THE EFFECT OF ANKLE BRACING AND RANGE OF MOTION ON LANDING BIOMECHANICS IN YOUNG NETBALLERS

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ATTESTATION OF AUTHORSHIP

I hereby declare that this submission is my own work and that, to the best of my knowledge and belief, it contains no material previously published or written by another person, nor material which to a substantial extent has been accepted for the award of any other degree or diploma of a university or other institution of higher learning, except where due acknowledgement is made.

Chapters two, three, and five of this thesis represent three separate papers that have been submitted to peer-reviewed journals for consideration for publication. My contribution and the contribution by the various co-authors to each of these papers are outlined at the beginning of the thesis. All co-authors have approved the inclusion of the joint work in this Master's thesis.

Anna Mason-Mackay

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CANDIDATE CONTRIBUTIONS TO CO-AUTHORED PAPERS

Chapter 2 Mason-Mackay, A.; Whatman, C., Reid, D. The Effect of Ankle Bracing on Lower Extremity Biomechanics during Landing: A review of the literature. Submitted to *Journal of Science and Medicine in Sport*	Mason-Mackay 80% Whatman 10% Reid 10%
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ETHICAL APPROVAL

Ethical approval for this research was granted by the Auckland University of Technology Ethics Committee (AUTEC) was granted on 30 June 2014 with a reference number of 14/167.

ABSTRACT

Introduction

Lower extremity injury is common in netball. Reduced ankle dorsiflexion range of motion has been linked to a number of these injuries although the biomechanical reasons for the association are as yet unclear. Ankle injury is particularly common in netball and netballers are encouraged to wear ankle braces to reduce the risk of ankle injury. These braces have the potential to predispose athletes to injury further up the kinetic chain as research has shown braces can restrict ankle dorsiflexion range of motion. The aims of this thesis were to investigate in young netballers: 1) The effect of restricted ankle dorsiflexion range of motion on landing biomechanics and 2) The effect of ankle bracing on landing biomechanics.

Methods

Landing biomechanics were investigated during a drop jump, drop land, and a netball-specific task involving a pass and a one-to-two landing style (unilateral initial landing with the second foot quickly brought down ahead of the first). Dependent variables included leg, knee and ankle stiffness, knee/ankle stiffness ratio, knee and ankle sagittal excursion, peak vertical ground reaction force (vGRF), time-to-peak vertical ground reaction force (TTP), and loading rate (LR) during landing. These variables were investigated using 3D motion capture and force plates. To investigate the association between dorsiflexion range of motion and landing biomechanics participants were divided into high and low dorsiflexion groups and results compared between groups. A within-subject design was used to investigate the effect of lace-up ankle braces with participants performing all tasks with and without braces.

Results

Ankle stiffness was moderately higher in the low DF group on the left during the drop land (ES=0.84) and in the lead limb during the netball jump (ES=0.87). The low DF group also had moderate reduction in ankle excursion on the left during the drop jump (ES=-0.55) and in the trailing ankle during the netball jump (ES=-0.97). Additionally they showed a large increase in knee excursion on the left during the drop jump (ES=1.91) and in the trailing limb during the netball jump (ES=1.85).

In the brace condition there was a small increase in bilateral leg stiffness during the drop land (ES=0.21, 0.22), a small increase in bilateral ankle stiffness during the drop jump (ES=0.37, 0.29), a small to moderate increase in bilateral ankle stiffness during the drop land (ES=0.40, 0.60), and small reductions in the knee/ankle stiffness ratio in all three tasks (ES=-0.22 to -0.45). Additionally, in the brace condition there were small decreases in bilateral ankle sagittal excursion during the drop jump (ES=-0.35,-0.53) and drop land (ES=-0.23,-46) and in the lead limb during the netball jump (ES=-0.36). Finally, in the brace condition there was a small reduction in knee excursion bilaterally during the drop jump (ES=-0.36,-0.40) and in the lead limb during netball task (ES=-0.59), and a small increase in lead limb TTP during the netball jump (ES=0.41).

Conclusion

Young netballers with low DF ROM may exhibit greater ankle stiffness, less ankle sagittal excursion and more sagittal knee excursion during landing than netballers with greater range. Lace-up ankle braces may result in greater leg and joint stiffness and reduced joint excursion during landing but do not appear to affect landing forces. These biomechanical changes may predispose young netballers to lower extremity injury and should be considered in the training and long term use of ankle braces in this group.

CHAPTER 1

INTRODUCTION AND RATIONALE

Background

In 2013 netball was New Zealand's leading women's sport (MyNetball). Netball is an explosive, dynamic, and physically demanding game which incorporates a high degree of fast jumping, hopping, and leaping in order to receive passes and evade opposition players (Hopper, Lo, Kirkham, & Elliott, 1992; Langeveld, Coetzee, & Holtzhausen, 2012; Mothersole, Cronin, & Harris, 2013). Due to game rules which do not allow players to step forward while holding the ball, these explosive jumps are combined with sudden deceleration on landing (Mothersole et al., 2013). The most common landing-style during netball is a one-two foot landing while run-through landings are one of the least-common (Ferdinand, Beilby, Black, Law, & Tomlinson, 2008). This suggests that most landings during a netball game require players to come to a sudden stop with landing forces initially managed on a single limb. Poor landing technique has been reported as one of the most common causes of lower-extremity injury in netball (Hopper & Elliott, 1993; Hume & Steele, 2000). With increasing participation over recent years the Accident Compensation Corporation (ACC) reported a 120% increase in the number of lower extremity netball injury claims from 2008/9 to 2012/13 resulting in an increased cost of almost 10 million dollars (ACC). Knee and ankle sprains, calf strains, and Achilles tendon injuries are the most commonly occurring netball-related injuries (Langeveld et al., 2012; Otago & Peake, 2006).

One possible contributor to poor landing technique is insufficient ankle dorsiflexion (DF) range of motion (ROM). DF restriction is associated with a number of acute and chronic lower-extremity injuries including plantarfaciitis (Kibler, Goldberg, & Chandler, 1991), ankle fractures (Tabrizi, McIntyre, Quesnel, & Howard, 2000), ankle ligament sprains (Hadzic et al., 2009; Tabrizi et al., 2000), achilles tendinitis (Kaufman, Brodine, Shaffer, Johnson, & Cullison, 1999), patellofemoral pain syndrome (Lun, Meeuwisse, Stergiou, & Stefanyshyn, 2003; Piva, Goodnite, & Childs, 2005), patellar tendinopathy (Backman & Danielson, 2011; Malliaras, Cook, & Kent, 2006), and ACL injury (Didier, 2011; Wahlsteadt & Rasmussen-Barr, 2014). The high prevalence of ankle sprains in netball may increase the risk of further injury as research has shown reduced DF following ankle sprain (Aiken, Pelland, Brison, Pickett, & Brouwer, 2008), as well as in athletes with chronic ankle instability (Delahunt, Monaghan, & Caulfield, 2006; Drewes, McKeon, Kerrigan, & Hertel, 2009), and following ankle ligament reconstruction (Baumhauer & O'Brien, 2002). It has been proposed that restricted DF ROM may contribute to poor landing technique as players make kinematic compensations for the reduced range, or by increasing landing stiffness resulting in increased ground-reaction forces (GRF) and loading-rates (LRs). Additionally, netball players are encouraged to wear ankle braces both to support existing ankle injuries and to prevent ankle injury occurring (Hume, 1998; Hume & Steele, 2000; MyNetball). Although ankle bracing has been found to be effective in the prevention of ankle injury (Papadopulos, Nicolopoulos, Anderson, Curran, & Athanasopoulos, 2005) some studies have found they reduce DF ROM (Eils et al., 2002; Eils, Völker, & Rosenbaum, 2007; Parsley, Chinn, Lee, Ingersoll, & Hertel, 2013), potentially increasing the risk of injury to other joints and tissues.

Restricted DF ROM may limit the ability of the leg to pass forwards over the foot (Bolgla, 2004; Mauntel et al., 2013; Piva et al., 2005; Tweed, Campbell, & Avil, 2008) and therefore inhibit the ability to lower the centre of mass during squat-type movements (Macrum, Bell, Boling, Lewek, & Padua, 2012). This restriction may then be compensated for via subtalar and midfoot pronation (Bolgla, 2004; Piva et al., 2005; Tweed et al., 2008) or knee valgus (Bell et al., 2012; Piva et al., 2005), increasing the risk of a number of lower-extremity injuries associated with these movements (Aminaka, Pietrosimone, Armstrong, Meszaros, & Gribble, 2011; Battaglia et al., 2009; Hewett et al., 2005; Lersch et al., 2012; Quatman et al., 2013; Wyndow, Cowan, Wrigley, & Crossley, 2010). Mechanical links between knee valgus and pronation suggest that utilisation of one of these movements may induce the other as knee valgus moves the line of weight bearing to fall medial to the subtalar joint inducing pronation (Barwick, Smith, & Chuter, 2012), and pronation may cause tibial abduction and thus contribute to knee valgus (Powers, 2003). Studies showing increased knee valgus in low DF participants during squat movements lend support to the theory that a similar compensation may occur on landing in the presence of reduced DF range (Bell et al., 2012; Mauntel et al., 2013). Several studies have found greater knee valgus on landing in female participants compared to males (Kernozek, Torry, Van Hoof, Cowley, & Tanner, 2005; McLean et al., 2007).

It is therefore important to increase our understanding of factors contributing to knee valgus in netballers as it is a female-dominated sport (Sport and Recreation New Zealand, 2009).

As DF restriction reduces available ankle sagittal excursion and potentially also reduces knee and hip sagittal excursion (Fong, Blackburn, Norcross, McGrath, & Padua, 2011; Wang, 2009) it may result in a stiffer landing style (Bisseling, Hof, Bredeweg, Zwerver, & Mulder, 2007, 2008; Fong et al., 2011) increasing the risk of injuries associated with higher GRFs and LRs (Bisseling et al., 2007, 2008; Cook, Khan, & Purdam, 2002; Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Hewett, Stroupe, Nance, & Noyes, 1996; Milner, Ferber, Pollard, Hamill, & Davis, 2006; Norcross et al., 2013; Radin, Yang, Riegger, Kish, & O'Connor, 1991). There are a number of proposed methods for calculating stiffness (see Appendix 1) as well as a number of different stiffness measurements including the resistance of the body to vertical displacement (vertical stiffness), resistance to change in leg-length (leg-stiffness), and resistance to angular joint displacement (joint stiffness) (Serpell, Ball, Scarvell, & Smith, 2012). Alterations in stiffness may occur during functional tasks in order to maintain joint stability and prevent limb collapse while attenuating impact forces (Butler, Crowell, & McClay Davis, 2003; Wang, 2009). Greater stiffness during jumping and running tasks contributes to joint stability (Duan, Allen, & Sun, 1997; Granata, Padua, & Wilson, 2002) while greater compliance allows landing forces to be dissipated (Wang, 2009; Williams, Davis, Scholz, Hamill, & Buchanan, 2004; Zhang, 1996). Research suggests that as landing forces increase the stiffness of the lower limb may decrease in order to prevent increased GRFs and loading-rates (LRs), resulting in increased joint-excursion (Wang, 2009). However, if landing forces become too high stiffness may not be able to be sufficiently decreased without compromising stability, leading to greater GRFs and LRs (Wang, 2009; Zhang, 1996). Stiffness may also be modulated in order to enhance functional performance, although the relationship between stiffness and performance during running and jumping tasks is not fully understood (Brughelli & Cronin, 2008). A degree of stiffness is required for utilisation of the stretch-shorten cycle and for the performance of controlled movement with higher-demand activities requiring greater stiffness (Butler et al., 2003). Specifically, stiffness has been found to increase with increased hopping and jumping frequency, running speed, and jump height (Butler et al., 2003).

Questions Addressed in this Thesis

Given the proposed links between restricted dorsiflexion range of motion, bracing and injury, and the limitations in the literature exploring these relationships (especially in netball players), the overall question of this thesis was "How do restricted dorsiflexion range of motion and ankle bracing affect landing biomechanics in young netball players?" The specific questions were:

I. What is the impact of restricted dorsiflexion range of motion on sagittal plane knee and ankle excursion, vertical ground reaction forces, loading rate, and lower extremity stiffness?

II. What is the impact of ankle bracing on sagittal plane knee and ankle excursion, vertical ground reaction forces, loading rate, and lower extremity stiffness?

Structure of the Thesis

This thesis consists of six chapters with an overall discussion in the final chapter. Chapters are written in journal article format as some are being submitted for publication, and as a result there is some repetition of information across the chapters particularly in the methods sections. A single reference list is provided at the end of the thesis rather than at the end of each chapter.

Chapter 2 reviews the current literature regarding the impact of restricted dorsiflexion range of motion on lower-extremity kinematics, kinetics, and stiffness during landing. This review highlights the lack of clarity as to the mechanisms underlying links between reduced dorsiflexion range of motion and biomechanics during landing. Although results were inconsistent between studies and dependent on the landing tasks investigated, there is evidence that restricted DF ROM may result in increased frontal plane ankle motion and knee valgus, reduced knee and hip sagittal excursion, and greater vertical GRF and LR on landing.

Chapter 3 reviews the current literature regarding the effect of ankle bracing on lower-extremity kinematics, kinetics, and stiffness during landing. Although results were again inconsistent and dependent on landing tasks and brace-types investigated, there is evidence that ankle bracing may restrict peak DF angle, knee and angle sagittal plane excursion, and increase GRFs and LRs.

Chapter 4 investigates the impact of low dorsiflexion range of motion on lower-extremity kinematics, kinetics, and stiffness during three different landing tasks. Although low participant numbers resulted in the majority of results being unclear, observed effects suggest that participants with low dorsiflexion range of motion may exhibit greater ankle stiffness, reduced ankle sagittal excursion, and greater knee excursion during landing. These biomechanical changes may predispose young netballers to injury and further investigation is warranted.

Chapter 5 investigates the impact of lace-up ankle braces on lower-extremity kinematics, kinetics, and stiffness during three different landing tasks. Overall findings suggest that lace-up ankle braces increase leg and joint stiffness, and decrease knee and ankle sagittal excursion during landing. These biomechanical changes may increase the risk of lower-extremity injury during netball and should be considered when recommending the long term use of ankle braces in young netballers.

Chapter 6 is an overall discussion of the key findings of the thesis, limitations of the thesis, areas for future research, and concluding statements.

CHAPTER 2

THE EFFECT OF REDUCED ANKLE DORSIFLEXION ON LOWER EXTREMITY BIOMECHANICS DURING LANDING: A SYSTEMATIC REVIEW OF THE LITERATURE

Overview

Restricted ankle dorsiflexion (DF) range of motion (ROM) has been linked to a number of chronic and acute lower-extremity injuries. As a result a number of theories have been proposed to explain how restricted DF may increase injury risk during landing-tasks. It is proposed a loss of DF leads to compensatory pronation and medial knee deviation, and reduced knee and hip excursion resulting in a stiffer landing-style and greater ground reaction forces (GRFs) and loading rates (LRs). In a search of the literature six studies were identified which investigated the effect of restricted DF ROM on landing biomechanics. The overall results were conflicting but there is evidence that restricted DF ROM may result in increased frontal plane ankle motion and knee valgus, reduced knee and hip sagittal excursion, and greater vertical GRF and LR on landing. This may increase the risk of injuries associated with these biomechanical patterns. The focus of studies on specific biomechanical variables rather than biomechanical patterns, analysis of pooled data means in the presence of differing compensation strategies between participants, variation in landing-tasks investigated in different studies, and lack of studies investigating goal-directed sport-specific landing tasks creates difficulty in interpreting results. These areas require further research.

Introduction

Many acute and chronic lower-limb injuries are associated with restricted ankle dorsiflexion (DF) range of motion (ROM) (Backman & Danielson, 2011; Didier, 2011; Hadzic et al., 2009; Kaufman et al., 1999; Kibler et al., 1991; Piva et al., 2005; Wahlsteadt & Rasmussen-Barr, 2014). In New Zealand from July 2012-June 2013 the Accident Compensation Corporation (ACC) received 172,461 new claims for lower-extremity injuries incurred during sport with costs exceeding $15 million NZD (ACC). Ankle injury sustained during sports participation may contribute to injury risk as reduced DF ROM has been reported following ankle sprain (Aiken et al., 2008) and ankle ligament reconstruction (Baumhauer & O'Brien, 2002), with chronic ankle instability (Eils et al., 2007; Parsley et al., 2013), and with ankle bracing (Eils et al., 2007; Parsley et al., 2013).

There are a number of theories as to the biomechanics behind the association between DF restriction and injury. Reduced DF may restrict both the ability to pass the leg forwards over the foot (Bolgla, 2004; Mauntel et al., 2013; Piva et al., 2005; Tweed et al., 2008) and to lower the centre of mass during squat-type movements (Macrum et al., 2012). This may be compensated for via subtalar and midfoot pronation (Bolgla, 2004; Piva et al., 2005; Tweed et al., 2008) or knee valgus (Bell et al., 2012; Piva et al., 2005) both of which have been linked to chronic and acute injury (Aminaka et al., 2011; Hewett et al., 2005; Lersch et al., 2012; Quatman et al., 2013; Wyndow et al., 2010). This theory is supported by studies reporting increased knee valgus during squat movements in participants with reduced DF ROM (Bell et al., 2012; Mauntel et al., 2013).

DF restriction may also increase injury risk by altering lower-extremity stiffness and landing forces. Decreased stiffness on landing results in greater lower-extremity joint-excursion and thereby reduces loading-rate (LR) and ground-reaction forces (GRFs) (Wang, 2009; Williams et al., 2004). Restricted DF and the associated reduction in hip and knee flexion (Fong et al., 2011; Wang, 2009) could therefore increase GRFs or LRs as the reduced joint excursion causes an increase in stiffness (Bisseling et al., 2007, 2008; Fong et al., 2011). Increased landing stiffness has therefore been speculated to increase injury-risk (Butler et al., 2003; Serpell et al., 2012) as increased injury-risk has been reported with both higher GRFs (Hewett et al., 1999; Hewett et al., 1996; Norcross et al., 2013) and higher LRs (Bisseling et al., 2007, 2008; Milner et al., 2006; Radin et al., 1991)

A further possibility is that DF loss is linked to injury via one of a number of compensatory biomechanical patterns rather than through a single common compensatory movement at a particular joint. Dynamical Systems theory approaches goal-directed movement from the perspective that there are multiple biomechanical degrees of freedom (DOF) which work in different patterns to achieve a consistent outcome (Davids, Glazier, Araújo, & Bartlett, 2003; Hamill, Palmer, & Van Emmerik, 2012). Restricted DF may represent a loss of DOF and force individuals into one of a number of alternative movement patterns which may be associated with various injuries. A measure such as stiffness which captures a number of variables into a single measure may be more beneficial for identifying changes in movement patterns than individual biomechanical variables (Lorimer, 2014).

Identifying predisposing factors to injury and the biomechanical factors linked to increased injury risk will assist clinicians in the development of treatment and prevention strategies. Given the above variation in rationale for a link between reduced dorsiflexion and injury incidence, the purpose of this review is to examine the evidence for the effect of DF ROM on peak DF angle, ankle, knee, and hip kinematics, peak vGRF, LR, time-to-peak (TTP) vGRF, and stiffness during landing.

Methods

Search

A preliminary database search (keywords: ankle, dorsiflex*, land*, mechanic*) was conducted on EBSCO Health Databases to identify keywords. A comprehensive search of the literature was then conducted on EBSCO Health Databases on 17/09/2014 (see table 1). Reference lists were scanned to identify further articles.

Table 1: Literature Review Search Strategy

Search		Keywords
Relationship between dorsiflexion ROM and lower extremity kinematic and kinetic variables		(dorsiflex* OR ankle OR talocrural) N8 (range OR ROM OR flex*)
	AND	mechanic* OR biomechanic* OR kinetic* OR kinematic* OR move* OR “ground-reaction force*” OR GRF* OR (force* AND (land* OR load)) OR stiff*
	AND	jump* OR land* OR hop*
	NOT	orthos* OR orthot* OR prosthet* OR prosthes* OR stroke OR "traumatic brain injury" OR "multiple sclerosis" OR disease OR surg* OR repair*

Study selections

Studies were included if they clinically assessed DF ROM goniometrically or via a standing lunge test, or investigated peak DF angle during a landing task. Studies were also required to include at least one of GRF, LR, time-to-peak (TTP) GRF, stiffness, or lower-extremity kinematics during landing included as an outcome. Studies were excluded if participants were injured, or if participant grouping introduced an important confounder such as comparing between genders, different landing-tasks, or in varying states of fatigue. Studies were also excluded if DF restriction was induced by bracing or strapping. Articles were restricted to full text in the English language, no publication date restrictions were imposed.

Data extraction

Data was tabulated under the headings study design, intervention, outcome measures and conclusions (see table 5). The terms ‘knee abduction’, ‘medial knee deviation’, ‘knee valgus’, and ‘knee frontal plane motion’ were considered synonymous. Where the foot model used was not stated and markers were placed at the malleoli, calcaneus, and metatarsal heads, it was assumed that a single-segment foot model was used. As not all studies reported confidence intervals (CIs) 90% CIs were calculated for mean differences and correlations where possible using an Excel spreadsheet (Hopkins, 2007a). For studies that did not report exact p-values the threshold value was used (e.g. $p \leq 0.05$) to calculate the CI. Where studies reported only that the p-value was above a given threshold (e.g. $p > 0.05$) it was not possible to calculate a CI.

Assessment of Quality

Articles were assessed for quality by two reviewers using the modified Downs and Black checklist which is a reliable tool for assessing RCTs and non RCTs (Downs & Black, 1998) (see table 4). Question 27 of the Downs and Black scale was altered to score 1 for sufficient sample size based on power calculation and score 0 for insufficient sample size or power not calculated.

After each study was critiqued, a Quality Index was derived to categorise methodological score. The Quality Index enables studies to be categorised as being of *poor, limited, moderate*, or *strong* quality (see table 2). This measure has been used by several systematic reviews which have rated methodology using the modified Downs and Black checklist (Hartling, Brison, Crumley, Klassen, & Pickett, 2004; Hignett, 2003; Hing, Bigelow, & Bremner, 2009).

Table 2: Categorisation of Quality Index Scores

Total modified Downs and Black checklist score (/28)	Percentage	Quality Index
21+	75% +	Strong
14-20	50-74%	Moderate
7-13	25-49%	Limited
<7	<25%	Poor

Adapted from Hartling et al. (2004), Hignett (2003), and Hing et al. (2009)

Table 3: Levels of Evidence

Strong	Consistent findings among multiple strong quality RCTs
Moderate	Consistent findings among multiple moderate quality RCTs and/or one strong quality RCT
Poor	Consistent findings among multiple low quality RCTs and/or CCTs and/or one moderate quality RCT
Limited	One low quality RCT and/or controlled clinical trial (CCT)
Conflicting	Inconsistent findings among multiple trials (RCTs and/or CCTs)

Adapted from van Tulder, Furlan, Bombardier, and Bouter (2003)

Evidence Synthesis

Levels of evidence were determined as outlined by van Tulder et al. (2003) (see table 3). Evidence was considered to be ‘consistent’ when at least 75% of articles agreed on the key outcomes (Reid, Rydwanski, Hing, & White, 2012).

Results

Studies included in the review

The database search yielded 268 articles of which six met the inclusion criteria (see figure 1). Scores ranged from 17-21/28 on the Downs and Black checklist (see table 4). The major quality issues were a lack of power calculations, not stating source populations, not stating the percentage of those approached who agreed to participate, and a lack of researcher and participant blinding.

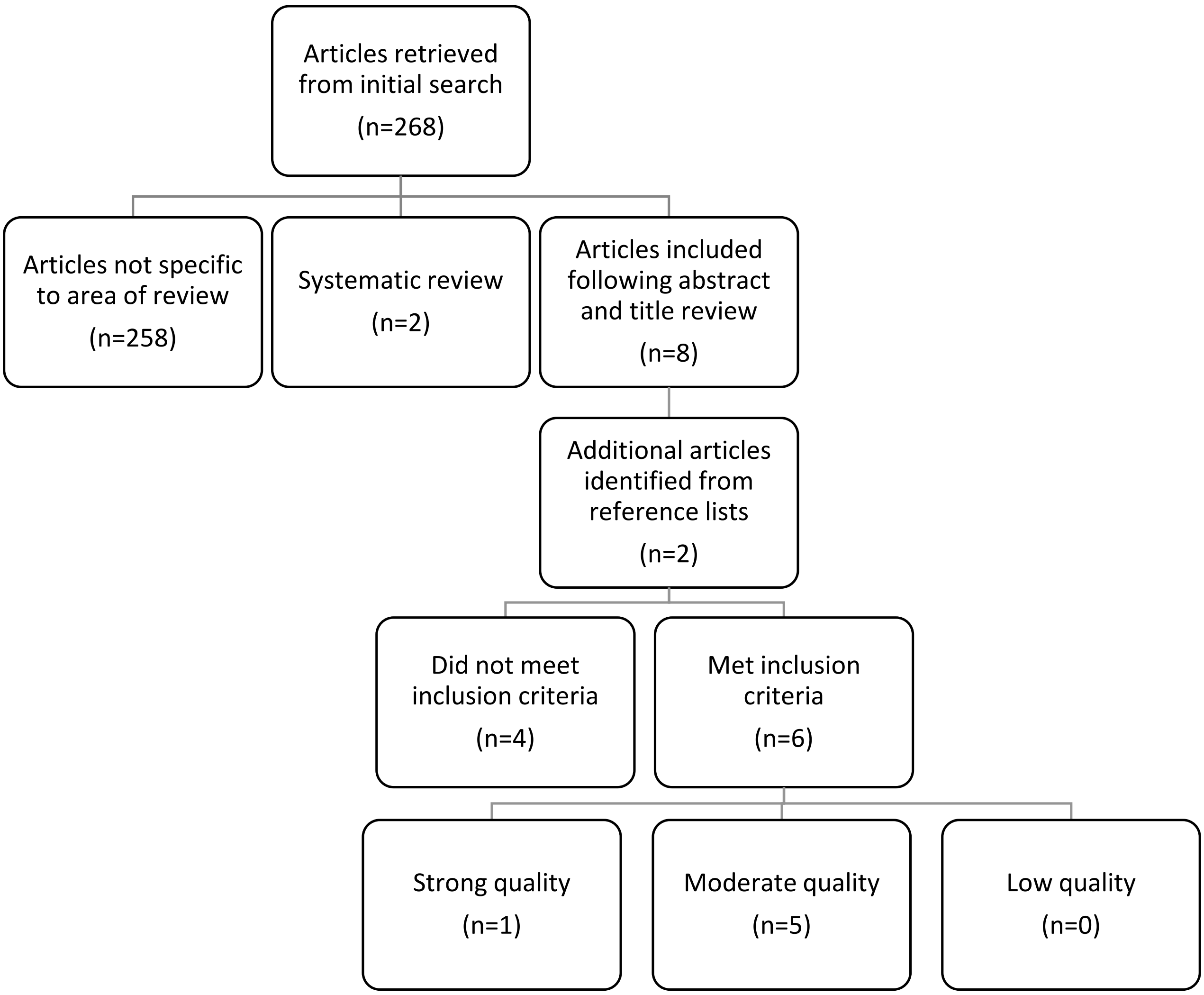

Figure 1: Search Results - Articles reporting on the effect of dorsiflexion ROM (≥21=strong quality, 14-20=moderate quality, 7-13=limited quality, <7=poor-quality)

Evidence classification

Level of evidence determination was based on the quality, consistency, and number of available research articles (van Tulder et al., 2003). Overall there is strong evidence that restricted DF ROM alters landing biomechanics. There is moderate evidence that restricted DF ROM does not reduce peak DF angle on landing and poor evidence for altered frontal plane ankle kinematics. There is moderate evidence that restricted DF ROM alters knee kinematics and poor evidence for altered hip kinematics. Evidence is conflicting regarding the effect of restricted DF ROM on peak vGRF and there is poor evidence for no effect on TTP vGRF. No studies investigated the effect of restricted DF ROM on LR or stiffness.

Table 4: Downs and Black Scores for Articles on Biomechanical Effect of Dorsiflexion Restriction (≥21=strong-quality, 14-20=moderate-quality, 7-13=limited quality, <7=poor-quality)

Study	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22	23	24	25	26	27	Total /28	Quality
Stiffler (2014)	1	1	1	1	2	1	1	0	1	1	1	0	n/a	0	0	1	1	1	1	1	1	1	0	n/a	1	1	1	21	Strong
Whitting (2011)	1	1	1	1	2	1	1	0	1	1	1	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	1	1	0	20	Moderate
Sigward (2008)	1	1	1	1	2	1	1	0	1	1	0	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	1	1	0	19	Moderate
Dill (2014)	1	1	1	1	2	1	1	0	1	1	0	0	n/a	0	0	1	1	1	1	1	1	1	0	n/a	1	1	0	19	Moderate
Malloy (2014)	1	1	1	1	2	1	0	0	1	1	0	0	n/a	0	0	1	1	0	1	1	1	1	n/a	n/a	1	1	0	17	Moderate
Fong (2011)	1	1	1	1	1	1	1	0	1	1	0	0	n/a	0	0	1	1	0	1	1	1	1	n/a	n/a	0	1	0	16	Moderate

Table 5: Effect of DF ROM on Landing Biomechanics

Article	Participants	Groups	Measurement			Landing-Task	Results		Conclusions
			Kinematics	Kinetics	DF		Kinematics	Kinetics	
Stiffler (2014)	Gender, no.: M,: 28 F. 69 Age (y): CON=20.3±1.5 MKD=20.2±1.4 Sport: Recreationally active	Participants grouped based on presence (MKD) or absence (CON) of medial knee deviation during an overhead squat	Standard video cameras in frontal and sagittal planes Videos used to score participants on the Landing Error Scoring System (LESS)	-	Open-chain pROM and aROM with knee extended and in 30° flexion using standard goniometer	Forward jump from 30cm box to a distance equal to 50% of participants' height, bilateral landing	Greater extended-knee DF pROM and aROM in CON than MKD group Mean difference (pROM) = 2.62°, CI = 0.48-4.8 Mean difference (aROM) = 3.58°, CI = 1.1-6 Significantly greater number of MKD participants displayed knee valgus at or medial to the great toe than CON participants Difference = 17.7% (29 participants) No significant difference in flexed-knee DF ROM or LESS score for IC KF or IC knee valgus angles between groups	-	Greater knee valgus displacement associated with reduced extended-knee DF aROM and pROM No difference in LESS score for IC knee flexion or IC knee valgus angles with extended-knee DF aROM or pROM No difference in LESS score for knee valgus displacement, IC knee flexion, or IC knee valgus angles with reduced flexed-knee DF ROM
Whitting et al (2011)	Gender, no.: M, 33 Age (y): 22.5±4.7 Sport: Physically active	Participants separated into high DF ROM (HDF) and low DF ROM (LDF) groups	3D motion capture Single-segment foot, unknown whether shoe or foot motion	Force platform (Peak vGRF, TTP vGRF)	Measured with a goniometer during standing lunge test	Unilateral drop-landing onto dominant leg from heights of 32cm and 72cm	No significant difference in peak DF angle or IC PF angle between groups Greater EV angle at time of peak Achilles tendon force in LDF than HDF group Mean difference = 3.8°, CI = 0.67-6.9	No significant difference in peak vGRF or TTP vGRF between groups	Greater ankle eversion angle at some time-points during landing with DF restriction No effect of DF restriction on peak ankle EV angle, peak vGRF, or TTP vGRF

			was measured (data used to create virtual ankle models from which sagittal and frontal plane joint angles and TTP angles were calculated)				Greater EV angle at time of peak DF in LDF than HDF group Mean difference = 4.2°, CI = 0.88-7.5 No significant difference in peak EV angle or EV angle at time of peak PF moment between groups		
Sigward et al (2008)	Gender, no.: F, 39 Age (y) 15.5±1.0 Sport: Soccer	Within-subject design	6-camera 3D motion capture (knee frontal plane kinematics)	-	pROM with knee flexed to 30° using a standard goniometer	Drop-jump from 46cm platform bilateral landing, immediately perform maximal vertical jump	DF ROM correlated with frontal plane knee excursion (*r*=-0.27), CI = -0.5-0	-	Increased frontal plane knee excursion with DF restriction
Dill (2014)	Gender, no.: M, 20 F, 20 Age (y): Normal group (NWB) =20.70±1.98	Participants grouped based on DF ROM during weight-bearing and non-weight-	Electromagnetic motion-tracking system Single-segment foot, measured	-	NWB test: Open-chain pROM measured with knee extended using a	Forward jump from 30cm box to distance equal to 50% participants' height, bilateral landing, immediately perform	No significant difference in peak DF, or knee sagittal, frontal or transverse plane excursion or peak angles, between groups	-	No effect of DF ROM on peak DF, or knee sagittal, frontal or transverse excursion

	Restricted group (NWB)=19.45±1.40 Normal group (WB) =20.70±1.95 Restricted group (WB)=19.45 ±1.43 Sport: Physically active	bearing tests Limited= ≤5° DF Normal= ≥15° DF	in-shoe foot motion (peak ankle DF, knee sagittal, frontal, and transverse plane kinematics)		standard goniometer WB test: Range measured with a digital inclinometer during a standing lunge	maximal vertical jump			
Malloy (2014)	Gender, no.: F, 23 Age (y): 19.4±0.84 Sport: College soccer	Within-subject design	14-camera 3D motion-capture Single-segment foot, unknown whether shoe or foot motion was measured (Peak ankle DF,	Force plate (peak vGRF)	Active-assisted DF ROM with knee extended using standard goniometer	Drop-jump from height equal to vertical displacement of PSIS during participants' maximal jump height, bilateral landing, immediately perform maximal vertical jump	DF ROM correlated with: Peak KF (*r*=0.385), CI = 0.04-0.65 Peak knee abduction (*r*=0.355), CI = 0.00-0.63 No significant correlation between DF ROM and peak DF angle	No significant correlation between DF ROM and peak vGRF	Reduced peak knee flexion and greater knee abduction with DF restriction No effect of DF ROM on peak DF angle or peak vGRF

			knee frontal and sagittal plane kinematics)						
Fong et al (2011)	Gender, no.: M, 17 F, 18 Age (y): 20.5±1.5 Sport: Physically active	Within-subject design	7-camera 3D motion-capture Single-segment foot, measured in-shoe foot motion (knee frontal and sagittal plane kinematics; hip and ankle sagittal plane kinematics)	Force plate (peak vGRF)	Open-chain pROM with knee extended and in 90° flexion using standard goniometer	Forward jump from 30cm box to a distance equal to 40% of participant's height, bilateral landing, dominant foot on force plate	Extended-knee DF ROM correlated with: KF excursion (*r*=0.464), CI = 0.21-0.66 HF excursion (*r*=0.357), CI = 0.08-0.58 No significant correlation between extended-knee DF ROM and KV excursion or DF excursion No significant correlation between flexed-knee DF ROM and KF excursion or KV excursion	Extended-knee DF ROM correlated with: Peak vGRF (*r*=-0.411), CI = -0.62--0.15 No significant correlation between flexed-knee DF ROM and peak vGRF	Reduced hip and knee sagittal excursion, and increased peak vGRF with DF restriction (measured with extended-knee) No effect of DF range on knee frontal plane excursion

IC=Initial Contact; DF = Dorsiflexion; PF = Plantarflexion;, EV = Eversion; KF = Knee flexion; KV = Knee valgus; HF = Hip flexion; ROM = Range of motion; pROM = Passive range of motion; GRF = Ground reaction force; vGRF = Vertical ground reaction force; LR = Loading rate; TTP = Time-to-peak, WB = Weight-bearing; NWB = Non-weight-bearing

Study Parameters

There were a total of 267 participants across all studies (112 female, 98 male) with an age range of 15.5±1.0 to 22.5±4.7 years. Kinematics were recorded via camera-based 3D motion-capture systems with the exception of Stiffler, Pennuto, Smith, Olson, and Bell (2014) who used standard video cameras in two planes, and Dill, Begalle, Frank, Zinder, and Padua (2014) who used an electromagnetic tracking system. Force plates were used to collect kinetic data. Kinematic variables investigated included sagittal and frontal plane ankle and knee motion, and sagittal plane hip motion. All studies investigated foot kinematics used a single-segment foot model, two (Dill et al., 2014; Fong et al., 2011) measured in-shoe foot motion and two (Malloy, Morgan, Meinerz, Geiser, & Kipp, 2014; Whitting, Steele, McGhee, & Munro, 2011) did not state whether shoes were worn. Kinetic variables included peak vGRF and TTP vGRF. All studies measured DF ROM with a standard goniometer with the exception of Dill et al. (2014) who used a digital inclinometer. DF ROM test positions included open-chain knee-extended, open-chain knee-flexed, and standing lunge test. No studies were identified which investigated lower-extremity stiffness. Three studies used a within-subject repeated-measures design (Fong et al., 2011; Malloy et al., 2014; Sigward, Ota, & Powers, 2008), two compared high and low DF ROM groups (Dill et al., 2014; Whitting et al., 2011), and Stiffler et al. (2014) compared medial-knee-deviators to non-medial-knee-deviators during a squat. All landing tasks involved dropping or jumping forward from a box ranging from 30-72cm in height with Malloy et al. (2014) basing box height on each participant's maximal vertical jump height. Three studies used a stop-landing-style (Fong et al., 2011; Stiffler et al., 2014; Whitting et al., 2011), three included a subsequent jump immediately on landing (Dill et al., 2014; Malloy et al., 2014; Sigward et al., 2008), and all landings were bilateral with the exception of Whitting et al. (2011) who investigated a unilateral landing.

Stiffler et al. (2014) scored participants on the Lower-Extremity Scoring System (LESS) during a bilateral forward jump from a 30cm box to a distance equal to 50% of participants' height. The LESS includes measures of knee valgus and IC knee angle which were reported individually. 97 recreationally active participants (28 male, 69 female) were split into those who displayed medial knee deviation past the first ray during an overhead squat (MDK group, age=20.2y±1.4) and those who did not medially deviate (CON group, age=20.3y±1.5). Kinematic data was collected via standard video cameras in the frontal and sagittal planes, and open-chain aROM and pROM was measured goniometrically both in full knee extension, and in 30° knee flexion. Mean values for DF ROM, overall LESS, and each LESS component during the landing task were analysed statistically between MKD and CON groups. The study was included in the review as it found a significant difference in both flexed and extended-knee DF ROM between groups and therefore allowed a comparison of groups based on DF range.

Whitting et al. (2011) investigated sagittal and frontal plane ankle motion, peak vGRF and time-to peak (TTP) TTP vGRF during unilateral drop-landings from heights of 32cm and 72cm. 48 physically active men were ranked from high to low DF ROM measured via a standing lunge test and the middle 15 participants were removed to form high and low DF groups (n=33, age=22.5y±4.7). Kinematic data obtained via 3D motion-capture (number of cameras not specified) was used to create virtual ankle models from with joint angles were calculated. Mean values for each dependant variable were analysed statistically between groups.

Sigward et al. (2008) investigated knee frontal plane motion during bilateral drop-jumps from a 46cm height. 39 female soccer players (age=15.5y±1.0) participated in the study. Kinematic data was collected via a 6-camera 3Dmotion capture system. Passive DF ROM was measured goniometrically with the knee flexed to 30°. Correlation coefficients were calculated between DF ROM and knee frontal plane excursion.

Dill et al. (2014) investigated peak ankle DF, and knee sagittal, frontal, and transverse plane motion during a bilateral forward jump from a 30cm box to a distance equal to 50% of participants' height with a maximal vertical jump performed immediately upon landing. 40 physically active participants (20 male, 20 female) participated in the study (see table 5 for participant age distribution). Kinematic data was collected via an electromagnetic motion-tracking system, open-chain DF pROM was measured goniometrically, and closed-chain DF ROM during a standing lunge test was measured via a digital inclinometer. Participants were divided into limited range (DF ROM≤5°) and normal range (DF ROM ≥15°) groups with each participant being grouped twice based on range achieved during the two different DF tests.

Mean values for dependent variables were analysed statistically between open-chain DF ROM groupings and closed-chain DF ROM groupings.

Malloy et al. (2014) investigated peak ankle DF, knee sagittal and frontal plane motion, and peak vGRF during bilateral drop-jumps. Participants jumped from a height equal to the vertical displacement of the right PSIS during a maximal jump for height. 23 female college soccer players participated in the study (age=19.4y±0.84). Kinematic data was collected via a 14-camera motion capture system. Active-assisted DF ROM was measured goniometrically with the knee extended. Correlation coefficients were calculated between DF ROM and dependent variables.

Fong et al. (2011) investigated ankle, knee and hip sagittal motion, knee frontal motion, peak vGRF and TTP vGRF during bilateral forward jumps from a 30cm height to a distance equal to 40% of participants' height. 35 physically active participants (17 male, 18 female) participated in the study (age=20.5y±1.5). Kinematic data was collected via a 7-camera 3D-motion capture system. Passive DF ROM was measured goniometrically both with the knee in full extension, and with the knee in 90°-flexion. Correlation coefficients were calculated between DF ROM and dependent variables.

Key findings

Five of six studies found changes in landing kinematics (Fong et al., 2011; Malloy et al., 2014; Sigward et al., 2008; Stiffler et al., 2014; Whitting et al., 2011). Four studies found no effect of DF ROM on peak DF angle (Dill et al., 2014; Fong et al., 2011; Malloy et al., 2014; Whitting et al., 2011) while Whitting et al. (2011) reported altered frontal plane ankle motion in association with reduced DF range. Malloy et al. (2014) found reduced peak knee flexion angle and Fong et al. (2011) found reduced sagittal knee excursion with DF restriction while Dill et al. (2014) found no effect of DF ROM on knee peak flexion angle or sagittal excursion. Stiffler et al. (2014) and Malloy et al. (2014) found greater peak knee valgus and Sigward et al. (2008) found greater frontal plane knee excursion in association with reduced DF ROM while Dill et al. (2014) found no effect of DF ROM on peak knee valgus and both Dill et al. (2014) and Fong et al. (2011) found no effect on knee frontal excursion. Dill et al. (2014) also found no effect of DF ROM on transverse plane knee peak angle or excursion. Fong et al. (2011) found reduced sagittal plane hip excursion with DF restriction.

Three studies investigated the effect of DF ROM on landing kinetics (Fong et al., 2011; Malloy et al., 2014; Whitting et al., 2011). Fong et al. (2011) found increased vGRF with DF restriction, while Whitting et al. (2011) and Malloy et al. (2014) found no effect of DF ROM on peak vGRF and Whitting et al. (2011) found no effect on TTP vGRF.

Discussion

Although five of the six reviewed studies found a significant association between DF range and landing biomechanics, results for each measured variable were inconsistent. This inconsistency may be due in part to variations in landing tasks investigated in each study and to variability in landing strategy within and between participants.

Kinematics - Ankle

Studies by Whitting et al. (2011), Fong et al. (2011), Dill et al. (2014), and Malloy et al. (2014) found that peak DF angle on landing was unaffected by DF ROM. Although these results call into question the impact of DF range on landing kinematics, all studies except Dill et al. (2014) found an association between DF range and other kinematic variables indicating that some degree of kinematic compensation was necessary. Fong et al. (2011) and Malloy et al. (2014) suggest that the lack of correlation is due to individual variations in landing technique and suggest that PF angle at initial contact (IC) may be adjusted in order to maintain adequate DF range in the presence of restricted mobility. Fong et al. (2011) noted that participants displayed a wide variation in IC ankle angle at IC (SD=±15°; range=60°) with some participants landing in a very plantarflexed position and thus maximising available DF range and others landing in more dorsiflexion. This coordinative variability may have resulted in a non-significant correlation between mean values while on an individual level there were important changes.

Dill et al. (2014) suggest that the lack of correlation between DF ROM and peak DF angle may have been due to their landing-task incorporating a jump immediately upon landing. They note that recoiling quickly in preparation for a second jump may reduce the amount of DF range utilised. This is supported by Arampatzis, Schade, Walsh, and Brüggemann (2001) who found that as ground-contact time between landing and a subsequent jump decreased sagittal displacement at the ankle decreased. Malloy et al. (2014) and Sigward et al. (2008) also investigated a landing-task with a subsequent jump, potentially contributing to the non-significant correlation.

The majority of studies measured DF range in the open-chain position which has been suggested to underestimate true functional DF range (Krause, Cloud, Forster, Schrank, & Hollman, 2011). Studies comparing DF ROM measurement in closed and open-chain positions have found significantly greater ranges during functional tests such as the standing lunge (Dill et al., 2014; Krause et al., 2011). Underestimating the DF range available as a result of landing-forces may further explain the lack of correlation found between DF range and peak angle in these studies. However, Dill et al. (2014) and Whitting et al. (2011) measured DF range during a standing lunge and although the authors found that the standing lunge test did yield greater DF range than an open-chain measurement, neither study found a correlation between DF ROM and peak DF angle on landing. It is possible that the standing lunge may still underestimate the DF range available during landing as landing-forces may allow greater DF angle to be achieved.

Excessive ankle eversion has been linked to Achilles tendon injury (Lersch et al., 2012; Wyndow et al., 2010) and is biomechanically linked to other movements associated with lower-extremity injury risk such as pronation (Hintermann & Nigg, 1998) and knee valgus (Petersen et al., 2013). Whitting et al. (2011) found that participants with low DF exhibited greater ankle eversion angle at peak Achilles tendon force (calculated by dividing internal PF moment by Achilles tendon moment arm) and at peak DF angle during a unilateral drop-landing. The authors speculated that increased eversion at times of greatest Achilles load and triceps surae end-range will further increase Achilles loading and predispose to tendon injury

Kinematics - Knee, Hip

Reduced knee sagittal excursion on landing may increase lower-extremity stiffness and increase injury risk (Bisseling et al., 2008; Wang, 2009; Williams et al., 2004) Although greater hip stiffness has not been found to increase GRFs or LRs (Wang, 2009), hip stiffness is a contributor to leg stiffness (Farley & Morgenroth, 1999). Hip stiffness may therefore have a bearing on injury risk as excessive leg stiffness is associated with greater GRFs and LRs and is speculated to increase injury-risk (Butler et al., 2003). Reduced extended-knee DF ROM was found to be moderately correlated with sagittal hip and knee excursion (Fong et al., 2011) and with peak knee flexion angle (Malloy et al., 2014), suggesting that sagittal knee and hip displacement may have been restricted by reduced DF range. Although Fong et al. (2011) found that the correlation between knee excursion and 90°-flexed-knee DF ROM was non-significant the CI suggests a small-to-moderate (Hopkins, 2000) correlation which may be important (r=0.33, CI=0.05-0.56). The limited size of these correlations may reflect movement variability as outlined by Dynamic Systems theory and the wide range of potential consequences of restricted joint range (Davids et al., 2003; Hamill et al., 2012). In contrast, Dill et al. (2014) found no significant difference in knee sagittal plane excursion, peak angle, or IC angle between high and low DF groups. It is possible that the incorporation of a vertical jump on landing contributed to the non-significant result in a similar manner to that described above regarding peak DF angle. Stiffler et al. (2014) also found no difference in IC knee angle between the low-DF MKD group and high-DF control group, although the authors suggested this may have been due to the study not being sufficiently powered.

Increased knee frontal plane excursion has been implicated in a number of lower-extremity injuries (Nakagawa, Moriya, Maciel, & Serrão, 2012; Quatman et al., 2013) and has been proposed to occur in compensation for restricted knee and ankle sagittal excursion (Bell, Padua, & Clark, 2008; Bell et al., 2012; Macrum et al., 2012; Mauntel et al., 2013). This theory is supported by the greater number of medial knee deviators in the low DF group reported by Stiffler et al. (2014) and the correlations between DF ROM and peak knee valgus and knee frontal plane excursion found by Malloy et al. (2014) and Sigward et al. (2008) respectively. The authors of all three studies theorised that DF restriction directly restricted knee flexion by limiting forward progression of the tibia and suggest that participants compensated via increased medial knee deviation. This theory is supported by the reduced knee sagittal excursion found by Fong et al. (2011) and increased ankle eversion found by Whitting et al. (2011). However as noted previously, Stiffler et al. (2014) found no reduction in peak knee flexion in the reduced DF group, possibly due to the study being underpowered for this variable. This study also grouped participants based on the presence or absence of medial knee deviation (MKD) during an overhead squat introducing the possibility that the MKD group might naturally preferre a valgus movement strategy during landing tasks. It is therefore possible that the knee valgus on landing demonstrated by the MKD group was due to factors other than the lower DF range found in this group. It should also be noted that the correlations found by Malloy et al. (2014) and Sigward et al. (2008) were moderate and weak respectively, highlighting the large number of possible compensations for restricted joint range.

Conversely, although Fong et al. (2011) found reduced knee sagittal excursion in association with restricted DF, the correlation with frontal plane knee excursion was non-significant. This study also found a negative correlation between DF ROM and GRFs suggesting that the lack of kinematic compensation for reduced sagittal range may have led to greater GRFs. Furthermore, low power and the CIs calculated for this variable suggest there may have been a small correlation with DF ROM (Hopkins, 2000).

Kinetics

Greater GRFs and LRs have been implicated in the aetiology of a number of lower-extremity injuries (Bisseling et al., 2007, 2008; Hewett et al., 1999; Hewett et al., 1996; Milner et al., 2006; Norcross et al., 2013; Radin et al., 1991). Consistent with the theory that restricted DF range increases landing stiffness and conseqently increases lower-extremity loading, Fong et al. (2011) found a small negative correlation between extended-knee DF ROM and peak vGRF while the flexed-knee DF correlation was non-significant. The authors contend that as the flexed-knee measurement eliminates the influence of gastrocnemius on ankle ROM the extended-knee measurement is a more valid representation of true DF ROM. However, the relevance of this to landing is debateable as the mean peak knee angle during landing was 80.2°±13.3° which will similarly reduce the influence of gastrocnemius on DF ROM. Although non-significant, a small-to-moderate correlation was found between flexed-knee DF ROM and vGRF. The authors noted that p-values approached significance suggesting a trend and that statistical power for these variables was insufficient. Furthermore, the CI for vGRF suggests a trivial-to-moderate correlation (Hopkins, 2007a).

In contrast, Whitting et al. (2011) and Malloy et al. (2014) found no effect of DF restriction on peak vGRF with Whitting et al. (2011) also reporting no difference in TTP vGRF between high and low-DF groups. The kinematic results of both studies indicate compensatory frontal plane movement with Malloy et al. (2014) reporting greater peak knee abduction angle and Whitting et al. (2011) reporting greater ankle eversion which is biomechanically associated with knee valgus (Petersen et al., 2013). The difference in kinetic results between these two studies and Fong et al. (2011) may be therefore due to the lack of kinematic compensation made by participants in Fong et al. (2011) leading to higher GRFs while participants in Malloy et al. (2014) and Whitting et al. (2011) may have altered landing kinematics in order to attenuate forces. However this is speculative as Whitting et al. (2011) did not measure hip or knee kinematics.

Landing-tasks

Differences in landing tasks between studies make comparisons difficult and may contribute to inconsistent results. There is evidence that the biomechanical demands of a landing task change with varying height, distance, goals (e.g. catching a ball), and landing style (e.g. unilateral vs bilateral), and result in participants utilising different landing strategies (Cruz et al., 2013; Mothersole et al., 2013). Caution must therefore be taken when comparing studies investigating different landing tasks. Although all studies investigated sagittal plane landings from a platform, Stiffler et al. (2014), Whitting et al. (2011) and Fong et al. (2011) investigated stop-landings while all other studies included a subsequent jump immediately upon landing. This may contribute to the inconsistent results as the shorter ground-contact time and need to recoil and prepare for a second jump may cause reduced joint excursion and increased stiffness compared with stop-landings (Arampatzis, Schade, et al., 2001). Furthermore, landings in Stiffler et al. (2014), Dill et al. (2014), and Fong et al. (2011) were from a 30cm height while Whitting et al. (2011) investigated heights of 32cm and 72cm and Sigward et al. (2008) investigated a 42cm height. Greater forces associated with greater height may have resulted in participants adopting kinematic compensation strategies in the frontal plane to attenuate landing forces while participants in Dill et al. (2014) and Fong et al. (2011) were not motivated to do so as forces were not sufficiently high. Malloy et al. (2014) based platform height on the maximal jump height of each participant, potentially making results more consistent between participants as the challenge of the landing was tailored to height and physical ability, and also more sport-specific as it mimics the height a participant would land from when jumping during competition. Whitting et al. (2011) was the only study to investigate unilateral landings. Attenuating forces through a single limb may have further contributed to in the kinematic results found this is study compared with those of Dill et al. (2014) and Fong et al. (2011). The landing-tasks in Stiffler et al. (2014), Dill et al. (2014), and Fong et al. (2011) also involved horizontal jumps from a height rather than straight vertical drops which may have affected both kinematic and kinetic results (Ali, Andersen, Rasmussen, Robertson, & Rouhi, 2013).

Coordinative variability

Individual landing strategies and interactions between biomechanical variables also create difficulty in interpreting results. The ability to achieve a consistent endpoint with a variety of movement patterns is referred to as coordinative variability and the particular strategy used can vary widely both within and between individuals (Davids et al., 2003; Hamill et al., 2012). When the number of available biomechanical degrees of freedom is constrained (e.g. reduction in DF ROM) the number of available movement patterns to complete a given task is reduced forcing individuals to select one of a number of alternative strategies (Davids et al., 2003). Davids et al. (2003) note that the analysis of pooled data means in the presence of coordinative variability can lead to non-significant results when biomechanical changes are in fact occurring and may contribute to inconsistencies between studies as different groups of participants prefer different strategies. It is possible within each study that some participants did not compensate kinematically and experienced greater GRFs or LRs while others increased their IC PF angle or altered knee or hip kinematics to attenuate forces. The number of possible compensations could result in too few participants utilising each one for a mean difference across participants to be found or in participants adjusting a combination of parameters to a small degree again resulting in no significant mean change in any single parameter. Unfortunately the studies reviewed investigated only a few biomechanical variables making it difficult to identify biomechanical patterns and reasons for conflicting results.

Areas for Future Research

The high degree of coordinative variability in landing and compensation strategies highlights the need for studies which investigate a large number of biomechanical variables across multiple joints in an attempt to identify pattern changes. It also highlights the need to analyse biomechanical changes in individuals rather than by pooling data and analysing means. The difficulty with analysing biomechanics in the presence of coordinative variability highlights the need for a measure which can identify when a biomechanical pattern changes allowing for conventional statistical analysis of results.

Although no studies were found which directly investigated the effect of DF ROM on lower-extremity stiffness the studies described support the theory that lower-extremity stiffness may be altered in compensation for DF restriction. Furthermore, as stiffness captures a number of biomechanical variables into a single measure it may have some utility in describing changes in movement patterns allowing for traditional statistical analysis of a highly individual and variable task (Lorimer, 2014). Further research is needed to investigate the effect of DF ROM on landing stiffness and the potential for stiffness measures to identify changes in movement patterns.

None of the landing-tasks investigated in the reviewed studies were sports-specific, limiting their applicability to injuries incurred during sporting tasks. Biomechanical studies which include goal-directed, sport-specific tasks are needed.

Conclusion

Restricted DF ROM may alter landing biomechanics in a manner which predisposes athletes to injury. There is some support for increased frontal plane ankle motion, knee valgus and frontal plane excursion, reduced knee and hip sagittal excursion, and increased peak vGRF, but results are inconsistent between studies. DF restriction does not appear to reduce peak DF angle on landing or TTP vGRF. Further studies are needed to investigate the effect of DF restriction on biomechanical patterns rather than on individual biomechanical variables, and to investigate sport-specific landing tasks.

CHAPTER 3

THE EFFECT OF ANKLE BRACING ON LOWER EXTREMITY BIOMECHANICS DURING LANDING: A SYSTEMATIC REVIEW OF THE LITERATURE

Overview

Ankle braces are commonly worn to prevent ankle sprains and are generally designed to limit inversion but may also restrict dorsiflexion (DF) range of motion (ROM). Restricted DF may alter landing biomechanics in a manner which increases injury-risk. Athletes may compensate via increased pronation and knee valgus, or be forced to land stiffly resulting in increased ground reaction forces (GRFs) and loading rates (LR). A search of the literature identified ten studies investigating the effect of ankle bracing on landing biomechanics. The overall evidence is conflicting but there is some support that some ankle braces restrict peak DF angle, knee and angle sagittal plane excursion, and increase GRFs and LRs. Variation in participant compensation strategies, the biomechanical variables investigated, the type of landing task, and the sport-specific nature of tasks investigated create difficulty in interpreting results. These areas require further research

Introduction

Ankle braces are commonly worn during sport to support or prevent ankle injury (Papadopulos et al., 2005). Some common brace types may restrict ankle dorsiflexion (DF) range of motion (ROM) (Eils et al., 2007; Parsley et al., 2013) which in turn may alter lower-extremity biomechanics during landing tasks in a manner which predisposes the athlete to injury (Fong et al., 2011; Macrum et al., 2012). Previous research has found reduced DF ROM with lace-up braces (Eils et al., 2007; Parsley et al., 2013) and Aircast-stirrup braces (Eils et al., 2002; Eils et al., 2007). However these studies measured DF range goniometrically in non-functional positions which has been shown to underestimate true maximal range (Krause et al., 2011). Therefore, although bracing appears to reduce available range during passive testing this effect may be different during landing tasks where greater forces may be needed to overcome the resistance of the brace.

Restricted DF ROM has been linked to a number of acute and chronic lower-extremity injuries (Backman & Danielson, 2011; Hadzic et al., 2009; Kaufman et al., 1999; Kibler et al., 1991; Lun et al., 2003; Malliaras et al., 2006; Piva et al., 2005; Tabrizi et al., 2000; Wahlsteadt & Rasmussen-Barr, 2014). The biomechanical reasons for these links remain unclear but it has been theorised that DF restriction limits the ability to pass the leg forwards over the foot (Bolgla, 2004; Mauntel et al., 2013; Piva et al., 2005; Tweed et al., 2008) and to lower the centre of mass during squatting movements (Macrum et al., 2012). This may be compensated for with subtalar and midfoot pronation (Bolgla, 2004; Piva et al., 2005; Tweed et al., 2008) or knee valgus (Bell et al., 2012; Piva et al., 2005), increasing the risk of associated injuries such as Achilles tendon injury, knee ligament injury, and patellofemoral pain syndrome (Aminaka et al., 2011; Hewett et al., 2005; Lersch et al., 2012; Quatman et al., 2013; Wyndow et al., 2010). The restricted sagittal excursion may also reduce time to attenuate landing-forces leading to increased loading-rates (LR) and ground-reaction forces (GRFs) (Bisseling et al., 2007, 2008; Fong et al., 2011). Increased injury risk has been reported with both higher GRFs (Hewett et al., 1999; Hewett et al., 1996; Norcross et al., 2013) and higher LRs (Bisseling et al., 2007, 2008; Milner et al., 2006; Radin et al., 1991). Furthermore, reduced sagittal excursion may increase lower-extremity stiffness which is also associated with increased GRFs and LRs (Bisseling et al., 2007, 2008; Fong et al., 2011; Wang, 2009) and is speculated to increase injury-risk (Butler et al., 2003; Serpell et al., 2012).

The link between reduced DF ROM and injury and the potential for braces to restrict DF suggests that ankle bracing may result in biomechanical compensations which predispose athletes to injury. Thus the purpose of this review is to examine the evidence for the effect of ankle bracing on lower extremity biomechanics during landing tasks.

Methods

Search

A comprehensive search of the literature was then conducted on EBSCO Health Databases using the key words identified in Table 6. Reference lists were scanned to identify further articles.

Table 6: Literature Review Search Strategy

Search		Keywords
Relationship between ankle bracing and lower extremity kinematic and kinetic variables:		brac* AND (ankle OR talocrural)
	AND	mechanic* OR biomechanic* OR kinetic* OR kinematic* OR move* OR "ground-reaction force*" OR GRF* OR (force* AND (land* OR load)) OR stiff*
	AND	jump* OR land* OR hop*

Study selection

Studies were included in the review if they compared biomechanical variables in braced and unbraced conditions with at least one of the following outcome measures: GRF, LR, time-to-peak (TTP) GRF, stiffness, or lower-extremity kinematics during a landing task. Studies were excluded if they included injured participants, or compared between genders, different landing-tasks, or in varying states of fatigue. Articles were restricted to full text in the English language, no publication date restrictions were imposed.

Data extraction

Data was tabulated under the headings study design, outcome measures, landing tasks, results, and conclusions (see Table 10). The terms 'knee abduction', 'medial knee deviation', 'knee valgus', and 'knee frontal plane motion' were considered synonymous. Where the foot model used was not stated and markers were placed at the malleoli, calcaneus, and metatarsal heads, it was assumed that a single-segment foot model was used. Dependent variables which were not a focus of the review were excluded from the table (e.g. effect of ankle strapping). Where the foot model used was not stated and markers were placed at the malleoli, calcaneus, and metatarsal heads, it was assumed that a single-segment foot model was used. As not all studies reported confidence intervals (CIs), additional analysis was performed to calculate 90% CIs using an Excel spreadsheet (Hopkins, 2007a). For studies that did not report exact P-values the threshold value was used (e.g. $P \leq 0.05$) to calculate the CI. Soft braces which used laces to secure the brace to the ankle were classified as 'lace-up', those using air-cells to splint the ankle were classified as 'Aircast-stirrup' braces, and those constructed from rigid or semi-rigid plastic with a hinge on the horizontal axis were classified as 'rigid-stirrup' braces.

Assessment of Quality

Articles were assessed for quality by two reviewers (AMM and CW) using the modified Downs and Black checklist (Downs & Black, 1998) (see Table 7). Question 27 of the Downs and Black checklist was altered to score 1 for sufficient sample size based on power calculation and score 0 for insufficient sample size or power not calculated.

After each study was critiqued, a Quality Index was derived to categorise methodological score. The Quality Index enables studies to be categorised as being of *poor, limited, moderate*, or *strong* quality (see Table 7). This measure has been used by several systematic reviews which have rated methodology using the modified Downs and Black checklist (Hartling et al., 2004; Hignett, 2003; Hing et al., 2009)

Table 7: Categorisation of Quality Index Scores

Total modified Downs and Black checklist score (/28)	Percentage	Quality Index
21+	75% +	Strong
14-20	50-74%	Moderate
7-13	25-49%	Limited
<7	<25%	Poor

Adapted from Hartling et al. (2004), Hignett (2003), Hing et al. (2009)

Evidence Synthesis

Levels of evidence were determined as outlined by van Tulder et al. (2003) (see table 8). For the purposes of this review the descriptor "High" has been replaced by "Strong", and "Limited" has been further divided into "Limited" and "Poor" (see table 8). Evidence was considered to be 'consistent' when at least 75% of articles agreed (Reid et al., 2012)

Table 8: Levels of Evidence

Strong	Consistent findings among multiple strong quality RCTs
Moderate	Consistent findings among multiple moderate quality RCTs and/or one strong quality RCT
Poor	Consistent findings among multiple low quality RCTs and/or CCTs and/or one moderate quality RCT
Limited	One low quality RCT and/or CCT
Conflicting	Inconsistent findings among multiple trials (RCTs and/or CCTs)

Adapted from van Tulder et al. (2003)

Results

Studies included in the review

The search for the effect of ankle bracing on landing biomechanics yielded 100 articles of which ten met the inclusion criteria (see figure 2).

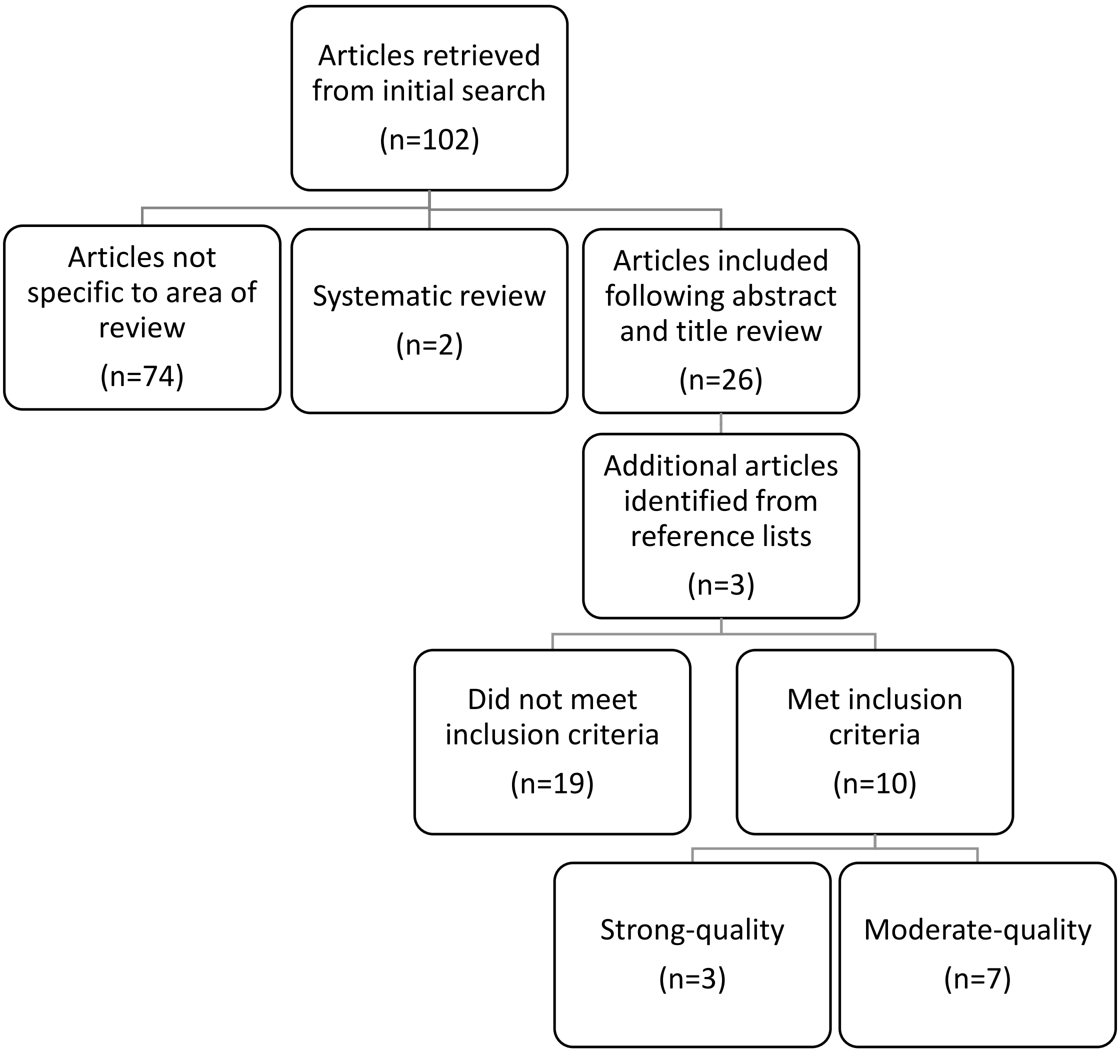

Figure 2: Search Results - Articles reporting on the effect of ankle bracing (≥21=strong-quality, 14-20=moderate quality, 7-13=limited-quality, <7=poor quality)

Quality

Scores ranged from 16 to 26/28 on the Downs and Black checklist with three articles classified as strong quality and seven as moderate quality (see table 9). The major quality issues were a lack of power calculations, not stating source populations, not stating the percentage of those approached who agreed to participate, and a lack of participant and researcher blinding.

Table 9: Downs and Black Scores for Articles on Effect of Ankle Bracing on Landing Biomechanics (≥21=strong-quality, 14-20=moderate-quality, 7-13=limited quality, <7=poor-quality

Study	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22	23	24	25	26	27	Total /28	Quality
Vanwanseele (2013)	1	1	1	1	2	1	1	0	1	1	1	1	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	1	1	1	22	Strong
Distephano et al (2008)	1	1	1	1	2	1	1	0	1	1	1	0	n/a	0	0	1	1	1	1	1	1	1	1	0	1	1	0	21	Strong
McCaw (1999)	1	1	1	1	2	1	1	0	1	1	1	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	1	1	1	21	Strong
Hodgson (2005)	1	1	1	1	2	1	1	0	1	1	1	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	1	1	0	20	Moderate
West (2014)	1	1	1	1	2	1	1	0	1	1	0	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	1	1	0	19	Moderate
Cordova (2010)	1	1	1	1	2	1	1	0	1	0	0	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	1	1	0	18	Moderate
Simpson et al (2013)	1	1	1	1	1	1	1	0	1	1	0	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	1	1	0	18	Moderate
Riemann (2002)	1	1	1	1	1	1	1	0	1	1	0	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	0	1	0	17	Moderate
Hopper (1999)	1	1	1	1	1	1	1	0	1	0	0	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	0	1	0	16	Moderate
Williams and Riemann (2009)	1	1	0	1	1	1	1	0	0	1	0	0	n/a	0	0	1	1	1	1	1	1	1	n/a	n/a	0	1	0	15	Moderate

Table 10: Effect of Ankle Bracing on Landing Biomechanics (articles organised in order of highest to lowest quality as per the Downs and Black checklist)

Article	**Participants**	**Brace**	**Study Characteristics**	**Measurement**		**Landing-Task**	**Results**		**Conclusions**
				Kinematics	Kinetics		Kinematics	Kinetics	
Vanwanseele et al (2013)	Gender, no.: F, 11 Age (y): 18.3±1.8 Sport: New South Wales Institute of sport Netball programme	Lace up brace (E-Professional)	Within-subject repeated measures design	14 camera 3D motion capture Two-segment foot (forefoot, rearfoot), measured in-shoe foot motion (Sagittal and frontal plane ankle kinematics)	Force plate (vGRF)	Forward jump from left leg to right leg while receiving a chest-pass. 5m straight-line approach to the jump at a self-selected speed, allowed one step forward with the left leg after landing	No significant difference in ankle peak angles or excursion in sagittal or frontal planes between conditions Reduced peak ankle EV and IV angles and frontal plane excursion with brace Mean difference (EV) = 2.37°, CI = 0.61-4.2 Mean difference (IV) = 1.58°, CI = 0.29-2.9 Mean difference (excursion) = 3.95°, CI = 1.4-6.5	No significant difference in peak vGRF between conditions	Reduced peak ankle EV and IV angles and frontal plane excursion with lace-up brace No significant difference in sagittal or frontal plane ankle motion with lace-up brace

DiStephano et al (2008)	Gender, no.: M, 22 F, 15 Intervention: Age(y) 19.63±0.72 Control: Age (y): 19.94±1.44 Sport: Recreational volleyball and basketball	Lace-up brace (ASO, Medical Specialities Inc.)	Within-between subject design Intervention group wore bilateral LUBs for 8 weeks	Electromagnetic 3D motion analysis system Single-segment foot, measured in-shoe foot motion (Sagittal plane ankle, knee kinematics)	Force plate (Peak vGRF, TTP vGRF)	Forward jump from 0.30m platform to a distance equal to half the participants' height, bilateral landing, dominant foot on force plate, immediately perform maximal vertical jump	No effect of wearing LUB for 8 weeks – significant results pooled for analysis Decreased PF angle at IC with brace Mean difference = 3°, CI = 0.92-5.1 Decreased peak DF angle with brace Mean difference = 1°, CI = 0.21-1.8 Decreased ankle sagittal excursion with brace Mean difference = 3°, CI = 1.6 – 4.4 Increased KF angle at IC with brace Mean difference = 3°, CI = 1.8-4.2 Decreased knee sagittal excursion with brace Mean difference = 3°, CI = 0.62-5.4 No significant difference in peak KF between conditions	No effect of wearing LUB for 8 weeks – significant results pooled for analysis No significant difference in peak vGRF or TTP vGRF between conditions	Reduced peak DF angle, IC PF angle, and ankle and knee sagittal excursion with lace-up brace Increased IC KF angle with lace-up brace No change in peak KF, peak vGRF,, or TTP vGRF with lace-up brace
McCaw (1999)	Gender, no.: F, 5 M, 9 Age (y): F 20±1 M 21±2 Sport: Basketball, volleyball	Lace-up brace (Swede-O, Inc.) Aircast stirrup brace (Aircast Sport Stirrup, Aircast, Inc)	Within-subject repeated-measures design	Single camera high-speed video recording Single-segment foot, measured shoe motion	-	Drop-landing from 0.59m platform, bilateral landing Two landing conditions: maximal knee flexion (soft landing); minimal knee flexion (stiff landing)	Reduced IC PF angle with LUB and ASB, no significant difference with RSB Mean difference (LUB) = 4.2°, CI = 0.76-7.6 Mean difference (ASB) = 4.1°, CI = 0.74-7.5 Reduced DF angle at peak knee flexion with LUB and ASB, no significant difference with RSB Mean difference (LUB = 2.2°, CI = 0.4-4 Mean difference (ASB) = 2.6°, CI = 0.47-4.7	-	Reduced IC PF angle, peak DF angle, and sagittal ankle excursion with lace-up and Aircast-stirrup braces No change in IC PF angle, peak DF angle, or sagittal ankle excursion with rigid-stirrup brace

		Rigid stirrup brace (Active Ankle Systems, Inc.)		(Sagittal plane ankle kinematics)			Reduced sagittal plane ankle excursion with LUB and ASB, no significant difference with RSB Mean difference (LUB) = 6.6°, CI = 1.2-12 Mean difference (ASB) = 6.8°, CI = 1.2-12		
Hodgson, Cobb, & Higbie (2005)	Gender, no.: F, 12 Age (y): 19.83±1.7 Sport: Division I college volleyball	Rigid-stirrup brace (Active ankle T2, Active Ankle Systems, Inc.)	Within-subject repeated-measures design	8-camera motion capture Single-segment foot, marker placement not stated (Sagittal plane ankle, knee, hip kinematics)	Force plate (P_1, P_2, TTP_1, TTP_2 vGRF, LR_1, LR_2)	Drop-jump, hanging from bar, feet 0.61m from floor, bilateral landing, right foot on force plate, immediately perform maximal vertical jump	Change in joint angle at P_1 (angle at P_1-angle at IC): Ankle - Reduced with brace (greater PF) Mean difference = 1.33°, CI = 0.52-2.1 No significant difference at hip or knee between conditions Change in joint angle at P_2 (angle at P_2-angle at inflection point between P_1 and P_2): No significant difference at hip, knee, or ankle between conditions Sagittal plane excursion (angle at P_2-angle at IC): No significant difference at hip, knee, or ankle between conditions	Greater P_1 with brace Mean difference = 0.19N/kg, CI = 0.096–0.28 Greater LR_1 with brace Mean difference = $1.72x10^4$N/s , CI = $1.02x10^4$-$2.40x10^4$ No significant difference in P_2, LR_2, TTP_1 or TTP_2 between conditions	Greater PF at IC with rigid-stirrup brace Greater P_1 and LR_1 with rigid-stirrup brace No change in IC hip or knee angles, peak angles or sagittal excursion at ankle, knee or hip, or P_2, LR_2, TTP_1 or TTP_2 with rigid-stirrup brace
West (2014)	Gender, no.: F, 15 Age (y): 22.7±3.30 Sport: Volleyball	Rigid-stirrup brace (Active ankle T2, Active Ankle Systems, Inc.)	Within-subject repeated-measures design	14-camera motion capture Single-segment foot, measure	Force plate (Peak vGRF, TTP vGRF)	Block-jump: Maximal effort jump while blocking a suspended volleyball with arms, bilateral landing Spike-jump:	No significant difference in sagittal plane knee excursion between conditions during any task	No significant difference in peak vGRF or TTP vGRF between conditions during any task	No change in sagittal plane knee excursion with rigid-stirrup brace No change in vGRF or TTP vGRF with rigid-stirrup brace

				d shoe motion (sagittal plane knee kinematics)		Maximal effort spike approach striking suspended volleyball, bilateral landing Block-jump push-off: Block-jump with lateral push-off immediately on landing Spike-and-cover: Spike-jump with lateral push-off immediately on landing			
Cordova et al (2010)	Gender, no.: M, 13 Age (y): 22±2 Sport: Recreational basketball	Rigid-stirrup brace (Ultra Ankle, McDavid Inc.)	Within-subject repeated measures design 5 trials each in brace and control	Single camera motion capture Single-segment foot, measured shoe motion	Force plate (Peak vGRF, TTP_1, TTP_2 vGRF)	Drop-landing from 0.30m platform, unilateral landing, keeping hands on iliac crests	Reduced sagittal ankle excursion with brace Mean difference = 8.9°, CI = 1.6-16 Reduced knee sagittal excursion with brace Mean difference = 2.5°, CI = 0.45-4.5 No significant difference in sagittal hip excursion between conditions	No significant difference in P_1 or P_2 between conditions TTP_1 and TTP_2 reduced with brace Reduced with brace Mean difference (TTP_1) = 3ms, CI = 0.55-5.5 Mean difference (TTP_2) = 4ms, CI = 0.73-7.3	Reduced knee and ankle sagittal excursion with rigid-stirrup brace Reduced TTP_1 and TTP_2 with rigid-stirrup brace No change in hip sagittal excursion, P_1

				(Sagittal plane ankle, knee, hip kinematics)					or P_2 with rigid-stirrup brace
Simpson et al (2013)	Gender, no.: F, 16 Age (y): 21.2±2.9 Sport: Basketball, soccer, volleyball	Lace-up brace (ASO, Medical Specialities, Inc.)	Within-subject repeated-measures design	7-camera 3D motion capture Single-segment foot, measured shoe motion (Sagittal, frontal, and transverse plane ankle, knee, hip kinematics)	Force plate (vGRF, TTP vGRF, LR)	Drop-landing, hanging from bar, feet 0.43m from floor, bilateral landing, right foot on force plate, keeping arms extended overhead	Ankle – Reduced IC PF angle with brace Mean difference = 8°, CI = 4.6-11 Reduced peak DF angle with brace Mean difference = 2°, CI = 0.81-3.2 Reduced sagittal excursion with brace Mean difference = 10°, CI = 5.7-14 No significant difference in IC or peak angles or excursion in frontal plane between conditions No significant difference in IC or peak angles in transverse plane between conditions Reduced transverse excursion with brace Mean difference = 2.5°, CI = 1.3-3.7 Knee – Greater IC KF angle with brace Mean difference = 3°, CI = 1.2-4.8 No significant difference in peak KF angle between conditions Reduced sagittal excursion with brace Mean difference = 3°, CI = 0.96-5	Greater peak vGRF with brace Mean difference = 3N/kg, CI = 1.7-4.3 Reduced TTP vGRF with brace Mean difference = 6.4ms, CI = 1.9-10 Greater LR with brace Mean difference = 39.1N/kg/s, CI = 22-56	Reduced IC PF and peak DF angles, ankle and knee transverse plane excursion, and ankle and knee sagittal excursion with lace-up brace Greater IC KF angle with lace-up brace No change in ankle frontal motion, IC or peak transverse plane ankle angles, peak KF, peak transverse plane knee angles, frontal plane knee motion, or sagittal frontal, and transverse plane hip motion between conditions Greater vGRF and LR, and reduced TTP vGRF with lace-up brace

							No significant difference in IC or peak angles or excursion in frontal plane between conditions No significant difference in IC or peak angles in transverse plane between conditions Reduced transverse excursion with brace Mean difference = 1.2°, CI = 0.5-1.9 Hip – No significant difference in IC or peak angles, or excursion in sagittal, frontal, or transverse planes between conditions		
Riemann et al (2002)	Gender, no.: M, 9 F, 5 Age (y): 17-26 Sport: Not stated	Aircast stirrup brace (AirSport, Aircast, Inc.)	Within-subject repeated measures design 6 stiff and 6 soft landings each in brace and control conditions performed prior to and following a 20min	-	Force plate (P_1, P_2, TTP_1, TTP_2 vGRF)	Drop-landing from 0.59m platform, bilateral landing, dominant foot on force plate	-	Pre-exercise: Reduced TTP_1 and TTP_2 with brace during stiff and soft landings Stiff landings - Mean difference (TTP_1) = 2.2ms, CI = 0.4-4.0 Mean difference (TTP_2) = 4.4ms, CI = 0.79-8.0 Soft-landings - Mean difference (TTP_1) = 1.5ms, CI =0.27-2.7 Mean difference (TTP_2) = 3.1ms, CI = 0.56-5.6 No significant difference in P_1 or P_2 between conditions during stiff or soft landings Post-exercise: Reduced TTP_1 and TTP_2 with brace during stiff and soft landings Stiff landings -	Reduced TTP_1 and TTP_2 with Aircast-stirrup brace both pre and post-exercise and during stiff and soft landing-styles No difference in P_1 or P_2 with Aircast-stirrup brace

			treadmill jog					Mean difference (TTP_1) = 1.8ms, CI = 0.32-3.3 Mean difference (TTP_2) = 3.4ms, CI = 0.61-6.2 Soft-landings - Mean difference (TTP_1) = 1.9ms, CI = 0.34-3.5 Mean difference (TTP_2) = 3.7ms, CI = 0.67-7.4 No significant difference in P_1 or P_2 between conditions during stiff or soft landings	
Hopper et al (1999)	Gender, no.: F, 15 Age (y): 22.6±4.2 Sport: Netball	Lace-up brace (Swede-O, Inc.)	Within-subject repeated measures design	Single camera motion capture Single-segment foot, no shoes worn (Frontal plane ankle position at IC)	Force plate (Peak vGRF, TTP vGRF)	Unilateral (dominant limb) forward jump to a distance equal to 1.25x leg-length	No significant difference in IC frontal plane rearfoot angle between conditions	No significant difference in peak vGRF or TTP vGRF between conditions	No change in IC rearfoot angle, peak vGRF, or TTP vGRF between conditions
Williams and Riemann (2009)	Gender, no.: M, 5 F, 14	Aircast stirrup brace (AirSport, AirCast Inc.)	Within-subject repeated measures design	-	Force plate (vGRF) Vertical stiffnes	30s bilateral hopping on a force plate at self-selected speed and at 3.0Hz.	-	No significant difference in peak vGRF or vertical stiffness between conditions	No significant difference in peak vGRF or vertical stiffness with an Aircast-stirrup brace

	Age (y): 'college age' Sport: Recreationally active	10 min run and cutting/ladder drills while wearing the brace prior to testing	s (calculated from vGRF and centre of mass displacement)

IC = Initial Contact; HC = Heel contact; DF = Dorsiflexion; PF = Plantarflexion; IV = Inversion; EV = Eversion; KF = Knee Flexion; ROM = Range of motion; pROM = Passive range of motion; aROM = Active range of motion; vGRF = Vertical ground reaction force; P_1 = First peak vGRF; P_2 = Second peak vGRF; TTP = time-to-peak vGRF; TTP_1 = time-to-P_1; TTP_2 = time-to-P_2; LR = Loading rate; LR_1 = LR at P_1; LR_2 = Loading rate at P_2; RSB = Rigid-stirrup brace; LUB = Lace-up brace; ASB = Aircast-stirrup brace

Evidence classification

Level of evidence determination was based on the quality, consistency, and number of available research articles (van Tulder et al., 2003). Overall there is strong evidence that lace-up and Aircast-stirrup braces alter landing biomechanics and conflicting evidence regarding rigid-stirrup braces. There is strong evidence for reduced peak DF angle with lace-up braces, moderate evidence for a reduction with Aircast-stirrup braces and moderate evidence for no reduction with rigid-stirrup braces. There is strong evidence for other alterations in ankle kinematics with lace-up braces, moderate evidence with Aircast-stirrup braces, and conflicting evidence regarding rigid-stirrup braces. There is moderate evidence for altered knee kinematics with lace-up braces, and conflicting evidence regarding rigid-stirrup braces. There is poor evidence that lace-up braces do not affect hip kinematics and moderate evidence that rigid-stirrup braces do not affect hip biomechanics. There is conflicting evidence regarding the effect of lace-up and rigid-stirrup braces on vGRF and TTP vGRF. There is moderate evidence for no effect of Aircast-stirrup braces on vGRF and poor evidence for reduced TTP vGRF and increased LR. There is conflicting evidence regarding the effect of rigid-stirrup braces on LR. There is poor evidence for no effect of Aircast-stirrup braces on vertical stiffness.

Participants

There were a total of 166 participants across all studies (108 female, 58 male) with an age range of 17 to 22.7years. Participants were involved in a number of sports including basketball, volleyball, netball, and soccer, with one study describing participants as 'recreationally active' (Williams & Riemann, 2009) and one not stating activity levels or sports (Riemann, Schmitz, Gale, & McCaw, 2002).

Outcome measures

Studies investigating kinematics used 3D motion capture systems, single-plane motion capture with one camera, or an electromagnetic 3D motion analysis system. Kinematic variables measured included ankle, knee and hip sagittal, frontal, and transverse motion and frontal plane ankle position at initial contact. Six studies (Cordova, Takahashi, Kress, Bruckner, & Finch, 2010; DiStephano, Padua, Brown, & Guskiewicz, 2008; Hodgson, Tis, Cobb, & Higbie, 2005; Hopper, McNair, & Elliott, 1999; McCaw, Stephen, & Cerullo, 1999; Simpson et al., 2013) used single-segment foot models while Vanwanseele, Stuelcken, Greene, and Smith (2013) used a two-segment model. Four studies (Cordova et al., 2010; McCaw et al., 1999; Simpson et al., 2013; West, Ng, & Campbell, 2014) measured shoe motion, and two (DiStephano et al., 2008; Vanwanseele et al., 2013) measured in-shoe foot motion, while Hopper et al. (1999) assessed participants barefoot. In one study (Hodgson et al., 2005) it was not possible to determine whether foot or shoe motion was measured. Kinetic variables were captured via force plates and included peak vertical GRF (vGRF), first and second peak vGRF (P_1, P_2), time-to-peak (TTP) vGRF, time-to first peak and second vGRF peaks (TTP_1, TTP_2), loading-rate (LR), and LR at P_1 and P_2 (LR_1, LR_2). Williams and Riemann (2009) calculated vertical stiffness using vGRF and COM displacement data.

Landing tasks

All studies used a within-subject design with all participants completing the same landing-tasks. DiStephano et al. (2008) included a between-subject component with an intervention group wearing ankle braces for eight-weeks prior to testing but no differences were found between groups during testing and results were pooled for statistical analysis. Cordova et al. (2010), McCaw et al. (1999), and Riemann et al. (2002) investigated drop-landings from heights of 0.30m, 0.59m, and 0.59m respectively. Cordova et al. (2010) investigated a unilateral landing while the other two studies investigated bilateral landings with McCaw et al. (1999) investigating both stiff and soft landing-styles. Hodgson et al. (2005) investigated a bilateral drop-jump from 0.61m. DiStephano et al. (2008) and Hopper et al. (1999) investigated forward jumps with DiStephano et al. (2008) investigating a bilateral jump from a 0.30m platform to a distance equal to half participants' height with a maximal vertical jump performed immediately after landing. Hopper et al. (1999) investigated a unilateral jump to a distance of 1.25x participants' leg-length. Vanwanseele et al. (2013) investigated a netball-specific forward jump from one foot to the other while receiving a chest pass, and one investigated 30s bilateral hopping both at a self-selected speed and at a standardised speed of 3Hz. West et al. (2014) investigated four volleyball specific tasks with bilateral landings including a block-jump, spike-jump, and block and spike-jumps with lateral push-off immediately on landing.

Key findings - Kinematics

Of the five studies investigating peak DF angle during a landing task three (DiStephano et al., 2008; McCaw et al., 1999; Simpson et al., 2013) found reduced peak DF with a lace-up brace with McCaw et al. (1999) also reporting reduced peak angle with an Aircast-stirrup brace. Conversely, Vanwanseele et al. (2013) found no change in DF angle with a lace-up brace, and Hodgson et al. (2005) and McCaw et al. (1999) found no change with rigid-stirrup braces.

Of the five studies investigating plantarflexion (PF) angle at initial contact (IC) three (DiStephano et al., 2008; McCaw et al., 1999; Simpson et al., 2013) found reduced IC angle with lace-up braces, with McCaw et al. (1999) also reporting reduced angle with an Aircast-stirrup brace. Conversely, Vanwanseele et al. (2013) found no effect of a lace-up brace on IC PF angle, McCaw et al. (1999) found no effect of a rigid-stirrup brace on IC angle and Hodgson et al. (2005) found IC PF angle was greater with a rigid-stirrup brace.

Of the six studies which measured total ankle sagittal excursion three (DiStephano et al., 2008; McCaw et al., 1999; Simpson et al., 2013) found reduced excursion in lace-up braces, one with an Aircast-stirrup brace (McCaw et al., 1999) and one with a rigid-stirrup brace (Cordova et al., 2010). Conversely, Vanwanseele et al. (2013) and Hodgson et al. (2005) found no change in ankle sagittal excursion with lace-up and rigid-stirrup braces respectively.

Of the three studies which measured frontal plane ankle motion, Vanwanseele et al. (2013) found reduced peak eversion and inversion angles and frontal excursion while Hopper et al. (1999) and Simpson et al. (2013) found IC angle was unaffected and Simpson et al. (2013) found no change in peak angle or excursion with lace-up braces. Simpson et al. (2013) also found reduced transverse plane excursion but no change in IC or peak angles in the transverse plane.

Of the five studies which investigated sagittal plane knee kinematics two found reduced excursion with lace-up braces (DiStephano et al., 2008; Simpson et al., 2013) and one with a rigid-stirrup brace (Cordova et al., 2010). Two studies found greater IC knee flexion angle with lace-up braces (DiStephano et al., 2008; Simpson et al., 2013). In contrast, two studies Hodgson et al. (2005) and West et al. (2014)found no difference in sagittal excursion with a rigid-stirrup brace with Hodgson et al. (2005) also reporting no difference in IC angle. All three studies measuring peak knee flexion angle found no significant difference with lace-up, rigid-stirrup, and lace-up braces respectively (DiStephano et al., 2008; Hodgson et al., 2005; Simpson et al., 2013). Simpson et al. (2013) also found reduced transverse excursion with a lace-up brace, but no change in peak transverse plane knee angles or frontal plane peak angles or excursion.

Three studies investigated hip kinematics and found no effect of lace-up (Simpson et al., 2013) or rigid-stirrup braces (Cordova et al., 2010; Hodgson et al., 2005).

Key findings - Kinetics

Of the nine studies investigating vertical ground reaction force (vGRF), only Simpson et al. (2013) found increased peak vGRF with a lace-up brace and one (Hodgson et al., 2005) found increased first peak vGRF (P_1) with a rigid-stirrup brace but no change in P_2. Conversely, several studies found no change in peak vGRF with lace-up braces (Hopper et al., 1999; Vanwanseele et al., 2013), Aircast-stirrup braces (Riemann et al., 2002; Williams & Riemann, 2009), or rigid-stirrup braces (Cordova et al., 2010; West et al., 2014).

Of the seven studies investigating time-to-peak (TTP) vGRF or loading-rate (LR) three found reduced TTP vGRF with rigid-stirrup (Cordova et al., 2010), Aircast-stirrup (Riemann et al., 2002), and lace-up braces (Simpson et al., 2013), one Simpson et al. (2013) found increased LR with a lace-up brace, and Hodgson et al. (2005) found increased LR at P_1 (LR_1) with a rigid-stirrup brace but no change in LR_2. Conversely, two studies (DiStephano et al., 2008; Hopper et al., 1999) found no change in TTP vGRF with lace-up braces and two (Hodgson et al., 2005; West et al., 2014) found no change with rigid-stirrup braces. The only study investigating stiffness (Williams & Riemann, 2009) found no effect of an Aircast-stirrup brace on vertical stiffness.

Discussion

Although the majority of studies found biomechanical changes in association with ankle bracing and a reduction in sagittal ankle range, results for each measured variable were inconsistent between studies. This inconsistency may be due in part to the different braces investigated, differing marker placements and food models, variations in landing tasks investigated in each study, variability in landing strategy within and between participants, and the focus of studies on individual biomechanical variables rather than on overall movement patterns (Hamill et al., 2012).

Ankle Kinematics - Dorsiflexion

Lace-up and Aircast-stirrup braces were found to reduce peak DF angle on landing (DiStephano et al., 2008; McCaw et al., 1999; Simpson et al., 2013) while rigid-stirrup braces did not (Hodgson et al., 2005; McCaw et al., 1999). This is consistent with brace design as rigid-stirrup braces leave the anterior and posterior aspects of the ankle open to allow sagittal plane movement while lace-up braces cover the anterior and posterior aspects of the ankle and Aircast-stirrup braces grip the ankle with air cells restricting sagittal movement (McCaw et al., 1999). The only exception was Vanwanseele et al. (2013) who found no reduction in peak DF with a lace-up brace. This was the only study to use a multi-segment foot model combined with measurement of in-shoe foot motion rather than shoe motion. The results of this study may therefore be more valid than the other studies mentioned (Deschamps et al., 2011; Sinclair, Taylor, Hebron, & Chockalingam, 2014). Alternatively, the conflicting results are likely explained by the landing task which was a netball-specific jump from one foot to the other allowing one further step forward after landing. Allowing a step forward after landing reduces the need to attenuate forces suddenly and therefore may also reduce the need to utilise maximal DF range in both braced and unbraced conditions. Participants may also have elected to approach the jump at a slow speed in order to reduce landing-forces in compensation for a restriction imposed by the brace.

The lack of reduction in peak DF angle with rigid-stirrup braces may be related to the landing tasks investigated. It has been reported that performing a jump immediately after landing causes participants to adopt a stiffer landing style with reduced joint excursion (Arampatzis, Schade, et al., 2001). Hodgson et al. (2005) investigated a drop-jump and instructed participants to jump as quickly as possible which may have resulted in a stiff landing style and utilisation of only a portion of available DF ROM regardless of brace condition. McCaw et al. (1999) instructed participants to use both minimal knee flexion (stiff landing) and maximal knee flexion (soft landing) and found greater peak DF angle during soft landings. However kinematic results were averaged across both techniques, potentially masking the effect of the brace during soft landings. Overall, although study results and brace design suggest that rigid-stirrup braces do not affect DF ROM it remains possible that peak DF angle on landing may be reduced by these braces.

Ankle Kinematics - Plantarflexion

Increased PF angle at IC is a possible compensation strategy which can maximise sagittal ankle excursion when DF range is restricted (Fong et al., 2011) while reduced IC PF angle is associated with knee injury and with greater vGRFs due to reduced time to attenuate forces (Ali, Rouhi, & Robertson, 2013). Initial contact PF angle was found to be restricted by lace-up braces (DiStephano et al., 2008; McCaw et al., 1999; Simpson et al., 2013) and Aircast-stirrup braces (McCaw et al., 1999) but not rigid-stirrup braces(McCaw et al., 1999). The reported reduction in peak DF suggests that lace-up and Aircast-stirrup braces both reduced DF angle and prevented the utilisation of one possible compensation strategy. As discussed previously, the contrasting lace-up brace results found by Vanwanseele et al. (2013) may be due to differences in landing-tasks. Interestingly, Hodgson et al. (2005) found IC PF angle was greater with a rigid-stirrup brace which the authors speculated was due to participants having a greater feeling of stability in the brace, making them confident enough to land in a more plantarflexed position.

Ankle Kinematics – Sagittal excursion

Reduced peak DF and IC PF angles may limit sagittal ankle excursion, potentially increasing injury-risk as a result of greater landing-stiffness(Bisseling et al., 2007; Fong et al., 2011), GRFs, and LRs(Bisseling et al., 2007, 2008; Fong et al., 2011). Reduced sagittal excursion was found with lace-up (DiStephano et al., 2008; McCaw et al., 1999; Simpson et al., 2013) and Aircast-stirrup braces (McCaw et al., 1999). Conflicting results for rigid-stirrup braces may again be related to the landing tasks investigated. The two studies reporting no difference in excursion (Hodgson et al., 2005; McCaw et al., 1999) investigated bilateral landing tasks while the unilateral landing investigated by Cordova et al. (2010) may have required greater joint-range for force attenuation making any restriction imposed by the brace more likely to alter landing kinematics. Furthermore, Hodgson et al. (2005) included a jump immediately on landing, potentially increasing lower-extremity stiffness and masking the effect of the brace as previously described(Arampatzis, Schade, et al., 2001).

Ankle Kinematics – Frontal and Transverse Planes

Pronation has been suggested as a compensatory mechanism for restricted DF ROM (Bolgla, 2004; Piva et al., 2005; Tweed et al., 2008). Vanwanseele et al. (2013) found reduced peak eversion (EV), inversion (IV) and frontal plane excursion, consistent with studies showing reduced eversion and inversion ROM in lace-up braces (Eils et al., 2007; Metcalf, Gretchen, Looney, & Renehan, 1997; Parsley et al., 2013). This restriction may impact the ability of the foot and ankle to utilise pronation as a compensatory mechanism for DF restriction and therefore has the potential to cause kinematic compensations at the knee and hip. The two studies reporting no change in frontal plane ankle kinematics (Hopper et al., 1999; Simpson et al., 2013) investigated highly controlled laboratory tasks rather than a goal directed sport-specific task, potentially explaining the differing results.(Cruz et al., 2013). As Vanwanseele et al. (2013) investigated a sport-specific jump and used a potentially more valid method of foot modelling and motion capture the results of this study may be more indicative of actual sport specific tasks.

Knee Kinematics

It has been suggested that low DF excursion restricts knee flexion during functional tasks resulting in greater GRFs(Fong et al., 2011) and compensatory frontal plane knee movement(Macrum et al., 2012). However two studies(DiStephano et al., 2008; Simpson et al., 2013) found no reduction in peak knee flexion angle with lace-up braces despite reduced peak DF angle. Reduced sagittal knee excursion found in these studies was instead a result of greater IC angle which may have occurred in response to reduced IC PF angle due to reflexive coordinative patterning between the two joints(Yeow, Lee, & Goh, 2011) or in an attempt to attenuate landing forces more effectively in compensation for the reduced ankle range (Begalle et al., 2015). Conflicting results for the rigid-stirrup brace may again be due to differences in landing-tasks

Increased knee frontal plane excursion has been implicated in a number of lower-extremity injuries (Nakagawa et al., 2012; Quatman et al., 2013) and is suggested to occur in compensation for restricted knee and ankle sagittal excursion (Bell et al., 2008; Bell et al., 2012; Macrum et al., 2012; Mauntel et al., 2013). Simpson et al. (2013) found reduced transverse plane knee excursion with a lace-up brace but no change in peak or IC angles in the transverse plane or in any frontal plane measurements. This study also found greater peak GRFs and LRs in the braced condition which may be a result of the observed lack of kinematic compensation for the reduction in ankle and knee sagittal excursion. Furthermore, these authors noted landing-strategy variation among participants with some adopting knee abduction and others adduction in the brace condition. While no significant difference was found between means for this variable, on an individual level there may have been an important effect of the brace.

Hip Kinematics

DF restriction has been suggested to reduce hip flexion excursion on landing (Fong et al., 2011; Wang, 2009) which may then be compensated for via increased hip internal rotation (Piva et al., 2005). However despite finding reduced peak DF angle and sagittal plane knee excursion, Simpson et al. (2013) found no changes in sagittal, frontal or transverse plane hip kinematics with a lace-up brace. Additionally, Cordova et al. (2010) and Hodgson et al. (2005) found no changes in sagittal plane hip kinematics with rigid-stirrup braces.

Kinetics

If kinematic compensations are not made, reduced DF excursion may increase GRFs or LRs as athletes have reduced time over which to attenuate landing-forces (Bisseling et al., 2007, 2008; Fong et al., 2011). This theory is supported by studies showing greater peak vGRF and LR (Simpson et al., 2013) or reduced TTP (Cordova et al., 2010; Riemann et al., 2002) with ankle bracing and reports of related decreases in knee and ankle sagittal excursion (Cordova et al., 2010; Simpson et al., 2013). In contrast, five studies found no kinetic changes in brace conditions. Two of these studies (Vanwanseele et al., 2013; Williams & Riemann, 2009) measured only vGRF magnitude and it is therefore possible that TTP or LR may have been affected by the braces. Furthermore, Vanwanseele et al. (2013) found that peak DF angle was not affected by the brace and accordingly no kinetic changes would be expected, and Williams and Riemann (2009) did not measure DF range and it is therefore possible that DF was not sufficiently restricted to cause kinetic changes. The continuous hopping task investigated by Williams and Riemann (2009) may have motivated participants to land stiffly regardless of brace condition in order to prepare for the next hop (Arampatzis, Schade, et al., 2001) resulting in no difference between conditions. Furthermore, none of these studies measured frontal or transverse hip and knee kinematics and it is therefore unknown whether compensations were made to prevent kinetic changes. Although Hodgson et al. (2005) found an increase in first peak vGRF (P_1) and LR at P_1 (LR_1) they found no change in P_2, or LR_2. The lack of a significant difference between conditions at P_2 is unsurprising as this study investigated a rigid-stirrup brace which was found not to restrict ankle ROM. The increase in P_1 and LR_1 may be due to other factors such as the observed reduction in IC PF angle or an effect of the brace on proprioception.

Stiffness

Williams and Riemann (2009) found no change in vertical stiffness with an Aircast-stirrup brace. This is inconsistent with studies which found reduced TTP vGRF, particularly those reporting associated reductions in sagittal excursion(Cordova et al., 2010; Simpson et al., 2013) suggesting increased stiffness(Fong et al., 2011). As mentioned previously the parameters of the study by Williams and Riemann (2009) may have resulted in either the brace not sufficiently restricting DF range or in full DF range not being utilised in either condition. Furthermore, this study investigated vertical stiffness rather than joint stiffness and may therefore have missed changes occurring at specific components of the kinetic chain. Finally, as kinematics were not investigated it is possible that compensations were made which prevented a change in vertical stiffness.

Coordinative variability

Individual landing and compensation strategies and interactions between biomechanical variables create difficulty in interpreting results. The ability to achieve a consistent endpoint with a variety of movement patterns is referred to as coordinative variability and the particular strategy used can vary widely both within and between individuals(Davids et al., 2003; Hamill et al., 2012). When the number of biomechanical degrees of freedom available to achieve a given movement goal is reduced (e.g. reduction in DF ROM due to bracing) individuals will select one of a number of alternative strategies which may be associated with various injuries (Davids et al., 2003). Davids et al. (2003)note that analysing the mean values from pooled data in the presence of coordinative variability can lead to non-significant results when biomechanical changes are in fact occurring. This may contribute to the inconsistencies seen between studies as different participant groups select different movement strategies. In the studies reviewed some participants may not have made kinematic compensations for range restriction and experienced greater GRFs, while others altered knee or hip kinematics to attenuate forces. The number of possible compensations may have resulted in too few participants utilising each one for a mean difference across participants to be found or, in participants adjusting a combination of parameters to a small degree again resulting in no significant mean change in any single parameter.

Landing-tasks

Differences in landing tasks between studies also make comparisons difficult and may contribute to inconsistent results. The biomechanical demands of a landing task change with varying height, distance, goals (e.g. catching a ball), and landing style (unilateral, bilateral etc.), and result in participants utilising different landing strategies (Cruz et al., 2013; Mothersole et al., 2013). Caution must therefore be taken when comparing studies investigating different landing tasks. The tasks investigated were also largely not sport-specific, limiting their applicability to injuries incurred during sports.

Areas for Future Research

The high degree of coordinative variability in landing and compensation strategies highlights the need for studies which investigate a large number of biomechanical variables across multiple joints in an attempt to identify pattern changes. It also highlights the need to analyse biomechanical changes in individuals rather than by pooling data and analysing means and the need for a measure which can identify changes in biomechanical patterns allowing for conventional statistical analysis of results. As stiffness measures capture a number of biomechanical variables into a single measure they may have some utility in describing changes in movement patterns allowing for traditional statistical analysis of a highly individual and variable task (Lorimer, 2014). Further research is needed to investigate this possibility and to investigate the effect of ankle bracing on leg and joint stiffness. Finally, studies are needed which investigate the effect of ankle bracing on landing biomechanics during sports-specific tasks.

Conclusion

There is evidence that ankle bracing can affect lower-extremity biomechanics during landing-tasks but the strength of evidence depends on the biomechanical variable of interest, the ankle brace in question and the landing task investigated. Lace-up and Aircast-stirrup braces have been shown to restrict peak DF angle and IC PF angle on landing while rigid-stirrup braces appear not to. Lace-up braces may restrict ankle range in the frontal plane, but the evidence is inconsistent and influenced by landing-task parameters. There is moderate evidence restricted ankle motion imposed by lace-up braces limits knee sagittal excursion but the influence of rigid-stirrup braces on knee motion is unclear. Additionally poor evidence suggests sagittal plane hip kinematics are unaffected by lace-up and rigid-stirrup braces and that frontal and transverse plane hip kinematics are unaffected by lace-up braces. Evidence regarding changes in GRFs, LRs, and TTP vGRF with ankle bracing is conflicted, while one study reported no effect on vertical stiffness. Further research is required in order to clarify the potential for ankle braces to affect lower extremity landing biomechanics.

CHAPTER 4

THE ASSOCIATION BETWEEN DORSIFLEXION RANGE OF MOTION AND LANDING BIOMECHANICS IN YOUNG NETBALLERS

Overview

Objective: To investigate the association between dorsiflexion range of motion on landing biomechanics.

Design: Cross-section observational

Methods: Twenty female high school netball players participated in this study (mean ±SD, age = 15.9 ±1.2 y; mass = 65.5 ±6.5 kg; height = 171.5 ±5.0 cm). Participants' ankles were separated into high and low-DF groups with ankles recording ≥130mm on the standing lunge assigned to the 'high-DF' group and an equal number of the lowest-scoring ankles assigned to the 'low-DF' group. Participants completed a drop jump, drop land, and a netball-specific task. Variables investigated included leg, knee and ankle stiffness, knee/ankle stiffness ratio, knee and ankle sagittal excursion, peak vertical ground reaction force (vGRF), time-to-peak vertical ground reaction force (TTP), and loading rate (LR). Mean differences between high and low DF groups were analysed using a spreadsheet for deriving effect sizes and magnitude based inferences.

Results: Ankle stiffness was higher in the low DF group on the left during the drop land (ES=0.84) and in the lead limb during the netball jump (ES=0.87). Ankle excursion was lower on the left during the drop jump (ES=-0.55) and in the trailing limb during the netball jump (ES=-0.97). Knee excursion was greater on the left during the drop jump (ES=1.91) and in the trailing limb during the netball jump (ES=1.85). All other results were unclear.

Conclusion: There is some evidence young netballers with low DF ROM exhibit greater ankle stiffness, less ankle sagittal excursion and more sagittal knee excursion during landing. Although unclear the observed effects for a number of other outcome variables suggest further research is warranted.

Introduction

Netball is one of the most popular women's sports in New Zealand (MyNetball). As the number of participants increases the number of injuries has also risen with a reported 120% increase in lower-extremity injury claims from 2008/9 to 2012/13 and an increased cost of almost 10 million dollars (ACC). The most commonly occurring netball injuries are knee and ankle sprains, calf strains, and Achilles tendon injuries (Langeveld et al., 2012; Otago & Peake, 2006).

Low dorsiflexion (DF) range of motion (ROM) has been associated with a number of lower-extremity injuries in landing and running sports including ACL injury (Wahlsteadt & Rasmussen-Barr, 2014), Achilles injury (Kaufman et al., 1999), and patellar tendon injury (Backman & Danielson, 2011; Malliaras et al., 2006). As a result, DF range of motion is commonly screened in athletes as one component of an injury-risk profile (Canavan, Roncarati, Lyles, & Kenney, 2012). The high incidence of ankle injury in netball may contribute to this increased risk as reduced DF ROM has been reported following ankle sprain (Aiken et al., 2008), ankle ligament reconstruction (Baumhauer & O'Brien, 2002), chronic ankle instability (Eils et al., 2007; Parsley et al., 2013), and with ankle bracing (Eils et al., 2007; Parsley et al., 2013).

The mechanisms underlying the association between reduced DF ROM and lower-extremity injury are as yet unclear. It has been proposed that low DF ROM may limit knee sagittal excursion on landing (Fong et al., 2011; Wang, 2009) resulting in a stiff landing style and greater peak vertical ground reaction force (vGRF) and loading rate (LR) (Bisseling et al., 2007, 2008; Fong et al., 2011). As stiff landing styles and the subsequent increase in vGRF and LR have been reported to increase injury risk (Milner et al., 2006; Mothersole et al., 2013; Norcross et al., 2013) the purpose of this study was to investigate the association between low DF ROM and landing stiffness, joint excursions, and forces. We hypothesised that low DF ROM would be associated with greater leg and joint stiffness, lower knee/ankle stiffness ratio, lower ankle and knee sagittal excursion, greater peak vGRF and LR, and lower time-to-peak (TTP) vGRF.

Methods

Participants

Twenty-six female high school netball players from a local secondary school were recruited. Participants were excluded if they had any injury or illness which had the potential to impact their ability to perform the landing tasks. Participants were considered uninjured if they were fully participating in trainings and games at the time of testing. Informed consent was obtained from all participants and legal guardians where appropriate prior to testing. Ethical approval was obtained from AUTEC (reference number 14/167). Sample size calculations were made using an Excel spreadsheet for estimating sample size (Hopkins, 2006a). Based on a standardised smallest important difference (Cohen's d) of 0.2 and the standard error of measurement for leg stiffness during running reported by Lorimer (2013) it was estimated that 13 participants were required to achieve 80% power.

Study Design

A between-subject design was used comparing high and low DF ROM groups. As both limbs were analysed separately grouping was assigned to individual limbs rather than to participants. All participants attended the motion analysis laboratory on one occasion and completed all tasks on the same day.

Instrumentation

Participants' height, weight, and bilateral trochanteric height were measured by a researcher trained in International Society for the Advancement of Kinanthropometry (ISAK) protocols (Stewart, Marfell-Jones, Olds, & de Ridder, 2011). Retroflective markers (10 mm) were then placed bilaterally at anatomical landmarks using a modification of three dimensional (3D) models described by, Besier, Sturnieks, Alderson, and Lloyd (2003), Tulchin, Orendurff, and Lori (2010), and Ferber, McClay Davis, and Williams (2003) (see Figure 3). Clusters of four retroreflective tracking markers on thermo-moulded plastic shells were placed at the sacrum, mid-thigh, and mid-shank. Individual tracking markers were placed bilaterally at ASISs, proximal and distal mid-line posterior calcaneus, medial and lateral anterior calcaneus, 1st and 5th metatarsal heads, and centre line of the forefoot between 2nd and 3rd metatarsal heads.

Anatomical markers were placed bilaterally at greater trochanters, medial and lateral femoral condyles, and medial and lateral malleoli. Foot markers were placed over the shoes with landmarks palpated through the shoes. The same physiotherapist palpated for anatomical landmarks and attached markers for all participants. Following a static standing calibration the femoral condyle and malleoli markers were removed.

Participants wore tight-fitting shorts and t-shirts or singlets which allowed easy palpation of bony landmarks and secure application of markers. All participants wore standard sports shoes of the same brand and style (Asics Gel-Kurow).

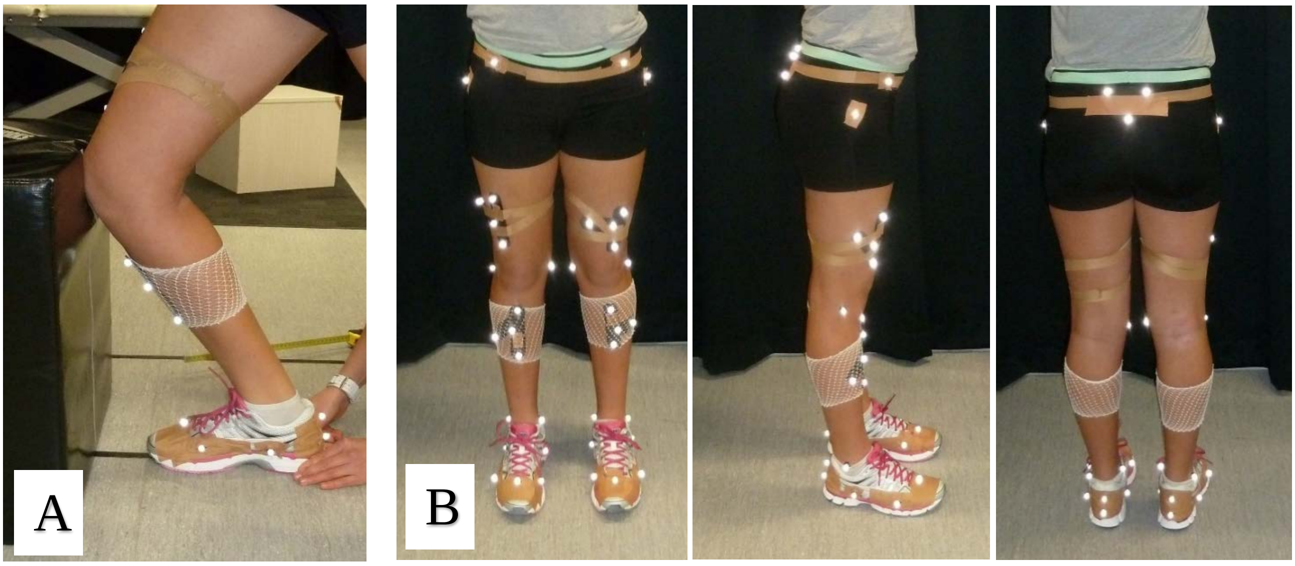

Figure 3: A - Standing lunge test; B - Marker placement

Testing procedures

Participants completed a warm-up consisting of three minutes on a stationary bike at a self-selected pace, five squats, and ten walking lunges. DF ROM was then measured using the standing lunge test as described by O'Shea and Grafton (2013). Participants stood in a lunge position with the test leg in front and foot aligned over a strip of tape aligned perpendicular to a box with the calcaneus and second toe centred on the tape (see Figure 3). Participants then lunged forward aiming their knee over the second toe and pushing the box forwards as far as possible without lifting their heel or losing knee contact with the box. The same researcher manually monitored the calcaneus for heel-lift and observed for correct sagittal plane knee movement for each participant. The distance from the tip of the great toe to the box was then measured. Both ankles were tested three times and the two closest measurements were averaged for analysis. Participants were tested while wearing shoes in order to ensure the result represented the DF range available in test conditions and sport situations. The standing lunge test has demonstrated high inter (ICC=0.99) and intra-rater (ICC=0.98-0.99) reliability (O'Shea & Grafton, 2013). Standing lunge tests have also been found to yield the greatest absolute DF range compared to open-chain measurement techniques and are postulated to give a more accurate measurement of functional, weight-bearing range (Krause et al., 2011). The test also mimics a landing position and is able to be easily used as a screening tool by coaches and clinicians. DF ROM was measured after warm-up to account for ROM increases occurring during exercise.

Participants then completed three successful trials of each landing task (see Table 11). Trials were considered unsuccessful if participants landed with any part of their foot off the force-plate, landed or approached the landing incorrectly as per task parameters, or lost their balance during landing. Tasks were first demonstrated and explained and participants were allowed to practice each task as often as needed until comfortable and consistent with the technique. Task order was randomised using a counterbalanced design with all possible task orders tabulated and randomly assigned to participants.

Landing tasks included a standardised drop jump and drop land allowing analysis of the effect of a subsequent jump on dependent variables (see Table 11). A netball-specific task involving a pass was also included to allow for analysis during a dynamic task with a sporting goal in which participants land instinctively with minimal instructions or restrictions. This task involved a one-to-two landing style (unilateral initial landing with the second foot quickly brought down ahead of the first) as it is the most common landing-style in netball (Ferdinand et al., 2008) and was found to be naturally preferred by participants during pilot testing. The majority of passes in netball are received bilaterally between chest and head-height and do not require players to reach, with players usually catching the ball while leaping or hopping forward (Hopper et al., 1992). For these reasons trials were considered unsuccessful if the pass was caught above the head or below the chest, required participants to reach further than arm-length, caught with a single hand, or dropped. All tasks required participants to land with one foot on each force plate.

Data Collection and Analysis

Outcome variables included ankle and knee sagittal excursion, leg, knee, and ankle stiffness, knee/ankle stiffness ratio, peak vGRF, TTP, and LR. Kinematic data was collected via a 9-camera (200 Hz) VICON motion analysis system (Oxford Metrics Ltd., Oxford, UK). A Bertec treadmill (BERTEC Corp, Worthington, OH, USA) set flush with the floor and locked to act as a force plate (1000 Hz) was used to collect kinetic data. Functional joint positions were determined using a custom built, MATLAB constrained optimization program (*Optimization Toolbox,* Mathworks Inc.; Natick, MA) (Besier et al., 2003). Visual3D software (*Visual 3D*, C-motion, Inc.; Germantown, MD) was used to calculate joint moments and foot centre of pressure locations via inverse dynamics. Anatomical co-ordinate systems were determined as described by Besier et al. (2003). The foot was modelled as a single-segment with the x-axis forming a line joining the two calcaneal markers, and y-axis forming a line from the proximal calcaneal marker to the forefoot midline marker. The z-axis was orthogonal to the x and y axes. Leg and joint (ankle and knee) stiffness were calculated from kinetic and kinematic data based on the validated mass-spring model as described in previous studies (Ambegaonkar et al., 2011; Schmitz & Shultz, 2010). The stiffness equations outlined in Table 12 were used as they have been found to be the most reliable stiffness calculations (Lorimer, 2013). Knee/ankle stiffness ratio was also calculated as it has been associated with injury (see Table 12) (Lorimer, 2013). Dorsiflexion ROM grouping was determined based on research indicating that participants with ≥130mm on the standing lunge test were at decreased risk of lower extremity injury (Dennis, Finch, McIntosh, & Elliott, 2008; Gabbe, Finch, Wajswelner, & Bennell, 2004). Ankles recording ≥130mm were assigned to the 'high-DF' group and an equal number of the lowest-scoring ankles were assigned to the 'low-DF' group. Group size was different for the netball jump as ankles were separated into leading and trailing limbs rather than left and right (see Table 13).

Table 11: Landing Tasks

Task	Description	Instructions
Drop jump	Drop from 35cm box placed at the edge of two side-by-side force plates with their dominant heel resting on the front edge of the platform. Maximal vertical jump immediately on landing.	"Step off the platform, land evenly on both feet with one foot on each force plate, then immediately jump straight up as high as you can".
Drop land	Drop from 35cm box placed at the edge of two side-by-side force plates with their dominant heel resting on the front edge of the platform.	"Step off the platform and land evenly on both feet with one foot on each force plate".
Netball jump	Running approach starting 3-4m behind the take-off point, jump forward off preferred limb, land with one foot on each force plate using a one-to-two landing style. During jump participants received a chest-pass delivered at a flat trajectory by a researcher standing 4m from the far edge of the force plate.at a 5° angle to the line of approach Exact starting point determined by what participants found to be the most comfortable in order to complete the task successfully.	"Imagine you are in a game situation running forward to receive a pass. Run forward to beat the defender to the ball, jump and catch the ball, then land, stop, and pass the ball back without stepping".

Leg stiffness (k_{Leg})	$k_{leg/dynamic} = \frac{\mathbf{F_{max}}}{\Delta\mathbf{L}}$	F_{max} = peak vertical force, ΔL = change in leg length from initial contact to peak vGRF
Joint stiffness (k_{Joint})	$k_{joint} = \frac{\Delta M}{\Delta\theta}$	ΔM = change in joint moment from initial contact to peak vGRF, Δθ = change in joint angle from initial contact to peak vGRF
Knee/ankle stiffness ratio (k_{KA})	$k_{KA} = k_{knee}/k_{ankle}$	

Table 12: Stiffness Equations

Statistical Analysis

Normality was tested via the Shapiro-Wilk test and visual inspection of histograms using SPSS version 22 (IBM Corp., 2013). Extreme outliers were defined as any data points lying more than three times the interquartile range away from the interquartile range and were removed prior to statistical analysis (Milner, Hamill, & Davis, 2007). The mean of three trials was used for analysis and as the data was not normally distributed it was log-transformed prior to analysis. A spreadsheet for comparing the means of two groups was used to derive magnitude based qualitative inferences (Hopkins, 2007c), as to the true effect of DF ROM on landing stiffness, joint excursions, and kinetics. An effect size (ES) of 0.20 was used as the threshold for substantial change (Hopkins, Batterham, Marshall, & Hanin, 2009). Where the effect had a >5% probability of being substantially positive and a >5% probability of being substantially negative the inference was stated as 'unclear' (Batterham & Hopkins, 2006). Otherwise the outcome was clear and the inference was based on the likelihood the true value of the ES was greater than 0.20 using the following scale: 25-75%, possibly; >75%, likely; >95%, very likely; >99.5%, most likely (Hopkins, 2007a; Hopkins et al., 2009). Magnitudes of observed ES's were interpreted based on the following scale: 0.2-0.59 (small), 0.6-1.19 (moderate), 1.2-1.99 (large), 2.0-3.99 (very large), ≥4.0 (extremely large) and their inverse (Hopkins et al., 2009).

Results

Prior to data collection three participants withdrew due to injury and one was unavailable on their scheduled testing day. Technical issues during data collection resulted in the exclusion of two further participants. Data analyses was therefore conducted with 20 participants (mean ±SD, age = 15.9 ±1.2 y; mass = 65.5 ±6.5 kg; height = 171.5 ±5.0 cm).

Six ankles were found to have ≥130mm on the standing lunge test and were assigned to the high-DF group for drop land and drop jump tasks. For the netball jump seven participants had ≥130mm in their leading limb and five had ≥130mm in their trailing limb and were assigned to the high-DF group. Differences in DF ROM between the high and low DF groups are summarised in Table 13.

Table 13: Differences in Dorsiflexion (DF) Range on the Standing Lunge Test Between the High-DF and Low-DF Groups

	DF group	Mean (mm) (SD)	Range (mm)	Mean difference (mm) (90% CI)	ES (90% CI)
Drop jump/drop land	High-DF (L)	145.50 (13.8)	130.0-162.5	60.5 (48.7-72.3)	5.72 (4.73-6.71)
	Low-DF (L)	85.00 (6.54)	74.0-93.0		
	High-DF (R)	144.42 (11.69)	134.5-162.0	58.0 (47.3-68.7)	4.42 (3.61-5.23)
	Low-DF (R)	86.42 (8.29)	73.0-998.0		
Netball jump	High-DF (lead)	144.36 (12.20)	130.5-162.5	54.1 (42.5-65.6)	3.16 (2.46-3.87)
	Low-DF (lead)	90.23 (11.86)	74.0-107.0		
	High-DF (trailing)	145.80 (13.63)	130.0-162.0	62.3 (48.7-75.9)	5.51 (4.47-6.54)
	Low-DF (trailing)	83.50 (6.52)	73.0-89.0		

R=right, L=left

Raw outcomes, effect sizes (ES), and inferences for the effect of low ankle DF ROM on stiffness, sagittal excursion, and kinetics are reported in Tables 14 to 16 Ankle stiffness showed a moderate increase on the left during the drop land (ES=0.84) and in the lead limb during the netball jump (ES=0.87). All other stiffness results were unclear. Ankle excursion showed a moderate decrease on the left during the drop jump (ES=-0.55) and in the trailing limb during the netball jump (ES=-0.97). Knee excursion showed a large increase on the left during the drop jump (ES=1.91) and in the trailing limb during the netball jump (ES=1.85). All other excursion results were unclear and all kinetic results were unclear.

Table 14: Effect of Low-DF ROM on Leg and Joint Stiffness

Variable (limb)	High DF group, mean (SD)	Low DF group, mean (SD)	Mean difference; ES (90% CI)	ES magnitude and qualitative inference†
Drop jump				
Leg (L) (kN/m/kg)	0.0270 (0.02)	0.0344 (0.02)	0.37 (-0.53-1.27)	Unclear
Leg (R) (kN/m/kg)	0.0398 (0.03)	0.0352 (0.02)	0.02 (-1.10-1.13)	Unclear
Knee (L) (Nm/°/kg)	0.0279 (0.01)	0.0267 (0.01)	-0.02 (-1.05-1.01)	Unclear
Knee (R) (Nm/°/kg)	0.0228 (0.01)	0.0204 (0.00)	-0.40 (-3.54-2.73)	Unclear
Ankle (L) (Nm/°/kg)	0.0260 (0.01)	0.0256 (0.01)	0.08 (-1.06-1.22)	Unclear
Ankle (R) (Nm/°/kg)	0.0238 (0.01)	0.0266 (0.01)	0.42 (-0.56-1.40)	Unclear
Ratio knee/ankle (L)	1.0801 (0.25)	1.0831 (0.28)	-0.02 (-0.84-0.81)	Unclear
Ratio knee/ankle (R)	0.9143 (0.29)	0.8732 (0.12)	-0.07 (-1.80-0.66)	Unclear
Drop land				
Leg (L) (kN/m/kg)	0.0208 (0.01)	0.0320 (0.02)	0.47 (-0.28-1.21)	Unclear
Leg (R) (kN/m/kg)	0.0251 (0.01)	0.0289 (0.01)	0.24 (-0.59-1.06)	Unclear
Knee (L) (Nm/°/kg)	0.0262 (0.01)	0.0238 (0.01)	0.21 (-0.75-1.17)	Unclear
Knee (R) (Nm/°/kg)	0.0177 (0.00)	0.0180 (0.00)	0.10 (-0.75-0.94)	Unclear
Ankle (L) (Nm/°/kg)	0.0172 (0.01)	0.0252 (0.01)	0.84 (-0.02-0.71)	Moderate** increase
Ankle (R) (Nm/°/kg)	0.0193 (0.01)	0.0202 (0.00)	0.43 (-1.03)-1.89	Unclear
Ratio knee/ankle (L)	1.4093 (0.38)	1.1830 (0.39)	-0.45 (-1.25-0.35)	Unclear
Ratio knee/ankle (R)	1.0273 (0.34)	0.9311 (0.21)	-0.23 (-1.62-1.16)	Unclear
Netball jump				
Leg (lead) (kN/m/kg)	0.2843(0.13)	0.3625 (0.18)	0.26 (-0.50-1.03)	Unclear
Leg (trailing) (kN/m/kg)	0.4535 (0.33)	0.4168 (0.34)	-0.05 (-1.11-1.00)	Unclear
Knee (lead) (Nm/°/kg)	0.3311 (0.35)	0.4498 (0.57)	-0.10 (-0.80-0.60)	Unclear
Knee (trailing) (Nm/°/kg)	0.0345 (0.01)	0.0309 (0.01)	-0.20 (-1.07-0.67)	Unclear
Ankle (lead) (Nm/°/kg)	0.1693 (0.11)	0.3975 (0.27)	0.87 (0.07-1.67)	Moderate** increase
Ankle (trailing) (Nm/°/kg)	0.2805 (0.39)	0.1596 (0.17)	-0.18 (0.91--1.27)	Unclear
Ratio knee/ankle (lead)	2.4047 (2.04)	2.7529 (2.35)	-0.13 (0.58--0.84)	Unclear
Ratio knee/ankle (trailing)	0.4535 (0.33)	0.4168 (0.34)	-0.05 (1.00--1.11)	Unclear

ES=effect size; L=left, R=right; CI=confidence interval; †ES magnitude interpreted based on the following scale: <0.2 (trivial), 0.2-0.59 (small), 0.6-1.19 (moderate), 1.2-1.99 (large), 2.0-3.99 (very large), ≥4.0 (extremely large) and their inverse; Inference based on the likelihood of ES >0.20 using the following scale: 25-75% possibly(*); >75% likely (**);>95% very likely (***)

Table 15: Effect of Low-DF ROM on Sagittal Joint Excursion

Variable (limb)	High DF group, mean (SD)	Low DF group, mean (SD)	Mean difference; ES (90% CI)	ES magnitude and qualitative inference†
Drop jump				
Ankle (L)	49.12 (6.99)	44.29 (12.78)	-0.35 (-1.07-0.37)	Unclear
Ankle (R)	46.63 (10.00)	45.97 (5.64)	-0.01 (-1.25-1.23)	Unclear
Knee (L)	22.10 (9.31)	37.44 (10.35)	1.91 (0.52-3.30)	Large*** increase
Knee (R)	39.57 (13.76)	34.20 (10.73)	-0.38 (-1.29-0.54)	Unclear
Drop land				
Ankle (L)	49.46 (8.47)	40.12 (13.72)	-0.55 (-1.28-0.17)	Moderate** decrease
Ankle (R)	46.23 (10.06)	43.29 (7.04)	-0.29 (-1.32-0.74)	Unclear
Knee (L)	33.76 (8.10)	33.43 (8.14)	-0.05 (-0.93-0.83)	Unclear
Knee (R)	9.60 (6.53)	31.55 (7.97)	0.15 (-0.66-0.95)	Unclear
Netball jump				
Ankle (lead)	22.03 (8.90)	30.19 (13.28)	0.52 (-0.24-1.29)	Unclear
Ankle (trailing)	25.12 (8.83)	13.37 (5.28)	-0.97 (-1.90--0.04)	Moderate** decrease
Knee (lead)	13.27 (3.27)	11.27 (5.17)	-0.40 (-1.14-0.33)	Unclear
Knee (trailing)	65.76 (19.85)	96.87 (16.49)	1.85 (0.42-3.29)	Large*** increase

ES=effect size; L=left, R=right; CI=confidence interval; †ES magnitude interpreted based on the following scale: <0.2 (trivial), 0.2-0.59 (small), 0.6-1.19 (moderate), 1.2-1.99 (large), 2.0-3.99 (very large), ≥4.0 (extremely large) and their inverse; Inference based on the likelihood of ES >0.20 using the following scale: 25-75% possibly(*); >75% likely (**);>95% very likely (***)

Table 16: Effect of Low-DF ROM on Kinetics

Variable (limb)	High DF group, mean (SD)	Low DF group, mean (SD)	Mean difference; ES (90% CI)	ES magnitude and qualitative inference†
Drop jump				
Peak vGRF (L) (N/kg)	10.64 (1.03)	10.55 (1.14)	-0.03 (-1.23-1.17)	Unclear
Peak vGRF (R) (N/kg)	10.54 (1.21)	11.22 (1.03)	0.58 (-0.44-1.60)	Unclear
TTP (L) (s)	0.400 (0.07)	0.382 (0.02)	-0.53 (-2.55-1.49)	Unclear
TTP (R) (s)	0.377 (0.04)	0.356 (0.03)	-0.56 (1.59--0.48)	Unclear
LR (L) (N/kg/s)	28.00 (4.70)	27.93 (4.74)	0.12 (1.30-1.53)	Unclear
LR (R) (N/kg/s)	28.88 (5.43)	31.81 (4.47)	0.59 (-0.45-1.64)	Unclear
Drop land				
Peak vGRF (L) (N/kg)	8.33 (0.94)	8.88 (0.77)	0.63 (-0.49-1.75)	Unclear
Peak vGRF (R) (N/kg)	8.81 (0.37)	9.44 (0.99)	0.50 (-0.22-1.22)	Unclear
TTP (L) (s)	0.350 (0.04)	0.337 (0.03)	-0.38 (-1.49-0.73)	Unclear
TTP (R) (s)	0.341 (0.02)	0.347 (0.02)	0.22 (-0.65-1.08)	Unclear
LR (L) (N/kg/s)	24.04 (3.64)	26.65 (3.87)	0.56 (-0.38-1.49)	Unclear
LR (R) (N/kg/s)	25.95 (1.41)	27.29 (3.15)	0.34 (-0.40-1.07)	Unclear
Netball jump				
Peak vGRF (lead) (N/kg)	12.47 (1.61)	11.81 (0.86)	-0.60 (-1.89-0.68)	Unclear
Peak vGRF (trailing) (N/kg)	11.83 (1.67)	11.79 (0.60)	0.05 (-1.93-2.02)	Unclear
TTP (lead) (s)	0.274 (0.01)	0.265 (0.01)	-0.55 (-1.39-0.28)	Unclear
TTP (trailing) (s)	0.324 (0.02)	0.327 (0.01)	0.20 (-1.07)-1.47	Unclear
LR (lead) (N/kg/s)	45.693 (5.74)	44.817 (3.81)	-0.16 (-1.28-0.97)	Unclear
LR (trailing) (N/kg/s)	36.795 (7.13)	36.072 (1.32)	-0.14 (-4.10-3.81)	Unclear

ES=effect size; L=left, R=right; CI=confidence interval; †ES magnitude interpreted based on the following scale: <0.2 (trivial), 0.2-0.59 (small), 0.6-1.19 (moderate), 1.2-1.99 (large), 2.0-3.99 (very large), ≥4.0 (extremely large) and their inverse; Inference based on the likelihood of ES >0.20 using the following scale: 25-75% possibly(*); >75% likely (**);>95% very likely (***)

Discussion

The current study investigated the effect of DF ROM on leg and joint stiffness, sagittal excursion, and ground-reaction forces during three different landing tasks in young netballers. Although unilateral increases in ankle stiffness and knee excursion and decreases in ankle excursion were seen in the low-DF groups for some tasks, the majority of results were unclear. Small to moderate effects observed for many of the unclear outcome variables indicate that further investigation is warranted.

Stiffness

The few clear stiffness differences supported our hypothesis, with the low-DF group demonstrating greater left ankle stiffness during the drop land and greater lead ankle stiffness during the netball jump. Although all other stiffness results were unclear, mean ESs suggest the possibility of an effect with greater leg, knee, and ankle stiffness observed in the low-DF group during the drop land, and greater lead limb leg and ankle stiffness observed during the netball jump. These observed effects suggest that further research is warranted. A causal relationship between lower extremity stiffness during landing tasks and injury risk has not been conclusively established due to a lack of research in this area (Brazier et al., 2014; Butler et al., 2003). Studies investigating stiffness during hopping and running tasks have reported greater forces with high stiffness and excessive joint motion with low stiffness, leading to speculations of greater injury risk when stiffness falls outside a given ideal range (Brazier et al., 2014; Butler et al., 2003). The greater stiffness values in the low-DF group suggest that low DF ROM may increase injury risk if stiffness values are excessively increased. Optimal stiffness ranges may be highly individual and task-dependent, further research is needed to establish the relationship between specific stiffness values and injury-risk. No other studies have directly investigated the association between DF ROM and landing stiffness. Based on the results of the current study further research is warranted to confirm the implications of DF ROM on landing stiffness and injury risk.

Joint excursion

The clear excursion results partially supported our hypotheses with the low-DF group demonstrating lower ankle excursion on the left during the drop land and in the trailing limb during the netball jump. Although other ankle inferences were unclear, the observation of lower ankle excursion on the right during the drop land and on the left during the drop jump in the low-DF group suggest that further research with greater participant numbers is warranted. The greater knee excursion seen in the low-DF group on the left during the drop jump and in the trailing limb during the netball jump was unexpected, and may have occurred in compensation for low ankle range. However as the majority of ankle results were unclear further research is needed to investigate this possibility. The only other study investigating the effect of low goniometric DF ROM on knee and ankle sagittal kinematics during a drop jump found no correlation between DF ROM and peak DF angle, while low DF ROM was correlated with reduced peak knee flexion angle (Malloy et al., 2014). However excursions were not reported making comparisons to the current study difficult. Reductions in sagittal ankle excursion during landing may increase injury risk as athletes adopt kinematic strategies to compensate. Reduced sagittal ankle joint excursion has been reported to result in compensatory subtalar and midfoot pronation (Bolgla, 2004; Piva et al., 2005; Tweed et al., 2008) or knee valgus (Bell et al., 2012; Piva et al., 2005; Pollard, Sigward, & Powers, 2010). Several studies have reported increased risk of injury with these movement patterns, particularly ACL injury (Hewett et al., 2005; Quatman et al., 2013), patellofemoral pain syndrome (Aminaka et al., 2011), and Achilles tendon injury (Lersch et al., 2012; Wyndow et al., 2010). Although frontal plane kinematics were not investigated in the current study the theory of compensatory frontal plane movement is supported by studies showing an association between low DF ROM and greater frontal plane knee motion or increased peak knee valgus (Malloy et al., 2014; Sigward et al., 2008; Stiffler et al., 2014) and ankle eversion (Whitting et al., 2011). The results of the current study and the potential injury implications indicate that further research is warranted in this area.

Kinetics

Although all kinetic results were unclear, observed effects suggest vGRF and LR may be higher in athletes with limited DF during jump/landing tasks. Previous studies have not supported a link between low DF and changes in vGRF's during landing. Malloy et al. (2014) found no correlation between DF ROM and peak vGRF during a drop jump, but did report greater peak knee abduction angle which has been hypothesised to be a compensatory movement adopted to prevent increased landing forces (Bell et al., 2012; Piva et al., 2005). Whitting et al. (2011) also found no effect of DF restriction on peak vGRF or TTP between high and low-DF groups during a unilateral drop land. However this study investigated a unilateral landing from heights of 32cm and 72cm which will have resulted in greater landing forces than in the current study, potentially causing participants not to utilise their full ROM. A drop land study by Wang (2009) found that when drop height increased from 40cm to 60cm leg stiffness was reduced in order to attenuate forces. However when the height was further increased to 80cm leg stiffness stabilised and landing forces increased. The authors concluded that when landing forces become excessively high the lower extremity is unable to reduced stiffness in order to buffer landing forces. The difference in results between Whitting et al. (2011) and the current study may therefore be the higher forces preventing participants increasing excursion in order to decrease stiffness. The results of the current study support further investigation in this area as increased lower-extremity injury risk has been reported in association with greater GRFs (Hewett et al., 1999; Hewett et al., 1996; Norcross et al., 2013) and LRs (Bisseling et al., 2007, 2008; Milner et al., 2006; Radin et al., 1991).

Limitations and future research

Low participant numbers in the current study resulted in wide confidence intervals and predominantly unclear inferences. However there were many observed differences between high and low-DF groups suggesting further research with greater participant numbers is warranted. Furthermore, there are a number of possible compensations for restricted DF ROM and different participants are likely to favour different strategies. Greater participant numbers might allow for the identification of multiple strategies.

The current study did not analyse frontal plane kinematics, or initial contact and peak angles which created some difficulty in interpreting results. There a large number of possible compensations for restricted DF ROM and the additional kinematic data may have allowed for a more complete profile of the compensations adopted. Further research is warranted to investigate these parameters.

Conclusion

The current study provides some evidence that young netballers with low DF ROM exhibit greater ankle stiffness, reduced ankle sagittal excursion and greater knee excursion during landing tasks. However many findings were unclear and further research with greater participant numbers is required.

CHAPTER 5

THE EFFECT OF ANKLE BRACING ON LANDING BIOMECHANICS IN YOUNG NETBALLERS

Abstract

Objective: To investigate the impact of lace-up ankle braces on landing biomechanics.

Design: Within-subject repeated measures.

Methods: Twenty female high school netball players (mean ±SD, age = 15.9 ±1.2 y; mass = 65.5 ±6.5 kg; height = 171.5 ±5.0 cm) completed three landing tasks in braced and unbraced conditions. Tasks included a drop jump, drop land, and netball-specific jump. Variables investigated included leg, knee and ankle stiffness, knee/ankle stiffness ratio, knee and ankle sagittal excursion, peak vertical ground reaction force (vGRF), time-to-peak vertical ground reaction force (TTP), and loading rate (LR). Mean differences between the braced and unbraced condition were analysed using a spreadsheet for deriving effect sizes and magnitude based inferences.

Results: In the brace condition leg stiffness increased bilaterally during the drop land (ES=0.21, 0.22), ankle stiffness increased bilaterally during the drop jump (ES=0.37, 0.29) and drop land (ES=0.40, 0.60), and knee/ankle stiffness ratio decreased in all three tasks (ES=-0.22 to -0.45). Leg and joint stiffness changes during the netball jump were unclear or trivial. In the brace condition ankle sagittal excursion decreased bilaterally during the drop jump (ES=-0.35,-0.53) and drop land (ES=-0.23,-46), and decreased in the lead limb during the netball jump (ES=-0.36). Knee excursion decreased bilaterally during the drop jump (ES=-0.36,-0.40) and in the lead limb during netball task (ES=-0.59). Changes in knee excursion during the drop land and trailing knee during the netball jump were trivial and results for the trailing ankle were unclear. Lead limb TTP was greater during the netball jump (ES=0.41) while all other kinetic results were trivial or unclear.

Conclusion: Lace-up ankle braces may result in greater leg and joint stiffness and reduced joint excursion during landing but do not appear to affect landing forces. The observed effect on landing biomechanics may predispose young netballers to injury.

Introduction

In 2013 netball was New Zealand's leading women's sport (MyNetball). With increasing participation over recent years the Accident Compensation Corporation (ACC) reported a 120% increase in the number of lower extremity netball injury claims from 2008/9 to 2012/13 resulting in an increased cost of almost 10 million dollars (ACC). The most commonly occurring netball injuries are knee and ankle sprains, calf strains, and Achilles tendon injuries (Langeveld et al., 2012; Otago & Peake, 2006). Poor landing technique and the high ground-reaction forces (GRFs) incurred as a result of sudden stop-landings may contribute to netball injury (Mothersole et al., 2013). Examples of poor landing techniques include stiff landings, low hip, knee, and ankle excursion, excessive hip and knee flexion at initial contact, excessive frontal plane knee excursion, and heel-to-forefoot ground contact pattern (Mothersole et al., 2013). Due to the high prevalence of ankle sprains in netball, players are encouraged to wear ankle braces both to support existing ankle injuries and to prevent ankle injury occurring (Hume, 1998; Hume & Steele, 2000; MyNetball). Although ankle bracing has been found to be effective in the prevention of ankle injury (Papadopulos et al., 2005) some studies have found reduced dorsiflexion (DF) range of motion (ROM) with lace-up braces (Eils et al., 2007; Parsley et al., 2013) and Aircast-stirrup braces (Eils et al., 2002; Eils et al., 2007). This reduction in range may subsequently restrict available knee and hip sagittal excursion on landing (Fong et al., 2011; Wang, 2009) resulting in a stiffer landing style and greater GRFs and loading rates (LRs) (Bisseling et al., 2007, 2008; Fong et al., 2011). As stiff landing styles and the subsequent increase in vGRF and LR have been reported to increase injury risk (Milner et al., 2006; Mothersole et al., 2013; Norcross et al., 2013) the purpose of this study is to investigate the impact of ankle bracing on landing stiffness. We hypothesised that ankle bracing would increase leg and joint stiffness, decrease knee/ankle stiffness ratio, decrease ankle and knee sagittal excursion, increase ground-reaction forces and loading rate, and decrease time-to-peak force.

Methods

Participants

Twenty-six female high school netball players from a local secondary school were recruited. Participants were excluded if they had any injury or illness which had the potential to impact their ability to perform the landing tasks. Participants were considered uninjured if they were fully participating in trainings and games at the time of testing. Informed consent was obtained from all participants and legal guardians where appropriate prior to testing. Ethical approval was obtained from AUTEC (reference number 14/167). Sample size calculations were made using an Excel spreadsheet for estimating sample size (Hopkins, 2006a). Based on a standardised smallest important difference (Cohen's d) of 0.2 and the standard error of measurement for leg stiffness during running reported by Lorimer (2013) it was estimated that 13 participants were required to achieve 80% power.

Study Design

A within-subject repeated measures design was used with participants completing all three landing-tasks with and without ankle braces. All participants attended the motion analysis laboratory on one occasion and completed all tasks in both braced and unbraced conditions on the same day.

Instrumentation

Participants' height, weight, and bilateral trochanteric height were measured by a researcher trained in International Society for the Advancement of Kinanthropometry (ISAK) protocols (Stewart et al., 2011). Retroreflective markers (10 mm) were then placed bilaterally at anatomical landmarks using a modification of three dimensional (3D) models described by, Besier et al. (2003), Tulchin et al. (2010), and Ferber et al. (2003) (see Figure 4). Clusters of four retroreflective tracking markers on thermo-moulded plastic shells were placed at the sacrum, mid-thigh, and mid-shank. Individual tracking markers were placed bilaterally on the ASIS, proximal and distal mid-line posterior calcaneus, medial and lateral anterior calcaneus, 1st and 5th metatarsal heads, and centre line of the forefoot between 2nd and 3rd metatarsal heads. Anatomical markers were placed bilaterally on the greater trochanter, medial and lateral femoral condyles, and medial and lateral malleoli. Foot markers were placed over the shoes with landmarks palpated through the shoes. The same physiotherapist palpated for anatomical landmarks and attached markers for all participants. Following a static standing calibration the femoral condyle and malleoli markers were removed.

Participants wore tight-fitting shorts and t-shirts or singlet's which allowed easy palpation of bony landmarks and secure application of markers. Brace-type was the McDavid lace-up 195-R which was selected as it is commonly used by netball players. All participants wore standard sports shoes of the same brand and style (Asics Gel-Kurow).

Testing procedures

Participants completed a warm-up consisting of three minutes on a stationary bike at a self-selected pace, five squats, and ten walking lunges. They then completed three successful trials of each landing task in braced and unbraced conditions (see Table 17). Trials were considered unsuccessful if participants landed with any part of their foot off the force-plate, landed or approached the landing incorrectly as per task parameters, or lost their balance during landing. Tasks were first demonstrated and explained and participants were allowed to practice each task as often as needed until comfortable and consistent with the technique. Task order was randomised using a counterbalanced design with all possible task orders tabulated and randomly assigned to participants. Tasks were completed in unbraced or braced condition and then repeated in the same order in the alternate condition. Brace condition order alternated with each consecutive participant.

Figure 4: Marker Placement

Landing tasks included a standardised drop jump and drop land allowing analysis of the effect of a subsequent jump on dependent variables (see Table 17). A netball-specific task involving a pass was also included to allow for analysis during a dynamic task with a sporting goal in which participants land instinctively with minimal instructions or restrictions. This task involved a one-to-two landing style (unilateral initial landing with the second foot quickly brought down ahead of the first) as it is the most common landing-style in netball (Ferdinand et al., 2008) and was found to be naturally preferred by participants during pilot testing. The majority of passes in netball are received bilaterally between chest and head-height and do not require players to reach, with players usually catching the ball while leaping or hopping forward (Hopper et al., 1992). For these reasons trials were considered unsuccessful if the pass was caught above the head or below the chest, required participants to reach further than arm-length, caught with a single hand, or dropped. All tasks required participants to land with one foot on each force plate.

Table 17: Landing Tasks

Task	Description	Instructions
Drop jump	Drop from 35cm box placed at the edge of two side-by-side force plates with their dominant heel resting on the front edge of the platform. Maximal vertical jump immediately on landing.	“Step off the platform, land evenly on both feet with one foot on each force plate, then immediately jump straight up as high as you can”.
Drop land	Drop from 35cm box placed at the edge of two side-by-side force plates with their dominant heel resting on the front edge of the platform.	“Step off the platform and land evenly on both feet with one foot on each force plate”.
Netball jump	Running approach starting 3-4m behind the take-off point, jump forward off preferred limb, land with one foot on each force plate using a one-to-two landing style. During jump participants received a chest-pass delivered at a flat trajectory by a researcher standing 4m from the far edge of the force plate at a 5° angle to the line of approach. Exact starting point determined by what participants found to be the most comfortable in order to complete the task successfully.	“Imagine you are in a game situation running forward to receive a pass. Run forward to beat the defender to the ball, jump and catch the ball, then land, stop, and pass the ball back without stepping”.

Data Collection and processing

Variables included ankle and knee sagittal excursion, leg, knee, and ankle stiffness, knee/ankle stiffness ratio, peak vertical ground reaction force (vGRF), time-to-peak vGRF (TTP), and loading rate (LR). Kinematic data was collected via a 9-camera (200 Hz) VICON motion analysis system (Oxford Metrics Ltd., Oxford, UK). A Bertec treadmill (BERTEC Corp, Worthington, OH, USA) set flush with the floor and locked to act as a force plate (1000 Hz) was used to collect kinetic data. Functional joint positions were determined using a custom built, MATLAB constrained optimization program (*Optimization Toolbox,* Mathworks Inc.; Natick, MA) (Besier et al., 2003). Visual3D software (*Visual 3D*, C-motion, Inc.; Germantown, MD) was used to calculate joint moments and foot centre of pressure locations via inverse dynamics. Anatomical co-ordinate systems were determined as described by Besier et al. (2003). The foot was modelled as a single-segment with the x-axis forming a line joining the two calcaneal markers, and y-axis forming a line from the proximal calcaneal marker to the forefoot midline marker. The z-axis was orthogonal to the x and y axes. Leg and joint (ankle and knee) stiffness were calculated from kinetic and kinematic data based on the validated mass-spring model as described in previous studies (Ambegaonkar et al., 2011; Schmitz & Shultz, 2010). The stiffness equations outlined in Table 18 were used as they have been found to be the most reliable stiffness calculations (Lorimer, 2013). Knee/ankle stiffness ratio was also calculated as it has been associated with injury (see Table 18) (Lorimer, 2013).

Table 18: Stiffness Equations

Leg stiffness (k_{Leg})	$k_{leg/dynamic} = \frac{F_{max}}{\Delta L}$	F_{max} = peak vertical force, ΔL = change in leg length from initial contact to peak vGRF
Joint stiffness (k_{Joint})	$k_{joint} = \frac{\Delta M}{\Delta \theta}$	ΔM = change in joint moment from initial contact to peak vGRF, Δθ = change in joint angle from initial contact to peak vGRF
Knee/ankle stiffness ratio (k_{KA})	$k_{KA} = k_{knee}/k_{ankle}$	

Statistical Analysis

Normality was tested via the Shapiro-Wilk test and visual inspection of histograms using SPSS version 22 (IBM Corp., 2013). Extreme outliers were defined as any data points lying more than three times the interquartile range away from the interquartile range and were removed prior to statistical analysis (Milner et al., 2007). Three outliers were identified and removed. The mean of three trials was used for analysis and as the data was not normally distributed it was log-transformed prior to analysis. A spreadsheet for comparing the means of two groups was used to derive magnitude based qualitative inferences (Hopkins, 2006) , as to the true effect of an ankle brace on landing stiffness, joint excursions, and kinetics. An effect size (ES) of 0.20 was used as the threshold for substantial change (Hopkins et al., 2009). Where the effect had a >5% probability of being substantially positive and a >5% probability of being substantially negative the inference was stated as 'unclear' (Batterham & Hopkins, 2006; Hopkins, 2007b). Otherwise the outcome was clear and the inference was based on the likelihood the true value of the ES was greater than 0.20 using the following scale: 25-75%, possibly; >75%, likely; >95%, very likely; >99.5%, most likely (Hopkins et al., 2009). Magnitudes of observed ES's were interpreted based on the following scale: 0.2-0.59 (small), 0.6-1.19 (moderate), 1.2-1.99 (large), 2.0-3.99 (very large), ≥4.0 (extremely large) and their inverse (Hopkins et al., 2009).

Results

Prior to data collection three participants withdrew due to injury and one was unavailable on their scheduled testing day. Technical issues during data collection resulted in the exclusion of two further participants. Data analyses was therefore conducted with 20 participants (mean ±SD, age = 15.9 ±1.2 y; mass = 65.5 ±6.5 kg; height = 171.5 ±5.0 cm).

Additionally technical issues during data collection resulted in the exclusion of all drop-land trials for one participant. A second participant switched leading limbs between braced and unbraced conditions during the netball jump and this data was also excluded.

Raw outcomes, effect sizes, and inferences for the effect of ankle bracing on stiffness, sagittal excursion, and kinetics are reported in Tables 19 to 21. The brace caused a small increase in leg stiffness during the drop land (ES L/R=0.21, 0.22) but trivial changes during the drop jump (ES L/R=0.18, -0.06). There was also a small increase in ankle stiffness in the brace condition during the drop jump (ES range=0.37, 0.29), and a moderate increase during the drop land (ES=0.40, 0.60). Changes in knee stiffness were trivial during the drop jump (ES L/R=0.05, -0.04), drop land on the right limb (ES=0.04) and trailing limb in the netball jump (0.11). Knee/ankle stiffness ratio showed a small decrease in the brace condition during the drop jump (ES L/R=-0.33, -0.45), drop land on the right limb (ES=-0.42), and netball jump (ES L/R=-0.38, -0.22). Results for leg and knee stiffness during the drop jump, right knee during the drop jump, and trailing knee during the netball jump were trivial (ES range=-0.19 to 0.18). All other changes in stiffness were unclear.

Ankle excursion showed a small decrease in the brace condition during the drop jump (ES L/R=-0.35, -0.53), and drop land (ES L/R=-0.23,-0.46) and in the lead limb during the netball jump (ES=-0.36). There was also a small decrease in knee excursion in the brace condition during the drop jump (ES L/R=-0.36, -0.40) and in the lead limb during the netball jump (ES=0.59), with trivial changes during the drop land (ES L/R=0.11, 0.13) and trailing limb in the netball jump (ES=0.15). The effect on trailing limb ankle excursion in the netball task was unclear.

TTP showed a small increase in the leading limb during the netball jump (ES=0.41). All other kinetic results were trivial (ES range=-0.15 to 0.14) or unclear.

Table 19: Effect of Ankle Bracing on Leg and Joint Stiffness

Variable (limb)	Brace; Mean (SD)	No brace; Mean (SD)	Mean Difference; ES (90% CI)	ES magnitude and qualitative inference†
Drop jump				
Leg (L) (kN/m/kg)	0.102 (0.31)	0.040 (0.03)	0.18 (-0.18-0.53)	Small* increase
Leg (R) (kN/m/kg)	0.035 (0.02)	0.038 (0.03)	-0.06 (-0.19-0.07)	Trivial***
Knee (L) (Nm/°/kg)	0.039 (0.03)	0.032 (0.02)	0.05 (-0.13-0.24)	Trivial**
Knee (R) (Nm/°/kg)	0.022 (0.01)	0.023 (0.01)	-0.04 (-0.21-0.13)	Trivial**
Ankle (L) (Nm/°/kg)	0.035 (0.01)	0.032 (0.02)	0.37 (-0.19-0.55)	Small* increase
Ankle (R) (Nm/°/kg)	0.031 (0.01)	0.029 (0.01)	0.29 (0.04-0.54)	Small* increase
Ratio knee/ankle (L)	0.946 (0.26)	1.057 (0.24)	-0.33 (-0.53--0.14)	Small*** decrease
Ratio knee/ankle (R)	0.732 (0.20)	0.854 (0.21)	-0.45 (-0.63--0.28)	Small** decrease
Drop land				
Leg (L) (kN/m/kg)	0.039 (0.03)	0.029 (0.02)	0.21 (-0.06-0.48)	Small* increase
Leg (R) (kN/m/kg)	0.039 (0.03)	0.029 (0.01)	0.22 (-0.02-0.47)	Small* increase
Knee (L) (Nm/°/kg)	0.06 (0.11)	0.05 (0.08)	0.07 (-0.34-0.47)	Unclear
Knee (R) (Nm/°/kg)	0.02 (0.01)	0.02 (0.01)	0.04 (-0.08-0.15)	Trivial***
Ankle (L) (Nm/°/kg)	0.028 (0.01)	0.026 (0.01)	0.40 (0.10-0.70)	Small** increase
Ankle (R) (Nm/°/kg)	0.026 (0.01)	0.022 (0.01)	0.60 (0.33-0.88)	Moderate*** increase
Ratio knee/ankle (L)	1.248 (0.47)	1.258 (0.37)	-0.19 (-0.53-0.15)	Small* decrease
Ratio knee/ankle (R)	0.840 (0.21)	1.015 (0.41)	-0.42 (-0.59--0.25)	Small*** decrease
Netball jump				
Leg (lead) (kN/m/kg)	0.28 (0.10)	0.34 (0.14)	-0.20 (-0.85-0.46)	Unclear
Leg (trailing) (kN/m/kg)	0.08 (0.04)	0.09 (0.05)	-0.14 (-0.56-0.27)	Unclear
Knee (lead) (Nm/°/kg)	0.34 (0.45)	0.41 (0.43)	-0.20 (-0.66-0.26)	Unclear
Knee (trailing) (Nm/°/kg)	0.04 (0.02)	0.03 (0.01)	0.11 (-0.05-0.28)	Trivial**
Ankle (lead) (Nm/°/kg)	0.38 (0.39)	0.29 (0.22)	0.12 (-0.27-0.52)	Unclear
Ankle (trailing) (Nm/°/kg)	0.31 (0.59)	0.44 (0.77)	-0.02 (-0.39-0.35)	Unclear
Ratio knee/ankle (lead)	1.53 (1.28)	2.357 (2.01)	-0.38 (-0.62--0.14)	Small** decrease
Ratio knee/ankle (trailing)	0.365 (0.32)	0.408 (0.35)	-0.22 (-0.53--0.08)	Small* decrease

ES=effect size; L=left, R=right; CI=confidence interval; †ES magnitude interpreted based on the following scale: <0.2 (trivial), 0.2-0.59 (small), 0.6-1.19 (moderate), 1.2-1.99 (large), 2.0-3.99 (very large), ≥4.0 (extremely large) and their inverse; Inference based on the likelihood of ES >0.20 using the following scale: 25-75% possibly(*); >75% likely (**);>95% very likely (***)

Table 20: Effect of Ankle Bracing on Sagittal Joint Excursion

Variable (limb)	Brace; Mean (SD)	No brace; Mean (SD)	Mean Difference; ES (90% CI)	ES magnitude and qualitative inference†
Drop jump				
Ankle (L)	41.91 (7.16)	45.91 (9.08)	-0.35 (-0.53--0.17)	Small** decrease
Ankle (R)	38.86 (6.82)	43.51 (8.95)	-0.53 (-0.78--0.28)	Small *** decrease
Knee (L)	37.80 (8.52)	40.33 (10.22)	-0.36 (-0.65--0.07)	Small ** decrease
Knee (R)	32.99 (6.05)	35.90 (9.53)	-0.40(-0.83-0.02)	Small ** decrease
Drop land				
Ankle (L)	39.26 (7.40)	43.57 (10.21)	-0.23 (-0.40--0.06)	Small* decrease
Ankle (R)	35.12 (7.69)	40.84 (9.10)	-0.46 (-0.67--0.24)	Small*** decrease
Knee (L)	32.68 (8.67)	32.55 (9.00)	0.11 (-0.19-0.42)	Small* increase
Knee (R)	30.62 (7.07)	30.67 (6.78)	0.13 (-0.14-0.40)	Small* increase
Netball jump				
Ankle (lead)	21.52 (8.77)	24.89 (10.96)	-0.36 (-0.09--0.62)	Small ** decrease
Ankle (trailing)	17.31 (7.36)	17.06 (9.38)	0.18 (-0.36-0.71)	Unclear
Knee (lead)	9.35 (3.29)	11.46 (4.32)	-0.59 (-1.00--0.18)	Small** decrease
Knee (trailing)	81.35 (17.80)	78.21 (20.3)	0.15-(-0.12--0.42)	Small* increase

ES=effect size; L=left, R=right; CI=confidence interval; †ES magnitude interpreted based on the following scale: <0.2 (trivial), 0.2-0.59 (small), 0.6-1.19 (moderate), 1.2-1.99 (large), 2.0-3.99 (very large), ≥4.0 (extremely large) and their inverse; Inference based on the likelihood of ES >0.20 using the following scale: 25-75% possibly(*); >75% likely (**);>95% very likely (***)

Table 21: Effect of Ankle Bracing on Kinetics

Variable (limb)	Brace; Mean (SD)	No brace; Mean (SD)	Mean Difference; ES (90% CI)	ES magnitude and qualitative inference†
Drop jump				
Peak vGRF (L) (N/kg)	11.06 (1.27)	10.89 (1.48)	0.14 (-0.06-0.35)	Small* increase
Peak vGRF (R) (N/kg)	11.00 (1.48)	11.22 (1.31)	-0.15 (-0.40-0.11)	Small* decrease
TTP (L) (s)	0.37 (0.04)	0.37 (0.05)	0.07 (-0.31-0.46)	Unclear
TTP (R) (s)	0.37 (0.04)	0.36 (0.03)	0.11 (-0.15-0.36)	Small* increase
LR (L) (N/kg/s)	30.54 (6.44)	30.50 (7.86)	0.04 (-0.24-0.32)	Unclear
LR (R) (N/kg/s)	30.91 (5.56)	31.49 (6.21)	-0.12 (-0.35-0.11)	Small* decrease
Drop land				
Peak vGRF (L) (N/kg)	9.07 (1.02)	8.99 (1.05)	0.02 (-0.25-0.30)	Unclear
Peak vGRF (R) (N/kg)	9.09 (0.81)	9.15 (0.79)	0.07 (-0.19-0.33)	Trivial**
TTP (L) (s)	0.34 (0.04)	0.32 (0.08)	-0.03 (-0.38-0.33)	Unclear
TTP (R) (s)	0.34 (0.04)	0.34 (0.03)	0.01 (0.35--0.36)	Unclear
LR (L) (N/kg/s)	27.42 (4.71)	26.87 (4.32)	0.07 (-0.20-0.34)	Unclear
LR (R) (N/kg/s)	27.49 (4.43)	27.38 (3.50)	0.06 (-0.19-0.31)	Trivial**
Netball jump				
Peak vGRF (lead) (N/kg)	12.17 (1.09)	12.15 (1.25)	-0.12 (-0.45-0.20)	Small* decrease
Peak vGRF (trailing) (N/kg)	11.54 (0.87)	11.61 (0.99)	0.05 (-0.24-0.35)	Unclear
TTP (lead) (s)	0.28 (0.01)	0.27 (0.01)	0.41 (0.04-0.79)	Small** increase
TTP (trailing) (s)	0.32 (0.02)	0.32 (0.02)	-0.13 (-0.66-0.40)	Unclear
LR (lead) (N/kg/s)	30.54 (6.44)	30.50 (7.86)	0.04 (-0.24-0.32)	Unclear
LR (trailing) (N/kg/s)	30.91 (5.56)	31.49 (6.21)	-0.12 (-0.35-0.11)	Small* decrease

ES=effect size; L=left, R=right; CI=confidence interval; †ES magnitude interpreted based on the following scale: <0.2 (trivial), 0.2-0.59 (small), 0.6-1.19 (moderate), 1.2-1.99 (large), 2.0-3.99 (very large), ≥4.0 (extremely large) and their inverse; Inference based on the likelihood of ES >0.20 using the following scale: 25-75% possibly(*); >75% likely (**);>95% very likely (***)

Discussion

The current study investigated the effect of ankle bracing on leg and joint stiffness, sagittal excursion, and kinetics during three different landing tasks. Our hypotheses were partially supported with increases in leg and joint stiffness and decreased sagittal excursion observed in the brace condition during a number of landing tasks. Contrary to our hypotheses, peak vGRF, TTP, and LR were largely unaffected by the brace.

Stiffness

Results for the drop jump and drop land partially support our hypotheses. The brace increased bilateral leg stiffness during the drop land, and increased bilateral ankle stiffness in both tasks. The lack of change in knee stiffness was unexpected, particularly given the reduction in knee excursion observed during the drop jump, and suggests that leg stiffness changes were primarily a result of changes at the ankle. The reduction in knee/ankle stiffness ratio for both tasks further supports this idea. The lack of knee stiffness changes combined with reduced knee excursion during the drop jump suggests that external knee flexion moment may have reduced. Lower flexion moment reduces the demand on the quadriceps to decelerate knee flexion, potentially reducing the risk of ACL (Sell et al., 2007) and patellar tendon injury (Bisseling et al., 2008; Van der Worp, de Poel, Dierks, van den Akker-Scheek, & Zwerver, 2014). The larger increases in ankle and leg stiffness during the drop land compared with the drop jump may be due to the drop jump requiring participants to recoil quickly in preparation for the second jump. Previous research has found that as ground-contact time between landing and a subsequent jump decreases, leg and joint stiffness increases (Arampatzis, Brüggemann, & Klapsing, 2001). If participants utilised a stiff landing style in both braced and unbraced conditions during the drop jump this may explain the smaller differences observed in the drop jump compared with the drop land. Although the effect on left leg stiffness was trivial in the drop jump, it approached the threshold for a small effect while the effect on the right leg was very likely trivial. This asymmetry may be due to the need to quickly perform a vertical jump for maximal height, introducing issues of jump performance and coordination. Jump performance is affected by alterations in leg and joint stiffness with excessive or insufficient stiffness affecting the ability to utilise the stretch-shorten-cycle during plyometric tasks (Brazier et al., 2014; Brughelli & Cronin, 2008). The specific stiffness parameters which allow for maximal performance are as yet unclear (Brughelli & Cronin, 2008; McMahon, Comfort, & Pearson, 2012) but the ability to utilise optimal stiffness is likely dependant on a number of factors including strength, power, and coordination (Brazier et al., 2014; McMahon et al., 2012). Differences in strength and coordination between dominant and non-dominant limbs may therefore contribute to asymmetrical stiffness parameters.

The wide confidence intervals and unclear inferences observed in the netball jump suggest a high degree of variation between participants. This is consistent with the parameters of the task which purposely allowed greater freedom in landing-style compared with the other two tasks. Although ankle stiffness results were unclear the decrease in the knee/ankle stiffness ratio supports increased ankle stiffness in the braced condition. Reduced ankle excursion found in the leading limb also supports an increase in ankle stiffness on this side. Although inferences for lead limb leg and knee stiffness were unclear, observed ESs suggest a decrease in both variables in the brace condition. If knee stiffness has decreased this may be a compensation for increased ankle stiffness, or alternatively, decreased stiffness may be a result of participants attempting to absorb more load through the trailing limb. This theory is supported by excursion results suggesting that in the brace condition participants may have utilised greater ROM in the trailing limb, allowing participants to take greater impact through the trailing limb without an increase in landing forces. If in the unbraced condition the trailing limb absorbs less load than it is safely capable of then adjusting landing strategy to increase the load taken by this limb may be an effective way to manage reduced joint ROM without substantially increasing injury-risk. However if landing forces become too high this strategy may have injury implications for the trailing limb.

The only other study investigating the effect of ankle bracing on stiffness found no effect of an Aircast-stirrup brace on vertical stiffness during continuous bilateral hopping (Williams & Riemann, 2009). However, as this study investigated a continuous task participants may have landed stiffly both with and without the brace in order to prepare for the next hop resulting in no difference between brace conditions. Furthermore, this study investigated vertical stiffness rather than lower-extremity or individual joint stiffness and may therefore have missed changes occurring at specific components of the kinetic chain.

The relationship between lower extremity stiffness during landing tasks and injury risk has not been conclusively established due to a lack of research in this area (Brazier et al., 2014; Butler et al., 2003). Studies investigating stiffness during hopping and running tasks have reported greater forces with high stiffness and excessive joint motion with low stiffness, leading to speculations of greater injury risk when stiffness falls outside a given ideal range (Brazier et al., 2014; Butler et al., 2003). The greater ankle stiffness found in the current study suggests that ankle bracing may increase injury risk if it causes stiffness values to exceed a given optimal range or alternatively may reduce injury risk if stiffness increases to within the optimal range. Optimal stiffness ranges may be highly individual and task-dependent. Thus further research is needed to establish the relationship between specific stiffness values and injury-risk. The relationship between injury and knee/ankle stiffness ratio has been investigated in a study by Lorimer (2014) who found that runners with a high ratio were at increased risk of Achilles tendon injury. Ankle bracing may therefore be protective against Achilles injury, but further research is required to investigate whether the same relationship is seen for landing tasks.

Excursion

Excursion results for the drop jump and drop land partially supported our hypothesis with reduced ankle and knee excursion during the drop jump and reduced ankle excursion during the drop land. Although the effect of the brace on knee excursion during the drop land was trivial, the inferences suggest the possibility of a small increase. As participants were not preparing for a second jump it is possible that they were able to increase knee excursion in compensation for the reduction in ankle excursion. Reduced excursion was also seen at the lead knee and ankle during the netball jump. The greater reduction in knee excursion during the netball jump compared with the drop jump and drop land suggests the effect of the brace may be more substantial during tasks more specific to playing netball. Although there was a trivial effect on the trailing knee during the netball jump, the inference suggests the possibility of a small increase. This may be a result of participants attempting to absorb greater load through the trailing limb in order to prevent an excessively heavy initial landing. Utilising greater knee joint excursion may have allowed participants to take greater impact through the trailing limb without an excessive increase in stiffness or landing forces.

The observed reduction in ankle and knee sagittal excursion may increase injury risk as athletes attempt to compensate via pronation (Bolgla, 2004; Piva et al., 2005; Tweed et al., 2008) or knee valgus (Bell et al., 2012; Piva et al., 2005). These movement patterns have been associated with increased risk of Achilles tendon injury, knee ligament injury, and patellofemoral pain syndrome (Aminaka et al., 2011; Hewett et al., 2005; Lersch et al., 2012; Quatman et al., 2013; Wyndow et al., 2010). Although there is limited evidence for compensatory pronation in association with reduced ankle sagittal excursion (Whitting et al., 2011) a previous study investigating lace-up braces during a sports-specific netball jump suggested the brace may reduce frontal plane motion and prevent this compensation strategy from being utilised (Vanwanseele et al., 2013). The current study did not investigate frontal plane kinematics but the theory of knee valgus compensation is supported by studies showing greater frontal plane knee motion or peak knee valgus during landing in association with low DF ROM (Malloy et al., 2014; Sigward et al., 2008; Stiffler et al., 2014). However, these studies investigated goniometric DF ROM rather than ankle excursion during landing and a number of studies have shown that low goniometric DF ROM does not necessarily result in reduced peak DF angle on landing (Dill et al., 2014; Fong et al., 2011; Malloy et al., 2014; Whitting et al., 2011). Furthermore, the only previous study investigating frontal plane knee motion with lace-up braces found no change in frontal plane knee motion during a drop land despite a reduction in knee and ankle sagittal excursion (Simpson et al., 2013). However this study reported increases in vGRF and LR, possibly due to the observed lack of frontal plane compensation while participants in the current study may have compensated in the frontal plane in order to prevent an increase in landing forces.

An alternative theory suggests that reduced ankle sagittal excursion on landing is associated with altered lower limb alignment at initial contact (IC) (Begalle et al., 2015). Begalle et al. (2015) found that low sagittal ankle excursion during landing was associated with greater IC hip and knee flexion angles, and reduced plantarflexion angle. The authors theorised that the more flexed IC position was adopted to reduce landing forces when joint range did not allow for sufficient excursion. They also suggest that this position contributed to larger IC knee varus and hip internal rotation angles which are associated with ACL injury and patellofemoral pain syndrome. Previous research into lace-up braces has found reduced IC plantarflexion angle (DiStephano et al., 2008; McCaw et al., 1999; Simpson et al., 2013) and greater IC knee flexion angle (DiStephano et al., 2008; Simpson et al., 2013). The current study did not investigate initial contact or peak angles, however it is possible that the observed reduction in sagittal excursion was due to increased IC flexion angles adopted in order to prevent increased loading in the presence of ankle restriction.

Kinetics

Contrary to our hypothesis there were minimal changes in kinetics as a result of ankle bracing, suggesting that changes in stiffness were primarily a result of reduced excursion. It is possible that participants increased frontal plane excursion or sagittal hip and trunk excursion or adopted a more flexed position at IC as described by Begalle et al. (2015), in compensation for the reduction in displacement at the hip and knee. It is also possible that multiple small compensations were made which were too minor to be detected. The increase in lead limb TTP during the netball jump combined with inferences suggesting reduced peak vGRF on this side support the theory of a compensatory increase in load absorption by the trailing limb. The many possible compensations for this task likely contribute to unclear results as different participants favoured different strategies.

Limitations and future research

There were a number of unclear inferences in the current study indicating that greater participant numbers may have been needed. Greater numbers might also allow for the identification of different compensation strategies between participants. The lack of analysis of IC and peak angles or frontal plane movement limits our ability to identify the compensation patterns adopted. Studies including these variables in their analysis may allow for more specific analysis of compensation strategies. As this study was conducted with female high school netball players, further research with a more diverse population involved in a variety of sports is required. Research into other sport-specific tasks is also required as different tasks may yield different results. As only one brace-type was investigated the results of this study are also limited to lace-up braces with locking straps, future research is needed to establish the impact into other braces and strapping tape on landing biomechanics.

Conclusion

During landing tasks lace-up ankle braces may increase leg and joint stiffness and alter lower extremity landing kinematics in a manner which predisposes young netballers to injury. However, results differed between tasks calling into question the applicability of results from standardised landing tasks to sport-specific situations. The number of unclear inferences suggest a high degree of variation between participants, particularly during the netball jump. The lack of kinetic changes combined with observed increases in ankle stiffness suggest that participants were able to compensate via changes in joint excursions to prevent increased loading.

CHAPTER 6

DISCUSSION AND CONCLUSIONS

Restricted ankle dorsiflexion (DF) range of motion (ROM) has been identified as a risk factor for a number of acute and chronic lower extremity injuries (Backman & Danielson, 2011; Didier, 2011; Hadzic et al., 2009; Kaufman et al., 1999; Kibler et al., 1991; Piva et al., 2005; Wahlsteadt & Rasmussen-Barr, 2014). Although the biomechanical reasons for this association are as yet unclear, alterations in landing technique have been suggested as a possible mechanism of injury (Bolgla, 2004; Fong et al., 2011; Macrum et al., 2012; Mauntel et al., 2013; Piva et al., 2005; Tweed et al., 2008). The high prevalence of ankle sprains in netball may place netballers at increased injury risk as reduced DF ROM has been reported following ankle sprain (Aiken et al., 2008), (Baumhauer & O'Brien, 2002), with chronic ankle instability (Eils et al., 2007; Parsley et al., 2013) Furthermore, in order to reduce the risk of ankle injury young netballers are encouraged to wear ankle braces which may themselves restrict DF ROM (Eils et al., 2007; Parsley et al., 2013).and increase the risk of injury to other joints. The sudden stop-landings involved in netball may also place these athletes at greater risk.

There are a number of theories as to the mechanisms by which reduced DF ROM alters landing biomechanics, however a literature review in this thesis identified only a small number of studies investigating the various mechanisms and results were inconsistent between studies. There was support for a number of biomechanical changes which are associated with increased injury risk, these included reduced knee and hip sagittal excursion, and increased peak vertical ground reaction force. As a number of studies have found that some types of ankle brace can restrict dorsiflexion range of motion, a second literature review was included in this thesis to establish the evidence for changes in landing biomechanics with ankle bracing. Overall findings of the review indicate that lace-up braces restrict sagittal plane ankle motion on landing and to possibly also reduce sagittal plane knee excursion, along with increased landing forces. However results were inconsistent between studies and in both reviews there were a lack of studies investigating sport-specific tasks such as netball. There are a large number of possible compensations for restricted range and ankle bracing and individuals are likely to favour different strategies. However the majority of studies investigated changes in individual biomechanical variables rather than biomechanical patterns, possibly contributing to variation in outcomes/results. It has been suggested that an alternative measure such as stiffness which captures a number of biomechanical variables into a single measure may allow for clearer outcomes (Lorimer, 2014). Thus this Master's thesis investigated the impact of both DF ROM and lace-up ankle braces on landing biomechanics in young netballers with the inclusion of stiffness as an outcome variable and a netball-specific landing task.

The main aim of chapter four was to identify the impact of low dorsiflexion range of motion on landing biomechanics. The primary outcome variable was stiffness with leg, ankle, and knee stiffness investigated as well as knee/ankle stiffness ratio. As stiffness is primarily modulated by changes in sagittal plane joint excursion and as changes in stiffness can affect landing kinetics, sagittal plane ankle and knee excursion were also investigated as well as peak vertical ground reaction force, time-to-peak ground reaction force, and loading rate. To the authors' knowledge this is one of the first studies to directly investigate the effect of ankle dorsiflexion range of motion on leg, knee, and ankle stiffness during landing tasks. Highly standardised drop jump and drop land tasks were investigated to allow comparison with other studies. A third, netball-specific task was also included to ensure results could be applied to sporting situations. The study recruited a group of high school netballers who were divided into high and low-DF groups based on research indicating that athletes who achieved ≥130mm on the standing lunge DF ROM test were at reduced risk of lower-extremity injury. Unfortunately the majority of findings were unclear, however the observed effects in the study indicated that low-DF participants exhibited greater ankle stiffness, reduced ankle sagittal excursion, and greater knee excursion during landing. The greater landing stiffness seen in low-DF participants may place them at a greater risk of injury. Excessive stiffness on landing has been suggested to increase the risk of certain injuries, such as stress-fractures (Butler et al., 2003), hamstring strain (Brazier et al., 2014), and patellar tendinopathy (Bisseling et al., 2007), but a causal relationship has not been established (Brazier et al., 2014; Butler et al., 2003). Reduced ankle excursion may also predispose athletes to injury as participants compensate for insufficient range with alternate movements such as increased knee valgus, pronation or increased ground reaction forces and loading rates. However, as frontal plane kinematics were not investigated in the current study and all kinetic results were unclear, it is not known whether these changes occurred. The observed increase in knee excursion may have been adopted in compensation for reduced ankle range in an attempt to keep leg stiffness constant and prevent an increase in landing forces.

Chapter five aimed to investigate the impact of lace-up ankle braces on landing biomechanics. Lace-up braces were chosen as they are commonly used by netballers both to support and prevent ankle injury and have been shown to restrict DF ROM. The same tasks and variables were investigated as in chapter four and a within-subject design was chosen with participants completing all tasks in both braced and unbraced conditions. To the authors' knowledge this is one of the first studies to directly investigate the effect of ankle bracing on leg, knee, and ankle stiffness during landing tasks. Overall results indicated increases in leg and joint stiffness and reduced joint excursion in the brace condition, while kinetics were unchanged. However results differed between tasks, with results for the netball jump suggesting the braces caused reduced ankle and knee excursion in the leading limb while excursion in the trailing limb increased, possibly reflecting a shift to increased loading in the trailing limb. Stiffness results were unclear for the netball jump, possibly due to the greater freedom in landing style inherent in the task. The differing results between the netball jump and other tasks call into question the applicability of results from standardised landing tasks to sporting situations. The greater stiffness and reduced joint excursion found in this study suggest lace-up ankle braces may contribute to increased risk of lower limb injures. This could occur because participants compensate kinematically for the restricted dorsiflexion movement.

Thesis Limitations

Several limitations in the studies presented should be taken into account when interpreting results.

- Low participant numbers created difficulty in interpretation of results as it was not possible to account for individual variation in landing and compensation strategies. Low numbers also contributed to a number of unclear results.
- The lack of investigation of the hip joint, frontal and transverse planes, and specific initial contact and peak angles limits the ability to explain all possible biomechanical compensations that could result due to low DF ROM and ankle bracing during landing.
- The investigation of only a single brace-type limits the generalisability of results.

- Although the studies in this thesis included a sport-specific jump, the applicability of the task is limited only to netball. The nature of laboratory testing also introduces the possibility that the effect of low DF range and ankle bracing may be different during actual sporting situations.

Recommendations for Future Research

The large number of possible biomechanical changes occurring as a result of reduced dorsiflexion range and ankle bracing necessitate further investigation into specific changes in initial contact and peak joint angles, frontal plane kinematic changes, and kinematic changes at the hip. The high degree of coordinative variability in landing and compensation strategies also necessitates studies which investigate changes in kinematic and kinetic parameters at multiple joints across all three planes to allow identification of changes in biomechanical patterns.

The studies in this thesis provide evidence for changes in lower-extremity stiffness during landing as a result of low DF ROM and ankle bracing. However, no other studies have investigated these relationships and therefore further research is needed to confirm these findings. The potential for stiffness measures to describe changes in movement patterns with clearer outcomes in a highly individual and variable task also warrants further investigation. Furthermore, the relationship between lower-extremity stiffness during landing and injury-risk has not been established. Prospective studies are required to investigate the relationship between stiffness and the risk of specific injuries during landing tasks.

The applicability of these results is limited to netball and to lace-up ankle braces. Further research investigating other types of goal-directed sporting tasks, and using other brace-types and strapping tape is required.

Conclusion

This thesis consists of two studies investigating the impact of low dorsiflexion range of motion and lace-up ankle braces on lower-extremity landing biomechanics in young netballers. These studies are the first to investigate lower-extremity stiffness during landing-tasks, and one of the few to investigate the impact of DF range and ankle bracing on landing biomechanics during a goal-directed sport-specific task.

Restricted dorsiflexion range of motion may alter landing biomechanics in a manner which predisposes young netballers to injury. Coaches, players, and clinicians should screen for dorsiflexion range and provide appropriate interventions to improve range when it is limited. While ankle bracing may be effective in reducing the risk of ankle injury, it may increase the risk of injuries at other parts of the kinetic chain. These risks need to be considered when recommending the use of ankle braces to young netballers.

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APPENDIX 1

Stiffness overview

The lower-extremity has been successfully modelled as a spring which stores elastic energy during loading and returns energy during the propulsive phase of gait and jumping tasks (Butler et al., 2003). In physics, the stiffness of a spring refers to a constant which describes the relationship between the force required to deform a material and distance the material is deformed and is described by the formula F = kx where F is the force required, x is the distance the material is deformed, and k is the proportionality constant which represents stiffness (Butler et al., 2003). In biomechanics this is generally translated into the relationship between displacement and ground reaction force (GRF) or joint-moment (Butler et al., 2003; Serpell et al., 2012). A number of different stiffness measurements can be taken during jumping and running tasks including the resistance of the body to vertical displacement (vertical stiffness), resistance to change in leg-length (leg-stiffness), and resistance to angular joint displacement (joint-stiffness) (Serpell et al., 2012). Studies disagree as to which joint has the greatest influence on overall lower-extremity stiffness with some weighting ankle stiffness as the most important and others knee stiffness (Serpell et al., 2012). It is likely that the relative contributions of each joint are highly task-dependent (Serpell et al., 2012).

A number of different methods of calculating lower-extremity stiffness have been proposed (Serpell et al., 2012). A reliability study by Lorimer (2015) found k_{leg}*/dynamic* be the most reliable formulae for calculating vertical and leg stiffness during unilateral hopping (see equations 1 and 2). Joint stiffness was found to be unreliable when calculated individually, however when stiffness values were combined reliability was rated as ‘good’ (see equations 3-6). This improved reliability may reflect interactions between joints during functional tasks (Lorimer, 2014)

(1)	$k_{vert/dynamic} = \frac{\boldsymbol{F_{max}}}{\Delta \boldsymbol{y}}$	F_{max} = peak vertical force, Δy = center of mass displacement from double integration F_{max}
(2)	$k_{leg/dynamic} = \frac{F_{max}}{\Delta L}$	F_{max} = peak vertical force, ΔL = change in leg length, Δy = center of mass displacement from double integration of force
(3)	$k_{joint} = \frac{\Delta M}{\Delta \theta}$	ΔM = change in joint moment, Δθ = change in joint angle
(4)	$k_{sumjoints} = k_{hip} + k_{knee} + k_{ankle}$	
(5)	$k_{hip+knee} = k_{hip} + k_{knee}$	
(6)	$k_{knee+ankle} = k_{knee} + k_{ankle}$	

APPENDIX 2

Participant information sheet

Participant Information Sheet

AUT UNIVERSITY
TE WĀNANGA ARONUI O TAMAKI MAKAU RAU

Date Information Sheet Produced:

15 May 2014

Project Title

The effect of ankle flexibility and ankle bracing on landing-style

An Invitation

My name is Anna Mason-Mackay, I am a Masters student at AUT conducting research involving netball players and I'd like to invite you to participate.

You and your parents should decide together whether or not to participate. You are not in any way obliged to participate and may withdraw at any time up until the end of data collection.

What is the purpose of this research?

The purpose of this study is to investigate how ankle flexibility and ankle bracing affects the way people land and how heavily they land. This will help us to understand how flexibility and braces might affect injury-risk.

This research is a component of my master's degree and the results will be submitted to be published in a scientific journal. It may also be presented at physiotherapy conferences.

How was I identified and why am I being invited to participate in this research?

I contacted the coaches of senior high school netball teams on the North Shore asking for their help to elite find players who do not have any recent injuries and might be interested in participating.

What will happen in this research?

You will be invited to the biomechanics laboratory at AUT and asked to perform four different jumps (described below) with markers placed on your ankles, knees, and hips. Video cameras will record your movement and the markers will help us to understand your particular landing style. For all four jumps you will land on a force plate which will record how heavily you land. You will then repeat the jumps again with braces on both of your ankles. The flexibility of your ankles will also be measured.

Running jump landing on both feet: You will run forward several steps, jump off one foot and land on both feet while catching a pass in the air.

Running jump with a 1-2 foot landing: You will run forward several steps, jump off one foot and land on one foot followed by the other while catching a pass in the air.

Drop-jump: You will jump off a platform landing on both feet and then quickly jump straight upwards as high as you can.

Hopping: You will hop continuously for 30 seconds in time with a beat, first on one foot, then the other.

What are the discomforts and risks?

The jumps will not be unreasonably difficult but there is the potential for injuries such as sprains and strains. Some participants may also be uncomfortable being watched while they perform the jumps.

How will these discomforts and risks be alleviated?

I am a trained physiotherapist and will be present at all research sessions to assess and treat any injuries which might occur. If it you appear likely to injure yourself during the jumps I will stop testing. If you have a current injury or for any other reason are concerned you might injure yourself during the jumps please let me know. If become concerned about injury during testing you may withdraw from the study.

You will be tested on your own, no other participants will be present while you are performing the jumps. However you may bring a support person with you to the session if you would like.

What are the benefits?

At the end of the research you will be given an assessment of your landing technique and an explanation of whether your technique might increase your risk of injury. This research will help netball coaches and players to understand how ankle flexibility and bracing affects injury-risk which will help in injury-prevention. You will also be given a petrol voucher to assist with transport costs to and from the university campus and go into the draw to win an iPod touch and iTunes vouchers.

Your participation in this project will also help me to complete my master's degree.

What compensation is available for injury or negligence?

In the unlikely event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by accident may be available from the Accident Compensation Corporation, providing the incident details satisfy the requirements of the law and the Corporation's regulations.

How will my privacy be protected?

All information collected will be used for research purposes only. Consent forms and contact details will be securely stored on the AUT campus and destroyed after a standard 6 year time period. Data will be completely anonymous to researchers and participants will not be identifiable in any published documents.

What are the costs of participating in this research?

There is no monetary cost but the research will require one hour of your time.

What opportunity do I have to consider this invitation?

You may take one week to consider the invitation before responding. Feel free to contact me during this time with any questions you may have about the project.

How do I agree to participate in this research?

If you would like to participate and are aged 16 years or older all you need to do is sign the attached consent form. If you are under 16 years your legal guardian will need to sign a consent form, while you will sign a slightly different form called an assent form. If you do not have the form you need please call me and I will get one to you, or ask your coach.

Will I receive feedback on the results of this research?

Yes, I will write up a brief summary of the overall results of the research and this will be given to your coach (this summary will not identify individual players). If you would like to receive a summary of your personal results you can indicate this when you complete the consent form and a summary will be given to you directly.

What do I do if I have concerns about this research?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor, Dr. Chris Whatman, chris.whatman@aut.ac.nz**,** 09 921 9999 ext 7037.

Concerns regarding the conduct of the research should be notified to the Executive Secretary of AUTEC, Kate O'Connor, *ethics@aut.ac.nz* , 921 9999 ext 6038.

Whom do I contact for further information about this research?

Researcher Contact Details:

Anna Mason-Mackay

a.mason.mackay@gmail.com

Project Supervisor Contact Details:

Dr. Chris Whatman (primary supervisor)

chris.whatman@aut.ac.nz

09 921 9999 ext 7037

Dr. Duncan Reid (secondary supervisor)

duncan.reid@aut.ac.nz

09 921 9999 ext 7806

Approved by the Auckland University of Technology Ethics Committee on *30 June 2014*, AUTEC Reference number *14/167*

APPENDIX 3

Consent form

Consent Form	AUT UNIVERSITY TE WĀNANGA ARONUI O TAMAKI MAKAU RAU

Project title: *The Effect of Reduced Ankle flexibility and Ankle Bracing on Injury Risk in Secondary School Netball Players*

Project Supervisor: *Chris Whatman (primary), Duncan Reid (secondary)*

Researcher: *Anna Mason-Mackay*

○ I have read and understood the information provided about this research project in the Information Sheet dated 15 April 2014.

○ I have had an opportunity to ask questions and to have them answered.

○ I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way.

o I understand that any information I give during this study will be confidential and my name will not be recorded on any collected data at any time.

○ I am currently injury-free and able to fully participate in netball training and games

○ I agree to take part in this research.

○ I wish to receive a copy of the report from the research (please tick one): Yes○ No○

Participant's signature: ....

Participant's name: ....

Participant's Contact Details:

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Date:

Approved by the Auckland University of Technology Ethics Committee on 30 June 2014 AUTEC Reference number 14/167

APPENDIX 4

Parent/guardian consent form

Parent/Guardian Consent Form

Project title: *The Effect of Reduced Ankle flexibility and Ankle Bracing on Injury Risk in Secondary School Netball Players*

Project Supervisor: *Chris Whatman (primary), Duncan Reid (secondary)*

Researcher: *Anna Mason-Mackay*

○ I have read and understood the information provided about this research project in the Information Sheet dated 15 April 2014

○ I have had an opportunity to ask questions and to have them answered.

○ I understand that I may withdraw my child/children and/or myself or any information that we have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way.

o I understand that any information my child/I give during this study will be confidential and my child's/my name will not be recorded on any collected data at any time.

○ If my child/children and/or I withdraw, I understand that all relevant information will be destroyed.

○ I agree to my child/children taking part in this research.

○ I wish to receive a copy of the report from the research (please tick one): Yes○ No○

Child/children's name/s : ………………………………………………………………………………………

……..

Parent/Guardian's signature: ……………………………………………………………………………………

Parent/Guardian's name: ……………………………………………………………………………………………

Parent/Guardian's Contact Details:

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Date:

Approved by the Auckland University of Technology Ethics Committee on 30 June 2014 AUTEC Reference number 14/167

Note: The Participant should retain a copy of this form.

APPENDIX 5

Assent form

Assent Form

Project title: *The Effect of Reduced Ankle flexibility and Ankle Bracing on Injury Risk in Secondary School Netball Players*

Project Supervisor: *Chris Whatman (primary), Duncan Reid (secondary)*

Researcher: *Anna Mason-Mackay*

○ I have read and understood the sheet telling me what will happen in this study and why it is important.

○ I have been able to ask questions and to have them answered.

○ I understand that while the information is being collected, I can stop being part of this study whenever I want and that it is perfectly ok for me to do this.

O I understand that any information I give during this study will not have my name on it and my name will not be recorded on any collected data at any time.

○ If I stop being part of the study, I understand that all information about me will be destroyed.

○ I agree to take part in this research.

Participant's signature: ....

Participant's name: ....

Participant Contact Details:

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Date:

Approved by the Auckland University of Technology Ethics Committee on 30 June 2014 AUTEC Reference number 14/167

Note: The Participant should retain a copy of this form.

APPENDIX 6

Ethical approval letter

AUTEC
SECRETARIAT

30 June 2014

Chris Whatman
Faculty of Health and Environmental Sciences

Dear Chris

Re Ethics Application: **14/167 The effect of reduced ankle dorsiflexion and ankle bracing on lower extremity mechanics in elite secondary school netball players.**

Thank you for providing evidence as requested, which satisfies the points raised by the Auckland University of Technology Ethics Committee (AUTEC).

Your ethics application has been approved for three years until 30 June 2017.

As part of the ethics approval process, you are required to submit the following to AUTEC:

- A brief annual progress report using form EA2, which is available online through http://www.aut.ac.nz/researchethics. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 30 June 2017;
- A brief report on the status of the project using form EA3, which is available online through http://www.aut.ac.nz/researchethics. This report is to be submitted either when the approval expires on 30 June 2017 or on completion of the project.

It is a condition of approval that AUTEC is notified of any adverse events or if the research does not commence. AUTEC approval needs to be sought for any alteration to the research, including any alteration of or addition to any documents that are provided to participants. You are responsible for ensuring that research undertaken under this approval occurs within the parameters outlined in the approved application.

AUTEC grants ethical approval only. If you require management approval from an institution or organisation for your research, then you will need to obtain this. If your research is undertaken within a jurisdiction outside New Zealand, you will need to make the arrangements necessary to meet the legal and ethical requirements that apply there.

To enable us to provide you with efficient service, please use the application number and study title in all correspondence with us. If you have any enquiries about this application, or anything else, please do contact us at ethics@aut.ac.nz.

All the very best with your research,

Kate O'Connor
Executive Secretary
Auckland University of Technology Ethics Committee

Cc: Anna Mason-Mackay a.mason.mackay@gmail.com