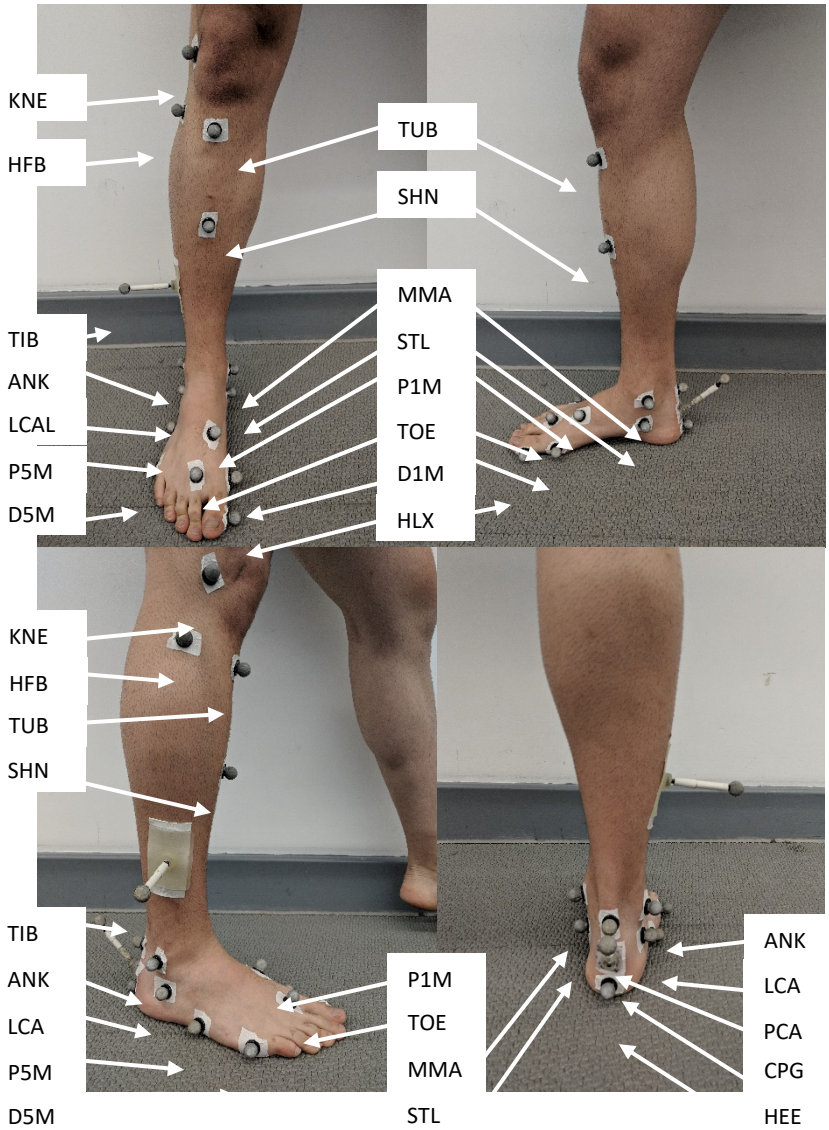

Figure 3. 3 Marker placement for the Oxford-Foot Model.

3.3.3 Footwear

Unless stated otherwise participants in experimental chapters ran barefoot, in minimal and structured-cushioned running shoes. In the minimal condition, participants ran in a VivoBarefoot® Stealth II, a minimal shoe with a non-cushioned and highly flexible 4mm EVA sole and thin mesh upper, and a 0mm heel-to-toe drop height (figure 3.3). The structured-cushioned running shoe was a Hoka One One Clifton 2, a shoe with an enlarged CMEVA midsole, a 29mm heel stack, a 24mm toe stack, and 5mm heel-to-toe drop (figure 3.3).

Figure 3. 4 Images of minimalist and structured-cushioned footwear conditions. Left: a minimalist VivoBarefoot® stealth II. Right: A structured-cushioned running shoe Hoka One One Clifton 2.

3.3.4 Running trial preparation

Once changed and in specified footwear, a static calibration was collected prior to the beginning of data collection in each respective footwear conditions. Participants were instructed to run at "an endurance pace that could be comfortably sustained for 45 minutes". This ensured that the speed chosen was representative of ER biomechanics. To calculate the average running speed for dynamic trials, participants ran down a 20-m track five times immediately after the five-minute warm up. Two sets of Brower timing gates (Brower timing gates, Utah, USA) recorded the time taken to run a central 10-m segment. This time was used to calculate running speed. The average of the five trials ± 5% represented acceptable limits for running speed in dynamic running trials.

3.3.5 Running trial procedure

For dynamic running trials in experimental chapters, participants were instructed to run along a predetermined 20-m runway. Unless stated otherwise, participants were assigned practice trials to determine their starting position. This ensured consistent-successful trials. According to Hopkins (2000a) three trials represent the number of trials necessary to gain a precise representation of reliability within a given population. Diss (2001) also demonstrated that three trials were sufficient to produce ICC values greater than 0.8 for running specific kinematics and kinetics, and with the exception of ankle eversion (4 trials), ICC's greater than 0.9 can be achieved with three trials. Furthermore, Diss (2001) used a 2D video camera with a sample rate of 50Hz. This relatively low sample rate compared to the 200Hz in the current

lab and might explain why the observed error in maximum ankle eversion required additional trials. In light of the available literature, three successful trials were captured for all experimental chapters.

A trial was deemed successful when both the entire stance phase of the dominant limb and average running speed ± 5% of the predetermined running speed were recorded. Only the dominant limb was investigated following this is in line with previous work investigating running biomechanics (Sobhani et al., 2017), 12 of 13 injuries occurring on the dominant limb in a study investigating the effects of knee joint load on injury rates (Dudley et al., 2017), and to assist processing given the timeline of the PhD the number of trials recorded in this thesis. Similar to participant preparation, average running speed was calculated using two sets of Brower timing gates (Brower timing gates, Utah, USA), but placed 2.7m apart representing the length of the kinetic data collection area (see figure 3.3 for floor plan of force platforms). Trials were excluded if the stance phase did not occur on a force platform, the calculated running speed was not ± 5%of the average-running speed, or they were perceived to be targeting the force platforms.

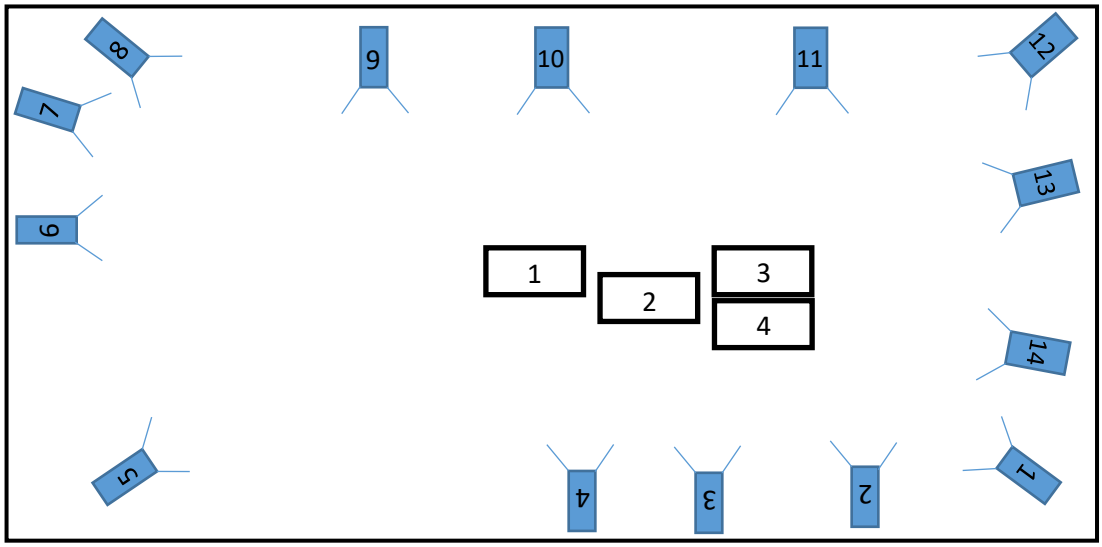

Figure 3. 5 Floor plan of the kinetic data collection area in the gait laboratory. Force platform dimensions were 464mm wide; 508mm long; 82.6mm deep. Anterior distance between force plates one and two, and two and three/four was 0.15 m. Numbered boxes represent cameras.

3.3.6 Data processing for three dimensional dynamic trials

In accordance with the default magnitude set for gait analysis, initial contact and toe-off events were produced when the magnitude of the GRF crossed a 20N threshold. The first frame in which the GRF exceeded 20N was identified as initial contact, and the first frame in which the GRF fell below 20N was identified as toe off.

All raw data for dynamic trials were processed in Vicon Nexus (Nexus 1.7, Vicon, Oxford, UK) by manually identifying marker trajectory gaps. Small gaps no larger than one frame were filled using a spline function. Larger marker trajectory gaps were filled using the pattern fill option, reflecting the trajectory of a marker with similar trajectory in the coinciding time frame, for example the left PSIS with the right PSIS. Following recommendations of Vicon Nexus (Vicon, 2010) the limit for gap filling was set to 30 frames. The dynamic models were then applied. Newton-Euler inverse dynamics approach was used to resolve external joint moments in the proximal segment co-ordinate system.

High frequency noise, soft-tissue movement artefact and double differentiation used to calculate segmental accelerations can all introduce errors into data. As a result, smoothing of data is necessary to provide an accurate characterisation of kinematics and joint moments. Typically, marker trajectory data is filtered at low frequencies and kinetic data is filtered at high frequencies (Ferber, Davis, Williams, & Laughton, 2002). However, Bisseling and Hof (2006) suggests this approach introduces an impact-like artefact in the calculation of moments proposed to arise from the attenuation of segmental accelerations in the inverse dynamic calculations (Bisseling & Hof, 2006; Edwards, Troy, & Derrick, 2011). In light of this evidence, Bisseling and Hof (2006) recommend the filtering of both kinetic and kinematic data at matched low-pass frequencies (20Hz). However, the impact proportion of the GRF during running has been reported to elicit frequencies ranging between 10Hz and 30Hz (Edwards et al., 2011). With this in mind, and following the methods of Chumanov, Heiderscheit, and Thelen (2011), both the kinematic and kinetic data were filtered at 25Hz, with a fourth order Butterworth filter with zero lag to mitigate the impact artefact and retain the high-frequency GRF data. Once processed in Vicon Nexus, dynamic trials for

experimental chapters were imported into Vicon Polygon (version 3.5.1, Vicon, Oxford, UK) where trials were normalised to 101 time points of the stance phase and joint moments were normalised to body mass ($Nm \cdot Kg^{-1}$).

Kinematic variables of interest were the hip, knee, and ankle angles at initial contact, midstance (50% of stance) and toe-off in the sagittal, frontal and transverse planes of motion. The maximum flexion/extension, abduction/adduction and internal/external joint angles, and their subsequent ROM for the lower-limb joints were also of interest (see figure 3.5 for illustration). Additionally, peak trunk lean, a measure of the thorax relative to the lab co-ordinate system, as well as the peak-forefoot pronation angle relative to the tibia were also of interest.

Kinetic variables of interest were the maximum hip, knee and ankle flexion/extension, adduction/abduction and internal/external rotation moments.

Discrete kinematic variables extracted from continuous waveforms were chosen for statistical analysis as previous research has used these variables to describe an individual's running technique and injury potential (Daoud et al., 2012; Tam, Wilson, Coetzee, van Pletsen, & Tucker, 2016). Discrete kinetic data extracted from continuous waveforms were chosen as previous research has demonstrated these to provide an understanding of joint loading and injury potential during ER (Dudley et al., 2017; Kerrigan et al., 2009), and secondly because peak-external joint moments represent a time when the muscles exert the largest turning moment on the skeletal structure; a time when the potential for repetitive strain injury is greatest (Lieberman, 2012b).

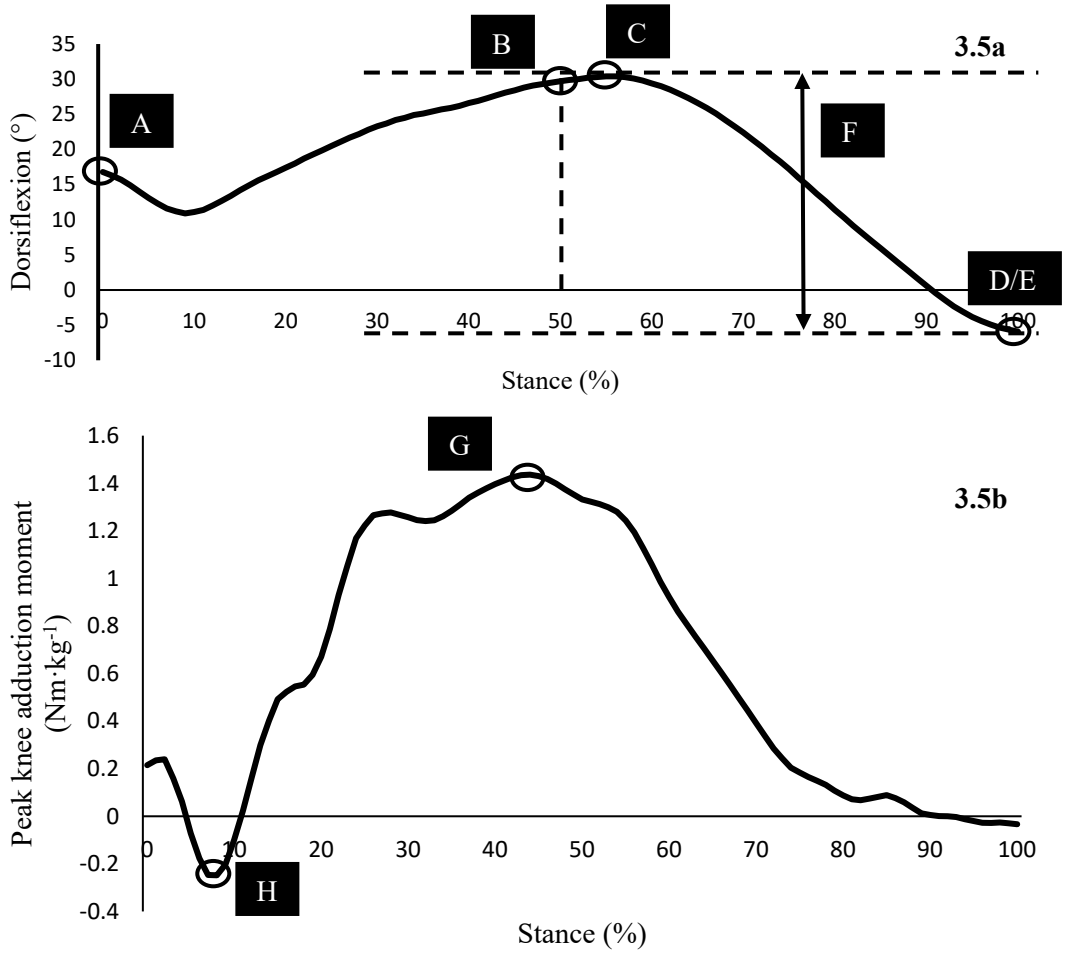

Figure 3. 6 Visualisation of key time points on example data to illustrate kinematic and kinetic variables of interest. 3.5a is a typical dorsiflexion graph during stance and 3.5b a typical external knee-adduction moment graph during stance. A: initial contact, B: mid-stance (50% of stance, dashed line), C: peak/maximum measure of interest (dorsiflexion), D: peak/ maximum in opposing direction (plantarflexion), E: toe-off, F: ROM over the data range (vertical solid arrow), G: peak joint moment in reported direction (adduction), H: peak joint moment in opposing direction (abduction).

4.0 Measurement error of 3-D kinematic and kinetic measures during over-ground endurance running in recreational runners within a test session and between two test sessions on a single day.

4.1 Introduction

Research suggests that natural selection has adapted *Homo sapiens* into ER specialists (Bramble & Lieberman, 2004; Lieberman, 2012b; Rolian et al., 2009). However, running injury rates are reported anywhere between 20% and 79% (Daoud et al., 2012; Taunton et al., 2003; van Gent et al., 2007). This observation suggests humans are prone to injury in an arguably species-specific movement pattern. Subsequently, biomechanics research has investigated effects of both footwear and running technique interventions on injury risk (Chan et al., 2018; Daoud et al., 2012). A common approach in these types of studies is 3-D kinematic and kinetic analysis. It is therefore important to quantify the associated measurement error to aid the interpretation of interventions and assist the planning of robust-future studies.

Currently there are few studies reporting the reliability of kinematic and kinetic measures in overground ER. While kinematic and kinetic data are available for walking (Stolze, Kuhtz-Buschbeck, Mondwurf, Jöhnk, & Friege, 1998; Wilken, Rodriguez, Brawner, & Darter, 2012), they cannot be applied to running due to the higher forces and speeds involved (Keller et al., 1996). Of the available kinematic and kinetic ER data, there is a trend for peak-joint angles and peak moments to be more reliable within-session than between-day (Ferber, Davis, Williams, et al., 2002; Queen, Gross, & Liu, 2006). Explanations for increased between-day variability include soft-tissue artefact, wand alignment (Plug-In Gait model) and anatomical-landmark identification (Della Croce, Leardini, Chiari, & Cappozzo, 2005; Leardini, Chiari, Della Croce, & Cappozzo, 2005); factors that also effect between-session data on a single day. Ferber, Davis, Williams, et al. (2002) suggest these factors influence absolute measures (e.g. joint angles at specific time points) more than relative measures such as joint excursions. This suggests that marker-placement errors between-sessions or days could introduce bias to joint kinematics, and that relative measures such as joint excursions will be more reliable than absolute measures.

Walking and running gait studies report the sagittal plane as the most reliable for kinematic data (Ferber, et al., 2002; Kadaba, Ramakrishnan, & Wootten, 1990; Manal, McClay, Stanhope, Richards, & Galinat, 2000). Della Croce et al. (2005) suggested that when joints predominantly operate in one plane, variability in rotations out of this plane are augmented by inaccurate anatomical identification. This suggests that inaccurate identification of anatomical landmarks underpins larger variability in the frontal and transverse planes. For this reason, Kadaba et al. (1990) concluded that ab/adduction and in/external rotations must be interpreted with caution, particularly at the knee, a joint not evolved to rotate in these planes. In addition, soft-tissue artefact also contributes to joint angle variability. In a study using bone pins as reference data, and an optimal marker configuration, Manal et al. (2000) demonstrated that error introduced by soft-tissue artefact was greater in the transverse plane than the sagittal plane (sagittal error ±2° and transverse ±4°). This provides further evidence that data in the frontal and transverse planes should be interpreted cautiously.

Similar to kinematic reports, there was a trend for sagittal-plane kinetic data to be less variable between-sessions in walking and running gait studies (Ferber, et al., 2002; Kadaba et al., 1989). However, Ferber, Davis, Williams, et al. (2002) reports conflicting findings for within-session ER kinetics, with peak-frontal-plane moments demonstrating smaller measurement error than peak-sagittal-plane moments. A possible explanation could be the biomechanical model used. Kadaba et al. (1989) used a hierarchical model reliant on wands that, when placed incorrectly, can introduce errors that propagate in a proximal-to-distal fashion (Buczek, Rainbow, Cooney, Walker, & Sanders, 2010). In contrast, Ferber, Davis, Williams, et al. (2002) used an over-determined marker set that independently tracks segments. This demonstrates that sagittal plane moments between-sessions were more reliable, but biomechanical models also influence the variability of the measures of interest.

While it has been argued that data in the frontal and transverse planes of particular joints should be interpreted with caution (Kadaba et al., 1990; Manal et al., 2000), few have provided comparisons between joints. The thigh segment has greater muscle mass than the lower leg, therefore markers attached to this location will logically have increased soft-tissue artefact. In

practice, Reinschmidt, Van Den Bogert, Nigg, Lundberg, and Murphy (1997) reports the soft-tissue artefact introduced by the lower leg as negligible, whereas soft-tissue artefact introduced by the thigh was the primary contributor to errors in measurement of knee joint rotation. This suggests measurement error at the hip will be greater than at the ankle as a result of increased soft-tissue artefact affecting the former. In addition, incorrect identification of anatomical landmarks is less likely at the ankle given the prominence of bones and lack of soft-tissue. This is yet to be examined in ER, however, if anatomical identification is the largest contributor to variability between sessions (Gorton, Hebert, & Gannotti, 2009), it is logical that ankle variability will be less than the hip and knee.

Currently, few studies address multiple measures in the stance phase at once. A study addressing multiple time points across stance using a controlled method is important, as many kinematic variables are used to examine mechanisms of running injury. For example, plantarflexion angle at initial contact has been used as a surrogate measure to assess a runner's potential for injury (Daoud et al., 2012). In addition, though the mean-peak loading of joints throughout stance has been reported (Ferber, Davis, Williams, et al., 2002), there are limited data quantifying measurement error of peak joint moments in the units of interest. Knowledge of this error in the units of the measurement tool is vital for interpreting the value of an intervention. Additionally, findings of within session, between session (same day) and between day comparisons are important for impending studies of this thesis. For example, when comparing footwear conditions it is yet to be decided whether conditions will be compared within a single day between multiple sessions or multiple sessions between days, this dependent on factors such as lab and participant availability. Furthermore, when undertaking a within-session acute-running intervention knowledge of natural variability within a single session is essential for sample size estimation. Knowledge of variability of within-session, between-session (same-day) and between day comparisons will be used to interpret the impact of interventions and contribute to sample sizes calculations.

Based on the evidence presented, the purpose of this investigation was to quantify measurement error of 3-D kinematic and kinetic variables at key stages in the gait cycle within

a session, between two sessions on the same day and between two days separated by 48 hours in recreational-endurance runners when running overground.

4.2 Methods

4.2.1 Participants

With approval from Northumbria University's Health and Life Sciences Ethics Committee, 23 volunteers participated. Due to the evolving nature of the thesis, 23 participants provided within-and between-session data and 13 of these 23 participants also provided between-day data. Within-and between-session participants comprised 18 males and five females with mean and SD age, stature and mass of 27 ± 6 yrs, 1.75 ± 0.09 m and 76.4 ± 10.5 kg. Between-day participants comprised of 10 males and three females with mean and SD age, stature and mass were 28 ± 6 yrs, 1.75 ± 0.07 m and 76.8 ± 10.0 Kg. Inclusion criteria were aged 18-45 years and participation in ER more than once per week as part of their exercise regime. Participants were excluded if they had an injury to the lower limbs in the previous six months, or any condition that could affect their normal running mechanics.

4.2.2 Experimental design

A repeated-measures design was used to assess measurement error in 3-D kinematic and kinetic measures within a single set of over-ground-running trials (within-session error), between two sets of trials separated by 30 minutes (between-session error) and between two sets of trials separated by 48 hours (between-day) on an indoor track and biomechanics lab. Within-session measurements were used to describe trial-to-trial variability, between-session measurements were used to quantify the effects of marker-placement error and between day comparisons were used to assess the effect of human day-to-day variability in addition to marker replacement. Each participant ran in their habitual running footwear to ensure data collected were representative of their running technique and not a result of novel footwear. Average running speed was determined and calculated as described in chapter three (3.3.4) and a successful trial was defined as described in chapter three, running trial procedure (3.3.5). Between sessions on the same day, markers were removed and 30 minutes was provided

between sessions to allow skin erythema to subside. When adhesive from markers were visible on the skin or shoe a wipe was used to remove left over residue. Mean and SD running speeds for day one and 48 hours later were 3.02 ± 0.36 $m \cdot s^{-1}$ and 2.81 ± 0.59 $m \cdot s^{-1}$, respectively.

4.2.3 Procedures

Participants were provided with appropriate clothing, anatomical measures taken, and a 3-D biomechanics analysis suite was calibrated as described in the chapter three (3.3.1 and 3.3.2). A series of retroflective markers were attached to participants in a 'Plug-In gait' formation (3.3.2) to facilitate the assessment of lower-limb biomechanics. Ten of the 23 participants had an additional series of markers attached to their lower-dominant limb in an 'Oxford-Foot Model' formation, as described in chapter three (3.3.2) to quantify forefoot pronation during stance. Marker locations of the 'Oxford Foot Model' are as described in chapter three (3.3.2).

Kinematic and kinetic data were captured by 14 calibrated infrared cameras (T10/20, Vicon MX, Oxford, UK) and four force plates (OR6-7, AMTI, Watertown MA, USA). Signals were captured and imported as described in chapter three, Three-dimensional gait laboratory calibration 3.3.1.

4.2.4 Data analysis

Data analysis and processing was undertaken in the 3-D motion analysis software in line with the processes described in chapter three, section 3.3.6. Peak-joint moment data and kinematic-joint angles of interest were calculated and were reported as described in chapter 3 (3.3.6). Data were then exported to Microsoft Excel (Microsoft, USA) for statistical analysis.

4.2.5 Statistical Analysis

After verification of underpinning assumptions of normality, linearity and equality of errors over the data range, average mean difference, intra-class correlation, least products regression and typical error were calculated. Least-products regression analysis cannot accommodate three variables and was used for between-session analysis only. Mean difference and average mean difference were chosen to quantify the systematic difference within a session (three

comparisons), between sessions (2 comparisons), and to aid comparisons to previous research. Intra-class correlation was chosen because it is a common metric used in other studies assessing reliability of 3-D kinematics and kinetics (Ferber, Davis, Williams, et al., 2002). Simple-linear regression is another commonly used metric to assess test-retest reliability, but simple linear regression assumes no error in the first testing occasion. Least-products regression overcomes this issue assuming error in both testing occasions and reports a coefficient that better represents data agreement on two separate testing occasions (Ludbrook, 1997). Typical error is suggested to be the most appropriate metric of reliability (Hopkins, 2000a) due to low bias and ease of interpretation (e.g. error is reported in the units of the measurement tool), therefore typical error was the primary metric of interest, with other metrics calculated to facilitate comparison to previous studies. Statistical analyses were performed using SPSS 24.0 (SPSS, Inc., Chicago, IL, USA) and Microsoft Excel (Microsoft, USA).

4.3 Results

4.3.1 Within-session

Tables 4.1 - 4.3 show kinetic and kinematic mean difference, typical error and intra-class correlations within a single session at key points during stance.

Table 4. 1 Within-session kinematic and kinetic measurement error of the hip joint during the stance phase of overground endurance running in 23 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)
Sagittal	IC	0.17 ± 1.31	1.91 (1.63 - 2.32)	0.94 (0.88- 0.97)
(°)	Midstance	-0.20 ± 1.82	2.04 (1.74 - - 2.48)	0.96 (0.91 - 0.98)
	TO	0.16 ± 1.11	1.61 (1.38 ± 1.96)	0.96 (0.91 ± 0.98)

	Peak flexion	-0.26 ± 1.89	2.22 (1.89 - 2.7)	0.93 (0.87 - 0.97)
	Peak extension	0.14 ± 1.12	1.61 (1.37 1.96)	0.96 (0.91- 0.98)
	ROM	-0.40 ± 1.73	2.68 (2.29 - 3.26)	0.79 (0.61 - 0.89)
Sagittal moments ($Nm \cdot kg^{-1}$)	Peak flexion	-0.02 ± 0.42	0.59 (0.50 - 0.72)	0.76 (0.55 - 0.87)
	Peak extension	0.00 ± 0.44	0.59 (0.50 - 0.72)	0.69 (0.44 - 0.84)
Frontal (°)	IC	0.16 ± 0.95	1.25 (1.07 - 1.52)	0.93 (0.86 - 0.97)
	Midstance	0.03 ± 0.88	1.25 (1.07 - 1.52)	0.94 (0.89 - 0.97)
	TO	0.00 ± 1.05	1.16 (0.99 - 1.4)	0.87 (0.74 - 0.93)
	Peak adduction	0.10 ± 1.02	1.38 (1.17 - 1.67)	0.94 (0.88 - 0.97)
	Peak abduction	-0.02 ± 1.07	1.18 (1.01 - 1.43)	0.86 (0.73 - 0.93)
	ROM	0.12 ± 1.18	1.29 (1.10 - 1.57)	0.94 (0.87 - 0.97)
Frontal moments ($Nm \cdot kg^{-1}$)	Peak adduction	-0.04 ± 0.26	0.27 (0.23 - 0.33)	0.75 (0.55 - 0.87)
	Peak abduction	-0.02 ± 0.15	0.19 (0.17 - 0.24)	0.61 (0.34 - 0.79)
Transverse (°)	IC	0.43 ± 2.16	2.66 (2.27 - 3.23)	0.89 (0.78 - 0.94)
	Midstance	0.15 ± 1.98	2.29 (1.95 - 2.78)	0.89 (0.79 - 0.95)
	TO	-0.11 ± 2.19	2.46 (2.10 - 2.99)	0.90 (0.81 - 0.95)
	Peak internal rotation	-0.15 ± 2.33	2.81 (2.40 - 3.41)	0.86 (0.74 - 0.93)
	Peak external rotation	0.05 ± 1.82	2.54 (2.17 - 3.09)	0.87 (0.75 - 0.93)

	ROM	-0.20 ± 1.82	2.44 (2.08 - 2.97)	0.81 (0.64 - 0.90)
Transverse moments (Nm·kg^{-1})	Peak internal rotation	0.01 ± 0.04	0.04 (0.03 - 0.05)	0.82 (0.67 - 0.91)
	Peak external rotation	0.01 ± 0.07	0.06 (0.05 - 0.07)	0.94 (0.88 - 0.97)

Table 4. 2 Within-session kinematic and kinetic measurement error of the knee joint during the stance phase of overground endurance running in 23 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)
Sagittal (°)	IC	-0.20 ± 1.39	1.98 (1.69 - 2.41)	0.82 (0.66 - 0.91)
	Midstance	-0.73 ± 1.87	1.95 (1.66 - 2.37)	0.96 (0.92 - 0.98)
	TO	-0.52 ± 1.39	2.12 (1.81 - 2.58)	0.91 (0.82 - 0.96)
	Peak flexion	-0.57 ± 1.58	1.60 (1.36 - 1.94)	0.89 (0.78 - 0.94)
	Peak extension	-0.23 ± 1.26	1.91 (1.63 - 2.32)	0.87 (0.76 - 0.94)
	ROM	-0.34 ± 1.89	2.10 (1.79 - 2.56)	0.85 (0.71 - 0.92)
Sagittal moments (Nm·kg^{-1})	Peak flexion	-0.04 ± 0.16	0.25 (0.21 - 0.31)	0.81 (0.64 - 0.90)
	Peak extension	-0.01 ± 0.10	0.16 (0.14 - 0.20)	0.63 (0.36 - 0.80)
Frontal (°)	IC	0.14 ± 0.72	0.86 (0.73 - 1.04)	0.96 (0.92 - 0.98)
	Midstance	0.14 ± 1.86	1.83 (1.56 - 2.23)	0.90 (0.81 - 0.95)

	TO	-0.04 ± 0.78	0.91 (0.77 - 1.10)	0.95 (0.90 - 0.98)
	Peak adduction	0.09 ± 1.78	1.90 (1.62 - 2.31)	0.89 (0.78 - 0.94)
	Peak abduction	0.23 ± 1.74	1.85 (1.57 - 2.24)	0.90 (0.81 - 0.95)
	ROM	-0.14 ± 1.11	1.33 (1.13 - 1.62)	0.86 (0.74 - 0.93)
Frontal moments ($Nm \cdot kg^{-1}$)	Peak adduction	-0.04 ± 0.18	0.19 (0.17 - 0.24)	0.90 (0.80 - 0.95)
	Peak abduction	-0.01 ± 0.09	0.10 (0.08 - 0.12)	0.77 (0.57 - 0.88)
Transverse (°)	IC	-0.12 ± 1.89	2.70 (2.31 - 3.29)	0.94 (0.88 - 0.97)
	Midstance	-0.01 ± 1.12	2.08 (1.77 - 2.52)	0.95 (0.90 - 0.98)
	TO	-0.27 ± 1.60	2.09 (1.78 - 2.54)	0.95 (0.89 - 0.97)
	Peak internal rotation	0.00 ± 2.10	2.11 (1.80 - 2.56)	0.96 (0.92 - 0.98)
	Peak external rotation	-0.37 ± 1.47	1.92 (1.64 - 2.33)	0.95 (0.90 - 0.98)
	ROM	0.37 ± 1.79	2.30 (1.96 - 2.79)	0.84 (0.69 - 0.92)
Transverse moments ($Nm \cdot kg^{-1}$)	Peak internal rotation	0.00 ± 0.03	0.04 (0.03 - 0.05)	0.87 (0.75 - 0.93)
	Peak external rotation	-0.01 ± 0.10	0.10 (0.09 - 0.12)	0.75 (0.55 - 0.87)

Table 4. 3: Within-session kinematic and kinetic measurement error of the ankle joint during the stance phase of overground endurance running in 23 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)

Sagittal (°)	IC	-0.28 ± 1.62	2.88 (2.45 - 3.50)	0.84 (0.69 - 0.92)
	Midstance	-0.12 ± 0.77	1.47 (1.26 - 1.79)	0.98 (0.97 - 0.99)
	TO	-0.51 ± 1.96	3.05 (2.60 - 3.71)	0.88 (0.76 - 0.94)
	Peak dorsiflexion	-0.32 ± 0.84	1.39 (1.19 - 1.69)	0.88 (0.77 - 0.94)
	Peak plantarflexion	-0.51 ± 1.96	2.97 (2.53 - 3.61)	0.88 (0.77 - 0.94)
	ROM	0.19 ± 2.60	3.41 (2.91 - 4.14)	0.82 (0.66 - 0.91)
Sagittal moments ($Nm \cdot kg^{-1}$)	Peak dorsiflexion	-0.01 ± 0.11	0.23 (0.20 - 0.28)	0.69 (0.45 - 0.84)
	Peak Plantarflexion	0.01 ± 0.07	0.09 (0.08 - 0.11)	0.74 (0.53 - 0.87)
Frontal (°)	IC	0.08 ± 0.32	0.46 (0.39 - 0.56)	0.96 (0.91 - 0.98)
	Midstance	0.09 ± 0.40	0.50 (0.42 - 0.60)	0.96 (0.92 - 0.98)
	TO	-0.02 ± 0.32	0.38 (0.33 - 0.46)	0.93 (0.87 - 0.97)
	Peak adduction	0.06 ± 0.41	0.48 (0.41 - 0.59)	0.98 (0.95 - 0.99)
	Peak abduction	-0.01 ± 0.31	0.38 (0.33 - 0.47)	0.93 (0.87 - 0.97)
	ROM	0.07 ± 0.51	0.59 (0.50 - 0.72)	0.95 (0.90 - 0.98)
Frontal moments ($Nm \cdot kg^{-1}$)	Peak adduction	0.00 ± 0.07	0.08 (0.07 - 0.09)	0.78 (0.60 - 0.89)
	Peak abduction	-0.01 ± 0.16	0.18 (0.16 - 0.22)	0.70 (0.47 - 0.84)

Transverse (°)	IC	-0.25 ± 1.38	2.20 (1.88 - 2.68)	0.95 (0.90 - 0.98)
	Midstance	-0.17 ± 1.26	1.71 (1.46 - 2.08)	0.97 (0.94 - 0.99)
	TO	0.07 ± 1.42	1.93 (1.65 - 2.35)	0.94 (0.89 - 0.97)
	Peak internal rotation	0.07 ± 1.39	1.90 (1.62 - 2.31)	0.95 (0.89 - 0.97)
	Peak external rotation	-0.09 ± 1.53	1.66 (1.42 - 2.02)	0.97 (0.94 - 0.99)
	ROM	0.15 ± 2.06	2.31 (1.97 - 2.80)	0.91 (0.82 - 0.95)
Transverse moments ($Nm \cdot kg^{-1}$)	Peak internal rotation	-0.01 ± 0.06	0.08 (0.07 - 0.10)	0.75 (0.53 - 0.87)
	Peak external rotation	0.00 ± 0.03	0.04 (0.03 - 0.05)	0.64 (0.41 - 0.80)

Typical error for all kinematic variables of interest within a single testing session of overground ER were equal to or less than 2.81°, 2.70° and 3.41° for the hip, knee and ankle joint, respectively. Specific to the hip, the largest typical error observed in the sagittal, frontal and transverse plane was 2.68°, 1.38° and 2.81°, respectively. At the knee joint, the greatest typical error in the sagittal, frontal and transverse plane was 2.12° 1.90° and 2.70°, respectively. Lastly, at the ankle joint, the largest typical error in the sagittal, frontal and transverse plane was 3.41°, 0.59°, and 2.31°, respectively.

The maximum recorded typical error for peak joint loading within a single testing session of overground running at the hip, knee and ankle was 0.59, 0.25 and 0.23 $Nm \cdot kg^{-1}$, respectively. Specific to the hip joint, the greatest typical error in peak joint loading in the sagittal, frontal and transverse plane was 0.59, 0.27 and 0.06 $Nm \cdot kg^{-1}$, respectively. At the knee joint, the greatest typical error in peak joint loading in the sagittal, frontal and transverse plane was 0.25, 0.19, 0.10 $Nm \cdot kg^{-1}$, respectively. At the ankle joint, the greatest typical error in peak joint

loading in the sagittal, frontal and transverse plane was 0.23, 0.18 and 0.08 $Nm \cdot kg^{-1}$, respectively.

4.3.2 Between-session

Tables 4.4 – 4.6 show the kinematics and kinetic mean difference, typical error, intra-class correlations and least products regression between two testing occasions, separated by 30 minutes.

Table 4. 4 Between-session kinematic and kinetic measurement error of the hip joint during the stance phase of overground endurance running, in 23 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Sagittal (°)	IC	0.16 ± 3.73	2.63 (2.12 - 3.52)	0.89 (0.78 - 0.94)	0.98 1.02
	Midstance	-0.78 ± 7.08	5.01 (4.03 - 6.69)	0.70 (0.45 - 0.84)	1.14 -4.11
	TO	0.16 ± 2.90	2.05 (1.65 - 2.74)	0.93 (0.86 - 0.96)	0.97 -0.06
	Peak flexion	0.37 ± 3.52	2.49 (2.01 - 3.33)	0.91 (0.82 - 0.95)	0.97 1.65
	Peak extension	0.09 ± 3.10	2.19 (1.76 - 2.93)	0.92 (0.84 - 0.96)	0.98 -0.03
	ROM	0.28 ± 2.11	1.49 (1.20 - 1.99)	0.92 (0.84 - 0.96)	0.92 4.04
Sagittal moments ($Nm \cdot kg^{-1}$)	Peak flexion	-0.10 ± 0.62	0.44 (0.35 - 0.58)	0.84 (0.69 - 0.92)	0.96 0.17

	Peak extension	0.04 ± 0.57	0.40 (0.32 - 0.54)	0.82 (0.66 - 0.91)	1.10 0.26
Frontal (°)	IC	-0.8 ± 4.37	3.09 (2.49 - 4.13)	0.68 (0.43 - 0.83)	1.37 2.37
	Midstance	-0.92 ± 4.55	3.22 (2.59 - 4.30)	0.66 (0.40 - 0.82)	1.21 -2.60
	TO	-0.78 ± 3.61	2.55 (2.05 - 3.41)	0.57 (0.27 - 0.77)	1.55 0.68
	Peak adduction	-0.58 ± 4.33	3.06 (2.47 - 4.09)	0.73 (0.52 - 0.86)	1.13 -2.02
	Peak abduction	-0.97 ± 3.57	2.52 (2.03 - 3.37)	0.58 (0.28 - 0.77)	1.58 0.59
	ROM	0.39 ± 1.67	1.18 (0.95 - 1.58)	0.95 (0.89 - 0.97)	1.07 -0.64
Frontal moments ($Nm \cdot kg^{-1}$)	Peak adduction	0.00 ± 0.36	0.25 (0.21 - 0.34)	0.83 (0.67 - 0.91)	1.40 -0.70
	Peak abduction	-0.06 ± 0.31	0.22 (0.18 - 0.29)	0.64 (0.37 - 0.81)	1.77 0.38
Transverse (°)	IC	-2.36 ± 7.15	5.05 (4.07 - 6.75)	0.56 (0.25 - 0.76)	1.00 -2.37
	Midstance	-1.63 ± 6.38	4.51 (3.63 - 6.03)	0.57 (0.27 - 0.77)	1.02 -1.56
	TO	-2.45 ± 6.89	4.88	0.57 (0.27 - 0.77)	0.96 -2.70

			(3.93 - 6.51)		
	Peak internal rotation	-1.79 ± 6.05	4.28 (3.44 - 5.71)	0.60 (0.31 - 0.78)	0.91 -1.61
	Peak external rotation	-2.16 ± 7.26	5.13 (4.13 - 6.85)	0.46 (0.13 - 0.70)	1.14 -0.75
	ROM	0.38 ± 2.23	1.58 (1.27 - 2.11)	0.91 (0.82 - 0.96)	1.05 -0.22
Transverse moments	Peak internal rotation	0.04 ± 0.12	0.08 (0.07 - 0.11)	0.66 (0.41 - 0.82)	2.02 -0.66
($Nm \cdot kg^{-1}$)	Peak external rotation	0.00 ± 0.11	0.08 (0.06 - 0.10)	0.91 (0.83 - 0.96)	1.18 0.92

Table 4. 5 Between-session kinematic and kinetic measurement error of the knee joint during the stance phase of overground endurance running, in 23 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Sagittal (°)	IC	0.58 ± 2.41	1.71 (1.37 - 2.28)	0.85 (0.71 - 0.92)	1.01 -0.39
	Midstance	-1.06 ± 10.77	7.62 (6.13 - 10.17)	0.15 (-0.21 - 0.48)	2.03 -48.21
	TO	1.40 ± 2.70	1.99 (1.53 - 2.55)	0.93 (0.85 - 0.96)	1.05 0.57
	Peak flexion	0.99 ± 2.52	1.78	0.85 (0.72 - 0.93)	1.09 -3.44

			(1.44 - 2.38)		
	Peak extension	0.83 ± 2.16	1.53 (1.23 - 2.04)	0.91 (0.82 - 0.96)	1.02 0.51
	ROM	0.16 ± 2.11	1.49 (1.20 - 1.99)	0.91 (0.83 - 0.96)	1.08 -2.44
Sagittal moments	Peak flexion	0.01 ± 0.31	0.22 (0.17 - 0.29)	0.84 (0.70 - 0.92)	1.08 -0.22
($Nm \cdot kg^{-1}$)	Peak extension	0.02 ± 0.15	0.10 (0.08 - 0.14)	0.79 (0.60 - 0.89)	1.00 0.02
Frontal (°)	IC	-0.01 ± 4.28	3.03 (2.44 - 4.04)	0.65 (0.38 - 0.81)	1.31 0.39
	Midstance	-0.70 ± 4.98	3.52 (2.84 - 4.71)	0.59 (0.30 - 0.78)	0.96 -0.76
	TO	-0.41 ± 3.41	2.41 (1.94 - 3.22)	0.69 (0.44 - 0.84)	1.10 -0.33
	Peak adduction	-0.62 ± 5.24	3.70 (2.98 - 4.95)	0.51 (0.20 - 0.73)	0.98 -0.57
	Peak abduction	-1.01 ± 4.86	3.44 (2.77 - 4.59)	0.61 (0.33 - 0.79)	0.99 -1.07
	ROM	0.39 ± 1.98	1.40 (1.13 - 1.87)	0.81 (0.64 - 0.90)	0.87 1.48
Frontal moments	Peak adduction	0.05 ± 0.24	0.17 (0.13 - 0.22)	0.91 (0.82 - 0.96)	0.90 0.22

(Nm·kg^{-1})	Peak abduction	0.00 ± 0.11	0.08 (0.06 - 0.10)	0.83 (0.68 - 0.92)	1.06 0.01
Transverse (°)	IC	3.37 ± 7.20	5.09 (4.10 - 6.80)	0.74 (0.52 - 0.86)	0.81 4.29
	Midstance	2.07 ± 7.47	5.28 (4.25 - 7.05)	0.68 (0.44 - 0.84)	1.03 1.52
	TO	4.15 ± 8.10	5.73 (4.61 - 7.64)	0.48 (0.16 - 0.71)	0.81 4.25
	Peak internal rotation	3.27 ± 7.91	5.60 (4.51 - 7.47)	0.69 (0.44 - 0.84)	0.91 5.43
	Peak external rotation	3.45 ± 7.26	5.14 (4.14 - 6.86)	0.57 (0.27 - 0.77)	0.84 3.39
	ROM	-0.18 ± 3.64	2.58 (2.07 - 3.44)	0.78 (0.60 - 0.89)	1.04 -1.10
Transverse moments	Peak internal rotation	0.00 ± 0.05	0.03 (0.03 - 0.04)	0.92 (0.84 - 0.96)	1.08 -0.01
(Nm·kg^{-1})	Peak external rotation	-0.02 ± 0.09	0.07 (0.05 - 0.09)	0.89 (0.78 - 0.95)	1.20 0.01

Table 4. 6 Between-session kinematic and kinetic measurement error of the ankle joint during the stance phase of overground endurance running, in 23 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Sagittal (°)	IC	0.33 ± 2.73	1.93 (1.56 - 2.58)	0.91 (0.82 - 0.96)	0.92 1.72
	Midstance	-1.12 ± 6.85	4.84 (3.90 - 6.46)	0.37 (0.03 - 0.64)	1.73 -22.91
	TO	0.66 ± 3.49	2.47 (1.99 - 3.29)	0.91 (0.81 - 0.95)	0.98 0.32
	Peak dorsiflexion	0.30 ± 1.94	1.37 (1.10 - 1.83)	0.89 (0.78 - 0.94)	1.07 -1.90
	Peak plantarflexion	0.69 ± 3.46	2.45 (1.97 - 3.27)	0.91 (0.81 - 0.95)	0.98 0.45
	ROM	-0.39 ± 3.54	2.50 (2.01 - 3.34)	0.88 (0.76 - 0.94)	0.98 0.34
Sagittal moments ($Nm \cdot kg^{-1}$)	Peak dorsiflexion	0.04 ± 0.19	0.13 (0.11 - 0.18)	0.86 (0.73 - 0.93)	0.94 0.21
	Peak plantarflexion	0.01 ± 0.10	0.07 (0.06 - 0.10)	0.79 (0.61 - 0.90)	1.05 0.02
Frontal (°)	IC	0.19 ± 0.93	0.66 (0.53 - 0.88)	0.90 (0.79 - 0.95)	0.87 0.40
	Midstance	0.14 ± 1.41	1.00 (0.80 - 1.33)	0.83 (0.67 - 0.91)	1.02 0.05
	TO	0.22 ± 0.95	0.67 (0.54 - 0.90)	0.78 (0.59 - 0.89)	0.96 0.20
	Peak adduction	0.49 ± 1.92	1.35 (1.09 - 1.81)	0.79 (0.61 - 0.89)	0.93 0.91
	Peak abduction	0.21 ± 0.92	0.65 (0.52 - 0.86)	0.79 (0.61 - 0.90)	0.95 0.18
	ROM	0.27 ± 1.55	1.10 (0.89 - 1.47)	0.81 (0.63 - 0.90)	0.93 0.73

Frontal moments (Nm·kg^{-1})	Peak adduction	0.00 ± 0.12	0.09 (0.07 - 0.11)	0.70 (0.47 - 0.85)	1.06 -0.01
	Peak abduction	-0.02 ± 0.18	0.13 (0.10 - 0.17)	0.81 (0.64 - 0.91)	1.03 -0.01
Transverse (°)	IC	-0.99 ± 3.99	2.82 (2.27 - 3.77)	0.89 (0.79 - 0.95)	0.79 -3.07
	Midstance	-0.33 ± 5.47	3.87 (3.11 - 5.16)	0.81 (0.64 - 0.90)	1.07 1.22
	TO	-0.83 ± 5.32	3.76 (3.03 - 5.02)	0.75 (0.55 - 0.87)	0.89 -0.68
	Peak internal rotation	-0.78 ± 5.13	3.63 (2.92 - 4.85)	0.77 (0.57 - 0.88)	0.89 -0.61
	Peak external rotation	-2.10 ± 4.00	2.83 (2.28 - 3.78)	0.90 (0.81 - 0.95)	0.89 -5.23
	ROM	1.32 ± 3.34	2.36 (1.90 - 3.16)	0.90 (0.80 - 0.95)	1.04 0.07
Transverse moments (Nm·kg^{-1})	Peak internal rotation	0.03 ± 0.08	0.06 (0.05 - 0.08)	0.85 (0.71 - 0.92)	1.06 0.00
	Peak external rotation	0.00 ± 0.04	0.03 (0.02 - 0.04)	0.84 (0.69 - 0.92)	1.62 0.04

Between two testing sessions separated by 30 minutes, the largest kinematic typical error for overground running at the hip, knee and ankle joint was 5.13°, 7.62° and 4.84°, respectively. The largest typical error for the hip joint in the sagittal, frontal and transverse plane was 5.01°, 3.22° and 5.13°, respectively. The greatest typical error reported for the knee joint in the sagittal, frontal and transverse plane was 7.62°, 3.70° and 5.73°, respectively. The largest typical error for the ankle joint in the sagittal, frontal and transverse plane at was 4.84°, 1.35°, 3.87°, respectively.

The maximum kinetic typical error between two testing occasions separated by 30 minutes at the hip, knee and ankle was 0.44, 0.22 and 0.13 Nm·kg^{-1}, respectively. The largest typical error reported for the hip joint in the sagittal, frontal and transverse plane was 0.44, 0.25, 0.08

Nm·kg^{-1}, respectively. The greatest typical error reported for the knee joint in the sagittal, frontal and transverse plane was 0.22, 0.17, 0.07 Nm·kg^{-1}, respectively. The maximum typical error reported for the ankle joint in the sagittal, frontal and transverse plane was 0.13, 0.13, 0.06 Nm·kg^{-1}.

4.3.3 Between-day

Tables 4.7 – 4.9 show mean difference, typical error, intra-class correlations and least-products regression of kinetic and kinematic measures at initial contact, midstance and toe off, maximum, minimum and ROM during overground running trials separated by 48hrs.

Table 4. 7 Between-day kinematic and kinetic measurement error of the hip joint during the stance phase of overground endurance running, in 13 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Sagittal (°)	IC	0.26 ± 3.08	2.18 (1.64 - 3.30)	0.89 (0.72 - 0.96)	1.15 -5.12
	Midstance	0.07 ± 2.75	1.95 (1.47 - 2.95)	0.94 (0.84 - 0.98)	1.18 -4.38
	TO	0.18 ± 2.53	1.79 (1.35 - 2.71)	0.92 (0.79 - 0.97)	0.87 -0.47
	Peak flexion	-0.11 ± 2.90	2.05 (1.55 - 3.11)	0.92 (0.78 - 0.97)	1.15 -5.57
	Peak extension	0.20 ± 2.48	1.76 (1.33 - 2.66)	0.92 (0.79 - 0.97)	0.88 -0.42
	ROM	-0.31 ± 2.28	1.61 (1.22 - 2.44)	0.89 (0.72 - 0.96)	1.17 -7.72

Sagittal moments (Nm·kg^{-1})	Peak flexion	0.37 ± 0.77	0.54 (0.41 - 0.82)	0.81 (0.55 - 0.93)	1.16 -0.07
	Peak extension	-0.25 ± 0.72	0.51 (0.39 - 0.77)	0.56 (0.12 - 0.82)	1.84 1.20
Frontal (°)	IC	-1.14 ± 2.65	1.88 (1.42 - 2.84)	0.81 (0.54 - 0.93)	0.95 -0.94
	Midstance	-1.15 ± 2.76	1.95 (1.48 - 2.96)	0.84 (0.61 - 0.94)	0.92 -0.46
	TO	-0.21 ± 2.63	1.86 (1.41 - 2.82)	0.69 (0.31 - 0.88)	0.78 -0.77
	Peak adduction	-0.21 ± 2.63	1.86 (1.41 - 2.82)	0.69 (0.31 - 0.88)	0.88 -0.13
	Peak abduction	-0.56 ± 2.38	1.68 (1.27 - 2.55)	0.74 (0.41 - 0.90)	0.79 -1.10
	ROM	-0.79 ± 1.88	1.33 (1.01 - 2.02)	0.86 (0.65 - 0.95)	0.93 0.20
Frontal moments (Nm·kg^{-1})	Peak adduction	0.08 ± 0.40	0.28 (0.21 - 0.42)	0.58 (0.14 - 0.83)	1.41 -0.58
	Peak abduction	0.05 ± 0.26	0.18 (0.14 - 0.28)	0.34 (-0.16 - 0.71)	1.41 -0.58
Transverse (°)	IC	-2.04 ± 8.56	6.05 (4.57 - 9.17)	0.44 (-0.05 - 0.76)	1.16 -1.39
	Midstance	0.10 ± 7.94	5.61	0.56 (0.11 - 0.82)	1.49 1.58

			(4.24 - 8.51)		
	TO	-2.30 ± 8.65	6.11 (4.62 - 9.27)	0.42 (0.08 - 0.75)	1.39 -0.28
	Peak internal rotation	-0.67 ± 8.35	5.90 (4.46 - 8.95)	0.60 (0.17 - 0.84)	1.33 -1.38
	Peak external rotation	-2.40 ± 8.71	6.16 (4.65 - 9.33)	0.37 (-0.13 - 0.72)	1.51 1.70
	ROM	1.73 ± 3.26	2.30 (1.74 - 3.49)	0.68 (0.30 - 0.87)	1.10 0.71
Transverse moments ($Nm \cdot kg^{-1}$)	Peak internal rotation	-0.01 ± 0.06	0.04 (0.03 - 0.06)	0.19 (-0.31 - 0.61)	0.76 0.01
	Peak external rotation	0.01 ± 0.13	0.09 (0.07 - 0.14)	0.57 (0.13 - 0.82)	1.09 0.05

Table 4. 8 Between-day kinematic and kinetic measurement error of the knee joint during the stance phase of overground endurance running, in 13 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Sagittal (°)	IC	0.23 ± 2.40	1.70 (1.28 - 2.57)	0.76 (0.44 - 0.91)	1.23 -3.83
	Midstance	-0.24 ± 2.02	1.42 (1.08 - 2.16)	0.83 (0.59 - 0.94)	0.95 2.06
	TO	1.32 ± 2.76	1.95	0.89	0.84

			(1.48 - 2.96)	(0.72 - 0.96)	4.38
	Peak flexion	0.62 ± 1.95	1.38 (1.04 - 2.09)	0.78 (0.49 - 0.92)	1.16 -7.27
	Peak extension	0.76 ± 2.26	1.60 (1.21 - 2.42)	0.80 (0.52 - 0.92)	1.00 0.75
	ROM	-0.14 ± 2.21	1.56 (1.18 - 2.37)	0.90 (0.74 - 0.96)	1.05 -1.88
Sagittal moments	Peak flexion	-0.08 ± 0.34	0.24 (0.18 - 0.36)	0.80 (0.53 - 0.93)	1.19 -0.63
($Nm \cdot kg^{-1}$)	Peak extension	0.02 ± 0.11	0.08 (0.06 - 0.12)	0.80 (0.53 - 0.93)	0.82 -0.08
Frontal (°)	IC	-0.12 ± 2.23	1.58 (1.19 - 2.39)	0.82 (0.56 - 0.93)	1.16 0.08
	Midstance	1.56 ± 5.46	3.86 (2.91 - 5.85)	0.59 (0.16 - 0.83)	1.44 2.49
	TO	0.17 ± 2.82	1.99 (1.51 - 3.02)	0.79 (0.50 - 0.92)	1.35 0.45
	Peak adduction	0.16 ± 4.48	3.17 (2.40 - 4.80)	0.66 (0.26 - 0.86)	1.26 -0.51
	Peak abduction	-0.85 ± 4.57	3.23 (2.44 - 4.90)	0.64 (0.23 - 0.85)	1.51 1.34
	ROM	1.01 ± 2.07	1.46 (1.10 - 2.21)	0.67 (0.29 - 0.87)	1.26 -0.78

Frontal moments	Peak adduction	-0.04 ± 0.53	0.37 (0.28 - 0.56)	0.49 (0.02 - 0.79)	1.18 -0.31
(Nm·kg^{-1})	Peak abduction	0.01 ± 0.10	0.07 (0.06 - 0.11)	0.63 (0.22 - 0.85)	1.48 0.12
Transverse (°)	IC	1.95 ± 5.67	4.01 (3.03 - 6.08)	0.82 (0.56 - 0.93)	0.96 2.16
	Midstance	0.91 ± 8.20	5.80 (4.38 - 8.79)	0.69 (0.32 - 0.88)	0.96 1.66
	TO	2.33 ± 6.31	4.46 (3.37 - 6.76)	0.75 (0.42 - 0.90)	1.26 1.87
	Peak internal rotation	1.31 ± 8.50	6.01 (4.54 - 9.10)	0.67 (0.28 - 0.87)	1.03 0.61
	Peak external rotation	2.19 ± 6.06	4.28 (3.24 - 6.49)	0.75 (0.43 - 0.91)	1.18 2.01
	ROM	-0.88 ± 4.14	2.93 (2.21 - 4.44)	0.50 (0.04 - 0.79)	1.01 -1.15
Transverse moments	Peak internal rotation	-0.04 ± 0.08	0.06 (0.04 - 0.08)	0.48 (0.01 - 0.78)	0.58 0.02
(Nm·kg^{-1})	Peak external rotation	0.02 ± 0.09	0.07 (0.05 - 0.10)	0.61 (0.18 - 0.84)	0.53 -0.04

Table 4. 9 Between-day kinematic and kinetic measurement error of the ankle joint during the stance phase of overground endurance running, in 13 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Sagittal	IC	0.59 ± 1.74	1.23	0.95	0.96
			(0.93 - 1.87)	(0.86 - 0.98)	1.38
(°)	Midstance	0.81 ± 1.84	1.30	0.87	0.91
			(0.98 - 1.97)	(0.68 - 0.95)	3.61
	TO	0.92 ± 4.75	3.36	0.82	0.94
			(2.54 - 5.09)	(0.57 - 0.93)	0.23
	Peak	0.63 ± 1.71	1.21	0.85	0.96
	dorsiflexion		(0.91 - 1.83)	(0.63 - 0.95)	1.89
	Peak	0.97 ± 4.70	3.33	0.82	0.95
	plantarflexion		(2.51 - 5.04)	(0.57 - 0.93)	0.40
	ROM	-0.34 ± 5.01	3.54	0.81	0.99
			(2.68 - 5.37)	(0.54 - 0.93)	0.30
Sagittal	Peak	-0.07 ± 0.19	0.13	0.89	1.02
moments	dorsiflexion		(0.10 - 0.20)	(0.71 - 0.96)	-0.13
($Nm \cdot kg^{-1}$)	Peak	0.02 ± 0.09	0.06	0.91	1.26
	plantarflexion		(0.05 - 0.09)	(0.76 - 0.97)	0.09
Frontal	IC	0.20 ± 1.49	1.05	0.69	0.93
(°)			(0.80 - 1.60)	(0.31 - 0.88)	0.31
	Midstance	0.28 ± 1.76	1.24	0.67	0.93
			(0.94 - 1.89)	(0.29 - 0.87)	0.55
	TO	0.22 ± 1.41	1.00	0.69	1.08
			(0.75 - 1.51)	(0.31 - 0.88)	0.26
	Peak	0.22 ± 1.80	1.27	0.74	0.97
	adduction		(0.96 - 1.93)	(0.41 - 0.90)	0.35
	Peak	0.23 ± 1.42	1.00	0.68	1.07
	abduction		(0.76 - 1.52)	(0.31 - 0.88)	0.26
	ROM	-0.01 ± 0.68	0.48	0.89	1.01
			(0.37 - 0.73)	(0.73 - 0.96)	-0.03

Frontal moments (Nm·kg^{-1})	Peak adduction	-0.06 ± 0.17	0.12 (0.09 - 0.18)	0.21 (-0.30 - 0.62)	0.49 0.01
	Peak abduction	0.00 ± 0.21	0.15 (0.11 - 0.22)	0.43 (-0.06 - 0.75)	0.61 -0.10
Transverse (°)	IC	-0.53 ± 9.22	6.52 (4.93 - 9.88)	0.56 (0.11 - 0.82)	0.92 -1.43
	Midstance	-1.40 ± 10.05	7.11 (5.37 - 10.77)	0.41 (-0.09 - 0.74)	0.88 -3.96
	TO	-0.83 ± 8.63	6.10 (4.61 - 9.24)	0.66 (0.26 - 0.86)	1.17 -0.83
	Peak internal rotation	-0.87 ± 8.66	6.12 (4.63 - 9.28)	0.65 (0.25 - 0.86)	1.17 -0.90
	Peak external rotation	-0.98 ± 10.41	7.36 (5.56 - 11.16)	0.46 (-0.02 - 0.77)	0.90 -3.54
	ROM	0.11 ± 3.15	2.23 (1.68 - 3.38)	0.81 (0.54 - 0.93)	0.74 6.68
Transverse moments (Nm·kg^{-1})	Peak internal rotation	0.02 ± 0.16	0.11 (0.08 - 0.17)	0.60 (0.17 - 0.84)	0.97 0.04
	Peak external rotation	0.01 ± 0.01	0.01 (0.01 - 0.02)	0.84 (0.60 - 0.94)	0.92 0.00

The largest typical error for the hip joint in the sagittal, frontal and transverse plane was 2.18°, 1.95° and 6.16°, respectively, at key positions of initial contact, midstance and peak-external rotation, respectively. The largest typical error for the knee joint in the sagittal, frontal and transverse plane was 1.95° 3.86° and 6.01°, respectively, at key positions of toe off, midstance and peak-internal rotation, respectively. The largest typical error for the ankle joint in the sagittal, frontal and transverse plane was 3.54°, 1.27° and 7.36°, respectively, for the following measures in stance, ROM, peak adduction and peak-external rotation, respectively.

The largest typical error for peak-joint moments at the hip joint in the sagittal, frontal and transverse plane was 0.54, 0.28 and 0.09 Nm·kg^{-1}, respectively. The largest typical error for peak-joint moments at the knee joint in the sagittal, frontal and transverse plane was 0.24, 0.37

and 0.07 Nm·kg^{-1}, respectively. The largest typical error for peak-joint moments at the ankle joint in the sagittal, frontal and transverse plane was 0.13, 0.15 and 0.11 Nm·kg^{-1}, respectively.

Table 4. 10 Within-session, between-session and between-day measurement error of peak-forefoot pronation angle during the stance phase of overground endurance running in 23 recreational runners.

Research design	Variable	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Within-session (°)	Peak forefoot pronation	0.32 ± 1.02	1.01 (0.79 - 1.39)	0.96 (0.87 - 0.99)	-
Between-sessions (°)	Peak forefoot pronation	1.70 ± 6.47	4.57 (3.34 - 7.53)	-0.13 (-0.64 - 0.45)	0.66 0.6
Between-day	Peak forefoot pronation	-0.32 ± 5.12	3.62 (2.64 - 5.96)	0.51 (-0.06 - 0.83)	1.03 -2.07

Table 4.10 reports peak-forefoot pronation angle, within a single-occasion, between two testing occasions separated by 30 minutes and between two days separated by 48 hours. Within a single-testing session the typical error was 1.01°. Between two testing occasions separated by 30 minutes the typical error was 4.57°. Between two test sessions separated by 48 hours the typical error was 3.62°.

4.4 Discussion

The purpose of this study was to quantify measurement error of three-dimensional kinematic and kinetic measures during overground ER, within a single testing session, between two testing sessions separated by 30 minutes on a single day and between two days separate by 48 hours. The largest typical error for kinematic variables of interest were 3.41°, 7.62° and 7.36° for within-session, between-sessions, and between-days respectively. The greatest typical error of kinetic variables of interest was 0.59 Nm·kg^{-1}, 0.44 Nm·kg^{-1} and 0.54 Nm·kg^{-1}, for

within-session, between-sessions, and between-day respectively. It is important to note that the interpretation of reliability is relative to the magnitude of the signal for a specific research question. Thus, conclusions whether data reported are reliable or not should be based on comparisons to previous work that report relative reliability (ICC) or work that reports an absolute magnitude of differences between two specific populations and how this compares to absolute data variability in this study.

4.4.1Within-session

Within-session 3-D kinematic measurement error was excellent for reported kinematic measures of interest during the stance phase of gait and were comparable to previous investigations that report excellent relative within-session reliability for ER kinematics (Ferber, Davis, Williams, et al., 2002; Queen et al., 2006). In a study investigating the reliability of overground ER, Ferber, Davis, Williams, et al. (2002) used intra-class correlations to assess reliability of peak angles and reported excellent reliability at the hip (>0.95) knee (>0.92) and ankle (>0.92). Although comparable, intra-class correlations were marginally smaller (tables 4.1, 4.2, 4.3). This could be explained by the difference in biomechanical models used. The 'Plug-In Gait' model uses small wands that extend from the lateral aspects of the thigh and lower leg. Unlike the strapped-tracking marker shells used in Ferber, Davis, Williams, et al. (2002), wands flex and resonate at impact explaining the trend for a small increase in artefact during stance. Collectively, within-session kinematic results reported in this study can be considered reliable despite small differences from a previous work using an alternative biomechanical model.

Peak joint moments also displayed good within-session reliability. However, the results of this study differ from those of Ferber, Davis, Williams, et al. (2002) who reported within-session kinetic intra-class correlation scores greater than or equal to 0.73, 0.86 and 0.85 for the hip, knee and ankle, respectively. These intra-class correlation values are higher than this study (see table 4.1, 4.2 and 4.3). The increased measurement error of peak-joint moments could be explained by inverse-dynamic calculations. Kinematic data in this study were more variable than that of Ferber, Davis, Williams, et al. (2002), and because kinematics are double

differentiated to calculate peak-joint moments, it follows that the increased measurement error in kinematics will augment measurement error in peak joint moments and explain the smaller intra-class correlations.

Although relative measurement error in this study is worse than that of Ferber, Davis, Williams, et al. (2002), it is important to remember when expressed in the units of interest, measures often used to evaluate changes in running technique such a plantarflexion angle at initial contact demonstrate excellent reliability. In a study by Heiderscheit et al. (2011) participants were asked to change their preferred stride frequency from -10% to +10% of their preferred stride frequency, following, there was a 6.7° increase in plantarflexion angle when running with a higher stride frequency. This is substantially larger than the 2.88° typical error reported in this study. Overall, within-session-kinetic data collected in this study were reliable and kinematic variables associated with change in running technique and injury had small within-test error when expressed in the units of the measurement tool.

4.4.2 Between-session

Three-dimensional kinematic-measurement error between two sessions separated by 30 minutes was generally greater than within-session measurement error. This is consistent with previous ER research (Ferber, Davis, Williams, et al., 2002; Queen et al., 2006). For example, when investigating the reliability of knee kinematics in both self-selected and speed-matched ER trials, Queen et al. (2006) reported larger measurement error in all planes between sessions. Specifically, sagittal, frontal and transverse plane correlations for peak-joint angles at the knee ranged from 0.81 - 0.97 within session, and 0.62 - 0.92 between sessions. This finding is supported by Ferber, Davis, Williams, et al. (2002) who also reported an increase in measurement error between sessions. Current relative reliability for between-session kinematics were similar to the findings of Ferber, Davis, Williams, et al. (2002). An explanation for increased variability between two sessions that does not pertain to within-session testing is the influence of erroneous-anatomical-landmark identification and wand alignment when reapplying retroflective markers (Della Croce et al., 2005; Leardini et al., 2005). For example, in a modelling study investigating the effects of systematically translating

marker positions in the anterior-posterior and vertical axis, Osis, Hettinga, Macdonald, and Ferber (2016) reported that an anterior-posterior translation of 10mm in the placement of the lateral-malleoli marker produced a peak-ankle angle change of 7.59° in the transverse plane. This observation is consistent with Della Croce et al. (2005) who reported that inaccurate identification of joint axis, as a result of poor marker placement, introduces cross talk to kinematic measurements. In summary, between-session kinematic data were similarly reliable when compared to previous work and small errors in marker re-application likely underpin the slightly larger variability between-sessions than within a session.

There was no consistent pattern dictating whether 3-D kinetic variables were more reliable within session or between sessions. Some kinetic variables were more reliable within a session than between sessions and vice versa. This is in contrast to previous research that consistently reports kinetics such the ground-reaction force and peak–joint moments as more reliable within session (Ferber, Davis, Williams, et al., 2002; Queen et al., 2006). This trend would make sense with errors introduced by erroneous-marker replacement exacerbated by inverse-dynamic calculations. A possible explanation for contrasting results is that, relative to the natural measurement error that exists within a single-testing session, the comparison of means between two sessions could have less error than previous between-session studies because of the short time between sessions (30 minutes). This work of Ferber, Davis, Williams, et al. (2002) and Queen et al. (2006) both separated testing sessions by one week. Separating testing by one week is likely to introduce a greater degree of physiological variability and could explain the general increase in measurement error between sessions. However, in context, the absolute value of both within-session and between-session typical error is small relative to previously reported minimal-important changes. For example, an increased peak-knee adduction moment of 0.39 $Nm{\cdot}Kg^{-1}$, differentiated injured and uninjured endurance runners (Dudley et al., 2017). This is far greater than within-session (0.19 $Nm{\cdot}Kg^{-1}$) and between session measurement error (0.17 $Nm{\cdot}Kg^{-1}$) reported in this study. In summary, depending on the variable of interest, kinetic data were either more or less reliable within-session or

between-session when separated by 30 minutes, however, when reported in absolute units, the error was small relative to thresholds associated with injury.

4.4.3 Between-day

Between-day kinematic error was comparable to previous work. For example, Queen et al. (2006) reported comparable coefficients for peak knee joint angles in the sagittal (0.92), frontal (0.62) and transverse (0.76) planes. Queen et al. (2006) and Ferber, Davis, Williams, et al. (2002) both reported that within-session measurement error of kinematics was generally smaller than between-day error. Specifically, when observing the effects of self-selected running speed on the reliability of knee joint kinematics, Queen et al. (2006) reported that measurement error was significantly higher in all planes between days than within-session trials on the same day. Although not consistent on all measures in this study, this trend is most clear when comparing within-session transverse plane measures to between-day transvers measures. A possible explanation for increased error between days is marker reapplication. As previously noted, Osis et al. (2016) demonstrated a 10mm anterior/posterior translation in marker location equated to a 7.59° change in peak-ankle joint angle; this demonstrates the sensitivity of measurement error to marker placement. As the maximum kinematic error in this study was less than the largest error reported by Osis et al. (2016), it can be concluded that measurement error was small. In summary, between-day measurement error for kinematic measures were small and comparable to previous work.

Between-day data kinetic measures were in some cases similar to within-session data but there was a trend for reduced reliability in non-sagittal plane kinetic measures. When reported in absolute terms, error in peak-joint moments were small between days. Intra class correlations calculated for comparative purposes were less than previously reported between-day-peak-joint-moment measurement error (Ferber, Davis, Williams, et al., 2002). As previuously disscussed a possible explanation might be the choice of biomechanical model. In this study, the 'Plug-In Gait' model used small wands that extend from the lateral aspects of the lower limbs. As there is no clear anatomical reference for the placement of wands, between-day measurement error could be larger as a result of problems identifying a consistent location

between days. Taking this further, if wand location is inconsistent, it is likely that the reported effects of phasic muscle action on wands will differ between-days (Manal et al., 2000). Remembering that inverse dynamic calculations exacerbate kinematic error, error in wand placement might explain the decreased intra-class correlations compared to previous work using a cluster marker system. Alternatively, from a statistical perspective, a reduced spread of scores would also decrease intra-class correlations; however, Ferber, Davis, Williams, et al. (2002) did not report standard deviations, therefore a comparison between spread of scores was not possible. Although intra-class correlations in this study were not as large as previous work, possibly because of differences in biomechanical models, when expressed in absolute terms, the reported kinetic error between days was small. In the context of 3-D kinetic measures in ER, absolute error reported was sufficiently small to detect magnitudes of difference associated with footwear manipulation. As reported by Sinclair (2014) when comparing footwear conditions and their impact on ankle joint loading and its relation to injury mechanics the plantarflexion moment significantly increased by 0.58 $Nm \cdot Kg^{-1}$ when changing from conventional to barefoot. This value is larger than between-day measurement error found in this study (0.13 $Nm \cdot Kg^{-1}$). Collectively, absolute between-day peak-joint-moment error was sufficiently small to detect differences that distinguish the effects of footwear conditions on ankle joint loads.

4.3.4 Comparison between joints

Results suggest the ankle joint has the least kinematic measurement error within session and between sessions. This is particularly apparent when comparing between-session-kinematic ankle data to the knee and hip. For example, in the transverse plane, measurement error of the ankle at initial contact, mid-stance and toe-off was equal to or less than 3.87°, whereas at similar time points, the knee and hip measurement error was less than or equal to 5.73° and 5.05° respectively. The between-session trend agrees with the findings of Ferber, Davis, Williams, et al. (2002) who reported greater intra-class correlation scores for the ankle compared to the hip. However, Ferber, Davis, Williams, et al. (2002) reported that the knee had the greatest intra-class correlation score. In a technical study investigating the differences

between skin and bone mounted markers, Reinschmidt et al. (1997) reported soft-tissue artefact at the thigh accounted for the majority of discrepancies between external-skin and skeletal-knee motion. With the addition of wand markers in the current study, this provides a technical explanation why the knee joint was not more reliable than the ankle joint. Contrastingly, between-day kinematic error was similar across the joints assessed. A possible explanation is that the combined effect of measurement error and physiological variability between days was large enough to supersede any joint-specific trend within a single day (Della Croce et al., 2005). Generally, there was a trend for peak-joint moments at the ankle joint to report the smaller measurement error compared to the knee and hip. In contrast, Ferber, Davis, Williams, et al. (2002) reported larger average intra-class correlations for peak moment data between days at the hip (0.86) and knee (0.84) compared to the ankle (0.7). Factors that might explain this finding are the reduced effects of soft-tissue artefact and wand placement error at the ankle. Increased soft-tissue mass at the thigh relative to the foot segment is likely to contribute to increased soft-tissue artefact for the hip-and knee-peak joint moments compared to ankle peak joint moments (Reinschmidt et al., 1997). Collectively, this suggests that when using a wand-based model and measuring multiple time points across stance, ankle kinematics within session and between sessions are the most reliable for reported comparisons because of reduced soft-tissue artefact and subsequent ease of accurate-marker placement. However, when separated by 48 hours the additional of human physiological variability was enough to mask a previously reported between-session trend. Ankle peak-joint moment data when reported in absolute terms was smaller than the hip or knee likely because of reduced mass and ease of anatomical landmark location.

In agreement with previous research, transverse-plane kinematic measures had the greatest measurement error within a session, between-sessions and between-day (Ferber, Davis, Williams, et al., 2002; Manal et al., 2000). In landmark work assessing the effects of erroneous marker placement and the subsequent effects on joint kinematics, Della Croce et al. (2005) concluded that when a joint predominantly performs in one plane, for example the sagittal plane when running, small rotations out of this plane are strongly influenced by erroneous

marker placement. This could explain why the transverse plane reported the largest kinematic measurement error between-sessions/days. In addition, Manal et al. (2000) also reported that soft-tissue artefact effects the transverse plane more than the primary plane of motion (sagittal error ±2° and transverse ±4°), citing phasic-muscle actions acting on mid-segment wands as the underlying cause of the increased-transverse-plane measurement error within a session. This suggests that similar to previous reports, the transverse plane has the largest measurement error as a result of an increased sensitivity to erroneous marker placement, soft-tissue artefact and phasic muscle contractions.

Relative measures (e.g. range of motion) are generally more reliable than absolute measures (e.g. angle at initial contact) for between session and between day comparisons. Research by Ferber, Davis, Williams, et al. (2002) reported similar findings, hypothesising the misalignment of markers between sessions as the primary explanation for this observation. Specifically, erroneous marker-placement introduces an offset between testing occasions for absolute measures. Consistent with this explanation, Kadaba et al. (1989) reported that when the hip-joint centre was translated by 10mm, an off-set in kinematic curves was observed, but curve shape was unchanged. This evidence supports the argument that between-sessions/days the reliability of absolute data decreases more than excursion data because of a kinematic offset introduced by erroneous marker placement. Conversely, within-session absolute measures demonstrated greater reliability than excursion data. A possible explanation is that absolute data without the error introduced by marker reapplication was more reliable than excursion data. This is logical given that the calculation of excursion data relies on two variables (maximum and minimum), not one. In summary, within-session excursion measurement error was greater than absolute data possibly because of the reliance on more than one measure, however, between-session or between-day, excursion data was more reliable, possibly because these measures are less affected by the offset introduced by erroneous marker placement.

4.4.5 Limitations

Unlike previous work, the footwear used by participants in this study was not controlled. However, as the main aim of this study was to assess the measurement error in ER biomechanics, and because the timescale of habituation to new footwear has not yet been reported, allowing participants to use familiar footwear ensured results could not be influenced by novel footwear. For data deduced from the 'Oxford-Foot Model', it is important to note that this data represents the motion of the shoe; therefore, some motion of the forefoot is likely occluded by footwear. A possible solution would have been to cut holes in participants' footwear, however, this was not feasible. It is also likely that the findings of this study are specific to the biomechanical models used. It could also be suggested that five trials might provide an improved representation of within participant reliability, and subsequently comparisons between time points. Conversely, Hopkins (2000) suggests three trials is enough to provide a precise estimation of reliability, and three trials was enough to produce reliable kinematic and kinetic data for Diss (2001) who was used a lower sampling frequency (50Hz) compared to the current study method (200Hz). In light of these limitations, extrapolation of these results to other studies that do not share a similar experimental design should be done cautiously.

4.4.6 Conclusion

The methods used in this study produce reliable measurements of 3-D kinetic and kinematic variables that can characterise overground ER. The typical error data reported can be used to infer minimal-detectable-change thresholds that define a 'true effect' and can also be used to inform sample size calculations for intervention studies using a similar population and timescale between sessions. Finally, peak-knee adduction moment, the key measure of interest of this thesis reported reliable data within-session, between-session and between-day when compared to work that describes mean differences between injured and uninjured samples.

5.1 Sagittal lower limb habituation in recreational runners performing overground endurance running while barefoot, in minimal and maximally-cushioned shoes.1

5.1 Introduction

Barefoot running and minimal-shoe running have been shown to reduce knee joint loading and the LR of the vertical component of the GRF (Divert et al., 2005; Lieberman et al., 2010; Sinclair, 2014; Squadrone & Gallozzi, 2009). Reduced LR and improved knee joint loading have been associated with reduced injury rates in runners (Dudley et al., 2017; Zadpoor & Nikooyan, 2011). Conversely, Willy and Davis (2013) and De Wit et al. (2000) report minimal footwear and barefoot running are associated with an increased vertical LR. Lack of a consensus on the effects of barefoot and minimal shoe running on measures associated injury complicate interpretations of the value of gait retraining protocols and footwear choices for injury avoidance.

A possible explanation for inconsistencies in findings might be a lack of consistency in habituation protocols in related studies. If participants are not habituated to a novel footwear condition, it is difficult to conclude whether differences observed between conditions are representative of habituated running mechanics or a participant's initial response to a novel-footwear condition. As previously reported, issues with running habituation protocols can be divided into one of three categories (Moore & Dixon, 2014), (1) failing to report habituation times (Fredericks et al., 2015; Hanson, Berg, Deka, Meendering, & Ryan, 2011; Squadrone & Gallozzi, 2009); (2) a set window of time to habituate to a footwear condition (Bonacci et al., 2014; Perl et al., 2012); or (3) participants providing verbal confirmation that they had habituated to a running condition (Riley et al., 2008). Time to habituate requires a definition of what 'habituated' means. This should be a predetermined level of within-trial variability that constitutes a 'stable' movement pattern. Such approaches are yet to be used to quantify time to habituation while running overground in novel footwear, but could help establish a consensus for practitioners and researchers alike.

In addition to studies of barefoot and minimal footwear, there has recently been an influx of maximally-cushioned footwear in the running-shoe industry. Maximally-cushioned running shoes are opposite to minimal footwear with excessive cushioning in the midsole (Sinclair et al., 2016). Maximally-cushioned running shoe research is in its infancy and the implications of such designs warrant further investigation. However, like minimal footwear, knowledge of the time for a participant to habituate to overground running in maximally-cushioned shoes does not exist, but is needed to ensure results are representative of maximally-cushioned running shoe biomechanics and not a participant's initial response to a novel-footwear condition.

A series of steps are necessary when running in a novel footwear condition to ensure data collected represents habituated biomechanics (Divert et al., 2005). However, time to habituation while performing overground running in novel footwear conditions are yet to be reported. In a study using participants unfamiliar with treadmill running, it was reported that participant's kinematics were considered stable after six minutes (Lavcanska, Taylor, & Schache, 2005). This suggests that when runners are exposed to running where only one variable is changed, in this case, the novelty of the treadmill, habituation can be achieved quickly. Conversely, when investigating the time to habituation while running barefoot on a treadmill, Moore and Dixon (2014) reported that 20 minutes of treadmill running was necessary for the majority of sagittal-plane kinematics to stabilise, reporting an *r* value greater than 0.8 and no significant difference between minute 20 and 21. This suggests that habituating to a novel footwear condition takes longer than simply acclimating to a treadmill without a change in footwear. Noteably, Moore and Dixon (2014) used treadmill running to examine habituation to barefoot running. Treadmills have been reported to substantially alter running kinematics compared to overground running (Nigg et al., 1995). An investigation of time to achieve stable kinematics in overground running in novel footwear is necessary for future research using novel footwear interventions. Time to habituation while running barefoot, in minimal shoes and maximally-cushioned shoes is yet to be reported in overground running.

The aim of this study was to investigate the time taken for 3-D lower limb kinematics to become stable in the sagittal plane for a sample of recreational runners during overground running while barefoot, in minimal and maximally-cushioned shoes, where 'stable' was defined as variability equal to previously-determined within-trial measurement error (chapter four). It was hypothesised that all measures of interest would be stable by the end of a 30-minute habituation run.

5.2 Method

5.2.1 Participants

With institutional ethics approved, 15 volunteers participated. Ten male and five female participants had mean and SD age, stature and mass of 25 ± 6 yrs, 1.74 ± 0.01 m and 69 ± 10.9 kg. Inclusion criteria were aged 18-45 years, no previous experience of barefoot, minimal, or maximally-cushioned shoe running, and participation in ER more than once per week as part of their exercise regime with one run lasting at least 30 minutes. Participants were excluded if they had an injury to the lower limbs in the previous six months or any condition that could affect their normal running gait.

5.2.2 Experimental design

A repeated measures design assessed kinematic habituation of recreational endurance runners in three novel footwear conditions (barefoot, minimal and maximally-cushioned shoes (See 3.3.3). Conditions were performed on separate days with sessions separated by 24 hours and conducted at a similar time of day within each participant. The order of sessions was counterbalanced and participants were instructed to be well rested before each testing session. Testing took place on an indoor running track and 3-D biomechanics lab, where they ran around the perimeter of a 56m straight running track that was 6m wide. Participants then ran from the indoor running track and through a calibrated 3-D biomechanics lab three times in one-minute windows every five-minutes during a 30-minute continuous run (0, 5, 10, 15, 20, 25 and 30 minutes). Participants were asked to run at a speed they would describe as "an endurance pace that could be comfortably sustained for 45 minutes".

5.2.3 Procedures

Participants were provided with appropriate clothing, anatomical measures taken, and a 3-D biomechanics analysis suite was calibrated as described in the chapter three (3.3.1 and 3.3.2).

A series of retroflective markers were attached to participants in a full-body 'Plug-In gait' and 'Oxford-Foot Model' formation as described in chapter three, participant preparation for 3-D analysis (3.3.2) to facilitate the assessment of lower-limb biomechanics.

Kinematic and kinetic data were captured by 14 calibrated infrared cameras (T10/20, Vicon MX, Oxford, UK) and four force plates (OR6-7, AMTI, Watertown MA, USA). Signals were captured and imported as described in chapter three, section 3.3.1.

5.2.5 Data analysis

Data analysis and processing was undertaken in the 3-D motion analysis software in line with the processes described in chapter three, section 3.3.6. Sagittal plane kinematic-joint angles of interest were derived as described in chapter 3 (3.3.6) and were then exported to Microsoft Excel (Microsoft, USA). Only data in the sagittal plane was analysed following sagittal plane kinematics reports the greatest reliability (Queen et al., 2006), and this being in line with previous work (Moore and Dixon, 2014). When GRF data was not available, and because trials were restricted to overground running in a one-minute window of time, initial contact and toe-off were in some cases identified by the visual identification of post-filtered marker trajectories (z-axis) from graphical outputs based on Vicon Nexus (Vicon, Oxford, UK) data.

5.2.6 Statistical analysis

Using SPSS (version 24.0, SPSS Inc., Chicago, IL), a Kaplan-Meier survival analysis was undertaken using each time a participant crossed a predetermined threshold of reliability as an event. As data were collected within a single session, within-session reliability data (chapter four) were used to set thresholds of stability. The Kaplan-Meier model, a model that provides a precise estimation of the average time taken for an event to occur within a given time period calculated the mean time taken for lower limb sagittal plane kinematics to demonstrate variability equal to or within that of previously collected within-session ER reliability data.

Additionally, after assumptions of normality and uniformity of error were verified mean differences in running speed between different running conditions as well as different time points (0, 15 and 30) were estimated for using 90% confidence intervals.

5.3 Results

Tables 5.1, 5.2 and 5.3 show the mean ± SE for time to habituation at the hip, knee and ankle joint, respectively, for a sample of participants performing overground running barefoot, in a minimal shoe and a maximally-cushioned shoe. One participant later reported experience running barefoot and in a minimal shoe and their data were only included for the maximally-cushioned shoe analysis. Example Kaplan-Meier plots at initial contact, midstance and peak dorsiflexion for the ankle joint are illustrated below (figure 5.1).

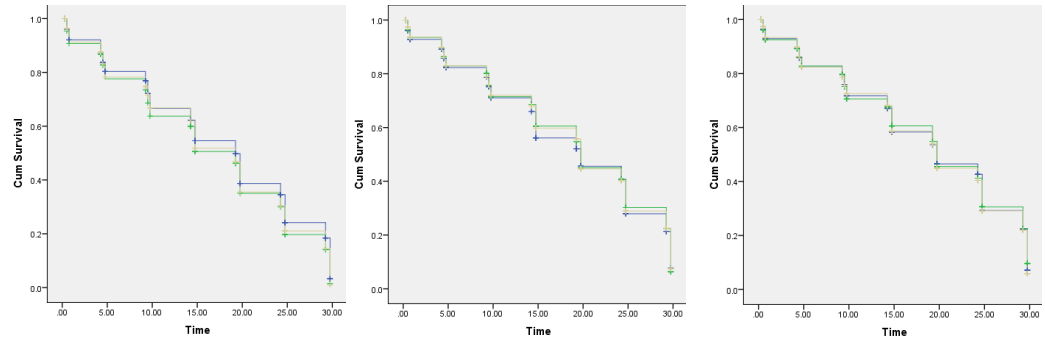

Figure 5. 1 Kaplan-Meier plot of a survival analysis illustrating the time sequence until participants ankle angle at initial contact, midstance and peak dorsiflexion (left to right) reported a level of trial-to-trial stability equal to or less than previously reported within-session reliability whilst running barefoot (blue), in minimal (green) and maximally-cushioned footwear (yellow).

Table 5. 1 Mean ± SE for time to habituation for hip kinematics in the sagittal plane. Note: BF: barefoot, MS: minimal shoe, MCS: maximally-cushioned shoe.

Parameter	Condition	Mean ± SE (minutes)
Hip flexion at initial contact	BF	18.69 ± 0.61
	MS	18.76 ± 0.61
	MCS	17.92 ± 0.59
Hip flexion at midstance	BF	20 ± 0.62
	MS	20.91 ± 0.62
	MCS	19.75 ± 0.60
Peak hip flexion	BF	18.03 ± 0.61
	MS	18.04 ± 0.61
	MCS	17.29 ± 0.59
Hip range of motion	BF	18.12 ± 0.63
	MS	18.95 ± 0.61
	MCS	18.15 ± 0.60

Habituation time at the hip ranged from 17.29 – 20.91 minutes. The range of habituation times were similar in maximally-cushioned footwear (17.29 – 19.75 minutes), minimal footwear (18.04 – 20.91minutes) and barefoot (18.03 – 20 minutes).

Table 5. 2 Mean ± SE for time to habituation for knee kinematics in the sagittal plane. Note: BF: barefoot, MS: minimal shoe, MCS: maximally-cushioned shoe.

Parameter	Condition	Mean ± SE (minutes)
Knee flexion at initial contact	BF	18.8 ± 0.62
	MS	20.39 ± 0.62
	MCS	20.26 ± 0.63
Knee flexion at midstance	BF	19.97 ± 0.65
	MS	20.15 ± 0.65
	MCS	19 ± 0.61
Peak knee flexion	BF	19.51 ± 0.64
	MS	18.49 ± 0.62
	MCS	18.10 ± 0.6
Knee range of motion	BF	18.44 ± 0.62
	MS	19.87 ± 0.63
	MCS	20.47 ± 0.61

Habituation time at the knee ranged from 18.44 – 20.47 minutes. Similar ranges of habituation time were observed for maximally-cushioned footwear (18.1 - 20.47 minutes), minimal footwear (18.49 – 20.39 minutes) and barefoot (18.44 – 19.97 minutes).

Table 5. 3 Mean ± standard error for time to habituation for ankle kinematics in the sagittal plane. Note: BF: barefoot, MS: minimal shoe, MCS: maximally-cushioned shoe.

Parameter	Condition	Mean ± SE (minutes)
Ankle dorsiflexion at initial contact	BF	17.49 ± 0.61
	MS	16.54 ± 0.6
	MCS	16.9 ± 0.58
Ankle dorsiflexion at midstance	BF	18.45 ± 0.62
	MS	18.84 ± 0.62
	MCS	18.76 ± 0.6
Peak dorsiflexion	BF	18.72 ± 0.62
	MS	18.78 ± 0.63
	MCS	18.69 ± 0.6
Ankle range of motion	BF	19.75 ± 0.63
	MS	19.62 ± 0.63
	MCS	19.7 ± 0.6

Habituation time at the ankle ranged from 16.54 – 19.75 minutes. Footwear did not clearly effect habituation times between conditions with similar values reported barefoot, in minimal, and in maximally-cushioned shoes (17.49 – 19.75; 16.54 – 19.62; 16.9 – 19.7 minutes respectively).

Running speed was clearly different between running conditions. Specifically, average barefoot running speed (2.44 $m \cdot s^{-1}$) was clearly slower when compared to minimal (2.59 $m \cdot s^{-1}$) (90% CI -0.24 to -0.07 $m \cdot s^{-1}$) and maximally cushioned footwear (2.62 $m \cdot s^{-1}$) (90% CI -0.25 to -0.12 $m \cdot s^{-1}$). Minimal and maximally-cushioned footwear were not clearly different.

Speed was clearly less at the beginning of the habituation period (2.5 $m \cdot s^{-1}$) compared to the mid-point (2.57 $m \cdot s^{-1}$) (90% CI -0.14 to -0.01 $m \cdot s^{-1}$) and the end of the habituation period (2.59 $m \cdot s^{-1}$) (90% CI: -0.16 to -0.02 $m \cdot s^{-1}$). There was no differences between the mid-point and the end of the habituation speed.

5.4 Discussion

This study investigated the time taken for sagittal-plane lower-limb kinematics to stabilise while performing overground running barefoot, in minimal and maximally-cushioned shoes, where 'stable' was defined as variability equal to or less than previously determined within-trial measurement error (chapter four). In line with the hypothesis of the study, all measures of interest stabilised within a 21-minute window. This suggests future work that provides 21 minute for measures of interest to stabilise will draw conclusions from stable kinematics. Mean times for stabilisation to occur at the hip, knee and ankle joint in the sagittal plane ranged from 17.29 – 20.91, 18.44 – 20.47 and 16.54 – 19.75 minutes, respectively.

In support of the study hypothesis, sagittal plane hip measures were stable after 21 minutes of overground running with similar ranges of time regardless of footwear condition (table 5.1). Findings suggest that at least 21 minutes of overground running is necessary for sagittal plane hip kinematics to stabilise when running barefoot, in minimal or maximally-cushioned footwear. This finding is in agreement with previous work by Moore and Dixon (2014) who report that sagittal plane kinematic hip measures were consistent after 20 minutes of barefoot treadmill running in a sample of runners who had little to no experience in barefoot running. This supports the opinion that future investigations should provide habituation times that that are appropriate to task complexity. For example, previous work has reported six minutes was necessary for treadmill habituation in novice treadmill runners (Lavcanska et al., 2005); whereas more complex tasks, such as footwear manipulation, as in this study, require more time. A potential explanation why long time periods were necessary to habituate is based on the respective decrease/increase in somatosensory feedback when running in maximally-cushioned footwear or barefoot/minimal footwear. A change in somatosensory feedback could elicit changes in ankle and knee joint kinematics and subsequently hip joint kinematics in

response to such changes in distal lower limb joints. In context, this data questions the interpretation of previous work such as Sinclair, Greenhalgh, Brooks, et al. (2013) that investigated sagittal hip angles while running barefoot and in minimal footwear where novice barefoot runners were provided with only five minutes to habituate. It could be argued that conclusions from work with insufficient habituation time might be representative of the initial response to barefoot running, capturing a learning effect and not habituated biomechanics.

Sagittal plane knee data also supported the study hypothesis with all knee measures achieving stability by 21 minutes with similar ranges across footwear conditions (table 5.2). This finding is again in line with previous work by Moore and Dixon (2014) that report familiarisation for knee joint kinematics by 20 minutes. Interestingly, the hip, knee and ankle joint required similar durations to stabilise. Previous work by Moore and Dixon (2014) suggested reduced variability in running mechanics might be a product of increased muscular fatigue as a function of time, however, this was accounted for in the current study by the instruction to run at an endurance pace that could be comfortably maintained for 45 minutes. An alternative explanation is that sagittal-plane hip, knee and ankle joint kinematics share an inherent interaction which saw one stabilise a short time after the other. Future work should investigate the relationship between hip, knee and ankle joint kinematics during habituation following similar times to stabilisation.

The knee joint is the most common site of injury for overground endurance runners (van Gent et al., 2007) and research often uses footwear as a means to reduce knee joint loads. For example, Bonacci et al. (2014) investigated the effect of overground barefoot running on a modelled patellofemoral joint load that relied on the sagittal knee joint angle during stance and concluded that running barefoot induced a reduced patellofemoral load. However, with only five overground running trials allocated before barefoot data capture began, conclusions from such a study design might be questionable in light of current study findings. Future work that relies on sagittal knee joint angles should quantify habituation in their sample to ensure biomechanics are representative of the footwear condition and not an initial response to a novel condition, or alternatively use data from the current study as a guideline.

Sagittal plane ankle data support the study hypothesis with all kinematic measures attaining stability by 20 minutes, with similar ranges of times across footwear conditions (see table 5.3). An explanation for such long time to stabilisation at the ankle joint could be based on localised change in somatosensory feedback underfoot when running barefoot and in minimal footwear. An increase in somatosensory feedback compared to a standard running shoe possibly caused large and immediate changes in running technique in an attempt to establish a technique that is less uncomfortable. An immediate response to the removal of cushioned footwear was demonstrated by Gruber et al. (2012) who reported when a running surface similar to a conventional-cushioned shoe was removed (EVA foam), the majority of barefoot participants made immediate changes from a RFS (80%) to a non-RFS (65%). This immediate transition is argued to be a response to the interaction between a non-compliant running surface and the subsequent increased pressures underfoot stimulating fast acting mechanoreceptors to induce increased plantarflexion and reduce plantar pressure at the heel (De Wit et al., 2000). Although immediate changes are likely an attempt to prevent discomfort, consistently larger trial to trial variability might have occurred in an attempt to optimise energy expenditure or reduce fatigue in specific muscles. Moore and Dixon (2014) support this suggestion reporting 20 minutes was necessary for sagittal plane ankle joint familiarisation for barefoot treadmill running. It is important to note larger trial-to-trial variability might also occur in maximally-cushioned footwear in an attempt to optimise energy expenditure or negate fatigue while running in a novel footwear condition. This potentially explains the similar range of time necessary for kinematic stability in maximally-cushioned footwear, however without a control group it is difficult to conclude. However, the immediate changes associated with barefoot running might not have occurred in the maximally-cushioned shoe following the unlikely perception of discomfort during impact. Future work investigating the effects of footwear on sagittal ankle joint kinematics should provide sufficient time to habituate to barefoot running, minimal and maximally-cushioned footwear, given that this study suggests long periods of time are necessary to produce stable sagittal ankle kinematics.

5.4.1 Limitations

The current study was limited to indoor overground running. The time for runners to habituate to other types of surface with different and varying levels of stiffness such as grass and outdoor running tracks warrant further investigation. The experimental design did not provide insight into how much habituation related to the footwear and how much related to the laboratory environment, however it did provide an analysis of the time taken to encapsulate both. The sample recruited in the current study were recreational endurance runners and the extrapolation of current findings to elite runners transitioning to either barefoot, minimal or maximal footwear should be made with caution. Running speed was clearly different between conditions, suggesting the barefoot running condition covered less distance compared to shod conditions. As participants speed was measured through the gait lab alone and not whilst running on the track, distance covered was not measured. Future studies should attempt to control the effect of distance ran and investigate whether this influences time to habituation. There is a variety of minimal and maximally-cushioned footwear companies in the current market and with this comes a variety of shoe designs. Future work should consider the consistency of the current findings across other minimal and maximally-cushioned shoe brands.

5.4.2 Conclusion

Results suggest that a 21-minute overground endurance run is sufficient for a sample of recreational runner to attain stable sagittal plane kinematics in a variety of novel footwear conditions. Hip, knee and ankle measures appear to stabilise after 21 minutes regardless of footwear condition. Care should be taken when interpreting the conclusions of work that does not report adequate habituation times as conclusions might be derived from the initial variable response to a novel running condition and not representative of stable kinematics in the novel footwear condition.

6.0 Differences in pronation and peak-knee adduction moment and their relationship with forefoot structure during overground running in barefoot, minimal and structured-cushioned shoes.

6.1 Introduction

Forefoot structure of habitually barefoot individuals is different from those who are habitually shod (D'AoUt et al., 2009; Shu et al., 2015) and is characterised by greater forefoot width and more equal distribution of pressure (D'AoUt et al., 2009). An abducted hallux and the associated greater distance between the hallux and second toe accounts for the increased forefoot width. Shu et al. (2015) reported a significantly more abducted hallux angle in a population that walked and exercised barefoot (3.42° ± 3.5), compared to a western shod population that walked and exercised in conventional footwear that compressed the metatarsals (10.3° ± 5.4). It has been suggested that hallux structure and position has evolved through natural selection to control the progression of the body weight during stance and that a compromised foot structure will impair stability and control of body weight (Wilkinson, Stoneham, & Saxby, 2018). This assertion is supported by Chou et al. (2009) who restricted the function of the hallux by splinting it in 30 degrees of dorsiflexion, and reported that 'directional control' (a measure of the ability to direct the COP) significantly worsened in the forward-left, forward, and forward-right directions in a single-leg balance task. This observation is logical given a mechanically-compromised hallux would no longer oppose the natural pronation of the foot when weight bearing. Additionally, phalange width and ball-of-foot width might also affect peak pronation as compressed foot width will compromise/reduce the functional-axis width. Hoogvliet et al. (1997) have previously reported that reduced 'functional-foot breadth' reduced a participant's ability to control the COP, invoking larger amplitudes of compensatory frontal-plane foot motion. Extrapolating these findings, compromised forefoot structure might compromise forefoot pronation and impair directional control of the GRF when running. Theoretically, as the foot could no longer effectively oppose pronation, the GRF would translate medially as the COP follows the natural lateral to medial shift during stance (De Cock, Vanrenterghem, Willems, Witvrouw, & De Clercq, 2008)

(Figure 6.1). This could potentially increase the peak-knee adduction moment by increasing the external moment arm, a loading pattern associated with injury Dudley et al. (2017). The interaction between forefoot structure, forefoot pronation and the peak-knee adduction moment has not been investigated.

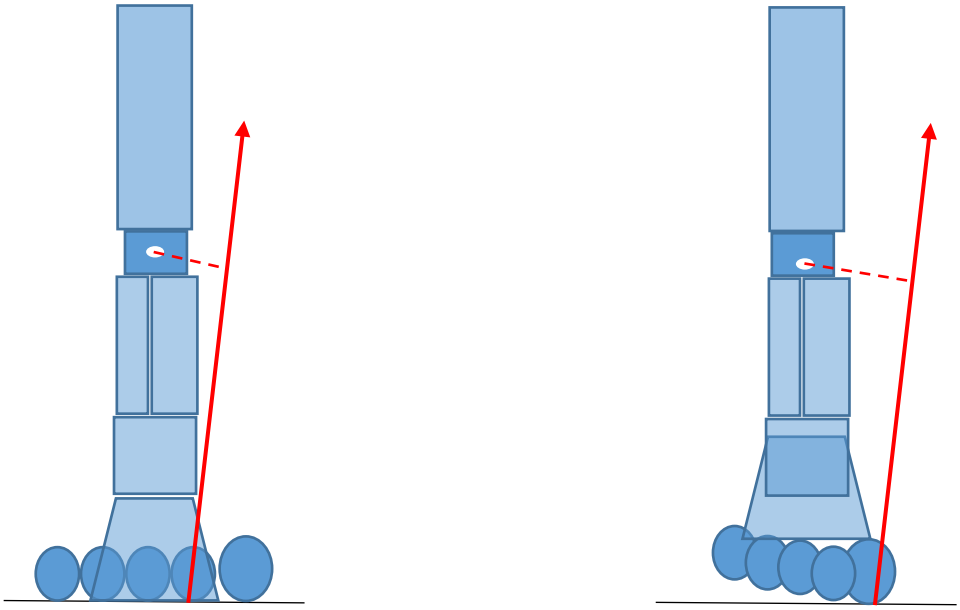

Figure 6. 1 Theoretical consequences of compromised forefoot structure in the frontal view. Natural abducted hallux and flailed toes (left) and compromised forefoot breadth and adducted hallux (right) in the frontal view. Solid red line represents the GRF, dashed line represents the external-knee-joint-moment arm and white circle represents the knee-joint centre. The compromised forefoot structure has a larger moment arm as a function of a medial shift in the COP.

In addition to forefoot structure, the effect different types footwear have on peak-forefoot pronation have not been investigated. Structured-cushioned running shoes are designed with a toe spring (upward curve of the toe box from the ground), and a symmetrical, narrow toe box that is a poor fit for natural-asymmetrical foot structure (Hoffmann, 1905; Willwacher et al., 2013). A symmetrical and narrow toe box compresses an abducted hallux towards the central apex of a shoe. So positioned, the hallux might no longer oppose forefoot pronation during stance and peak-forefoot pronation angle could increase, compromising the directional control of the GRF during stance. Equally, a toe spring might also compromise pronation by raising the toes into a dorsi-flexed position such that they are not in contact with the ground at midstance, reducing the active base of support and the ability of the hallux to direct body

weight in the transverse plane (Chou et al., 2009; Morton, 1935). However, following the recent definition of minimal shoes (Esculier et al., 2015), minimal-shoe design theoretically overcomes these restrictive issues seen in structured-cushioned running shoes. Minimal footwear are proposed to permit natural function of the foot due to high flexibility and lack of motion control and stability devices (Esculier et al., 2015) and have been recommended as tools to reduce injury risk (Sinclair et al., 2016). Comparisons of forefoot kinematics while running in structured-cushioned running shoes, minimal shoes and barefoot conditions have not been made.

This study had three aims: a) to investigate relationships between forefoot structure and forefoot pronation during overground running; b) to examine differences in forefoot pronation during overground running in different types of footwear (barefoot, minimal shoe and structured-cushioned running shoe) and; c) to explore the influence of pronation on the peak-knee adduction moment during overground running.

6.2 Method

6.2.1 Participants

Sample size, participant characteristics and inclusion/exclusion criteria were as described in chapter five (5.2.1).

6.2.2 Experimental design

A within-participant design was used to assess the relationship between foot structure and forefoot pronation, differences between footwear conditions (barefoot, minimal shoe and structured-cushioned running shoe for peak-forefoot pronation), and relationships between peak-forefoot pronation and peak-knee adduction moment. Participant testing sessions were separated by 24 hours. Participants were prepared as described in the general method section (3.3.2). Foot structure was assessed on the first day of testing and prior to running. The average speed from the first five data collection trials post habituation of each participant determined their average running speed ± 5% for that session. Electronic timing gates (Brower timing gates, Utah, USA) were used to record speed in each trial. The average running speeds for

barefoot, minimal and structured-cushioned footwear were 2.48 ± 0.38 $m \cdot s^{-1}$, 2.60 ± 0.43 $m \cdot s^{-1}$ and 2.68 ± 0.37 $m \cdot s^{-1}$ respectively.

6.2.3 Procedures

Foot structure

Five functional foot measures (see table 6.1) were recorded using anthropometric callipers (Harpenden Anthropometer, Holtain, Crosswell). All measures were relevant to shoe design and the functional capacity of the foot (Hoogvliet et al., 1997; Lee, Lin, & Wang, 2014; Mauch, Grau, Krauss, Maiwald, & Horstmann, 2009; Shu et al., 2015). Measurement definitions and their anatomical illustrations can be found in figure 6.2 and table 6.1.

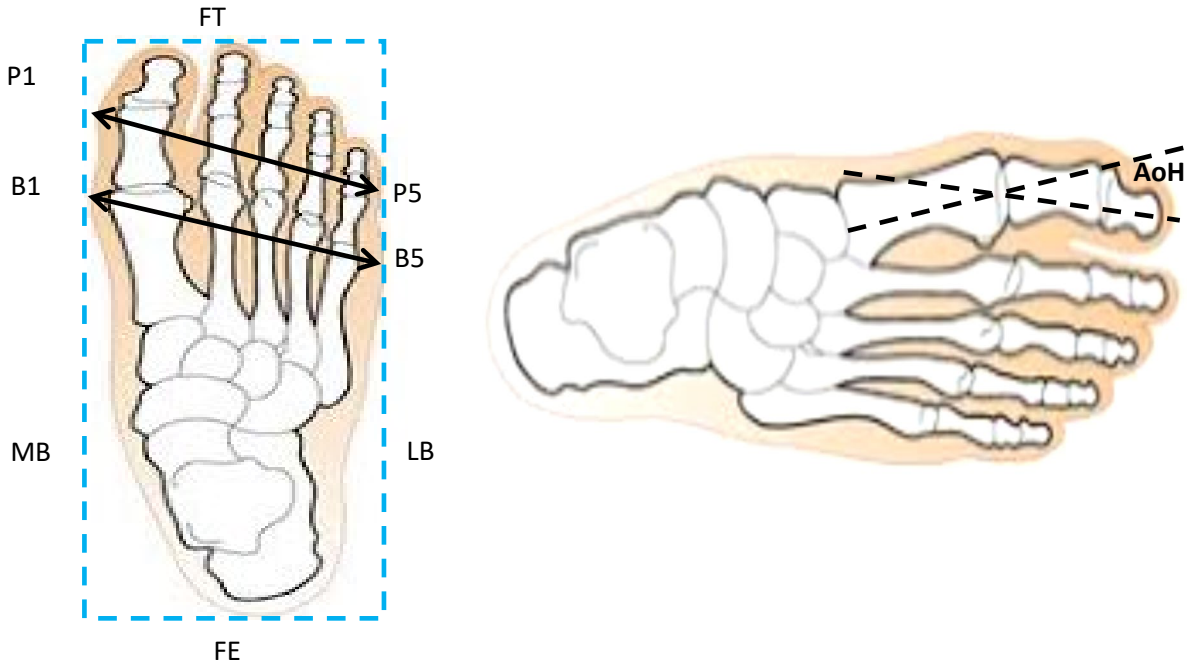

Figure 6. 2 An illustration of foot measures of interest. The image on the left depicts the following: ball-of-foot length, ball-of-foot width, phalange width, length of foot and width of foot. The image on the right illustrates the calculation of hallux angle.

Table 6. 1 Anatomical and notational definitions of foot structure measures of interest.

Variable	Image notation	Anatomical definition
		Image 1
Ball-of-foot length	$FE – B_1$	Distance between foot end (FE) (heel) and the 1st metatarsophalangeal protrusion (MTP) (B1).
Ball of foot width	$B_1 – B_5$	Connection line between 1st metatarsophalangeal (MTP) joint (B1) and 5th MTP (B5) (ball line).
Phalange width	$P_1 – P_5$	Connection line between the medial aspect of the hallux interphalangeal joint (P1) and the lateral aspect of the 5th interphalangeal joint (P5) (Phalange line).
Foot length	'Length'	Distance between FE (heel) and foot tip (FT) (anterior point of the most protruding toe)
Foot width	'Width'	Distance between the furthest most medial (MB) and lateral border (LB) of the foot.
Coefficient of spreading	'Width' / 'Length'	Widest aspect of the forefoot ('Width') divided by foot length ('Length').
		Image 2
Hallux angle	AoH	Angle created by the deviation of the hallux line (the line connecting the first metatarsal head and the central aspect of the interphalangeal joint of the hallux) away from the 1st metatarsal line (the line joining the 1st metatarsal head and base)

To collect foot structure data, participants were asked to stand barefoot on top of a 0.35m high platform covered in graph paper. Participants placed their non-dominant foot on the platform first, keeping the most posterior aspect of their foot aligned with a horizontal reference line

on the graph paper. Participants then placed their dominant foot on the platform, shoulder width apart from the other foot, and with the most posterior aspect of the foot on the same horizontal reference line. Following this, the second metatarsal head of the dominant foot was aligned with a line at right angles to the horizontal reference line, this represented the longitudinal axis of the foot. The foot was then palpated to identify the first metatarsal proximal-and distal-dorsal protrusions, and the central and dorsal point of the interphalangeal joint of the hallux. These anatomical locations were marked. A single set of foot structure measures were collected following definitions in table 6.1. A digital camera (CX240, Sony, Japan) raised 30cm above the platform using a tripod was aligned with the first metatarsophalangeal joint, and the zoom function was adjusted so the bony prominences defining the hallux angle were visible. An image was then recorded and saved for analysis.

Kinematics and kinetics

In addition to anthropometric measures (3.3.2) and retroflective markers were attached to participants in a full-body 'Plug-In gait' and 'Oxford-Foot Model' formation as described in chapter three, section 3.3.2. Kinematic and kinetic data were captured by 14 calibrated infrared cameras (T10/20, Vicon MX, Oxford, UK) and one of four force plates (OR6-7, AMTI, Watertown MA, USA). The measure forefoot pronation is as described in chapter one, section 1.3, injury rates, and illustrated in figure 1.1. Signals were captured and imported as described in chapter three, three-dimensional gait laboratory calibration (3.3.1).

6.2.4 Data analysis

Data analysis and processing was undertaken in the 3-D motion analysis software in line with the processes described in chapter three, section 3.3.6. Peak-joint moment and peak-forefoot pronation data were then tabulated in Microsoft Excel (Microsoft, USA) following methods described in 3.3.6. Foot structure images were loaded to Dartfish ClassroomPlus (version 7.0, Fribourg, Switzerland) where hallux angles were measured using the angle tool. Centre of pressure offset, defined as the mediolateral distance from the CoP to the longitudinal axis of the foot (heel marker to toe marker) at the time of peak-forefoot pronation were extracted

following previous work (Hinman, Bowles, Metcalf, Wrigley, & Bennell, 2012). Data were collectively tabulated in SPSS (version 24.0, SPSS Inc., Chicago, IL) for statistical analysis.

6.2.5 Statistical analysis

Following verification of assumptions of normality, linearity and uniformity of errors, Pearson's correlations were used to assess the relationship between individual measures of foot structure and peak-forefoot pronation angle. Pearson's correlations were also used to assess the relationships between significant predictor variables. If predictors correlated with each other with an r value above 0.8, the predictor with the weakest association to peak-forefoot pronation angle was discarded (Newell, Aitchison, & Grant, 2014). The remaining predictor variables were entered into a multiple regression to determine the amount of variance in peak-forefoot pronation angle that could be explained by foot-structure measures. After assumptions of normality and uniformity of error were verified, means of each footwear condition were adjusted for speed (by using speed as a covariate). Mean differences between footwear conditions for peak-knee adduction moment and pronation angle were then estimated using 90% confidence intervals. After assumptions of normality, uniformity of error and linearity were verified, the relationship between peak-forefoot abduction angle and peak-knee adduction moment was assessed with a Pearson's correlations in each footwear condition. After assumptions of normality, uniformity of error and linearity were verified the relationship between peak-forefoot pronation angle and COP offset were also assessed with Pearson's correlations in each footwear condition. Significance for correlations and multiple regression was accepted at $P < 0.05$.

6.3 Results

6.3.1 Foot structure and function relationships.

Foot structure data are presented in table (6.2). Hallux angle and phalange width were associated with peak-forefoot pronation angle in the barefoot condition ($r = 0.52$, $P = 0.047$; $r = 0.52$, $P = 0.046$, respectively) but were not related to each other ($r = 0.22$, $P = 0.422$). No foot-structure measures were related to peak forefoot-pronation angle in the minimal or

structured-cushioned running shoe conditions. Together, hallux angle and phalange width accounted for 35% of variance in peak-forefoot pronation angle in the barefoot condition (adjusted r^2 = 0.35, SEE = 5.35, $F_{2, 12}$ = 4.80, P = 0.03). Beta coefficients showed that for every one degree increase hallux adduction and 1mm reduction in phalange width, pronation increased by 0.24 degrees. Participants' predicted peak-forefoot pronation angle was modelled as -46.74 + (0.46) phalange width + (-0.463) hallux angle. A positive hallux angle represents a compromised/adducted hallux, an increased phalange width represents a wider spread of the phalanges, and a negative forefoot pronation angle represents a pronated forefoot.

Table 6. 2 Table to report the mean ± SD of the recorded foot structure measures as described in table 6.1. A positive hallux angle corresponds to an adducted hallux.

	Foot Length	Foot Width	Ball-of-Foot Length	Ball-of-Foot Width	Phalange width	Co-efficient of spreading	Hallux Angle
Average	261.13	98.33	189.53	98.53	100.07	0.38	9.47
(SD)	(19.37)	(6.29)	(14.13)	(7.12)	(6.18)	(0.02)	(6.1)

6.3.2 Differences between shoe conditions.

Peak-forefoot pronation angle (mean ± SD) at 2.59m·s^{-1} was worse in the minimal shoe (-9.16 ± 5°) than when barefoot (-5.46 ± 5.62°) (90% CI -7.05 to -0.35°) and in the structured-cushioned running shoe (-2.39 ± 5.5°) (90% CI -10 to -3.54°). Barefoot and structured-cushioned running shoes were similar (90% CI -0.71 to 6.85°). Mean differences are shown in figure 6.3. There was no significant relationship between peak-forefoot pronation angle and COP offset relative to the longitudinal axis of the foot in any running condition (r = -0.39, P = 0.155; r = 0.37, P = 0.174; r = 0.30, P = 0.275, for barefoot, minimal and structured-cushioned shoes respectively). Mean ± SD peak-forefoot pronation angle occurred at a similar percentage of stance for barefoot, minimal and maximally-cushioned (41.42 ± 6.78 %; 38.4 ± 9.23 %; and 39.11 ± 7.04 %).

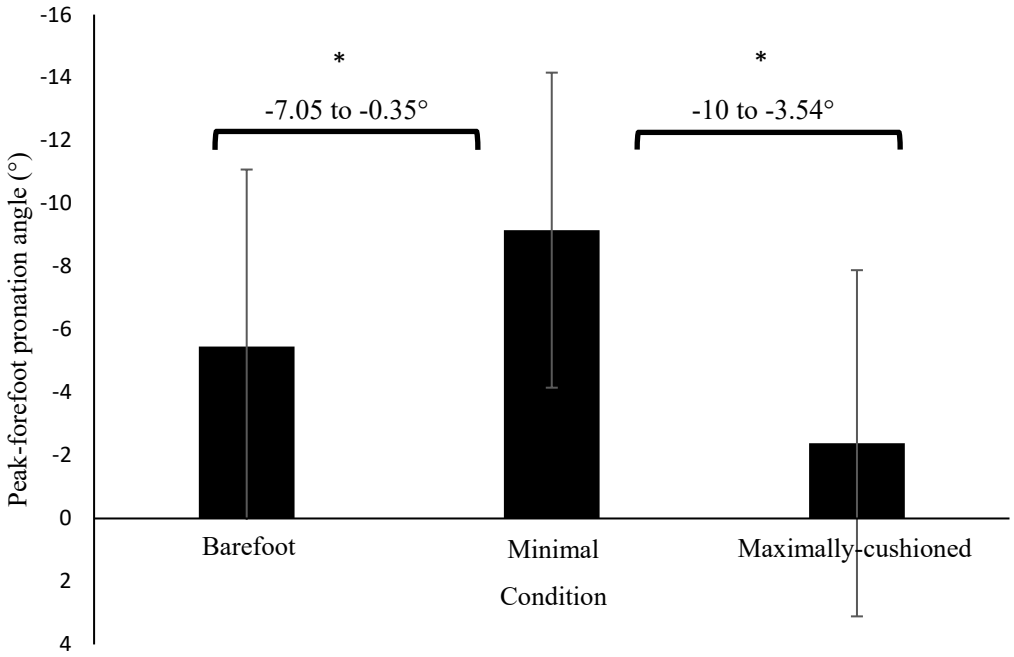

Figure 6. 3 Comparisons of the mean peak-pronation angle of 15 recreational-endurance runners during overground endurance running. Means are adjusted to a speed of 2.59 $m \cdot s^{-1}$. Differences between footwear conditions are indicated by *. Bars are mean ± the standard deviation.

Peak-knee adduction moment was negatively correlated with peak-forefoot pronation angle in all footwear conditions (r = -0.57, P = 0.027; r = -0.77, P = 0.001; r = -0.61, P = 0.015 for barefoot, minimal and structured-cushioned shoes respectively). However when adjusted to 2.59$m \cdot s^{-1}$ there was no clear difference for peak-knee adduction moment between barefoot (0.76 ± 0.34 $Nm \cdot Kg^{-1}$), minimal shoe (0.79 ± 0.31 $Nm \cdot Kg^{-1}$) and maximally cushioned shoe running (0.98 ± 0.33 $Nm \cdot Kg^{-1}$) (figure 6.4). Mean ± SD peak-knee adduction moment occurred at a similar percentage of stance for barefoot, minimal and maximally-cushioned running conditions (31.64 ± 11.19 %; 25.02 ± 10.67 % and 33.73 ± 10.24 %).

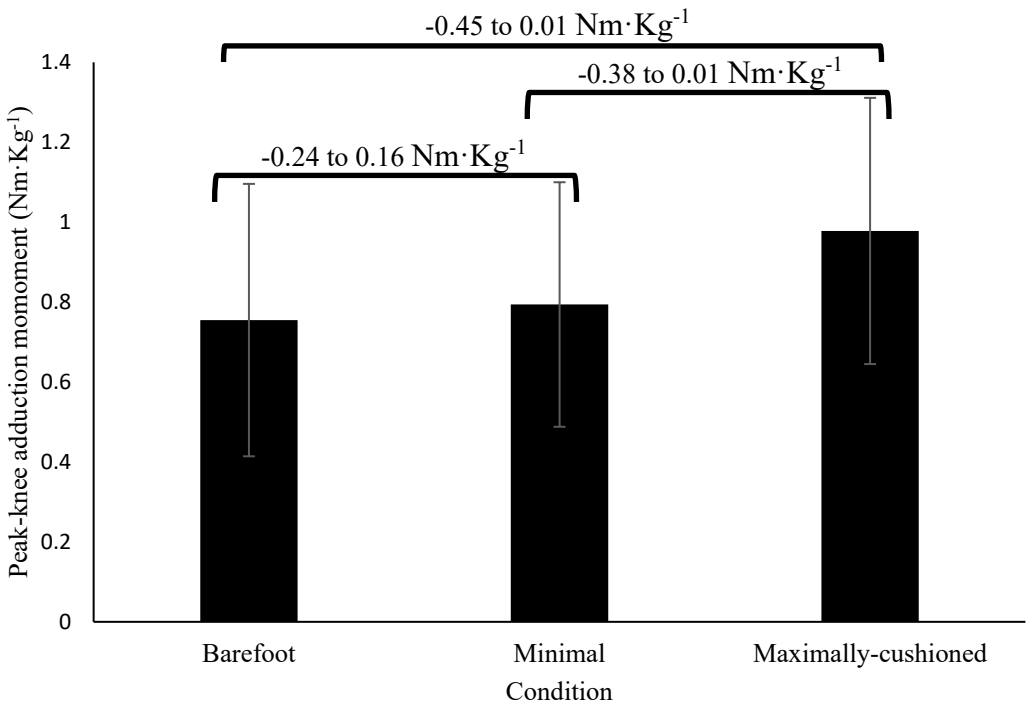

Figure 6. 4 Comparisons of the mean peak-knee adduction moment of 15 recreational-endurance runners during overground endurance running. Means are adjusted to a speed of 2.59 $m \cdot s^{-1}$. Bars are mean ± SD.

6.4 Discussion

The aims of this study were: to investigate relationships between foot structure and peak-forefoot pronation angle during overground running; to examine differences in peak-forefoot pronation angle during overground running in different types of footwear (barefoot, minimal and structured-cushioned shoes) and; to explore the influence of peak-forefoot pronation angle on the peak-knee adduction moment during overground running. Key findings were that metrics of foot structure predicted peak-forefoot pronation angle when running barefoot but not in shoes, that peak-forefoot pronation angle was statistically larger in minimal shoes than when barefoot and in structured-cushioned shoes, and that peak-forefoot pronation angle correlated with the peak-knee adduction moment irrespective of footwear condition.

6.4.1 Structure-function relationships

Foot structure predicted peak-forefoot pronation angle in the barefoot running condition. This investigation is the first of its kind to demonstrate such an interaction between forefoot

structure and peak-forefoot pronation angle in overground running. This observation is in line with previous balance research that reports compromised control of the GRF following compromised foot structure in a single-leg balance task. Both Chou et al. (2009) and Hoogvliet et al. (1997) demonstrate that the control of the COP was impaired when the function of the hallux was removed or the functional-foot breadth was reduced, respectively. As the role of the hallux in an uncompromised foot is to oppose excessive forefoot pronation (Morton, 1935), it is logical that at an adducted hallux would be unable to oppose forefoot pronation and that this might compromise loading at joints proximal to the foot. However, there was no relationship between any measures of foot structure and peak-forefoot pronation angle when running in the minimal or structured-cushioned shoes. A possible explanation might be that peak values for forefoot pronation were occluded by placing the markers on the surface of the shoe. For example, Sinclair, Greenhalgh, Taylor, et al. (2013) reported markers attached to footwear resulted in an underestimation of tibial-calcaneal kinematics, suggesting that foot structure might continue on its pre-contact trajectory, in this case, excessive forefoot pronation after foot-flat in shod conditions. An alternative approach was to create windows in the footwear to attach markers to the skin, however this might have compromised the restraint of the footwear on foot structure.

6.4.2 Footwear effects on forefoot pronation

Peak-forefoot pronation angle was statistically larger in the minimal shoe compared to the barefoot and structured-cushioned shoe conditions. This was unexpected as the minimal footwear in this study is marketed as providing a wide toe box to accommodate natural hallux position. A possible explanation is that although wide at the ball of the foot, the apex of the minimal shoe was still medially placed, merging towards a point. This could cause the hallux to converge towards an adducted position, compromising the capacity to oppose excessive forefoot pronation relative to the barefoot condition where hallux position was unconstrained (Chou et al., 2009; Morton, 1935). Larger peak-forefoot pronation angle in the minimal shoe than the structured-cushioned running shoe (a shoe that elevates the hallux and compresses it towards a central apex) is in contrast to previous work that suggests footwear that excessively

constrains the hallux to an adducted and raised position would compromise the functional capacity to control foot motion (Chou et al., 2009; Hoogvliet et al., 1997). Indeed, mean peak-forefoot pronation angle was smallest in the structured-cushioned shoe condition (though statistically similar to the barefoot condition). A possible explanation is the thick and stiff mid-sole design of the structured-cushioned running shoe. A stiff mid-sole design would act to buttress excessive forefoot pronation by acting in sequence with the foot, providing a wider functional axis across the phalanges, and in doing so, opposing excessive forefoot pronation. This contrasts the highly flexible design seen in minimal footwear (Esculier et al., 2015).

However, comparisons between footwear and barefoot running results are limited in light of previous work by Bishop, Thewlis, Uden, Ogilvie, & Paul, (2011) who highlighted the potential for error in placing markers on shoes when compared to skin mounted markers. When comparing the error of markers that define the forefoot region in the current model an error of up to 6.9 mm was observed. Additionally, Osis et al. (2016) reported a change of 10mm in the placement of the lateral ankle marker induced a change of 7.59°. However, Osis et al. (2016) also reported a change in marker position at the distal aspects of the forefoot (distal 1st and 5th metatarsal head) induced changes of <0.5° in ankle and foot rotations. If 0.5° was applied to current confidence intervals there would be be no clear difference between barefoot and minimal shoe running, but differences would still be clear between minimal and structure-cushioned footwear. However, if the largest potential error, 6.9 mm and therefore potenitally 5.24° was factored into peak-forefoot pronation results, there would be no clear difference between conditions. However, it was the compression of the forefoot region in an intact shoe that was the aim of this comparison, and it was believed that cutting holes in shoes would compromise the ecological validity of the study, therefore holes were not cut in the shoe. Collectively, results suggest footwear might have influenced peak-forefoot pronation angle, with the largest peak-forefoot pronation angle in the minimal shoe, possibly because of its flexibility and toe-box design. However, while statistics report clear differences, findings should be interpreted in light of potential error introduced by placing markers on the shoe.

6.4.3 Forefoot pronation and knee loading

Peak-forefoot pronation angle was moderately associated with the peak-knee adduction moment irrespective of footwear condition. This is particularly important, as increased peak-knee adduction moment has been shown to differentiate injured and uninjured runners (Dudley et al., 2017), predict a rapid progression of medial-compartment osteoarthritis (Sharma et al., 1998) as well as differentiate those with patellofemoral joint pain (Willy et al., 2012). It has been suggested that a medial shift of the COP (the origin of the GRF) as a function of increased peak-forefoot pronation as weight bearing shifts medially might have explained the observed increase in the peak-knee adduction moment. This was supported by a review by Reeves and Bowling (2011) who investigated strategies to reduce knee osteoarthritis where a medially translated COP was associated with an increased peak-knee adduction moment. A medially translated GRF would increase the external moment arm of the knee joint and increase the peak-knee adduction moment proportionally. However, results from the current study report no association between COP offset and peak-forefoot pronation angle. Following, it seems that an increase in peak-forefoot pronation does not induce a medial shift in the COP. Therefore compared to previous walking gait research other variables beyond COP offset underpin the observed relationship between peak-forefoot pronation and peak-knee adduction moment when performing dynamic tasks such as ER. An explanation for this observation requires further investigation. Future investigations should explore this line of questioning further using a pressure insole to examine the interactions between centre of pressure magnitude and knee-adduction moment.

6.4.4 Limitations

Results of minimal and structured-cushioned running shoes are limited with biomechanical representation of the forefoot relying on the motion of the shoe, but not forefoot structure itself. Although cutting holes in footwear was an option, this might have compromised the restraint footwear applied to foot structure, misrepresented forefoot kinematics in overground running and therefore compromised ecological validity. Specifically, it has been argued that if skin to shoe marker placement error can be as large as 6.9 mm in the forefoot (Bishop et al.,

2011), and a 10 mm change in marker position can induce a change of up to 7.59°, then when scaled, reported values could change by as much as 5.24°. Following, aforementioned statistically clear differences between minimal and other running conditions might not be clear. However, it was also noted that 10 mm changes in forefoot specific markers induced a change of less than 0.5° in ankle and foot rotations. Consequently, although statistical results report clear differences, these differences should be interpreted in light of potential error. Future studies should investigate the relationship between forefoot pronation and COP data using a pressure insole to deduce magnitudes of pressure. Finally, it is also important to consider that other factors beyond forefoot structure also play a role in the peak-knee adduction moment, factors such as knee valgus and rearfoot abduction (Hurwitz, Ryals, Case, Block, & Andriacchi, 2002). Future work should consider such factors in addition to the relationships reported in this chapter.

6.4.5 Conclusion

The results suggest that hallux angle and phalange width influence peak-forefoot pronation during overground running when the foot is unconstrained (barefoot), but not in minimal or structured-cushioned shoes. Results further suggest that peak-forefoot pronation is worse in a minimal shoe than barefoot and in structured-cushioned shoes. However, comparisons should be made in light of error introduced when placing markers on shoes. An increased peak-forefoot-pronation angle was not associated with COP position. Future studies examining or aiming to reduce the peak-knee adduction moment should consider the influence of footwear choice and forefoot structure.

7.0 Kinematics and kinetics of recreational runners during overground running when barefoot and in minimal and maximally-cushioned running shoes.

7.1 Introduction

Following previous discussions on injury rates, research has attempted to address injury rates by intervening with running technique and footwear choice to reduce surrogate measures associated with injury. Barefoot and minimal shoe running have received particular attention following arguments that humans evolved to run barefoot (Bramble & Lieberman, 2004).

A commonly reported kinematic adaptation to running barefoot and in minimal shoes is increased plantarflexion at initial contact (Gruber et al., 2012; Squadrone et al., 2015). The potential implications of foot strike strategy were highlighted by Daoud et al. (2012) who reported habitual rear-foot strikers incur approximately twice as many repetitive stress injuries as individuals who FFS. A more flexed knee joint at initial contact when running in minimal compared to conventional footwear has also been reported (Willy & Davis, 2013). Increased knee flexion at initial contact has been shown to increase knee flexion at midstance, reducing the effective mass (Derrick, 2004) and impact peak by 68N per degree of flexion (Gerritsen et al., 1995). These findings demonstrate that barefoot and minimal-shoe conditions can change running technique and potentially alter injury rates.

Barefoot and minimal-shoe running have been reported to improve joint-specific loading patterns associated with injury. As discussed in the literature review (2.7.1-3) these running modalities can reduce peak-knee joint moments (flexion and adduction) and patellofemoral joint stress, however, at the same time peak-plantarflexion moment increased suggesting a trend to shift loading to the ankle. This potentially increasing the stress on anatomical structures of the ankle, however, longitudinal studies are needed. Reducing peak-knee flexion and adduction moment is advantageous given that higher adduction moments differentiate injured from uninjured overground runners (Dudley et al., 2017), Sharma et al. (1998) reported increased peak-knee adduction predicted patients with advanced medial tibiofemoral osteoarthritis and poor joint health (reduced joint space width) and a review by Reeves and

Bowline (2011) discusses that increased peak-knee adduction moment was intimately linked with the severity and progression of medial knee osteoarthritis. Furthermore, increased peak-knee flexion moment also coincides with increased patellofemoral joint stress (Bonacci et al., 2014). This evidence suggests that running barefoot and in minimal shoes can reduce surrogate-knee-joint loading measures associated with sporting injury and poor joint health (osteoarthritis) relative to conventional running shoes; however, comparisons to other types of footwear such as maximal footwear are warranted. In contrast, research on maximally-cushioned shoes is in its infancy and their effects on running gait warrant investigation. The effects of maximally-cushioned footwear on kinematic and kinetics of overground running in comparison to barefoot and minimalist footwear has received little attention.

To date, no studies have investigated the effects of footwear on upper body kinematics in overground ER. Work investigating trunk lean is inconsistent, with some advocating a more upright trunk lean (Preece et al., 2016) and others encouraging increased trunk lean (Teng & Powers, 2014). Theoretically, footwear has the potential to effect peak trunk lean. The VPP model (2.7.6) predicts that because running in cushioned footwear increases stride length (Kerrigan et al., 2009) the foot would be projected anterior to the hip. An anterior shift in the location of the foot and therefore the COP would subsequently increase the hip flexion external moment arm responsible for trunk lean during impact. An increase in the external moment arm would proportionally increase the resultant external-hip flexion moment and increase peak trunk lean. The influence of different types of footwear on peak trunk lean have not been investigated.

The aim of this study was to compare the kinematics and kinetics of overground ER between barefoot, minimal and maximally-cushioned footwear conditions in a sample of recreational endurance runners. We hypothesised that maximally-cushioned footwear would: 1) increase knee extension and ankle dorsiflexion at initial contact; 2) increase peak-joint moments in the frontal and sagittal plane at the knee joint; 3) reduce the peak-dorsiflexion moment and; 4) increase peak trunk lean compared to the barefoot and minimal-shoe conditions.

7.2 Methods

7.2.1 Participants

Sample size, participant characteristics and inclusion/exclusion criteria were as described in chapter five (5.2.1).

7.2.2 Experimental design

A within-participant design was used to assess the kinematic and kinetic differences between footwear conditions (barefoot, minimal and maximally-cushioned footwear (see 3.3.3). Participants were prepared as described in the experimental design of chapter three (3.3.2) and data was collected immediately after a 30-minute habituation run in the relevant footwear for that session (chapter five). Similarly to chapter six, participants ran on separate days in a counterbalanced order separated by 24 hours. Participants were instructed to be well rested and run at a speed described in section 3.3.4. Average running speed was calculated as described in chapter six (6.2.2). Average running speed for barefoot, minimal and maximally-cushioned footwear was 2.48 ± 0.38, 2.60 ± 0.43, 2.68 ± 0.37 $m \cdot s^{-1}$, respectively.

7.2.4 Procedure

Anthropometric measures were recorded and retroflective markers were attached to participants in a full-body 'Plug-In gait' and 'Oxford-Foot Model' formation, as described in participant preparation for 3-D analysis (3.3.2) to facilitate the assessment of lower-limb biomechanics and additional measures in a previous chapter. Kinematic and kinetic data were captured by 14 calibrated infrared cameras (T10/20, Vicon MX, Oxford, UK) and one of four force plates (OR6-7, AMTI, Watertown MA, USA). Signals were captured and imported with equipment described in chapter three, section 3.3.1.

7.2.5 Data analysis

Data analysis and processing was undertaken in the 3-D motion analysis software in line with the processes described in chapter three, section 3.3.6. Contact time, joint angles and peak-

joint moments were derived as described in section 3.3.6 and tabulated in SPSS (version 24.0, SPSS Inc., Chicago, IL) for statistical analysis.

7.2.6 Statistical analysis

After assumptions of normality and uniformity of error were verified, the means of each footwear condition were percentage adjusted for speed (by using speed as a covariate) and normalising comparisons to a common speed of 2.59 $m \cdot s^{-1}$. Mean ± SE and mean difference in lower limb joint angles at initial contact, midstance, peak values and ROM, in the sagittal, frontal and transverse plane, as well as peak lower-limb joint moments in the sagittal, frontal and transverse plane were estimated using 90% confidence intervals between footwear conditions. Differences in peak trunk lean and contact time were also estimated using 90% confidence intervals between footwear conditions.

7.3 Results

Tables 7.1, 7.2 and 7.3 show speed-adjusted mean ± SE and 90% confidence interval comparisons between barefoot, minimal and maximally-cushioned footwear conditions for hip, knee and ankle kinematics in the sagittal, frontal and transverse plane. Table 7.4 shows speed-adjusted mean ± SE and 90% confidence interval comparisons between footwear conditions for peak-lower-limb-joint moments at the hip, knee and ankle in the sagittal, frontal and transverse plane while performing overground ER.

Table 7. 1 Speed-adjusted mean ± SE and 90% confidence intervals for kinematic comparisons at the hip joint when barefoot, in minimal and maximally-cushioned shoes during overground endurance running in recreational runners (n=15). Note: BF: Barefoot, MS: minimal shoe, MCS: Maximally-cushioned shoe.

Plane	Parameter	BF	MS	MCS	BF – MS (°)	MS – MCS (°)	BF-MCS (°)
		Mean ± SE	Mean ± SE	Mean ± SE	90% CI	90% CI	90% CI
Sagittal	IC	38.082 ± 0.787	38.089 ± 0.7	39.467 ± 0.77	-1.811 to 1.825	-0.374 to 3.129	-0.663 to 3.432
(Flexion +/	Midstance	24.04 ± 0.817	22.885 ± 0.727	24.893 ± 0.799	-3.043 to 0.731	0.191 to 3.826*	-1.272 to 2.978
Extension -)	Peak flexion	38.585 ± 0.847	38.709 ± 0.754	40.126 ± 0.828	-1.831 to 2.081	-0.467 to 3.301	-0.661 to 3.744
	ROM	39.336 ± 0.503	40.137 ±0.447	40.606 ± 0.491	-0.360 to 1.962	-0.649 to 1.587	-0.037 to 2.577

Frontal	IC	5.067 ± 0.712	5.370 ± 0.633	6.503 ± 0.696	-1.340 to 1.947	-0.451 to 2.715	-0.415 to 3.287
(Adduction +/	Midstance	10.711 ± 0.575	9.654 ± 0.512	9.225 ± 0.562	-2.394 to 0.262	-1.699 to 0.859	-2.981 to 0.010
Abduction -)	Peak adduction	12.635 ± 0.594	12.354 ± 0.528	12.833 ± 0.581	-1.652 to 1.091	-0.843 to 1.800	-1.347 to 1.743
	ROM	12.46 ± 0.438	13.303 ± 0.39	13.785 ± 0.428	-0.169 to 1.854	-0.493 to 1.456	0.185 to 2.463*
Transverse	IC	-3.640 ± 1.844	-4.406 ± 1.641	0.624 ± 1.803	-5.026 to 3.493	-0.321 to 7.885	-1.781 to 7.812
(Internal +/	Midstance	-2.007 ± 1.486	-1.289 ± 1.322	1.174 ± 1.453	-2.714 to 4.152	-0.844 to 5.769	-0.684 to 7.047
External -)	Peak internal rotation	1.860 ± 1.581	2.381 ± 1.406	5.011 ± 1.546	-3.130 to 4.172	-0.887 to 6.147	-0.960 to 7.262
	ROM	10.149 ± 0.625	10.249 ± 0.556	9.209 ± 0.611	-1.342 to 1.543	-2.430 to 0.350	-2.564 to 0.685

* indicates a clear difference between comparison.

Table 7. 2 Speed-adjusted mean ± SE and 90% confidence intervals for kinematic comparisons at the knee joint when barefoot, in minimal and maximally-cushioned shoes during overground endurance running in recreational runners (n=15). Note: BF: Barefoot, MS: minimal shoe, MCS: Maximally-cushioned shoe.

Plane	Parameter	BF	MS	MCS	BF – MS (°)	MS – MCS (°)	BF – MCS (°)
		Mean ± SE	Mean ± SE	Mean ± SE	90% CI	90% CI	90% CI
Sagittal	IC	20.999 ± 0.827	19.889 ± 0.735	18.721 ± 0.808	-3.019 to 0.799	-3.007 to 0.670	-4.428 to -0.129*
(Flexion +/	Midstance	42.596 ± 0.802	43.931 ± 0.714	46.304 ± 0.784	-0.517 to 3.188	0.589 to 4.157*	1.623 to 5.794*
Extension -)	Peak flexion	45.516 ± 1.114	47.794 ± 0.991	50.419 ± 1.089	-0.296 to 4.850	0.147 to 5.104*	2.005 to 7.800*
	ROM	25.858 ± 1.029	28.772 ± 0.915	32.659 ± 1.006	0.539 to 5.289*	1.599 to 6.176*	4.127 to 9.476*
Frontal	IC	0.256 ± 0.637	-0.785 ± 0.566	-0.620 ± 0.622	-2.512 to 0.429	-1.251 to 1.582	-2.532 to 0.780
(Adduction +/	Midstance	-0.774 ± 0.976	-1.072 ± 0.869	1.296 ± 0.955	-2.553 to 1.957	0.195 to 4.539*	-0.470 to 4.608
Abduction -)	Peak adduction	2.483 ± 0.845	2.192 ± 0.751	3.653 ± 0.826	-2.196 to 1.705	-0.418 to 3.340	-0.981 to 3.412
	ROM	6.549 ± 0.51	6.798 ± 0.454	6.986 ± 0.499	-0.929 to 1.428	-0.948 to 1.323	-0.891 to 1.764
Transverse	IC	-5.195 ± 2.484	-5.604 ± 2.210	-4.419 ± 2.429	-6.146 to 5.329	-4.342 to 6.711	-5.685 to 7.237

(Internal +/	Midstance	6.034 ± 1.995	5.369 ± 1.775	6.258 ± 1.951	-5.273 to 3.943	-3.550 to 5.328	-4.965 to 5.413
External -)	Peak internal rotation	8.417 ± 2.198	8.479 ± 1.955	8.687 ± 2.148	-5.013 to 5.137	-4.680 to 5.097	-5.445 to 5.985
	ROM	16.167 ± 0.694	18.055 ± 0.617	17.607 ± 0.678	0.286 to 3.491*	-1.992 to 1.096	-0.364 to 3.245

* indicates a clear difference between comparison.

Table 7. 3 Speed-adjusted mean ± SE and 90% confidence intervals for kinematic comparisons at the ankle joint when barefoot, in minimal and maximally-cushioned shoes during overground endurance running in recreational runners (n=15). Note: BF: Barefoot, MS: minimalist shoe, MCS: Maximally-cushioned shoe.

Plane	Parameter	BF	MS	MCS	BF – MS (°)	MS – MCS (°)	BF – MCS (°)
		Mean ± SE	Mean ± SE	Mean ± SE	90% CI	90% CI	90% CI
Sagittal	IC	2.360 ± 1.201	11.885 ± 1.069	16.130 ± 1.174	6.751 to 12.300*	1.572 to 6.917*	10.646 to 16.894*
(Dorsiflexion +/	Midstance	26.022 ± 0.942	28.807 ± 0.838	29.433 ± 0.921	0.609 to 4.960*	-1.468 to 2.722	0.962 to 5.860*
Plantarflexion -)	Peak Dorsiflexion	26.066 ± 0.861	29.143 ± 0.766	30.602 ± 0.841	1.089 to 5.064*	-0.455 to 3.375	2.298 to 6.775*
	ROM	39.772 ± 0.635	39.301 ± 0.565	35.926 ± 0.621	-1.937 to 0.996	-4.787 to -1.962*	-5.497 to -2.194*

Frontal (Adduction +/ Abduction -)	IC	0.211 ± 0.310	-0.631 ± 0.275	0.072 ± 0.303	-1.557 to -0.127*	0.014 to 1.391*	-0.944 to 0.666
	Midstance	2.137 ± 0.335	1.183 ± 0.298	2.724 ± 0.328	-1.728 to -0.180*	0.796 to 2.286*	-0.284 to 1.459
	Peak adduction	2.462 ± 0.330	1.591 ± 0.294	3.396 ± 0.323	-1.633 to -0.108*	1.070 to 2.539*	0.075 to 1.792*
	Peak abduction	-1.051 ± 0.333	-1.555 ± 0.297	-0.967 ± 0.326	-1.273 to 0.266	-0.153 to 1.330	-0.782 to 0.952
	ROM	3.514 ± 0.276	3.147 ± 0.245	4.362 ± 0.270	-1.004 to 0.270	0.602 to 1.829*	0.131 to 1.566*
Transverse (Internal +/ External -)	IC	-1.649 ± 1.937	2.115 ± 1.755	-1.889 ± 1.929	-0.791 to 8.321	-8.393 to 0.384	-5.370 to 4.891
	Midstance	2.680 ± 1.281	-11.359 ± 1.140	-14.605 ± 1.252	-16.998 to -11.081*	-6.096 to -0.396*	-20.617 to -13.954*
	Peak external rotation	-16.23 ± 1.798	-14.115 ± 1.599	-16.722 ± 1.757	-2.030 to 6.272	-6.606 to 1.391	-5.161 to 4.189
	ROM	22.97 ± 0.685	23.157 ± 0.609	19.573 ± 0.669	-1.394 to 1.768	-5.107 to -2.061*	-5.177 to -1.616*

* indicates a clear difference between comparison.

Table 7. 4 Speed-adjusted mean ± SE and 90% confidence intervals for peak-joint moment comparisons at the hip, knee and ankle joint when barefoot, in minimal and maximally-cushioned shoes during overground endurance running in recreational runners (n=15). Note: BF: Barefoot, MS: minimal shoe, MCS: Maximally-cushioned shoe.

Parameter	BF	MS	MCS	BF - MS	MS – MCS	BF – MCS
	Mean ± SE	Mean ± SE	Mean ± SE	90% CI	90% CI	90% CI
Peak hip flexion moment ($Nm \cdot kg^{-1}$) (Flexion +/ Extension)	2.678 ± 0.187	2.686 ± 0.166	2.679 ± 0.183	-0.423 to 0.440	-0.422 to 0.409	-0.484 to 0.487
Peak hip adduction moment ($Nm \cdot kg^{-1}$) (Adduction +/ Abduction -)	1.499 ± 0.055	1.490 ± 0.049	1.501 ± 0.054	-0.136 to 0.117	-0.111 to 0.134	-0.141 to 0.145
Peak hip external rotation moment ($Nm \cdot kg^{-1}$) (Internal +/ External -)	-0.36 ± 0.015	-0.36 ± 0.013	-0.399 ± 0.015	-0.035 to 0.034	-0.072 to -0.005*	-0.078 to 0.000
Peak knee flexion moment ($Nm \cdot kg^{-1}$) (Flexion +/ Extension)	2.511 ± 0.062	2.668 ± 0.055	2.813 ± 0.061	0.015 to 0.301*	0.006 to 0.282*	0.141 to 0.463*

Peak knee adduction moment ($Nm \cdot kg^{-1}$) (Adduction +/ Abduction -)	0.755 ± 0.088	0.794 ± 0.079	0.978 ± 0.086	-0.164 to 0.244	-0.013 to 0.380	-0.006 to 0.453
Peak knee internal rotation moment ($Nm \cdot kg^{-1}$) (Internal +/ External -)	0.093 ± 0.011	0.105 ± 0.010	0.113 ± 0.011	-0.013 to 0.038	-0.017 to 0.032	-0.009 to 0.049
Peak dorsiflexion moment ($Nm \cdot kg^{-1}$) (Dorsiflexion +/ Plantarflexion -)	2.570 ± 0.037	2.655 ± 0.033	2.338 ± 0.036	0.000 to 0.171	-0.399 to -0.235*	-0.328 to -0.136*
Peak ankle abduction moment ($Nm \cdot kg^{-1}$) (Adduction +/ Abduction -)	-0.192 ± 0.016	-0.258 ± 0.014	-0.153 ± 0.016	-0.103 to -0.029*	0.070 to 0.141*	-0.003 to 0.081
Peak ankle internal rotation moment ($Nm \cdot kg^{-1}$) (Internal +/ External -)	0.293 ± 0.029	0.366 ± 0.025	0.326 ± 0.028	0.008 to 0.140*	-0.104 to 0.023	-0.041 to 0.108

* indicates a clear difference between comparison.

Key results show the knee joint in the sagittal plane at initial contact was significantly more extended in the maximally-cushioned shoe (18.721 ± 0.808°) when compared to barefoot running (20.999 ± 0.827°). This coincided with a clear increase in dorsiflexion at initial contact when moving from barefoot (2.360 ± 1.201°), to minimal (11.885 ± 1.069°) and then maximally-cushioned shoes (16.130 ± 1.174°). Following this trend, there were also clear increases in peak dorsiflexion and dorsiflexion at midstance when barefoot (26.066 ± 0.861°; 26.022 ± 0.942°) when compared to maximally cushioned footwear (30.602 ± 0.841°; 29.433 ± 0.921°). These trends were reflected in sagittal plane peak-joint moments. There was a clear trend to increase peak-knee flexion moment when transitioning from barefoot (2.511 ± 0.062 $Nm \cdot kg^{-1}$), to minimal-shoe (2.668 ± 0.055 $Nm \cdot kg^{-1}$) and maximally-cushioned footwear (2.813 ± 0.061 $Nm \cdot kg^{-1}$). There was also a clear trend for maximally-cushioned footwear to reduce peak dorsiflexion moment (2.338 ± 0.036 $Nm \cdot kg^{-1}$) compared to barefoot (2.570 ± 0.037 $Nm \cdot kg^{-1}$) and minimal footwear (2.655 ± 0.033 $Nm \cdot kg^{-1}$). There was no clear difference between speed adjusted peak trunk lean values when changing from barefoot (6.84 ± 0.39°), to minimal (7.07 ± 0.35°), or maximally-cushioned shoes (7.67 ± 0.38°). Contact time in the barefoot condition (0.272 ± 0.003 s) was clearly less than both minimal (0.296 ± 0.003 s) (90% CI: -0.031 to -0.015 s) and maximally-cushioned footwear (0.299 ± 0.003s) (90 CI: -0.035 to -0.018 s) condition.

7.4 Discussion

This study compared kinematics and kinetics of overground running barefoot and in minimal and maximally-cushioned shoes in a sample of recreational runners. It was hypothesised that the maximally-cushioned footwear would increase knee extension and ankle dorsiflexion at initial contact. It was also hypothesised that peak-lower limb joint moments at the knee in the frontal and sagittal plane would increase, and the peak-dorsiflexion moment would decrease while running in the maximally-cushioned shoe. Additionally, it was hypothesised that running in a maximally-cushioned shoe would increase peak trunk lean. Key findings were a more extended knee joint at initial contact in the maximally-cushioned shoe compared to the barefoot condition, a clear increase in dorsiflexion from barefoot to minimal and then

maximally-cushioned footwear, an increased peak-knee flexion moment from barefoot to minimal to the maximally-cushioned shoe, and an increase in the minimal shoe peak-dorsiflexion moment compared to the maximally-cushioned condition.

In support of hypothesis one, there was an increase in dorsiflexion at initial contact when moving from barefoot to minimal and then maximally-cushioned footwear. This agrees with previous work by Sinclair, Greenhalgh, Brooks, et al. (2013) who compared barefoot, minimal and conventional shoes, and reported significant increases in plantarflexion at initial contact when running barefoot (-4.9 ± 8.26) compared to minimal (4.47 ± 7.35) and conventional-running footwear (7.64 ± 6.07). As suggested by De Wit et al. (2000) this change was likely a response to the lack of sensory insulation and therefore detection of high pressures under the heel when running with the conventional RFS strategy observed in the MCS (foot strike angle: 16.130 ± 1.174°). The adaptation of a more plantarflexed ankle is also logical when running barefoot given the potential to convert the translational energy of a rear-foot strike, into the rotational energy of a FFS (Lieberman et al., 2010), a foot strike strategy associated with reduced effective mass and in some cases injury rates (Daoud et al., 2012; Lieberman et al., 2010). Additionally, midstance and peak values for the ankle joint in the sagittal plane showed increased dorsiflexion when changing from barefoot to minimal and maximally-cushioned footwear. An increased t_c is proposed as a potential explanation. Following a comparison between barefoot, multiple types of minimal footwear, and conventional footwear, Squadrone et al. (2015) reported minimal footwear to significantly increase t_c compared to barefoot, and no minimal footwear contact times were significantly larger than conventional footwear. This trend was also observed in the current data with barefoot t_c reporting a clear reduction compared to the minimal and maximally-cushioned condition. An increased t_c would provide more time weight bearing during stance and potentially explain increased peak dorsiflexion and midstance values when running in maximally-cushioned footwear.

There was increased knee flexion at initial contact as participants changed from maximally-cushioned shoes to barefoot. The trend of increased knee flexion at initial contact when running barefoot as compared to footwear with cushioning is consistent with previous work

(Squadrone et al., 2015; Willy & Davis, 2013). Squadrone et al. (2015) reported the knee was more extended in footwear with greater heel thickness, with such postural adaptations thought to be associated with increased landing stiffness. In support of this, Derrick (2004) reported a more extended knee angle at initial contact was associated with increased peak impact force and effective mass. Additionally, midstance and peak knee flexion increased from barefoot and minimal to maximally-cushioned footwear, and a clear increase was shown for all comparisons in knee flexion ROM when moving from barefoot, to minimal and maximally-cushioned shoes. In line with t_c results, moving from barefoot to running footwear facilitated more time weight bearing and potentially explains the reported increases in peak flexion and ROM values. Collectively, this demonstrates footwear has the potential to statistically influence peak-forefoot pronation angle (chapter six) and kinematics more proximal in the kinetic chain (chapter seven). Future studies should examine the relationship between spatiotemporal values, footwear choice and their implications for injury.

Following kinematic trends, peak-knee flexion moment increased as participants changed from barefoot, to minimal, to maximally-cushioned footwear. In accordance with Kerrigan et al. (2009) who compared barefoot to conventional-cushioned shoe running, there was no clear difference in the peak-sagittal hip moment, but large reductions in the peak-sagittal knee joint moment when changing from a cushioned shoe to barefoot. However, the magnitude of change in Kerrigan et al. (2009) was somewhat larger for peak-knee flexion moment (36%) compared to the current study (BF - MCS; sagittal: 12%). This observed change was likely caused by the reduction in midsole cushioning, or lack of, in barefoot and minimal conditions compared to the maximally-cushioned shoe which induced an increased peak dorsiflexion moment (as a function of increased plantarflexion at IC) (De Wit et al., 2000) and subsequently reduced the peak sagittal knee moment. A potential explanation for the difference in % difference is the habituation period afforded to participants. In the current study, participants were provided 30 minutes to habituate, however, Kerrigan et al. (2009) provided three to five minutes following work that argues kinetic stabilisation after five minutes of running (Riley et al., 2008). In contrast, previous work suggests this is not the case for lower-limb kinematic measures, with

Moore and Dixon (2014) and Arnold, Weeks, and Horan (2018) reporting barefoot running habituation requiring 20 and 8 minutes, respectively, and this thesis reporting 21 minutes was necessary for habituation to novel footwear conditions. Accordingly, it could be argued that such a large difference in peak-sagittal knee moment might be a product of participants initially perceiving barefoot running as injurious and overcompensating for injury potential before refining their technique for barefoot running. Additionally, there was no clear increase in peak-knee adduction moment when changing from barefoot to maximally-cushioned footwear. This contrasts previous work by Kerrigan et al. (2009) that reported changing from conventional-cushioned footwear to barefoot reduced peak-knee adduction moment ($P < 0.001$). Inconsistencies in time allocated for habituation might explain such differences. The effects of footwear, subsequent changes in overground running technique and its interaction with peak-knee adduction moments warrant further investigation following its association with injury rates in overground ER (Dudley et al., 2017).

Following increases in the peak-knee flexion moment when moving from barefoot, to minimal, and then maximally-cushioned shoes, barefoot and minimal-shoe running produced higher peak-dorsiflexion moments compared to maximally-cushioned footwear. This is in line with previous work by Sinclair (2014) that reported a significant reduction in the peak-knee flexion moment was accompanied by a significant increase in the peak-external dorsiflexion moment when running barefoot or in minimal shoes. As described by Lieberman (2012b) a plantarflexed foot would create a large external dorsiflexion moment arm and explain an increased peak-dorsiflexion moment. This explanation is further supported by the observed increase in plantarflexion when changing from maximally-cushioned footwear to minimal and then barefoot. In contrast to previous work showing cushioned shoes increase ankle abduction moments because of an increased external moment arm as a function of a lateral heel flare (Altman & Davis, 2012), minimal shoes produced a larger peak-ankle abduction moment than barefoot and maximally-cushioned footwear. A potential explanation is that the slight protective cushioning provided by the minimal shoe allowed participants to land on the lateral border of their foot and perceive this as safe, however, without the addition of a cushioned

midsole, impact forces might have been larger explaining increased peak-ankle adduction moment. Collectively footwear influences peak-forefoot pronation (chapter six), and lower-limb kinematics and kinetics. This highlights the importance of footwear choice for recreational runners.

From an injury perspective, simultaneous reductions in both the peak-sagittal-knee-and ankle-joint moments would likely compromise lower-limb function, as a reduction in both would likely overload the hip joint. Therefore, to avoid injury, a manageable mechanical distribution of load within the capacity of lower-limb anatomical structures is necessary to avoid overload of anatomical structures susceptible to injury. However, it can be argued that a shift in the sagittal load from the knee to the ankle joint might reduce injury rates and potentially improve performance. This is because the knee is the most common site of injury and the Achilles tendon when conditioned to a forefoot strike pattern (following an incremental transition period) has the capacity to deal with high loads in the sagittal plane by exchanging translational energy to rotational energy (Lieberman et al., 2010). Beyond this, increasing Achilles tendon load by changing running technique secondary to footwear choice offers substantial energy restitution. The work of Ker et al. (1984) reported the Achilles tendon returns approximately 35% of the energy stored in its structure. In context, if a runner suffers from a joint specific injury underpinned by the demands placed upon it, for example patellofemoral joint pain, an injury characterised by increased sagittal knee joint load (Sinclair et al., 2016), then a shoe that induces a shift in loading from the knee to the ankle might prove beneficial or reduce injury prevalence.

There was no clear difference in peak-trunk lean between the footwear conditions, with values in the current study similar to work by Dos Santos, Nakagawa, Nakashima, Maciel, and Serrão (2016) (9.44 ± 5.19°; 2.67 ± 0.39 $m \cdot s^{-1}$). The predicted increased peak trunk lean when changing from barefoot to maximally-cushioned shoe was expected following increased knee extension at initial contact and the subsequent projection of the foot further in front of the body. A potential explanation for a lack of clear difference might be that the sample size for this study was based precision of estimation for differences in peak-knee adduction moment,

the primary measure in this thesis, and not for peak trunk lean. It is possible that there was insufficient precision to establish clear differences in trunk lean. A larger sample might have elicited a clear difference. This warrants further investigation.

7.4.1 Limitations

As part of the experimental design participants were instructed to run at a speed they could comfortably sustain for 45 minutes. However, during both habituation and data collection the metabolic cost of running was not monitored. Following, there was no way to know running conditions were metabolically equivalent, therefore runners might have been more fatigued in one condition than another. Future studies undertaking similar lines of questioning should consider monitoring the metabolic cost and therefore fatigue when comparing footwear conditions and consider this when interpreting findings.

7.4.2 Conclusion

In conclusion, relative to barefoot the maximally-cushioned shoe induced a more extended knee, dorsiflexed ankle at initial contact compared to both minimal and barefoot, increased the peak-knee flexion moment, and reduced the peak-dorsiflexion moment. Footwear had no clear effects on peak trunk lean. This demonstrates that footwear choice is an important one with chapter six reporting statistical differences in peak-forefoot pronation and the current chapter reporting clear differences in lower-limb kinematics and kinetics. Future work should consider the interactions between foot landing position, footwear choice and lower-limb joint loading.

8.0 The effects of barefoot, minimal and maximally-cushioned running shoes on overstride and peak-knee adduction moment and the relationship between them during overground endurance running in recreational runners.

8.1 Introduction

Previous work has manipulated spatiotemporal variables such as stride length and stride frequency to improve surrogate measures associated with injury (Edwards et al., 2009; Firminger & Edwards, 2016; Heiderscheit et al., 2011). A recent systematic review confirmed that an increased stride frequency resulted in improved shock attenuation, reduced GRF and reduced energy absorbed at the hip, knee, and ankle joints (Schubert et al., 2014). Furthermore, Firminger and Edwards (2016) reported running at 90% of preferred stride length significantly reduced peak-knee flexion moment, among other kinetic knee variables while peak-dorsiflexion moment was unchanged. Additionally, Edwards et al. (2009) showed reduced strain from reducing stride length outweighed negative effects of increased loading cycles. Reducing stride length seems to have the potential to decrease knee-joint loading and potentially reduce injury risk at this frequently injured joint.

Stride length can be altered by altering the distance that the foot lands in front of the body. This is commonly termed 'overstride' (Lieberman, Warrener, et al., 2015). There are a variety of ways that an individual can overstride, yet have an identical stride length (figure 1.3). Lieberman, Warrener, et al. (2015) was first to quantify overstride as the position of the ankle of the lead leg relative to the hip and knee, reporting positive associations between overstride relative to the hip and posteriorly directed braking force ($P = 0.0005$), and between overstride relative to the knee and magnitude ($P = 0.0001$) and rate of loading ($P = 0.07$) of the vertical component of the GRF impact peak. These findings are important in the context of injury risk following an increase in the magnitude of the GRF would increase peak-knee joint moments, measures often associated with running injury etiology and joint health (Dudley et al., 2017; Sharma et al., 1998; Sinclair et al., 2016). Furthermore, Kerrigan et al. (2009) reported a positive association between stride length and peak-knee adduction moment ($r = 0.29$, $P = 0.02$). This demonstrates that reducing stride length can reduce peak-knee adduction moments

and therefore injury risk. However, as identical stride length can be produced in a variety of gait configurations, investigations into overstride relative to the hip and knee are essential to better understand possible effects on joint loading.

Reductions in stride length when running barefoot and in minimal footwear have been consistently reported (Bonacci et al., 2014; Kerrigan et al., 2009). Alterations in GRF characteristics and joint loading patterns have also been reported, though the alteration has not always been a reduction (Sinclair et al., 2015). Evidence for reductions in injury risk with transition to barefoot and minimal shoe running is also mixed (Lieberman, 2012b; Murphy, Curry, & Matzkin, 2013; Sinclair et al., 2015). In opposition to the barefoot/minimal concept, there has been a recent influx of maximally-cushioned footwear in the running-shoe market and the biomechanical effects of such designs are yet to be fully explored. Compared to conventional running shoes, barefoot and minimal shoes reduce stride length and peak-knee adduction moment (a variable associated with injury at the knee) (Dudley et al., 2017; Kerrigan et al., 2009). At the opposite end of the cushioning spectrum, maximally-cushioned shoes could increase stride length and peak-knee adduction moment. The effect of maximally-cushioned running footwear on overstride relative to the hip and knee.

The aim of this study was to examine overstride at the ankle relative to the hip and knee and the relationship between overstride and peak-knee adduction moment during overground ER performed barefoot, in minimal and in maximally-cushioned shoes. It was hypothesised that: 1) overstride relative to the hip would increase from barefoot to minimal to maximally-cushioned shoes; 2) overstride relative to the knee would increase from barefoot to minimal to maximally-cushioned shoes and; 3) overstride relative to the hip and knee would be positively correlated with peak-knee adduction moment.

8.2 Method

8.2.1 Participants

Sample size, participant characteristics and inclusion/exclusion criteria were as described in chapter five (5.2.1).

8.2.2 Experimental design

A repeated-measures design was used to assess the effect of footwear condition (barefoot, minimal and maximally-cushioned shoes (see 3.3.3) on overstride at the ankle relative to the hip and knee, and relationships between overstride and peak-knee adduction moment. Participants were prepared as described in section 3.3.2 and data was collected immediately after a 30-minute habituation run in the relevant footwear for that session (chapter five). Similarly to chapter six, participants ran on separate days in a counterbalanced order separated by 24 hours. Participants were instructed to be well rested and run at a speed described in section 3.3.4. Average running speed was calculated as described in chapter six (6.2.2). Average running speed for barefoot, minimal and maximally-cushioned footwear was 2.48 ± 0.38, 2.60 ± 0.43, 2.68 ± 0.37 $m \cdot s^{-1}$, respectively.

8.2.4 Procedure

Anthropometric measures were recorded and retroflective markers were attached to participants in a full-body 'Plug-In gait' and 'Oxford-Foot Model' formation, as described in section 3.3.2 to facilitate the assessment of lower-limb biomechanics and additional measures in a previous chapter. Kinematic and kinetic data were captured by 14 calibrated infrared cameras (T10/20, Vicon MX, Oxford, UK) and one of four force plates (OR6-7, AMTI, Watertown MA, USA). Signals were captured and imported with equipment described in chapter three, three-dimensional gait laboratory calibration 3.3.1.

8.2.5 Data analysis

Data analysis and processing was undertaken in the 3-D motion analysis software in line with the processes described in chapter three, section 3.3.6. Overstride relative to the hip and knee were defined as the anterior distance between the ankle joint centre and the hip and knee joint centre at initial contact, respectively. Peak-knee adduction moments were calculated as described in section 3.3.6. Data were tabulated in SPSS (version 24.0, SPSS Inc., Chicago, IL) for statistical analysis.

8.2.6 Statistical analysis

After assumptions of normality and uniformity of error were verified, the mean data of each footwear condition were adjusted for speed (by using speed as a covariate), normalising comparisons to a common speed of 2.59 $m \cdot s^{-1}$. Mean difference in overstride relative to the hip and knee were assessed using 90% confidence intervals. After assumptions of normality, uniformity of error and linearity were verified relationships between overstride relative to the hip and knee and the peak-knee adduction moment were assessed using Pearson's correlations in each footwear condition (significance accepted at $P < 0.05$).

8.3 Results

Overstride relative to the hip and knee for the three footwear conditions are illustrated in figures 8.1 and 8.2.

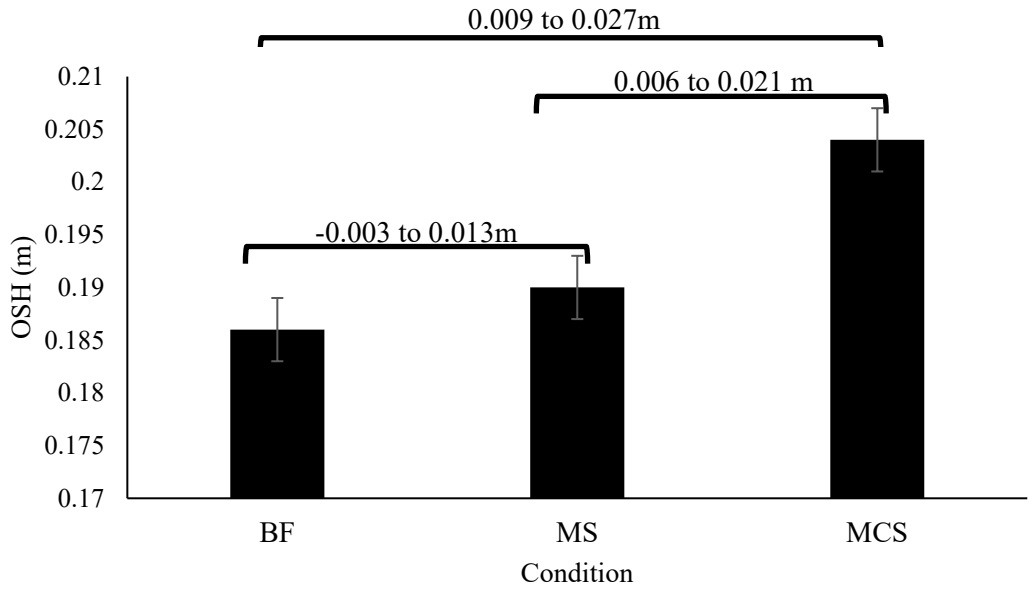

Figure 8. 1 Mean overstride relative to the hip of 15 recreational-endurance runners during overground-endurance running when normalised to a speed of 2.59 $m \cdot s^{-1}$. Values above comparisons are 90% confidence intervals for the population mean difference between conditions. Columns and error bars are mean ± SE. Note, OSH: overstride relative to the hip.

Mean overstride relative to the hip in the maximally-cushioned shoe (0.204 m ± 0.003m) was larger than the minimal shoe (0.190 ± 0.03m) (90% CI 0.006 and 0.021m) and the barefoot condition (0.186 ± 0.003m) (90% CI 0.009 and 0.027m). There was no clear difference in overstride relative to the hip between the minimal and barefoot conditions (90% CI -0.003 and 0.0313m).

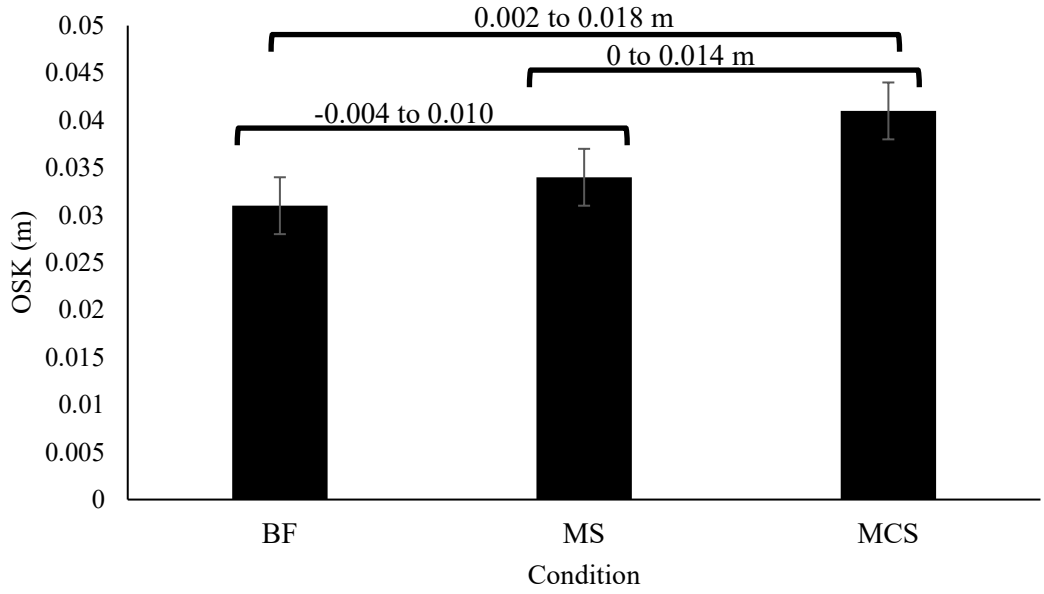

Figure 8. 2 Mean overstride relative to the knee of 15 recreational-endurance runners during overground-endurance running when normalised to a speed of 2.59 $m \cdot s^{-1}$. Values above comparisons are 90% confidence intervals for the population mean difference between conditions. Columns and error bars are mean ± SE. Note, OSK: overstride relative to the knee.

Mean overstride relative to the knee was larger in the maximally-cushioned shoe (0.041 ± 0.003m) than when barefoot (0.031 ± 0.003m) (90% CI 0.002 and 0.018m) but not different from the minimal shoe condition (0.034 ± 0.003m) (90% CI 0 and 0.014m). Overstride relative to the knee was similar in the minimal and barefoot conditions (90% CI -0.004 and 0.010m).

Correlations between the peak-knee adduction moment, and overstride relative to hip and knee are shown in table 8.1.

Table 8. 1 Pearson's correlations between overstride at the hip and knee and the peak-knee adduction moment of 15 recreational runners during overground running while barefoot, in minimal and in maximally-cushioned running shoes. Note, OSK: overstride relative to the knee; OSH: overstride relative to the hip.

Condition	Measure of overstride	Peak-knee adduction moment ($Nm \cdot Kg^{-1}$)		
		r	SEE	*P* value
Barefoot	OSH	0.616	0.319	*P* = .014
	OSK	0.808	0.239	*P* = .000
Minimal shoe	OSH	0.618	0.451	*P* = .014
	OSK	0.818	0.330	*P* = .000
Maximally-cushioned shoe	OSH	0.247	0.407	*P* = .374
	OSK	0.565	0.346	*P* = .028

Moderate to strong and positive associations existed between peak-knee adduction moments and overstride relative to the hip knee in the barefoot and minimal shoe conditions. Only overstride relative to the knee was associated with peak-knee adduction moment in the maximally-cushioned shoe.

8.4 Discussion

The aim of this study was to examine overstride at the ankle relative to the hip and knee, and the relationship between overstride and peak-knee adduction moment during overground ER performed barefoot and in minimal and maximally-cushioned shoes. Results showed that footwear influenced overstride, with overstride relative to the hip being greater in the maximally-cushioned shoe compared to both the minimal and barefoot conditions. Overstride relative to the knee was also larger in the maximally-cushioned shoe compared to the barefoot condition. Moreover, there were positive associations between peak-knee adduction moment

and overstride relative to both the hip and knee in the barefoot and minimal shoe conditions, and between peak-knee adduction moment and overstride at the knee only in the maximally-cushioned shoe.

In support of hypothesis one, as participants changed from maximally-cushioned shoes to minimal or barefoot there was a progressive decrease in overstride relative to the hip. This in line with trends observed in chapter seven where the knee joint was clearly more flexed at IC when transitioning from maximally-cushioned to barefoot running. The biological imperative provides a possible explanation for the effect of footwear on overstride relative to the hip. In the context of locomotion, Kram and Taylor (1990) have shown that the cost of running is inversely related to ground-contact time and that the metabolic cost of running is paid per step. It therefore follows that it is energetically favourable to cover a given distance with fewer steps, facilitated by an extended stride and a resulting longer ground-contact time. However, while metabolically less costly, this movement strategy might not be the least injurious. Increased cushioning, in this case moving from the barefoot, to minimal, to the maximally-cushioned shoe likely insulates mechanoreceptors at the calcaneus from the true forces acting upon the foot, and allows runners to perceive increased overstride relative to the hip as safe (Robbins & Gouw, 1991; Robbins et al., 1994). This was supported by results from chapter seven where knee flexion at IC increased when transitioning from maximally cushioned to barefoot, a measure that likely projected the foot further in front of the hip. This finding is important as previous work has associated increased overstride relative to the hip with increased kinetic measures. In a study investigating overstride and joint mechanics, Heiderscheit et al. (2011) showed increased overstride relative to the COM was associated with increased posteriorly-directed braking impulse and a simultaneous increase in the energy absorbed at the knee joint. Furthermore, Lieberman, Warrener, et al. (2015) reported an increased posteriorly-directed braking force as a function of increased overstride relative to the hip when confounding factors such as limb length were controlled. Runners should therefore exercise caution when running in maximally-cushioned footwear as evidence suggests an increase in overstride relative to the hip, with consequent increases in posteriorly

directed braking force, braking impulse and energy absorbed at the knee that might increase the likelihood of injury.

In support of hypothesis two, overstride relative to the knee was greater in the maximally-cushioned shoe than when barefoot, but there were no clear differences between the minimal shoe and barefoot or maximally-cushioned footwear. This was in line with chapter seven where the knee was more extended when transitioning from barefoot to maximally-cushioned footwear. However, minimal footwear does not clearly alter overstride relative to the knee. The larger overstride relative to the knee to in the maximally-cushioned shoes compared to barefoot was again in line with line with biological imperative argument. Similar to overstride at the hip, participants might increase overstride at knee because they perceive this movement strategy to be safe (due to decreased sense of impact) and are attempting to reduce the cost of locomotion. Lieberman, Warrener, et al. (2015) highlighted the importance of overstride relative to the knee and its association with increased magnitudes of the vertical component of the GRF impact peak. Increased impact magnitudes might increase knee joint moments, given joint moments are the product of both the magnitude of the GRF and the perpendicular distance from the GRF and the respective joint centre.

In support of the third hypothesis, there were moderate-to strong positive associations between overstride and peak-knee adduction moment in all footwear conditions. These findings support previous work by Kerrigan et al. (2009) who investigated the kinematic and kinetic differences between barefoot and conventional-shoe running in a sample of habitually-shod runners. When barefoot, runners reduced their stride length ($P < 0.01$). Reduced stride length correlated with a reduced peak-knee adduction moment ($r = 0.29$, $P = 0.02$). Lieberman, Warrener, et al. (2015), provide a potential mechanistic explanation with the positive association between overstride relative to the knee and the vertical component of the GRF impact peak. As an external joint moment is product of the GRF and the perpendicular distance from the respective joint centre, it follows that an increased overstride relative to the knee might increase the peak-knee adduction moment as a function of an increased GRF. Because increased peak-knee adduction moment is associated with increased injury rates in endurance

runners and general joint health (Dudley et al., 2017; Sharma et al., 1998), the associations between overstride relative to the hip and knee and peak-knee adduction moments suggest future gait retraining interventions should target overstride (particularly relative to the knee).

8.4.1 Limitations

Similar to previous chapters, although participants were asked to run at a speed they could comfortably sustain for 45 minutes, the metabolic cost was not quantified, therefore it was impossible to know if one running condition induced a greater metabolic cost than another did. Following, the influence of fatigue on each running condition was not known. Future studies investigating a similar line of questioning should consider this in their study design and how this might effect overstride.

8.4.2 Conclusion

Changing from maximally cushioned, to minimal shoes or barefoot, reduced overstride relative to hip. Overstride relative to the knee decreased from maximally cushioned shoes to barefoot. There were moderate to strong positive correlations between overstride and the peak-knee adduction moment in all conditions. This further highlights the importance of footwear choice following previous chapters showing footwear statistically influences forefoot pronation (chapter six), lower-limb kinematics and kinetics (chapter seven) and now overstride. Future investigations aiming to reduce the peak-knee adduction moment should consider reducing overstride relative to the hip and knee.

9.0 The effect of a 30-minute coach--led gait retraining intervention on peak-knee adduction moment in recreational runners

9.1 Introduction

As previously noted, running injury rates are high in ER populations, with the knee cited as the most common site of injury (Taunton et al., 2002). Aberrant biomechanics have been proposed as the underpinning etiology behind these injury rates (Napier, Cochrane, Taunton, & Hunt, 2015; Ryan, MacLean, & Taunton, 2006). For example, increased peak-knee adduction moment has differentiated injured from uninjured overground runners (Dudley et al., 2017).

Given the association between abnormal biomechanics and injury rates, studies have aimed to reduce overloading of the lower limbs using interventions including barefoot running (da Silva Azevedo, Mezêncio, Amadio, & Serrão, 2016), minimal shoes (Firminger & Edwards, 2016), and manipulation of spatiotemporal variables such as stride length and frequency (Napier et al., 2015). Manipulating spatiotemporal variables has received much attention with consistent evidence for a reduction in stride length reducing surrogate measures associated with running injury etiology (Edwards et al., 2009; Napier et al., 2015; Schubert et al., 2014).

As discussed in chapter eight, runners can produce an identical stride length with a variety of lower limb configurations, highlighting the need to quantify overstride relative to hip and knee joints. Overstride has been shown to be associated with increased braking forces (Heiderscheit et al., 2011; Lieberman, Warrener, et al., 2015). Chapter eight reported moderate to strong associations between overstride the peak-knee adduction moment. As such, it can be hypothesised that if a runner reduced overstride they might reduce their peak-knee adduction moment and potential for injury (Dudley et al., 2017).

Trunk lean is associated with overstride because of the reflex to increase overstride and improve dynamic stability in response to an anteriorly projected COM as the trunk flexes (Horak & Nashner, 1986; Preece et al., 2016). As such, reduced trunk lean should reduce overstride. This is important as chapter nine showed positive associations between overstride

and peak-knee adduction moment, and Dudley et al. (2017) reported an increased peak-knee adduction moment was associated with injured runners. A coaching intervention designed to reduce overstride by also manipulating trunk lean and with the overarching aim to reduce peak-knee adduction moment has not been undertaken.

The aim of this study was to perform an acute 30-minute coach-led gait retraining intervention to reduce peak-knee adduction moment as a function of reduced overstride relative to the hip and knee, and trunk lean in a sample of recreational runners. It was hypothesised that post intervention, peak-knee adduction moment, overstride relative to the hip and knee and peak trunk lean and would be smaller in the intervention group compared to a control group.

9.2 Method

9.2.1 Participants

With institutional ethics approved, 12 volunteers participated. Eight male and four female participants had mean and SD age, stature and mass of 26 ± 5 yrs, 1.76 ± 0.1 m and 71 ± 14.5 kg. Sample size was estimated for a 90% confidence interval for the difference between experimental and control group post-test mean peak-knee adduction moment to exclude mean differences smaller than within-session typical error from reliability analysis (0.19 $Nm \cdot kg^{-1}$) and to include a smallest worthwhile difference for peak-knee adduction moment of 0.39 $Nm \cdot kg^{-1}$ (Dudley et al., 2017). Inclusion criteria were aged 18-45 years and participation in ER more than once per week as part of habitual-exercise regime, with one run lasting > 30 minutes. Participants were excluded if they had an injury to the lower limbs in the previous six months or any condition that could affect their normal running gait.

10.2.2 Experimental design

A randomised-control-trial (RCT) design was used to assess the effects of a 30-minute gait retraining intervention to reduce peak-knee adduction moment. Participants were randomly allocated to receive either a 30-minute coaching intervention or to simply run for 30 minutes with the running coach using an online coding system (GraphPad, 2017) (six in each group).

Participants were instructed to be well rested before testing. Once consent was attained, participants were provided with a short-sleeved compression top and shorts to improve skeletal representation in biomechanical modelling. For consistency and to eliminate any cofounding effects of footwear choice, each participant was provided with a pair of neutral-cushioned running shoes (Asics Gel Pulse 9 men's and women's respectively). As this model of shoe was a conventional, structured, cushioned and neutral running shoe, a prior habituation period was not necessary for participants. Participants then performed a five-minute indoor overground endurance run as described in 3.3.4 to assess habitual running speed. Data trials were conducted over 20 meters with participants running from an indoor running track through a biomechanics lab as described. Participants ran both before and after either a control 30-minute run or a 30-minute coaching intervention. Running speed post control and post intervention were matched to pre control and pre intervention run speed ±5%. The average running speed for the control group was pre: 2.91 ± 0.28; post: 2.84 ± 0.32. The average running speed for the intervention group was pre: 2.85 ± 0.30; post: 2.84 ± 0.31.

9.2.3 Kinematics

Anthropometric measures were recorded and retroflective markers were attached to participants in a full-body 'Plug-In gait' model, as described in section 3.3.2 to facilitate the assessment of lower-limb biomechanics and trunk lean. Kinematic and kinetic data were captured by 14 calibrated infrared cameras (T10/20, Vicon MX, Oxford, UK) and one of four force plates (OR6-7, AMTI, Watertown MA, USA). Signals were captured and imported with equipment described in chapter three, section 3.3.1.

9.2.4 Coaching intervention

After three acceptable running trials (see data analysis 3.2.4), a coaching intervention was undertaken by a qualified running technique coach. The coaching intervention began with slow running on a motorised treadmill (Woodway ELG2, Germany) at a speed that participants could no longer walk where participants performed three small double-footed jumps every fourth step to reduce overstride and t_c (5 minutes). The jumps serve to trigger

subconscious adjustment of posture (more vertical trunk) to avoid the sensations of unbalance while jumping, as having a greater trunk lean at this time would increase gravitational torque (Romanov & Fletcher, 2007). This then progressed to running at the previously recorded running speed with single small double-footed jumps (five minutes). This was followed by running while keeping cadence in time with a metronome set to 175 beats per minute and with a weighted bar (5Kg) overhead to encourage decreased stride length subsequent to increased cadence and an upright running posture respectively (five minutes). The use of a metronome to manipulate and maintain stride frequency is in line with previous work (Bonacci et al., 2018; Heiderscheit et al., 2011; Roper et al., 2016). Once a vertically aligned trunk and the required cadence was established on the treadmill, participants progressed to overground running with metronome-guided cadence and a bar overhead for 10 minutes. For the final five minutes, participants ran overground without a bar overhead and with metronome guided cadence. During the latter 15 minutes, the coach provided verbal ques to maintain vertically-aligned posture and reduced overstride while matching ground contacts with a metronome. Following the 30-minute intervention, participants performed threes successful post-intervention running trials.

9.2.5 Control group

Control group participants initially ran on a treadmill for 15 minutes at a speed that was recorded in the participant preparation phase. Similar to the intervention group, participants then ran overground at a comfortable speed for the remaining 15 minutes. To replicate coach-participant interaction, the coach spoke to the participant about running related topics, but running technique was not discussed.

9.2.6 Data analysis

Data analysis and processing was undertaken in the 3-D motion analysis software in line with the processes described in chapter three, section 3.3.6. Overstride relative to the hip and knee were defined as described in 8.2.5. Peak-knee adduction moments and trunk lean data were

derived as described in section 3.3.6. Data were then tabulated in SPSS (version 24.0, SPSS Inc., Chicago, IL) for statistical analysis.

9.2.7 Statistical analysis

After assumptions of normality and uniformity of error were verified, differences between intervention and control group means at post-test were estimated with 90% confidence intervals after adjusting for baseline scores as a covariate using SPSS (version 24.0, SPSS Inc., Chicago, IL). Differences in speed that might confound between-group comparisons were examined with independent t tests and found not to be different.

9.3 Results

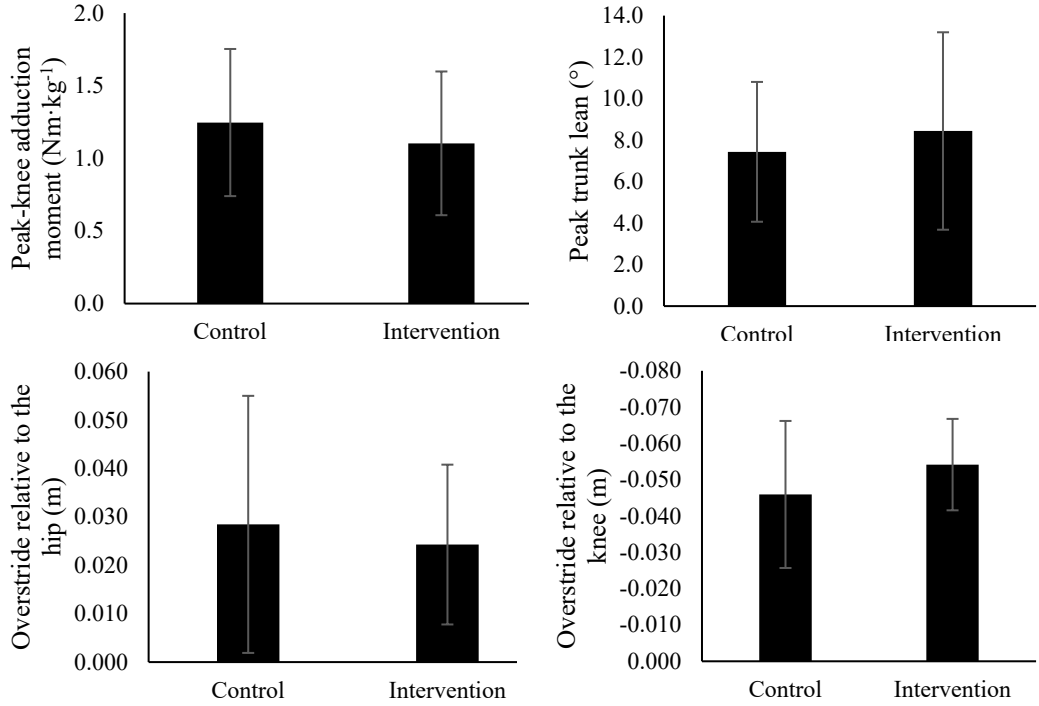

Figure 9. 1 Comparisons between control and intervention group means post 30 minutes of overground endurance running (control) and 30 minutes of gait retraining (intervention). Top left: peak-knee adduction moment; top right: peak trunk lean; bottom left: overstride relative to the hip; bottom right: overstride relative to the knee. Positive overstride values indicate an anterior projection of the ankle joint centre relative to the respective joint centre, negative overstride values indicate the joint centre of interest was anterior to the ankle joint centre. Bars represent the means ± SD.

Despite sample mean effects in the hypothesised direction, after adjusting for baseline values, post intervention peak-knee adduction moment (1.103 ± 0.495 $Nm \cdot kg^{-1}$) was not clearly reduced when compared to the control group (1.246 ± 0.507 $Nm \cdot kg^{-1}$) (90% CI -0.123 to 0.245 $Nm \cdot kg^{-1}$). There was no clear difference in overstride relative to the hip post-intervention (0.024 ± 0.017) compared to control (0.028 ± 0.027 m) (90% CI -0.03 to 0.008 m). Overstride relative to the knee was also similar post-intervention (-0.054 ± 0.013) compared to the control group (-0.046 ± 0.02) (90% CI -0.02 and 0.002 m). Additionally, there was no clear difference in post-intervention peak trunk lean in the intervention group (8.439 ± 4.758°) compared to control the control group (7.435 ± 3.369°) (90% CI -3.627 to 0.197°). Contact time was also

clearly reduced post intervention (0.253 ± 0.023s) compared to the post-control (0.276 ± 0.023s) (90% CI -.037 to -0.008s).

9.4 Discussion

The current study investigated the effects of a 30-minute coach-led gait retraining intervention to reduce peak-knee adduction moment as a function of a reduced overstride relative to the hip and knee and reduced peak trunk lean. Despite sample mean effects in the hypothesised direction, results suggest no statistically clear reduction in peak-knee adduction moment, overstride relative to the hip and knee or peak trunk lean.

Peak-knee adduction moment was not clearly different between groups when comparing post-control and post-intervention measures suggesting the acute gait-retraining did not reduce peak-knee adduction moment by a magnitude sufficient to produce a clear effect. Reported values for peak-knee adduction moment in the current study were marginally less than Kerrigan et al. (2009) who compared conventional shod runners when barefoot and in conventional running shoes (1.43 $Nm \cdot kg^{-1}$). A potential explanation is average running speed differences. Kerrigan et al. (2009) instructed participants to run at a self-selected running speed, but reported a faster average running speed (3.2 $m \cdot s^{-1}$) compared to the current control and intervention groups. It has been previously reported that decreased running speed is associated with a decreased GRF (Nilsson & Thorstensson, 1989). Given that joint moments are the product of the GRF and the perpendicular distance from the joint centre, it is logical that a reduced running speed would reduce the GRF and peak-knee adduction moment. A potential explanation for lack of differences between post-control and post-intervention peak-knee adduction moment is based on the biological imperative and the short time allocated to gait retraining. Following previous work by Kram and Taylor (1990) that reports the cost of running is inversely related to ground-contact time and that the metabolic cost of running is paid per step. It follows, when running in a conventional running shoe that masks the true magnitude of the impact transient (Robbins et al., 1994), the subconscious drive to adopt a running technique that is most efficient yet potentially injurious might have led some participants to revert to their original running technique during data collection trials. However,

t_c was clearly reduced in the post-intervention group. This suggests change was induced in the intervention group, however, the magnitude of this change was not large enough to induce change in other measures such as peak-knee adduction moment. Another contributing factor might have been the short window of time allocated for gait retraining. Thirty minutes might not have been enough for some participants to learn and consistently replicate the desired changes in gait. Collectively, a 30-minute coach-led gait retraining intervention did not clearly reduce the peak-knee adduction moment. Future work should investigate this type of gait retraining after longer sessions performed on multiple occasions.

Trunk lean was not clearly reduced when comparing post-control and post-intervention group means. Values reported in the current study were similar to previous work by Dos Santos et al. (2016) who reported that when performing normal running at a similar speed (2.67 ± 0.39 $m \cdot s^{-}1$) peak trunk lean was 9.44 ± 5.19°. The lack of clear difference between post-control and post-intervention scores could be explained by movement of the compression clothing. It is possible that the ballistic nature of running caused markers to translate relative to the trunk segment during impact. Indeed, if the markers had moved during impact, the kinematics of the trunk would have been misrepresented and potentially explain the high variability for post-control and post-intervention scores. Additionally, sample size for the current study was based on the measurement error of peak-knee adduction moment and not peak trunk lean. Trunk lean is a highly variable measure with coefficients of variation of 45% and 56% for the post-control and post-intervention scores, respectively, a larger sample would be required to reveal a clear effect magnitude and direction. Another explanation is that the time allocated for gait retraining was not sufficient to induce a consistent change in posture, this supported by the large standard deviations in post-control/intervention scores. Following these limitations, future work investigating peak trunk lean should have participants run topless (male) or in a sports bra (female) for a series of coaching sessions following the theoretical framework that an upright posture would reduce the need to overstride, and reduced overstride is associated with reduced peak-knee adduction moment (chapter nine).

Overstride relative to the hip and knee was not clearly different between post-control and post-intervention group scores. A potential explanation is based on the observation that peak trunk lean did not clearly change post intervention. Given that overstride relative to the hip and knee might be dependent on the anterior COM position as a function of a flexed trunk to increase dynamic stability (Horak & Nashner, 1986; Preece et al., 2016), it is logical that if the change in peak trunk lean was unclear then then overstride will be likewise. Notably, overstride relative to the knee produced negative values for post-control and post-intervention groups. This suggests that running at speeds reported in this study (post-intervention: 2.84 ± 0.31; post-control: 2.84 ± 0.32 $m \cdot s^{-1}$) allowed the foot to fall in a near vertical position below the knee. This contrasts previous work at a 3.0 $m \cdot s^{-1}$ reporting positive values for overstride relative to the knee at a range of stride frequencies (Lieberman, Warrener, et al., 2015). This suggests that faster running speeds induce a more anteriorly projected foot relative to the knee in an attempt to increase stride length to satisfy a predetermined running speed, or alternatively faster speeds induce a more flexed trunk and thus a larger overstride to maintain dynamic stability. Future work should consider the effects of posture based interventions on overstride relative to the knee at speeds ≥ 3.0 $m \cdot s^{-1}$ given the reported anterior position of the foot relative to the knee at these speeds, the association between overstride and peak-knee adduction moment (chapter nine), and the association between peak-knee adduction moment and injured runners (Dudley et al., 2017).

9.4.1 Limitations

Although previous work has shown that variables such as stride length and overstride can be successfully manipulated in a single session (Heiderscheit et al., 2011) more coaching sessions might have helped engrain targeted changes in running technique and helped create a clear reduction in peak-knee adduction moment. Future works should undertake a similar intervention, but over a longer period following chapter nine reported associations between overstride and peak-knee adduction moment. In addition, compression clothing might have underpinned the large variability in trunk lean. Following, future work should undertake similar investigation, but attach markers directly to the skin.

9.4.2 Conclusion

There was no clear reduction in peak-knee adduction moment after a 30-minute, coach-led gait retraining intervention. Change in overstride relative to the hip and knee, and peak trunk lean were also unclear. Future studies should attempt to repeat the intervention used here with longer session durations and with multiple sessions, as the rationale for reducing peak knee adduction moment by reducing overstride and trunk lean remains sound and evidence based.

10.0 General discussion

10.1 Key findings

- The method and procedures of chapter four produced reliable measures of 3-D kinematic and kinetic variables that characterise overground ER. Typical error data can be used to set minimal-detectable change thresholds for future comparative and intervention studies. These results were used to inform sample sizes for chapters 4-9.
- A 21-minute overground habituation run is sufficient to establish stable sagittal plane ER kinematics when barefoot, in minimal shoes or maximally-cushioned shoes, where stability is defined as variability equal to or less than a predetermined level of within-session variability.
- Foot structure, specifically increased hallux valgus angle and reduced phalange width are associated with increased peak-forefoot pronation when barefoot, and increased peak-forefoot pronation is associated with increased peak-knee adduction moment when barefoot, in minimal shoes and in maximally-cushioned shoes.
- Maximally-cushioned footwear increased overstride relative to the hip compared to minimal shoes and barefoot running, and overstride relative to the knee compared to barefoot.
- Overstride relative to the knee is associated with peak-knee adduction moment for all footwear conditions and overstride relative to the hip is associated with peak-knee adduction moment when running barefoot and wearing minimal shoes.
- An acute, 30-minute, coach-led gait retraining intervention is not sufficient to clearly reduce peak-knee adduction moment, overstride or trunk lean.

10.2 Discussion of key findings

The primary aims of the thesis were to determine the effects of foot structure on peak-forefoot pronation, how footwear influenced peak-forefoot pronation, if peak-forefoot pronation was associated with peak-knee adduction moment, how footwear changed running technique and, based on the findings, to undertake a running intervention with the aim of reducing peak-knee adduction moment, a variable previously associated with injury and joint health. Key findings and their relationship to foot structure, footwear choice and running technique are discussed below.

10.2.1 The foot

Work in chapter four reported the measurement error of peak-knee adduction moment, and was used to estimate sample size in subsequent chapters that examined the relationship between foot structure and peak-forefoot-pronation angle and the relationship between peak-forefoot-pronation angle and peak-knee adduction moment. Peak-knee adduction moment was the primary measure of interest following an increase in this variable has been associated with poor joint health and injury in endurance runners (Dudley et al., 2017; Sharma et al., 1998; Willy et al., 2012). The findings of this thesis reported that forefoot structure shared a relationship with peak-forefoot-pronation angle when performing barefoot overground ER. Specifically, chapter six demonstrated that when running barefoot, forefoot structure (hallux angle and phalange width) predicted peak-forefoot-pronation angle. Additionally, chapter six showed that, irrespective of footwear condition, increased peak-forefoot-pronation angle was positively associated with peak-knee adduction moment. This is a novel and important contribution to knowledge following an increased peak-knee adduction moment has been associated with poor knee-joint health (reduced joint width) (Sharma et al., 1998), increased patellofemoral joint pain (Willy et al., 2012) and increased injury rates in overground ER (Dudley et al., 2017). These findings relate well to walking gait and unipedal balance literature that reported a reduced forefoot width would reduce the functional axis of the forefoot to control foot motion (Hoogvliet et al., 1997) and landmark work by Morton (1935) that reported an adducted hallux would compromise pronation and likely the loading of lower-

limb joints. These findings are important for recent work that compares peak moments of the knee joint when barefoot and wearing conventional running footwear (Kerrigan et al., 2009). In light of the findings of chapter six, forefoot structure, like shoe design should be considered as a potential contributing factor for differences in variables such as peak-knee adduction moment following the reported relationships between forefoot structure, peak-forefoot-pronation angle and peak-knee adduction moment. Moreover, as foot structure is difficult to manipulate in a short period of time future work might consider examining the effects of long-term footwear use that permits the foot to respond to natural loading in a shoe with minimal constraint and whether this influences forefoot structure, forefoot pronation and subsequently lower-limb joint loading.

10.2.2 Footwear

To date findings from footwear investigations are equivocal with some advocating the barefoot and minimal-footwear movement (Squadrone & Gallozzi, 2009) and others advising against such footwear choices (Willy & Davis, 2013). A logical explanation for this inconsistency and subsequent variable recommendations are variable habituation protocols used by different research designs. Chapter five reported that 21 minutes was necessary for runners to report variability within previously recorded within-session variability in each footwear condition. This is consistent with other work that investigated the time taken to habituate to a novel running condition (barefoot) (Moore and Dixon, 2014). Twenty-one minutes is more than what is typically prescribed in overground-running research (Perl et al., 2012). This highlights when sufficient habituation periods are not provided some previous works will have drawn conclusions from samples unaccustomed to novel footwear that likely represent a habituating response to novel footwear. Following, future studies that attempt to address questions that examine novel-footwear designs should provide a minimum of 21 minutes for habituation before collecting data. Future work should also compare this finding on other running surfaces such as grass.

Footwear also reported statistically clear differences in peak-forefoot-pronation angle when comparing minimal to barefoot and structure-cushioned running footwear. The observation of

increased peak-forefoot-pronation angle in the minimal shoe compared to the maximally-cushioned shoe and barefoot condition was in contrast to previous work in this area that reported the elevation of toes and compression of the hallux to compromise the ability to control forefoot motion (Chou et al., 2009; Hoogvliet et al., 1997). A potential explanation for chapter six observations was that the barefoot condition was unconstraint and the minimal shoe might have still restrained the natural spreading of the Hallux underweight, and the structured-cushioned shoe with a stiff and broad mid-sole might have acted in sequence with the forefoot to provide a wider functional axis. However, this finding must be interpreted in light of potential marker placement error. This was introduced by placing markers on the shoe surface compared to the skin. This issue is discussed in greater depth in the limitations section (10.4). Subsequently, chapter six comparisons should be interpreated in light of potential marker error. Following the potential for footwear choice to statistically influence peak-forefoot-pronation angle and this measure reporting a relationship with peak-knee addcution moment, future work should take care when selecting footwear for interventions. This is because footwear design will likely influence the geometry of the forefoot and potentially the peak-forefoot pronation angle, a measure associated with peak-knee adduciton and subsequently running injury and joint health (Dudley et al., 2017; Sharma et al., 1998; Willy et al., 2012).

Footwear also influenced kinematics and kinetics proximal to the foot. Findings from chapter seven are important and unique with all participants undergoing a habituation protocol prior to data collection. Key findings were as participants transitioned from barefoot, to minimal footwear, to a maximally-cushioned shoe ankle dorsiflexion at initial contact increased and likewise peak-and midstance dorsiflexion increased when transitioning from barefoot to both minimal and maximally-cushioning footwear. The knee was more extended at initial contact in the maximally-cushioned shoe compared to barefoot running, and peak knee flexion moment increased when transitioning from barefoot, to minimal shoe, to maximally-cushioned shoe running. These trends are in line with previous work that report a more extended knee and dorsiflexed ankle at initial contact when transitioning from barefoot, to

minimal, to conventional and maximally-cushioned footwear (Sinclair et al., 2013; Squadrone et al., 2015), and a shift in peak-joint moment from the ankle joint to the knee joint (Sinclair, 2014). This demonstrates that footwear is an important consideration in study design as footwear choice influences lower-limb kinematics and kinetics and footwear can shift loading from one joint to another. This an important consideration for future work when attempting to manipulate joint loading variables that underpin a specific injury mechanism. Future work should also investigate the long-term effects of wearing minimal and maximal footwear.

10.2.3 Technique

Following work in chapter seven that reported increased knee extension at initial contact when transitioning from barefoot to maximally-cushioned, chapter eight investigated how overstride changes as a function of running condition. Chapter eight then investigated how this technique change related to peak-knee adduction moment. Results reported overstride relative to the hip was clearly greater in the maximally-cushioned shoe relative to the minimal shoe and barefoot condition, and overstride relative to the knee was greater in the maximally-cushioned shoe than the barefoot condition. These findings are important in the wider literature as an increased overstride relative to the hip and knee have been associated with higher magnitudes of posteriorly directed braking force and the vertical component of the GRF, respectively (Lieberman et al., 2015). Results also showed that irrespective of footwear condition overstride relative to the knee, and overstride relative to the hip in the barefoot and minimal condition were associated with peak-knee adduction moment. At present, this suggests increased overstride increases the vertical component of the GRF (Lieberman et al., 2015), a key component in the calculation of external lower-limb peak moments, and the peak-knee adduction moment (chapter eight). Future work should be undertaken with the aim of reducing overstride given its association with peak-knee adduction moment, and this measures relation to injury and joint health (Dudley et al., 2017; Sharma et al., 1998; Willy et al., 2012).

Following findings in chapter eight that overstride at the knee and hip were positively associated with peak-knee adduction moment, a 30-minute gait-retraining intervention was designed to reduce both measures. Although peak-knee adduction moment was anticipated to

reduce as a function of reduced overstride and trunk lean, and changes were observed in the hypothesised direction there were no clear difference compared to the control group for any measure of interest. The short intervention time (30 minutes) and not all participants maintaining the desired technique change during data collection might potentially explain a lack of clear differences. Future work should undertake similarly designed interventions but over a greater period of time in an attempt to permanently engrain such changes in participants running technique.

10.3 Original contribution to knowledge

- Recreational endurance runners with no experience running barefoot, in minimal shoes or in maximally-cushioned shoes achieve stability in sagittal plane kinematic measures within a 21-minute overground endurance run.
- When barefoot, foot structure is associated with peak-forefoot pronation angle, with compromised foot structure increasing peak-forefoot pronation.
- Increased peak-forefoot pronation is associated with increased peak-knee adduction moment when running barefoot, in minimal shoes and in maximally-cushioned shoes.
- When comparing maximally-cushioned footwear to barefoot running and minimal footwear, recreational runners land with a more extended knee and dorsiflexed ankle joint, with a shift in sagittal lower-limb loading from the ankle to the knee joint in the maximally-cushioned shoe.
- When wearing maximally-cushioned footwear, overstride relative to the hip is greater than when running barefoot and in minimal shoes, and overstride relative to the knee is greater than when running barefoot.
- Overstride relative to the knee is positively associated with peak-knee adduction moment when running barefoot, in minimal and maximally-cushioned shoes. An increased overstride relative to the hip is associated with peak-knee adduction moment in barefoot and minimal shoe running.
- Thirty minutes of coach-led gait retraining in insufficient to clearly alter overstride, trunk lean and peak-knee adduction moment.

10.4 Limitations

A limitation of the current thesis was the application of markers on footwear and not foot structure. As discussed in chapter six Bishop et al. (2011) highlighted the potential for reported differences when comparing skin mounted and shoe mounted markers, with maximum error for markers used to calculate pronation of 6.9mm. Osis et al. (2016) also reported that a 10 mm change in marker placement could induce a kinematic change up to 7.59°. Scaling this error, marker placement error could increase/decrease confidence intervals by as much as 5.24°. This subsequently removing the statistically clear differences reported in chapter six between footwear conditions. However, Osis et al. (2016) also reported change in marker position of the distal aspects of the forefoot (distal 1st and 5th metatarsal head) induced changes of less than 0.5° in ankle and foot rotations. This suggesting that the forefoot is not as susceptible to change as other markers (lateral malleoli, 7.59°) and clear differences would still exist between the structured-cushioned and minimal shoe. Schultz (2012) has suggested that cutting holes in shoes does not compromise the shoe structure, however, this was a case study design and examined walking gait, not running, where forces exerted on shoe materials are larger. Beyond this, it was more important to maintain ecological validity and examine the kinematics of the forefoot structure in an intact shoe compared to one that might misrepresent forefoot motion whilst under the greater forces associated with running.

The results of this thesis were specific to recreational runners and overground ER. The extrapolation of these findings to elite endurance runners and/or to treadmill running should be done cautiously. Additionally, all trials in this thesis were performed overground and trials were accepted at ±5% of a pre-recorded running speed. This means in chapter four (reliability chapter) small difference in speed might have influenced results. A solution would have been to use an instrumented treadmill, but this equipment was not available during data collection. Additionally, despite runners being instructed to run at a speed they could comfortably sustain for 45 minutes, running speeds in the habituation chapter were different between conditions. Following, it was likely that runners ran further in some footwear conditions than others. Future work should perform similar investigations on a treadmill were running speed and

distance can be controlled. However, as the aim of this study was to investigate overground running, overground running represents a more ecologically valid approach. In addition, the metabolic demand between different footwear conditions was not measured, therefore runners might have been more fatigued in one condition than another. As physiological testing equipment was not used in this thesis it was impossible to know the difference in physiological demand between each condition, however, as instructions were consistent between conditions it is likely that the metabolic demand was similar between conditions.

10.5 Future directions

Chapter five reports a 21-minutes overground run was sufficient for recreational runners to achieve stable technique in novel running conditions. However, runners often run outdoors on surfaces with different and varying levels of compliance. Future work should investigate whether these findings are consistent across different running surfaces such as grass. Additionally, there are a variety of minimal and maximally-cushioned footwear on the market varying in design. Future work should consider the consistency of the current findings across other minimal and maximally-cushioned shoe brands.

Chapter six reported that compromised forefoot structure was associated with increased peak-forefoot pronation angle when barefoot, however, increased peak forefoot pronation angle was not associated a medial shift in the COP relative to the longitudinal axis of the foot. Future work should use plantar-pressure insoles to explore the relationship between magnitudes of peak pressure and attempt to explain the observed relationship between peak-forefoot pronation angle and peak-knee adduction moment.

The intervention undertaken in chapter nine was unable to elicit a statistically clear reduction in peak-knee adduction moment, despite changes in the proposed direction. A potential explanation was that 30 minutes was insufficient for participants to learn and consistently replicate the changes in gait. Future work should repeat the intervention, but over a series of longer sessions to allow participants to consistently produce the desired kinematic and kinetic changes.

10.6 Conclusion

The results of this thesis suggest that foot structure influences forefoot pronation when barefoot, with compromised structure associated with increased forefoot pronation. Furthermore, this thesis revealed footwear is an important consideration, with footwear choice statistically influencing peak-forefoot pronation which was subsequently associated with increased peak-knee adduction moment, a measure associated with running injury. Footwear choice altered lower-limb kinematics and overstride measures, with the maximally-cushioned shoes increasing the anterior projection of the ankle relative to barefoot and minimal-shoe conditions. This finding was notable given the associations between overstride and peak-knee adduction moment.

It is hoped that the findings of this thesis will inform the practice of coaches and scientists working in fields that address foot structure, running footwear, running technique and lower-limb joint loads. Specifically, it is hoped that work of this thesis inspires a continued line of questioning into the associations between foot structure and running biomechanics related to injury. Finally, if the output of this thesis makes the currently complex body of work investigating foot structure, footwear, and running technique and their collective association with running-injury biomechanics more informed then it will have been a worthwhile endeavour.

11.0 Appendices

11.1 Appendix A - Ethical approval letters for experimental study one.

Hi Rich,

The project below has been granted approval. Please keep this message for your records.

Regards

Mick

HLSRS250216

Marker placement reliability of an active population aged 18-45 while running at a preferred endurance velocity.

11.2 Appendix B - Ethical approval letters for experimental study two.

The study listed below has been granted ethical approval. Please keep this message for you records.

Regards,

Mick

HLSRS080216

Relationship between metrics of foot shape and static and dynamic postural stability while barefoot, in minimal-running shoes and in conventional-cushioned-running shoes

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11.3 Appendix C - Ethical approval letters for experimental study three.

20/09/2018

Dear richie.stoneham,

Submission Ref: 1010

Following independent peer review of the above proposal, I am pleased to inform you that **APPROVAL** has been granted on the basis of this proposal and subject to continued compliance with the University policies on ethics, informed consent, and any other policies applicable to your individual research. You should also have current Disclosure & Barring Service (DBS) clearance if your research involves working with children and/or vulnerable adults.

The University's Policies and Procedures are here

All researchers must also notify this office of the following:

- Any significant changes to the study design, by submitting an 'Ethics Amendment Form'
- Any incidents which have an adverse effect on participants, researchers or study outcomes, by submitting an 'Ethical incident Form'
- Any suspension or abandonment of the study.

Please check your approved proposal for any Approval Conditions upon which approval has been made.

Use this link to view the submission: View Submission

Research Ethics Home: Research Ethics Home

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