The effects of ankle protectors on lower-limb kinematics of association football players: A comparison to braced and unbraced ankles.

By

Robert William Graydon

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Abstract

Association football (football) players have a high risk of injuring the lower extremities with ankles being one of the predominantly injured sites. To reduce the risk of ankle-contusion injuries ankle protectors made from ethylene-vinyl acetate foam can be used. To reduce the risk of ankle-inversion injuries ankle braces, which contain rigid plastic polymer strips located along the medial and lateral sides of the support, can be worn. However, athletes can only wear one of these devices at a time. Ankle protectors have previously been found to be effective at reducing the risk of contusion injuries by reducing forces being transferred to the ankle. Ankle braces have previously been found to reduce the risk of ankle-inversion injuries by reducing ankle-inversion angle, and ankle-inversion velocity. However, ankle protectors effect on lower-limb kinematics has previously had no attention. As the location of ankle protectors are the same as ankle braces there is a possibility that they reduce the risk of ankle-inversion injuries as well as protect against contusion injuries. Therefore, this thesis aimed to assess the effects of ankle protectors on lower-limb kinematics during sporting movements that commonly occur in football and compare them to braced and unbraced ankles. The four movements selected to be investigated were; running, a countermovement vertical jump (CMVJ), 45˚ cutting manoeuvre, and on the stance limb during kicking a football. Kinetic and kinematic data were collected from male and female participants in three test conditions for each movement; wearing ankle protectors (PROTECTOR), wearing ankle braces (BRACE) and with uncovered ankles (WITHOUT). All kinematic data obtained within this thesis were recorded using an eight-camera Qualysis motion capture system and a single Kistler force plate was used to collect kinetic data. As the main aim of this thesis was to assess the effects of ankle protectors on lower-limb kinematics, and not to investigate gender differences, throughout this thesis the data for each movement is grouped by gender and analysed separately using repeated measures ANOVAs.

The running study found that for males' ankle protectors provided very little restriction to the ankle and did not restrict the ankle like ankle braces. Although no restrictions were seen in the coronal plane there were reductions in sagittal plane motion; ankle protectors significantly ($P \leq 0.01$) reduced angle at toe-off (WITHOUT = -23.65°, PROTECTOR = -21.69°, & BRACE = -21.32°), absolute range of motion (ROM) (WITHOUT = 42.66°, PROTECTOR = 40.15°, & BRACE = 38.34°), and peak plantarflexion velocity (WITHOUT = -665.97°/s, PROTECTOR = -619.33°/s, & BRACE = -595.27°/s) when compared to uncovered ankles. For the females significant ($P \leq 0.01$) restrictions were found in the coronal plane when wearing ankle protectors for relative ROM (WITHOUT = 15.50°, PROTECTOR = 14.37°, & BRACE = 11.13°) and peak inversion velocity (WITHOUT = 159.90°/s, PROTECTOR = 140.67°/s, & BRACE = 122.58°/s) when compared to uncovered ankles. Additionally, there were significant ($P \leq 0.01$) reductions found in sagittal plane motion for the angle at toe-off (WITHOUT = -27.35°, PROTECTOR = -25.07°, & BRACE = -23.86°), absolute ROM (WITHOUT = 44.83°, PROTECTOR = 41.99°, & BRACE = 39.33°), peak dorsiflexion velocity (WITHOUT = 348.29°/s, PROTECTOR = 339.88°/s, & BRACE = 313.78°/s), and peak plantarflexion velocity (WITHOUT = -651.55°/s, PROTECTOR = -593.63°/s, & BRACE = -563.13°/s) when compared to uncovered ankles. The running study concluded that for both males and females' ankle protectors are only effective at reducing the risk of contusion injuries and cannot protect against ankle-inversion injuries. However, the sagittal plane reductions could possibly increase energy demand needed for locomotion and affect performance of other football related movements. The CMVJ study found for males' ankle protectors did not restrict any plane of motion for the ankle, knee, or hip during take-off or landing and did not decrease jump height. For females' ankle protectors were found to significantly ($P \leq 0.01$) reduce jump height (WITHOUT = 0.35m, PROTECTOR = 0.34m, & BRACE = 0.33m) and this reduction was likely due to the significant ($P \leq 0.01$) restrictions found in the sagittal plane for the angle at take-off (WITHOUT = -36.38˚, PROTECTOR = -33.44˚, & BRACE = -31.50˚), absolute ROM (WITHOUT = 63.36˚, PROTECTOR = 59.85˚, & BRACE = 54.99˚), and peak plantarflexion velocity (WITHOUT = -839.34˚/s, PROTECTOR = -794.05˚/s, & BRACE = -733.10˚/s) when compared to uncovered ankles. During landing ankle protectors did not restrict any plane of motion for the ankle, knee, or hip when used by the females. It was concluded that for both males and females during a CMVJ ankle protectors are only effective at reducing the risk of contusion injuries and cannot protect against ankle-inversion injuries during this manoeuvre. The 45˚ cutting manoeuvre study found ankle protectors do not restrict any plane of motion for the ankle, knee, or hip for either the dominant or non-dominant limb for males or females. This study again concluded that ankle protectors are only effective at reducing the risk of contusion injuries and cannot protect against ankle-inversion injuries. The final study investigating the effects on the stance limb during kicking a football again found ankle protectors did not restrict any plane of motion for the ankle, knee, or hip when used by males or females. This study also concluded that ankle protectors are only effective at reducing the risk of contusion injuries and cannot protect against ankle-inversion injuries during this manoeuvre.

Overall the key finding of this thesis is that ankle protectors can only protect against contusion injuries and cannot protect against inversion injuries. Additionally, the current "one size fits all" design should be re-evaluated as it can cause significant alterations to sagittal plane kinematics of the ankle for some footballing related movements. The current design of ankle protectors, could benefit from changes in material construct to either make them better at dissipating forces, by using newer materials, or by the introduction of firmer materials which are integrated into the foam to protect against both contusion and inversion injuries.

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Publications

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Abbreviations

2D – Two dimensional

3D – Three dimensional

ANOVA – Analysis of variance

ASIS – Anterior superior iliac spine

Absolute ROM - Absolute range of motion

BRACE - Wearing ankle braces

BW – Body weights

CAST – Calibrated anatomical system technique

CMVJ – Countermovement vertical jump

COGV – Centre of gravity velocity

COM – Centre of mass

EVA – Ethylene-vinyl acetate

FAM – Functional ankle method

GCS – Global co-ordinate system

GRF – Ground reaction force

Hz – Hertz

ICC – Intra-class correlation

mm – Millimetres

ms – Milliseconds

$m.s^{-1}$ – Metres per second

N – Newtons

$p\eta^2$ - Partial Eta^2

PGM – Plug in gait method

PROTECTOR – Wearing ankle protectors

PSIS - Posterior superior iliac spine

QTM – Qualysis track manager

ROM – Range of motion

Relative ROM – Relative range of motion

s - second

SCS – segment co-ordinate system

TMM – Two marker method

VGRF – Vertical ground reaction force

WITHOUT – With uncovered ankles

˚ - Degrees

˚/s – Degrees per second

% - percentage

1. Introduction

Association football (football) is an immensely popular sport with an estimated 265 million participants' worldwide (FIFA Communications Division, 2007). This popularity has allowed the sport to flourish into an extremely wealthy industry which is valued in England at approximately €2.9billion (Jones et al., 2014). The generation of large revenues by clubs has led to vast amounts of money being spent to secure the services of the best players in the world. These players become valuable assets to the club and as with any business it is of paramount importance to protect these assets. Unfortunately, as with any sport, there is an inherent risk of injury to participants and football is no exception. Figures for injury incidences vary among studies due to differing methodologies, time frames observed, competitions observed, ability, gender, age, and playing position of participants but all conclude there is a high risk of injury during both competitive match play and during training (Arnason, et al., 2004, Falese, et al., 2016, Faude, et al., 2005, Giza, et al., 2005, Jacobson & Tegner, 2006, Peterson, et al., 2000, Van Beijsterveldt, et al., 2015).

1.1 Injury risk in association football

Professional male football players are at risk of between 5.6 and 6.2 injuries per 1000 exposure hours (Peterson, et al., 2000, Van Beijsterveldt, et al., 2015), whilst semi-professionals have a slightly higher risk of 8.9 injuries per 1000 exposure hours (Peterson, et al., 2000). The highest injury risk is to amateur players who have a risk of 9.6-20.2 injuries per 1000 exposure hours (Peterson, et al., 2000, Van Beijsterveldt, et al., 2015). Older players have a higher risk of injury than younger players (Arnason, et al., 2004) with further exploration of this showing that under 25 year olds are at risk of 13.5 injuries per 1000 match hours, 25 to 29 year olds are at risk of 14.3 injuries per 1000 match hours, and over 29 year olds are at risk of 18.9 injuries per 1000 match hours (Falese, et al., 2016). Playing position can also affect injury risk with goalkeepers being at least risk (11.5 per 1000 hours), followed by midfielders (14 per 1000 hours), then defenders (16.1 per 1000 hours), whilst forwards are at the highest risk of injury (17.2 per 1000 hours) (Falese, et al., 2016). Players are most likely to be injured during a competitive match (Professional; 31.8 per 1000, Amateur; 20.4 per 1000, Veteran; 24.7 per 1000) than during training (Professional; 2.1 per 1000, Amateur; 3.9 per 1000, Veteran; 4.5 per 1000) (Hammes, et al., 2015, Van Beijsterveldt, et al., 2015)

Professional female football players have a similar risk of injury to male players with between 1.93 and 9.6 injuries per 1000 exposure hours (Faude, et al., 2005, Giza, et al., 2005, Jacobson & Tegner, 2006), again with the risk being higher during competitive matches (12.6-23.6 per 1000) than when training (1.17 -8.4 per 1000) (Faude, et al., 2005, Giza, et al., 2005, Jacobson & Tegner, 2006, Tegnander, et al., 2008). The risk of injury to semi-professional and amateur female players has had very little attention. A study looking at a mixture of elite and non-elite players, without grouping them separately, found 14.3 injuries per 1000 game hours and 3.7 injuries per 1000 practice hours (Ostenberg & Roos, 2000). Whilst another study on amateur Caribbean women found 30.8 injuries per 1000 hours of match hours (Babwah, 2014). Similar to the males it appears that amateur players are at a higher risk of injury than the professional players. Tegnander, et al. (2008) found midfield players to be at the highest risk of injury (42.4 per 1000 hours), followed by defenders (23.5 per 1000 hours), strikers (22.7 per 1000 hours), wing players (15.2 per 1000 hours), and goalkeepers to be at least risk (12.1 per 1000). Giza, et al. (2005) also found midfield players to sustain more injuries however, Faude, et al. (2006) found defenders to be at the highest risk (9.4 per 1000 hours), followed by strikers (8.4 per 1000 hours), goalkeepers (4.8 per 1000 hours), and midfielders to be the least at risk (4.6 per 1000 hours). These discrepancies might be due to the playing style of the teams observed by the different studies.

Very few studies have compared the injury frequency between elite male and female football players. Two studies that have found that male football players were overall at a higher risk of injury than female players (Larruskain, et al., 2017, Hagglund, et al., 2009). However, the risk of injury to both genders is high and the risk is higher for the less skilled players. Depending on the type and severity of an injury sustained can lead to lengthy periods of time when a player is unable to play. Losing integral team members through injury can lead to a reduced chance of winning competitive matches and furthermore lead to loss of major trophies (Hägglund, et al., 2013). This can eventually equate to loss of earnings for the club, affecting the business and stability of the club. Therefore, an understanding of the common types of injuries sustained by players and also methods to reduce the rate of injury occurrence is a high priority for football clubs.

1.2 Injuries most commonly sustained by football players

Footballing injuries mainly occur to the lower extremities (Peterson, et al., 2000) with the majority of these injuries being muscle strains, ligament sprains, and contusion injuries (Árnason, et al., 1996). The ankle is one of the most predominately injured sites amongst players (Junge & Dvorak, 2013, Peterson, et al., 2000) with ankle-inversion injuries and contusion injuries accounting for a large proportion of the total amount of ankle injuries (Peterson, et al., 2000, Waldén, et al., 2013). During a two season period ankle sprains accounted for a total of 2033 matches being missed by players due to injury (Woods, et al., 2003) with a 25 man professional football team suffering on average seven ankle injuries per season (Waldén, et al., 2013). The average time loss due to ankle ligament sprains is 16±27 days with the most severe sprains leading to 43±33 days lost (Waldén, et al., 2013). According to JLT Specialty (2017), who are a specialist insurance broker and risk consultant, the average cost of ankle injuries for premier league clubs is £253,000 a season. Once a player has suffered an ankle-inversion injury they have an increased risk of reinjuring the ankle (Árnason, et al., 1996, Arnason, et al., 2004, Thacker, et al., 1999). Although playing ability affects the overall risk of being injured it does not affect the incidence of ankle injuries with high and low skilled players having an equal risk (Ekstrand & Tropp, 1990). The majority of studies investigating injury epidemiology frequently focus on a single season or during a single tournament and few have done more longitudinal studies. One study followed 23 elite European teams for seven consecutive seasons and found the same trend as the studies looking at just single seasons, that the vast majority of injuries occur to the lower extremities with muscle strains, ligament sprains and contusions being the most frequent across all seasons (Ekstrand, et al., 2011). Although males overall are at a higher risk of injury (Hagglund, et al., 2009) the overall risk of muscle, joint, or ligament injuries is not significantly different between genders (Larruskain, et al., 2017). The increase risk to males is due to the nearly five times higher incidence of contusion injuries than females (Larruskain, et al., 2017). Due to the ankle being one of the main sites of injury, ways of reducing the risk of injury to this site is of high priority. However, first an understanding of the ankle joint complex and how ankle injuries affect these complexes is needed.

1.3 The anatomy of the ankle

The ankle joint is possibly the most complex joint in the human body as it is made up of multiple facets that allow multidirectional articulations. The ankle joint in its simplest form is made up of the distal ends of the shank, and the foot and its purpose is to allow the foot to accommodate the uneven surfaces it frequently comes into contact with. To do this three differing joints must act together to allow this to occur; the talocalcaneal (subtalar), tibiotalar (talocrural) and transverse-tarsal (talocalcaneonavicular) joints. The following descriptions of the joints are summarised from Brockett & Chapman (2016).

The talocalcaneal joint (subtalar)

The talocalcaneal joint allows mostly inversion and eversion of the ankle and is made up of the talus resting on the anterior part of the calcaneus. The ligaments that make up this joint are the interosseous talocalcaneal ligament, the lateral talocalcaneal ligament, and the anterior talocalcaneal ligament. The joint is also supported by the lateral collateral ligament, and the tibiocalcaneal ligament of the deltoid. Additional support is also provided by the long tendons of peroneus longus, peroneus brevis, flexor hallucis longus, tibialis posterior, and flexor digitorum longus.

The tibiotalar joint (talocrural joint)

The tibiotalar joint allows predominately dorsiflexion and platarflexion motion, although the construct of this joint indicates that it doesn't necessary work solely as a hinge joint. This is due to the design of this joint which is made up of the distal ends of the tibia and fibula forming a mortise in which the talus bone fits within. The joint is stabilised by three groups of ligaments; the first is the tibiofibular syndesmosis which consists of three parts, the anterior tibiofibular ligament, the posterior tibiofibular ligament and the interosseous tibiofibular joint which limit the motion between the tibia and fibula. The second is the medial collateral ligaments (or deltoid ligaments) which are made up of the anterior and posterior tibiotalar ligaments, the tibionavicular ligament and the tibiocalcaneal ligament. The third is the lateral collateral ligaments which are made up of the anterior and posterior talofibular ligaments and the calcaneofibular ligament. Additionally, the inferior tibiofibular joint plays a role in the tibiotalar joint. The joints role is as a stabiliser and not as an articulating joint as it connects the two distal portions of the fibula and tibia.

The transverse tarsal joint

The transverse tarsal joint contributes to inversion and eversion movement of the foot and is considered as part of the same functional unit as the tibiotalar joint. This joint is made up of the anterior part of the talus where it meets with the posterior part of the navicular as well as the calcaneocuboid joint which is the joint between the calcaneus and cuboid.

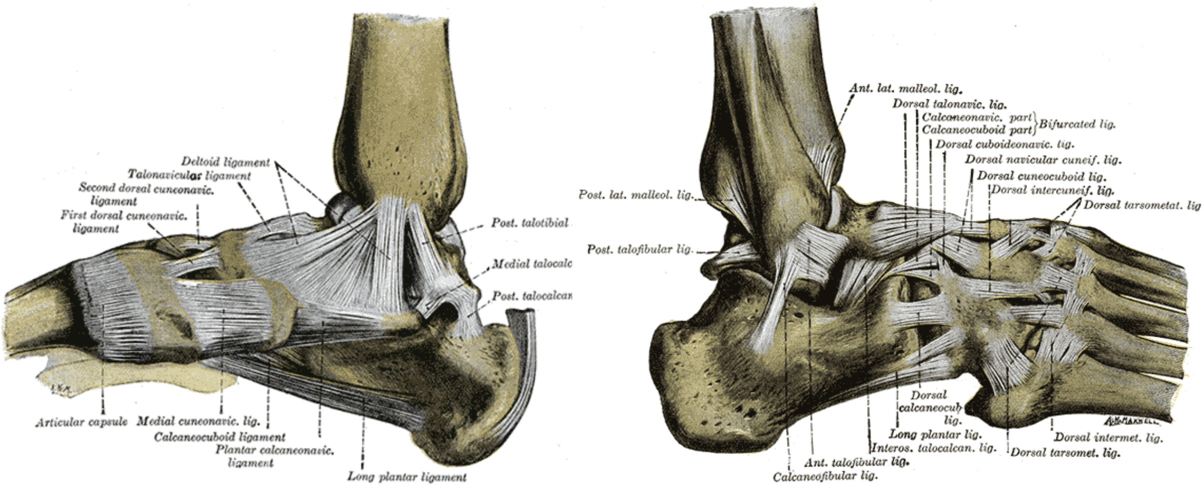

Figure 1.1 Medial ligments of the ankle (left picture) and lateral ligments of the ankle (right picture) picutres obtained from https://commons.wikimedia.org/w/index.php?curid=537826 and https://commons.wikimedia.org/wiki/File%3AGray355.png

1.4 The biomechanics of ankle-contusion and ankle-inversion injuries.

As ankle-contusion and ankle-inversion injuries are two of the most common ankle injuries (Peterson, et al., 2000) it is important to understand the biomechanics of each so that training methods and equipment can be implemented to reduce the risk of each. The cause of ankle injuries in football can be split into two categories; contact and non-contact injuries. Contact injuries involve player to player contact which result in the injury and non-contact injuries usually occur during the interaction between the player and playing surface.

Ankle-contusion injuries are predominantly caused by player to player contact (Árnason, et al., 1996) but can also occur due to contact with the football or goalpost. The impact can cause damage to capillaries causing blood to accumulate under the skin which manifests as a purple/blue discoloration of the skin commonly referred to as a bruise. The severity of these injuries is dependent on the speed of impact, how fast the muscle is compressed, and the size/depth of the area of blood vessels that are affected (Anderson, et al., 2000). Additionally, the state of the muscle during impact plays a role in the risk and severity of contusion injuries with a contracted muscle lessening the risk and severity of a contusion injury (Beiner & Jokl, 2001).

Few studies can offer insight into quantification of the forces that cause contusion injuries. Those currently available have used animal models (Beiner & Jokl, 2001, Crisco, et al., 1994, Stauber, et al., 1990). Crisco, et al. (1996) found that the average pressure to cause contusions when impacting rats leg muscles reached 9000 kPa. Based on this data Ankrah (2002) investigated the tolerable threshold that human participants could withstand when applying force to the ankle and found that the maximum pressure was 419 kPa. From their data, and the data from previous studies on rats, it was extrapolated that approximately 500-1000 kPa may cause discomfort, 1000 - 3000 kPa may cause moderate bruising, and 3000 + kPa could cause severe bruising and or fracture to humans.

Ankle-inversion injuries are more complex and can occur by both contact and non-contact situations. There are two main mechanisms that cause them in contact situations. The first is an impact on the medial aspect of the leg by another player at the moment of foot strike, or just before, causing the ankle to land in an inverted position due to a laterally directed force. The second is by forced plantar flexion where the player kicks an opponent's foot when trying to kick the ball (Andersen, et al., 2004, Árnason, et al., 1996). Video analysis of sporting events has offered insights into the mechanics of non-contact ankle-inversion injuries. Mok, et al. (2011) analysed two ankle-inversion injuries from the 2008 Beijing Olympics, one for a high jumper and one for a hockey player. The study concluded that the injuries were a result of excessive internal rotation and inversion during footstrike with inversion velocity contributing to the mechanics behind the injuries. Another study, this time looking at five inversion injuries recorded during televised tennis competitions, also found that a sudden inversion and internal rotation of the ankle during ground contact appeared to be the cause of the injury (Fong, et al., 2012). Although these studies offer indicators of the mechanics behind non-contact inversion injuries these studies are not specific to football and are using pre-recorded footage that has been recorded for entertainment purposes and not investigatory purpose.

Purposefully causing an ankle-inversion injury in a laboratory setting is unethical however computer simulation models can give an insight into the mechanics behind non-contact inversion injuries. One study, using a computer simulation model, investigated the effects of different foot positions at footstrike on ankle-inversion risk (Wright, et al., 2000). It was found that the talocalcaneal joint angle was not found to influence inversion risk. However, increased plantarflexion did increase risk of inversion injury. This study suggests that reducing plantarflexion at footstrike might reduce ankle injury risk for people with a history of ankle injury (Wright, et al., 2000). Although computer simulations can predict risk factors they are constrained by the computer program that has created them and therefore might not be entirely representative of the multifactorial mechanics behind ankle-inversion injuries. It is important then to understand the mechanics during sporting movements. Chu, et al. (2010) investigated the difference in ankle motion between normal sporting movements and simulated sprains. The study found little difference for inversion angle between the conditions but did find the inversion velocity to be greater in the simulated sprains. The study suggested inversion velocity can be used to differentiate between normal sporting movements and sprains and further suggested that a threshold of 300 °/s can be used to identify ankle sprains.

Fortunately for biomechanists, but unfortunately for the participants involved, there have been cases where during sports specific tasks in a biomechanics laboratory accidental ankle-inversion injuries have occurred (Fong, et al., 2009, Gehring, et al., 2013, Kristianslund, et al., 2011). These have offered unprecedented insight into the mechanics behind non-contact ankle-inversion injuries. One of the first lab based studies that accidently captured an ankle-inversion injury found that at footstrike the ankle exhibited more inversion and internal rotation in the injury trial than the non-injury trials (Fong, et al., 2009). In the injury trial at 0.11s from footstrike there was a marked increase in inversion angle and internal rotation as well as an increase in inversion velocity which was noted as the injury phase. The peak inversion angle in the injury trial reached 48°. Another study by Kristianslund, et al. (2011) found that there was a marked difference between inversion angle (16° in the injury trial and 5° & 6° in the non-injury trials) and internal rotation angle (8° in the injury trial and 4° & -1° in the non-injury trials) during the injury trial when compared to the non-injury trials. Also inversion velocity was a lot higher in the inversion injury trial compared to the two no-injury trials (559°/s in the injury trial and 166°/s & 221°/s in the non-injury trials). One further lab based accidental inversion injury published by Gehring, et al. (2013) again found an increase in inversion angle and internal rotation angle in the injury trial compared to the non-injury trials during the first 60 ms of ground contact. As with the other studies an increased inversion velocity was observed (1290˚/s). One difference between this study and the other studies is that plantarflexion in the injury trial was noted to be higher than the non-injury trials. This might be due to the participant in this study previously sustaining an inversion injury on the same ankle which might have caused changes in their gait which is supported by the computer simulation study by Wright, et al. (2000). For all three of the accidental injuries the inversion velocities were higher than the threshold suggested by Chu, et al. (2010) supporting the use of 300˚/s or below to determine normal sporting movement.

Contact and non-contact inversion injuries most frequently affect the anterior talofibular ligament which has been reported to have been damaged in 82.8% of all ankle sprains (Fallat, et al., 1998). However, more often than not when the anterior talofibular ligament is injured so too is the calcaneofibular ligament, the calcaneofibular ligament is very rarely the only ligament to be damaged (Fallat, et al., 1998). Due to the construct of the ankle, injury to the posterior talofibular ligament will only occur during dislocation of the ankle. One explanation for the anterior talofibular ligament being the most likely to rupture before the calcaneofibular ligament is that it has been found to bear the lowest maximum load before failure (138.9 ± 23.5 N), followed by the posterior talofibular ligament (261.2 ± 32.4 N), and the calcaneofibular ligament (345.7 ± 55.2 N) (Attarian & Devito, 1985).

From the review of literature, it would appear that non-contact ankle-inversion injuries are mainly caused by excessive inversion and rotation of the ankle as well as an increase in ankle-inversion velocity. To a lesser extent excessive plantarflexion also appears to factor into the risk of people with previous inversion injuries risk of being reinjured. There appears to be a high risk of damage to the anterior talofibular ligament during both contact and non-contact ankle-inversion injuries. Therefore, methods to reduce these motions of the ankle might prove beneficial for reducing the risk of injuring this ligament whilst playing football. Furthermore, anklecontusion injuries can cause bruising or swelling to the area of impact and it might be beneficial to reduce force being transferred to the ankle to reduce the risk of these injuries.

1.5 Methods of reducing ankle injury and the impact on injury risk

To reduce the risk of contusion injuries ankle protectors can be worn. To date, only two studies have investigated the effectiveness of them on reducing force and pressure transference to the body (Ankrah & Mills, 2002, Ankrah & Mills, 2004). The first found that the domed protective shells, which covers the malleoli, can effectively transfer loads to the underlying foam constructs which in turn can spread the load over a larger surface area of the ankle. Due to no specific data quantifying the amount of force required to cause bruising of the underlying structure to compare to it was unable to confirm if the loads were diminished enough to reduce the risk of a contusion injury (Ankrah & Mills, 2002). The second found that the current designs of ankle protectors made from ethylene vinyl acetate and low-density polyethylene shells were effective at protecting against low impact kicks but greater impacts, especially with a football stud contact, could not prevent bruising of the ankle (Ankrah & Mills, 2004). The study suggested the current design could be improved by introducing thicker foams of higher modulus and domed shells of higher stiffness.

To reduce the risk of ankle-inversion injuries ankle braces can be worn, the ankles can be taped, or a neuromuscular training program can be utilised. Using tape to support the ankle has been found to be ineffective after approximately fifteen minutes of use (Lohkamp et al 2009) and expensive (Olmsted et al 2004) whereas neuromuscular training programs have been found to be effective but take long periods of time to implement (Emery & Meeuwisse 2010). This makes ankle braces an attractive alternative because they are easy to put on, don't need to be regularly replaced, and have been found to reduce the risk of ankle-inversion injury (Pedowitz et al 2008). Additionally they have been found to be superior than neuromuscular facilitation training at reducing risk of re-injury of a previous ankle-inversion injury (Janssen, et al., 2014).

Studies looking at the effectiveness of ankle braces on ankle-inversion risk in football have found a reduction in the frequency of ankle injuries when braces are used by male players who have previously injured the ankle (Tropp, et al., 1985). Another study, again using male football players, found that using a semi-rigid orthosis significantly reduced the risk of ankle injury in players that have previously injured the ankle (Surve, et al., 1994). These findings have also been replicated when using female football players with a previous ankle injury (Sharpe, et al., 1997). It was found that using an ankle brace significantly reduced the risk of reinjuring the ankle when compared to both taping and no preventative method. These results have also been replicated in other sports. McGuine, et al. (2011) found the use of a lace-up ankle brace by both male and female basketball players reduced the risk of injury in athletes with and without a previous history of an ankle injury. In the sport of American football, McGuine, et al. (2012) found that ankle braces reduced the risk of ankle injuries and also their use did not increase the likelihood of sustaining a knee injury. Unfortunately, due to the application of both ankle braces and ankle protectors, only one of these devices can be used at any one time. This selection is usually dependent on whether the wearer wants to reduce acute or chronic ankle injuries.

1.6 Effects of ankle braces and ankle protectors on lower-limb kinematics.

Currently no research has been conducted on the effects of ankle protectors on ankle kinematics, however the effects of ankle braces has been extensively researched. Using cadavers Omori, et al. (2004) investigated the effect of an ankle brace on the motion of the tibiotalar joint after severing the anterior talofibular and calcaneofibular ligaments. Upon severing these ligaments inversion and internal rotation significantly increased. However, after applying an ankle brace the inversion displacement resembled its pre-severed state but the internal rotation displacement did not. The study concluded that ankle braces primarily reduce inversion of the ankle. However, ankle-inversion injuries involve inversion, plantarflexion, and internal rotation and therefore do not restrict all the necessary planes of motion.

Studies into the effects of ankle braces on athletes frequently use ankle-inversion tables, which are devices that simulate sudden inversion by one side of the device dropping away, to test the effects of ankle braces on ankle movement during these situations. Vaes, et al. (1998) investigated the effects of ankle braces on talar tilt when used by individuals with functional ankle instability. The study found significant reductions in speed of talar tilt when using an ankle-inversion table and suggested that a slower inversion velocity is advantageous for the wearer as it allows more time for muscular activation to prevent the ankle injury. A more recent study by Tang, et al. (2010) investigated the effects of a semi-rigid ankle brace on sudden ankle-inversion when worn by healthy males. This study used a three dimensional motion capture system to track the motion of the ankle and found the semi-rigid ankle brace significantly reduced ankle angular displacement and angular velocity. Another study by Podzielny & Henning (1997) found similar results when investigating the effects of four different ankle braces on inversion velocity using an inversion platform. Interestingly this study found three of the ankle braces significantly reduced inversion angle and inversion velocity whereas one did not. The brace that showed no significant difference was an elasticated support whereas the other three braces were more rigid. This finding suggests that only the more rigid styles of ankle brace can protect against inversion injuries. Eils, et al. (2002) further explored this by comparing the effects of ten different ankle braces on passive ankle range of motion and during simulated ankle-inversion using an inversion platform. The study also compared soft braces to semi-rigid braces to assess if there was a difference between the style of brace. It found all the braces significantly reduced ankle range of motion in both the passive and simulated inversion conditions when compared to no brace. Furthermore, it was found that the semi-rigid braces provided significantly more reductions in inversion than the soft braces. These findings suggest that semi-rigid ankle braces are more efficient at reducing ankle-inversion velocity. This is supported by Cordova, et al. (2000) who conducted a meta-analysis on 19 publications to compare the effects of ankle taping, lace-up ankle braces, and semi-rigid ankle braces on ankle kinematics before and after exercise in healthy individuals. The analysis found semi-rigid braces provided the greatest restrictions for inversion and eversion both pre-exercise and post-exercise. It also found the semi-rigid braces did not affect dorsiflexion or plantarflexion range of motion.

The findings from the above studies suggest that ankle braces are effective at reducing inversion injuries by reducing peak inversion, and peak inversion velocity. Although this section has covered the effects of ankle braces on ankle kinematics using inversion tables, the effects of ankle braces on specific sporting motions will be covered in the relevant chapter introductions later in the thesis.

1.7 Rationale

Ankle injuries are common in football (Peterson, et al., 2000) and once an ankle injury is sustained there is a high probability of re-injury at the same place (Arnason, et al., 2004). Ankle braces have been found to be effective at reducing the risk of ankle-inversion injuries for both male (Surve, et al., 1994) and female (Sharpe, et al., 1997) football players. Ankle protectors have been found to be effective at reducing the risk of contusion injuries (Ankrah & Mills, 2004). Unfortunately, due to ankle braces and ankle protectors aiming to reduce differing injuries at the same location only one of these devices can be used at any one time. This selection is dependent on whether the wearer wants to reduce the risk of acute or chronic ankle injuries. Ankle-inversion injuries are caused by excessive ankle-inversion, rotation, inversion velocity, and to a lesser extent excessive plantarflexion (Fong, et al., 2012, Gehring, et al., 2013, Kristianslund, et al., 2011). Ankle braces have been found to reduce peak inversion, and peak inversion velocity which are believed to be the main mechanics behind ankle braces success at reducing the risk of ankle-inversion injuries (Tang, et al., 2010, Vaes, et al., 1998). However, the effects of ankle protectors on ankle kinematics has, to the author's best knowledge, had little to no attention. As the location of ankle protectors are the same as ankle braces there is a possibility that they inadvertently act like ankle braces during sporting tasks by reducing the amount of inversion and inversion velocity of the ankle. Therefore, the overall aim of the thesis is to assess the effects of ankle protectors on lower-limb kinematics during sporting movements that commonly occur in football and compare them to braced and unbraced ankles.

1.8 Aims and Objectives

The main aim of the thesis is to assess the effects of ankle protectors on lower-limb kinematics during sporting movements that commonly occur in football and compare them to braced and unbraced ankles. To accomplish the aim of the thesis the following objectives will be undertaken:

I. Establish the best method to determine the ankle joint centre using the Visual 3D software.

II. Investigate the effects of ankle protectors on lower-limb kinematics of male and female football players during running.

III. Investigate the effects of ankle protectors on lower-limb kinematics of male and female football players during the take-off and landing phase of a countermovement vertical jump.

IV. Investigate the effects of ankle protectors on lower-limb kinematics of both the dominant and non-dominant limb of male and female football players during a 45˚ cutting manoeuvre.

V. Investigate the effects of ankle protectors on the kinematics of the stance limb of male and female football players during kicking a football.

2. Methodology

2.1 Introduction

Throughout this thesis a combination of both kinematic and kinetic data acquisition methods were used to investigate the aims and objectives outlined in section 1.8. However, before any data acquisition was conducted the reliability and accuracy of the data collection systems and methodologies of processing were quantified and justified. This section will cover the general methods utilised and any development of methods undertaken during the thesis. However, each main study contained within the thesis will define the protocols, data processing, and analysis specific to that study.

2.1.1 Kinematic data acquisition

Developing an understanding of human movement is the basis of kinematic data acquisition. By developing this understanding practitioners can better understand pathologies which cause injury and develop ways of reducing the risk. To be able to analyse the movement patterns of the body, ways of recording the movements are necessary. Kinematic data can be collected in one of two ways. The first is using a two dimensional (2D) camera configuration system and the second is to use a three dimensional (3D) camera configuration system. 2D camera systems are the least expensive of the two set ups but are restricted by the ability to only record data in one plane of motion (Richards, et al., 2008). Whereas 3D motion capture systems usually cost considerably more than the 2D set ups but are advantageous in the ability to collect multi-planar data, and are not affected by perspective or cross-planar errors (Richards, et al., 2008). This makes 3D configurations the more desirable for the investigation of human movement.

All kinematic data obtained within this thesis were recorded using an eight-camera Qualysis motion capture system (Qualisys Medical AB, Goteburg, Sweden). This system tracks retro reflective markers, in view of the cameras using passive infrared technology. All eight cameras were the Oqus 310 series model and the capture frequency of the cameras throughout the thesis was set at 250Hz. This frame rate was selected for two reasons; the first was based on Nyquist-Shannon sampling rate criterion which states a sampling rate should be at least twice that of the maximum motion frequency (Nyquist, 1928, Shannon, 1949) and the second was that previous research using the same camera set up have used the same capture frequency and found it sufficient to record these motions (Sinclair, et al., 2015, Vanrenterghem, et al., 2012).

2.1.2 Kinetic data acquisition

Kinematic data on its own only tells part of the story of what is occurring during human movement. Another part is made up of the forces that initiate movement and exploration of these can give insights as to the effects of these forces on the body.

Throughout this thesis, a single Kistler force plate (Kistler Instruments Ltd., Alton, Hampshire) sampling at 1000 Hz was used to record ground reaction forces. This sampling frequency was selected for two reasons; the first was that a lower boundary of 200 Hz has been established as a point at which error attributed to sampling rate starts to increase in magnitude once lower than this threshold (Hori, et al., 2009) and the second was that a sampling rate of 1000 Hz is frequently used in biomechanical research (Cordova, et al., 2010, Vanwanseele, et al., 2014, West & Campbell, 2014). The force plate measured 600mm length by 400mm width and was embedded in the floor in the centre of the view of all eight of the Qualysis cameras as shown in figure 2.1.

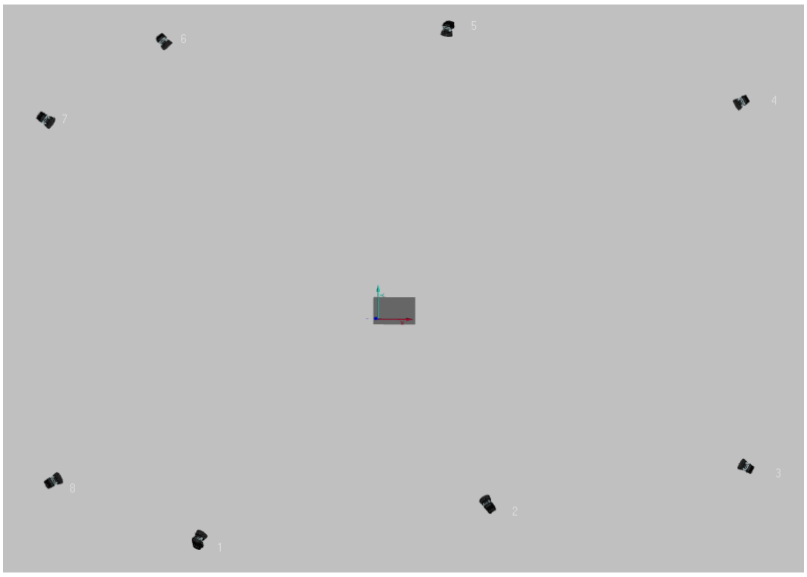

Figure 2.1 Camera positioning around the force plate using the eight camera set-up.

2.1.3 Data collection, format, and analysis.

The software used to synchronise and collect the 3D kinematic and ground reaction data was Qualysis track manager (QTM) which records files in a .QTM file format. Within the QTM software each marker in view of the camera was manually identified for each recorded trial before the data were saved and exported as a .C3D file. To analyse the data the .C3D files were imported into Visual 3D software (C-Motion, Germantown, MD, USA) so that joint angles, joint velocities, and ground reaction data could be calculated. These calculated metrics were then exported to Microsoft excel (Microsoft Corp., Redmond, WA, USA) to allow data to be formatted before being inputted into SPSS (SPSS Inc., Chicago, IL, USA) for statistical analysis. Each step in this process will be further explained in subsequent sections found within this chapter.

2.2 Ankle braces and ankle protectors

Throughout the thesis the ankle protectors used were a pair of Nike ankle shield 10 (Nike Inc, Washington County, Oregon, USA) and the ankle braces used were a pair of Aircast A60 (DJO, Vista, CA, USA).

Figure 2.2. On the left a pair of Nike ankle shield 10 ankle protectors and on the right a pair of Aircast A60 ankle braces.

The ankle protectors come in a ‘one size fits all’ design and were selected based on them being the most popular ankle protector available on Amazon at the time of commencing the thesis. The ankle protectors contain an ethylene-vinyl acetate (EVA) foam construct that starts on the medial side of the protector, runs around the posterior side, and finishes on the lateral side of the protector. The ankle protectors are designed so that there is a left and right protector and the foam construct is thicker around the lateral malleolus location to provide additional protection in this location.

The ankle brace is a semi-rigid ankle brace, which consists of two rigid plastic polymer strips located along the medial and lateral sides of the support. The ankle brace also has a support strap that starts low on the lateral side of the brace and spirals up and around the support into a locking mechanism attached to the medial side of the ankle brace which allows the support to be tightened. Figure 2.3 shows the application of the ankle brace. The ankle braces were purchased in three sizes; small, medium, and large for both the right leg and left leg as the design of the brace is leg specific. The small size is designed to fit UK shoe size up to a 6.5, the medium UK shoe sizes 7-11, and the large UK shoe sizes 11.5+. The Aircast A60s were selected due to previous research establishing semi-rigid ankle braces being far superior at reducing inversion angle than other types of ankle brace (Eils, et al., 2002).

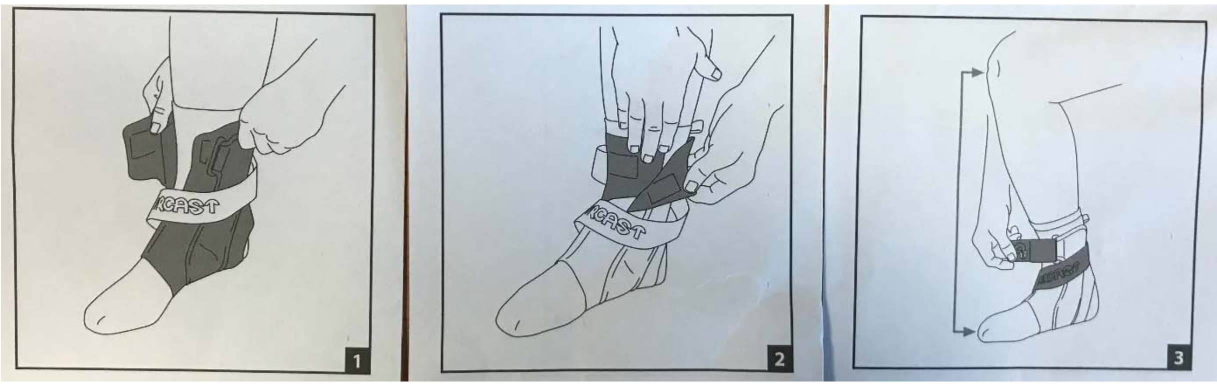

Figure 2.3. Photos of the application process for the Aircast A60 ankle brace. These pictures are contained within the instruction manual that accompanies the ankle brace when purchased.

2.3 Pilot study 1: Optimal Calibration of the Qualysis system

2.3.1 Introduction

When utilising a 3D camera system to record data, which will be investigated for small changes in the orientation of markers, it is important to establish an accurate calibration method to be able to collect reliable and accurate 3D kinematic data. To do this the camera system being used must have the global coordinate system (GCS) defined to allow it to accurately position objects in space and time. This is achieved by positioning a static L-Frame in view of the cameras. Additionally, a T shaped wand of a known length is moved through an area in view of the cameras to define a volume of area in which data will be collected. This study set out to compare differing calibration techniques to produce a standardised method to calibrate the system which would be used throughout the thesis.

2.3.2 Methodology

2.3.2.1 Equipment

Eight Oqus 310 cameras sampling at 250 Hz were used to collect the data. One T shaped wand and one L-frame both precision manufactured to size were also used (Figure 2.4).

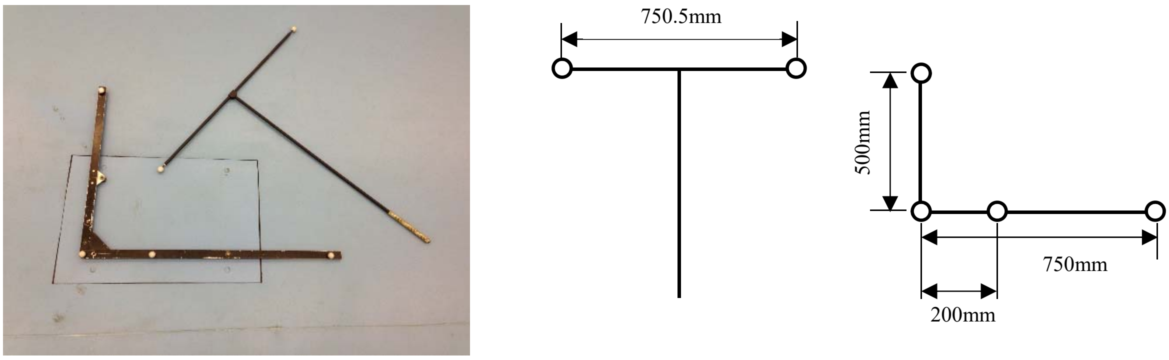

Figure 2.4. T shaped wand and L-frame (Left) and dimensions of the T shaped wand and L-frame (Right)

2.3.2.2 Procedure

The first three calibrations were collected using a stationary position (static) close to the Kistler force plate located at the centre of the camera system. The three movements used to calibrate from the stationary position were linear motions (forwards/backwards, side to side, up and down), spinning motions (Twirling the calibration wand) and a combination of both linear and spinning motions. The second three calibrations were collected whilst moving around the Kistler force plate (dynamic). Again the three movements used were linear motions, spinning motions and a combination of both motions. Each of these six calibrations were collected using 30 seconds of data capture.

Using the dynamic combination technique, the effects of differing calibration lengths were assessed. The time lengths used were; 15 seconds, 30 seconds, 45 seconds, and 60 seconds.

2.3.3 Results

Table 2.1. The average data points, average residuals and standard deviation of wand length obtained for each calibration method.

Calibration Method	Average Points	Average Residuals (mm)	Standard Deviation of wand length (mm)
Static Linear	4689.75	0.44	1.17
Static Spin	4529.25	0.52	1.31
Static Combination	4714.00	0.57	1.50
Dynamic Linear	5008.50	0.35	0.25
Dynamic Spin	4691.38	0.35	0.26
Dynamic Combination	4758.63	0.37	0.25

Table 2.2 The average data points, average residuals and standard deviation of wand length obtained for varied lengths of calibration time using the dynamic combination technique.

Calibration Method	Average Points	Average Residuals (mm)	Standard Deviation of wand length (mm)
15 Seconds	2384.50	0.36	0.26
30 Seconds	4758.63	0.37	0.25
45 Seconds	7208.88	0.37	0.26
60 Seconds	8011.50	0.36	0.25

2.3.4 Conclusions

The results show that the dynamic calibrations produced lower average residuals than the static calibrations. The dynamic combination calibration produced the lowest standard deviation of wand length and has been selected as the method to be used to calibrate the Oqus 310 camera system during testing throughout the thesis. A calibration time of 30 seconds has been selected as all of the calibration lengths produced similar average residuals and standard deviation of wand lengths therefore it was selected based on it being the first condition where the average points recorded were over 4000.

2.4 Pilot study 2: Accuracy of the Qualysis system

2.4.1 Introduction

To ensure the kinematic data obtained using a 3D motion capture system is accurate the accuracy of the system itself needs to be assessed. Only by first establishing how accurate the system is can it be ensured that the results of any subsequent studies using the system are reliable. Therefore, this study set out to assess how accurate the Qualysis motion capture system being used is.

2.4.2 Methodology

2.4.2.1 Equipment

Eight Oqus 310 cameras sampling at 250 Hz were used to collect the data. One T shaped wand precision manufactured to size was also used.

2.4.2.2 Procedure

The camera system was calibrated using the criteria outlined in section 2.3. For the purpose of this study the T shaped wand, which is used to calibrate the system, was also used as the reference frame for this study. The distance between the two markers at either side of the head of the wand are 750.5mm apart. The wand was moved up and down five times, side to side 5 times and in a figure of eight five times in three different locations; in the centre of the calibrated area, away from the centre of the calibrated area in the positive X direction with reference to the global coordinate system, and away from the centre of the calibrated area in the positive Y direction with reference to the global coordinate system.

2.4.3 Results

Table 2.3. Mean distance between the two markers attached to the T shaped wand in three different locations with mean error, standard deviation, and max error.

	Centre	Positive X	Positive Y
Mean (mm)	750.49	750.72	750.60
Mean Error (mm)	0.06	0.22	0.13
Std. Dev	0.17	0.37	0.28
Max Error (mm)	1.24	2.62	1.60

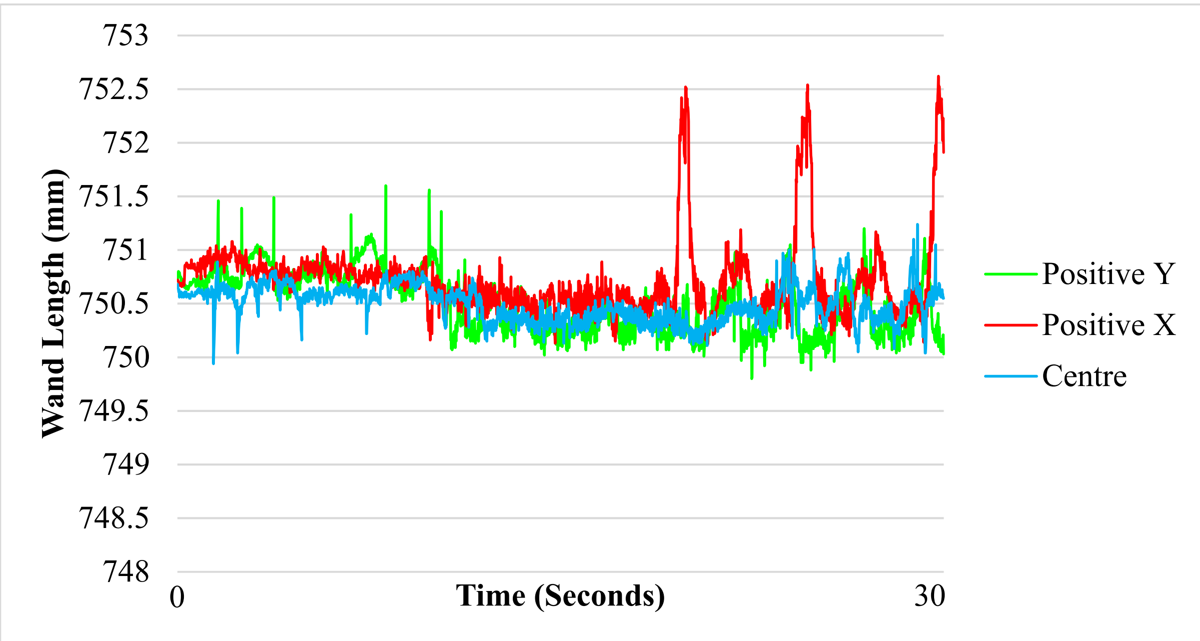

Figure 2.5. Fluctuation of the distance between the two markers attached to the T shaped wand for the three differing positions.

2.4.4 Conclusion

The results of this investigation suggest that measurements taken in the centre of the calibrated area are more accurate than measurements taken in either of the Positive X or Positive Y directions from the centre. The mean error in the centre of the calibrated area is 0.06 mm with a maximum error of 1.24 mm demonstrating that the Qualisys camera system being used throughout this thesis is highly accurate and thus allows meaningful conclusions to be drawn from data obtained.

2.5 Modelling and tracking of the lower limbs

To track the motion of the participants using the 3D software markers made from 20mm wooden balls covered in 3M retroreflective tape were used. These took two forms, the first was a singular ball marker, referred to as an anatomical marker, and the second was a group of four ball markers attached to a rigid plastic mount, referred to as a tracking cluster, which were affixed to specific locations on the participants' bodies.

Figure 2.6. Anatomical marker (left picture) and tracking cluster (right picture) which were attached to the body of the participants.

Throughout this thesis, the calibrated anatomical system technique (CAST) was utilised to model the anatomy of the lower limbs for data acquisition (Cappozzo, et al., 1995). The CAST system uses specific anatomical landmarks and tracking markers to model each body segment in 6 degrees of freedom. The anatomical markers are used to define the proximal and distal portions of the segment they are attached to as well as define the medial and lateral aspects of the segment. These positions are then associated to the positions of the tracking markers to allow tracking of motion in view of the 3D camera system. Each segment used in this thesis and how it was tracked in the 3D software is shown in figure 2.7 and defined on the next page.

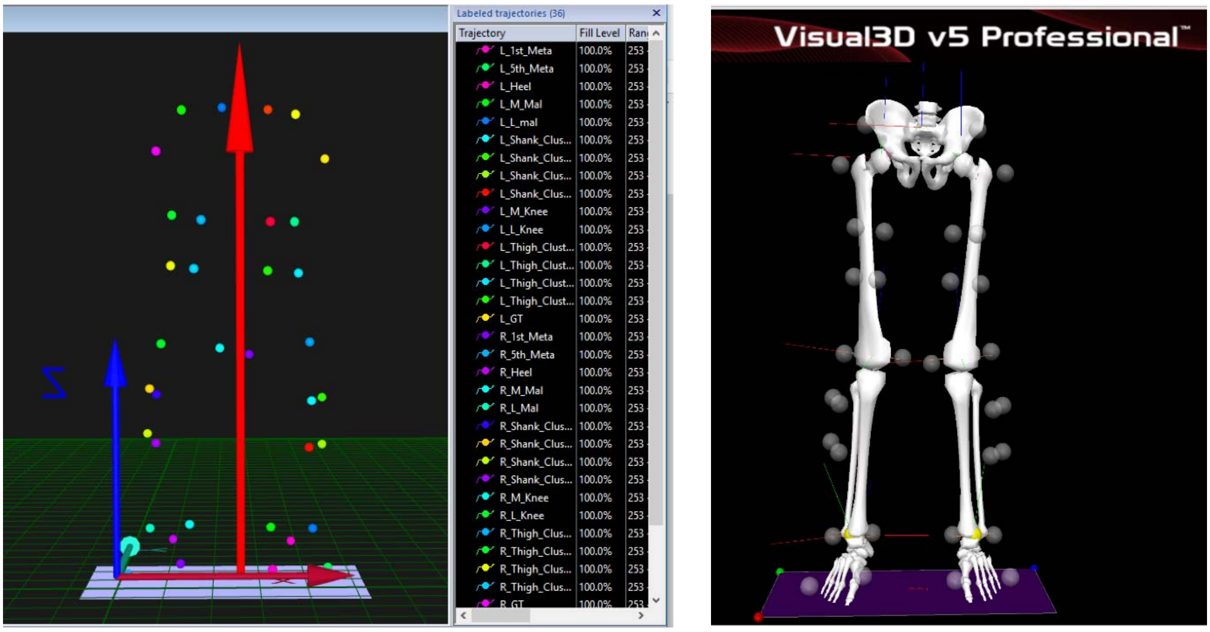

Figure 2.7. Anatomical and tracking markers attached to a participant's lower body segments shown in the QTM software (left picture) and once built into a skeletal model in the Visual 3D software (right picture).

2.5.1 Foot segment

To define the foot segment anatomical landmarks were placed on the 1st metatarsal, 5th metatarsal, medial malleolus, lateral malleolus, and calcaneus. The metatarsals were used to define the distal end of the segment whilst the malleoli were used to define the proximal end. To track the foot segment the 1st and 5th metatarsal heads and the calcaneus were used. Additionally, it must be noted that throughout this thesis the foot was considered to be a rigid segment.

2.5.2 Shank segment

To define the shank segment anatomical landmarks attached to the medial and lateral malleoli were used to define the distal end whilst anatomical landmarks placed on the medial and lateral femoral epicondyles were used to define the proximal end. Rigid plastic mounts with four markers on each were also attached to the midpoint between the proximal and distal ends of the shank and were secured using elasticated bandage. These were used as tracking markers for the shank segment.

2.5.3 Thigh segment

To define the thigh segment the anatomical landmarks attached to the medial and lateral femoral epicondyles were used to define the distal end whilst the proximal end was defined using the hip joint centre. The method of establishing the hip joint centre is contained within section 2.6. Rigid plastic mounts with four markers on each were also attached to the midpoint between the proximal and distal ends of the thigh and were secured using elasticated bandage. These were used as tracking markers for the thigh segment.

2.5.4 *Pelvis segment*

To define the pelvis segment anatomical markers were attached to the left and right anterior superior iliac spines (ASIS) and the left and right posterior superior iliac spines (PSIS). The segment was then constructed using the CODA option in the visual 3D software. To track the hip segment the right/left ASIS and right/left PSIS were used.

2.6 Joint centre location techniques

Accurate joint centre identification is imperative for the collection of reliable and accurate lower-limb kinetic and kinematic data for gait analysis (Kirkwood, et al., 1999, Piazza, et al., 2001). One of the key sources of measurement ambiguity in 3D kinematic analyses using surface marker placement is the definition of the joint centre about which segmental rotations are considered to occur. Methods to accurately identify the hip joint centre have been extensively researched (Bell, et al., 1989, Besier, et al., 2003, Sinclair, et al., 2014, Siston & Delp, 2006) and to a lesser extent methods of accurately identifying the knee joint centre have been researched (Davis, et al., 1991, Sinclair, et al., 2015, Thewlis, et al., 2008). The most reliable method of identifying the hip joint centre has been established to be the anatomical method (Sinclair, et al., 2014). This method is based on work conducted by Bell, et al., (1989) and Bell, et al., (1990) which constructs the joint centre using the following regression equation:

Hip joint centre = (0.36 x Distance between ASIS markers medial to the ASIS marker, 0.19 x Distance between ASIS markers posterior to the ASIS marker, 0.3 x Distance between ASIS markers inferior to the ASIS marker)

This method has been found to be superior for its test-retest reliability when compared to functional and projection methods (Sinclair, et al., 2014). The most reliable method for establishing the knee joint centre is the two marker method with the anatomical markers either affixed to the medial/lateral femoral epicondyles or medial/lateral femoral condyles (Sinclair, et al., 2015). This method locates the joint centre using the midpoint between the two affixed markers and has been found to be superior for its test-retest reliability when compared to functional and plug-in-gait methods (Sinclair, et al., 2015). Therefore, throughout this thesis when defining hip and knee joint centres within the visual 3D software the hip joint centre will be located using the anatomical method and the knee joint will be defined using the two marker method with the markers being located on the medial/lateral femoral epicondyles. Although the accuracy of both the hip joint centre and knee joint centre location techniques have been investigated the best method for ankle joint centre identification technique has received very little attention. Therefore, a pilot study to identify the best ankle joint centre technique was undertaken.

2.6.1 Pilot study 3: The test-retest reliability of different ankle joint centre location techniques

2.6.1.1 Introduction

There are currently three main methods of identifying the ankle joint centre when using Visual 3D software. These methods are the two-marker-model (TMM), plug-in-gait model (PGM) and functional ankle model (FAM). Each method relies on differing methodology to define the ankle joint centre. TMM uses the markers on the medial and lateral malleoli to define the joint centre (Nair, et al., 2010), PGM uses several markers to identify the joint centre by first identifying the hip joint centre followed by the knee joint centre and finally identifies the ankle joint centre based on the locations of the previous two joint centres (Vicon®, 2002) and FAM uses the rotation of the foot relative to the shank to estimate the ankle joint centre (Schwartz &

Rozumalski, 2005). For the purpose of this study only TMM and FAM will be compared for test-retest reliability because PGM has previously been found to be more likely to produce errors in defining the ankle joint centre when compared to TMM (Nair, et al., 2010). The TMM relies on accurate identification of the malleoli to create the ankle joint centre; however, when the ankle is covered by a brace or protector the identification of these bony protrudes might be impaired. Therefore, it is important to establish the test-retest reliability of the TMM method when the ankle is covered by a brace or protector as well as when the ankle is not covered. The aim of the current pilot study is to assess which method, TMM or FAM, is the most reliable method to define the ankle joint centre. Kinematic data will be compared using Intra-class correlation analyses to identify which method is the most reliable.

2.6.1.2 Methodology

2.6.1.2.1 Participants

Ten participants (8 females and 2 males) all with size six feet took part in the current investigation (aged: 24 ± 2.63 years, height: 166.73 ± 3.24 cm, body mass: 62.54±6.56 kg, and BMI: 22.48 ± 2.14). All were free from injury at the time of data collection and provided written consent.

2.6.1.2.2 Procedure

Participants completed five walking trials striking an embedded force platform (Kistler Instruments Ltd., Alton, Hampshire) which sampled at 1000 Hz. The start of the stance phase during the walking trials was determined as the point at which the force plate first recorded a vertical ground reaction force that exceeded 20N (Sinclair, et al., 2011). Kinematic and ground reaction force data were obtained during the right leg stance phase. Kinematic data were recorded using an eight camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) tracking retro-reflective markers at a sampling rate of 250 Hz. Using the calibrated anatomical system technique (CAST) (Cappozzo, et al., 1995) the retro-reflective markers were attached to the locations outlined in section 2.5.

Before dynamic trials were captured a static trial of the participant stood in the anatomical position was captured in three conditions; wearing ankle braces (BRACE), wearing ankle protectors (PROTECTOR) and without a brace or protector (WITHOUT). The static trial was used to define the ankle joint using the TMM using the medial and lateral malleoli markers. Also a FAM was delineated without any brace or protector (FUNCTIONAL). The FAM trial involved the participant standing on their left leg, raising their right leg in the air and dorsiflexing followed by plantarflexing the foot five times. The dorsi-plantarflexion range of motion was typically around 60°. The ankle joint centre was taken as the stationary point relative to the shank and foot segments (Schwartz & Rozumalski, 2005). Once the dynamic trials were captured, the medial and lateral malleoli markers were removed and then reapplied and a static trial of each test condition was again recorded.

2.6.1.2.3 Ankle brace and protector

The ankle protectors and ankle braces used for the pilot study were the ones outlined in section 2.2.

2.6.1.2.4 Data Processing

Anatomical and tracking landmarks were identified within the Qualisys Track Manager software and then exported as C3D files to be analysed using Visual 3D (C-Motion, Germantown, MD, USA) software. The walking trials were filtered at 6 Hz using a low pass 4th order zero-lag filter Butterworth filter (Winter, 1990). Two methods of defining the ankle joint centre were utilised and applied to the walking trials; the first used the medial and lateral malleoli markers to define ankle joint centre and the second used the functional movement dynamic trial to calculate the ankle joint centre. Data were normalized to 100% of the stance phase then processed gait trials were averaged. 3D kinematics of the ankle joint were calculated using an XYZ cardan sequence of rotations. 3D ankle joint kinematic measures which were extracted for further analysis were 1) angle at footstrike, 2) angle at toe-off, 3) angular range of motion (ROM) from footstrike to toe-off during stance, 4) peak angle during the stance phase.

2.6.1.2.5 Statistical analyses

To compare pre-post differences paired samples t-tests were employed. Significance was accepted at the $p \leq 0.05$ level. Intra-class correlations (ICC) were used to compare test and retest sagittal, coronal and transverse plane waveforms of the ankle for each ankle joint centre location technique. Statistical analyses were conducted using SPSS 21.0 (SPSS Inc, Chicago, USA).

2.6.1.3 Results

The results indicate that the test and retest 3D kinematic waveforms measured as a function of each ankle joint centre configuration were qualitatively similar and quantitatively showed a high level of similarity (ICC ≥0.779). It should be noted however that some statistically significant differences in discrete kinematic parameters were observed. Table 2.4 shows the similarity between test and retest waveforms for each ankle configuration and tables 2.5-2.8 and figures 2.8-2.11 present the discrete ankle joint kinematics and 3D waveforms for each configuration.

Table 2.4. Intraclass correlations for 3D joint waveforms.

	ICC test/ retest
Brace	
X	0.984
Y	0.997
Z	0.779
Without	
X	1.000
Y	0.997
Z	0.818
Protector	
X	0.999
Y	0.994
Z	0.995
Functional	
X	0.985
Y	0.987
Z	0.806

Notes: X = Sagittal, Y = Coronal and Z = Transverse plane.

Table 2.5. Kinematic data (means and stand deviations) for the ankle obtained during stance phase of the walking gait without a brace or protector using the two marker method.

WITHOUT	**Test**		**Retest**		
	Mean	*SD*	*Mean*	*SD*	
Sagittal plane (+ = dorsiflexion/ - = plantarflexion)					
Angle at footstrike (˚)	66.74	5.08	66.62	3.84	
Angle at toe-off (˚)	45.20	3.85	44.12	3.16	*
Peak dorsiflexion (˚)	72.94	2.97	71.78	3.13	
ROM (˚)	21.54	3.60	22.50	3.55	
Coronal plane (+ = inversion/ - =eversion)					
Angle at footstrike (˚)	-2.13	6.39	-1.63	5.41	
Angle at toe-off (˚)	4.64	6.06	5.47	4.59	
Peak eversion (˚)	-9.37	6.64	-8.35	4.72	
ROM (˚)	7.39	3.75	7.97	3.29	
Transverse plane (+ = external/ - =internal)					
Angle at footstrike (˚)	-8.37	3.60	-9.69	3.86	
Angle at toe-off (˚)	0.98	4.93	-3.19	4.99	*
Peak external rotation (˚)	-0.82	4.30	-1.88	3.92	
ROM (˚)	9.35	5.08	6.50	3.33	

Notes: * = significant difference ($p<0.05$)

Table 2.6. Kinematic data (means and stand deviations) for the ankle obtained during stance phase of the walking gait whilst wearing ankle protectors.

PROTECTOR	**Test**		**Retest**		
	Mean	*SD*	*Mean*	*SD*	
Sagittal plane (+ = dorsiflexion/ - = plantarflexion)					
Angle at footstrike (˚)	66.31	3.97	66.65	3.83	
Angle at toe-off (˚)	43.42	4.36	43.82	3.81	
Peak dorsiflexion (˚)	71.43	4.25	71.75	4.55	
ROM (˚)	22.89	3.63	22.83	3.62	
Coronal plane (+ = inversion/ - =eversion)					
Angle at footstrike (˚)	1.93	5.69	1.29	5.07	
Angle at toe-off (˚)	8.38	4.67	7.86	4.60	
Peak eversion (˚)	-5.02	4.50	-5.59	4.36	
ROM (˚)	7.52	2.78	7.45	3.07	
Transverse plane (+ = external/ - =internal)					
Angle at footstrike (˚)	-9.63	2.86	-9.48	3.45	
Angle at toe-off (˚)	-2.14	3.19	-2.09	3.79	
Peak external rotation (˚)	-1.15	2.57	-1.13	3.40	
ROM (˚)	7.49	3.70	7.39	3.46	

Notes: * = significant difference ($p<0.05$)

Table 2.7. Kinematic data (means and stand deviations) for the ankle obtained during stance phase of the walking gait whilst wearing ankle braces.

BRACE	**Test**		**Retest**		
	Mean	*SD*	*Mean*	*SD*	
Sagittal plane (+ = dorsiflexion/ - = plantarflexion)					
Angle at footstrike (˚)	65.88	4.96	64.34	4.35	*
Angle at toe-off (˚)	43.45	4.72	41.96	3.79	*
Peak dorsiflexion (˚)	71.06	4.54	69.50	4.98	*
ROM (˚)	22.43	3.56	22.38	3.53	
Coronal plane (+ = inversion/ - =eversion)					
Angle at footstrike (˚)	-2.70	6.26	-3.08	5.32	
Angle at toe-off (˚)	4.69	4.99	4.39	5.20	
Peak eversion (˚)	-9.35	4.99	-9.75	4.43	
ROM (˚)	8.27	3.16	8.24	3.31	
Transverse plane (+ = external/ - =internal)					
Angle at footstrike (˚)	-6.46	3.07	-6.71	2.55	
Angle at toe-off (˚)	-0.39	3.83	-0.65	3.41	
Peak external rotation (˚)	1.02	3.08	0.64	2.66	
ROM (˚)	6.07	3.70	6.06	3.27	

Notes: * = significant difference ($p<0.05$)

Table 2.8. Kinematic data (means and stand deviations) for the ankle obtained during stance phase of the walking gait without a brace or protector using the functional ankle method.

FUNCTIONAL	**Test**		**Retest**		
	Mean	*SD*	*Mean*	*SD*	
Sagittal plane (+ = dorsiflexion/ - = plantarflexion)					
Angle at footstrike (°)	67.66	4.79	69.23	4.56	*
Angle at toe-off (°)	45.20	3.85	46.74	3.98	*
Peak dorsiflexion (°)	72.94	2.97	74.43	3.06	*
ROM (°)	22.47	3.51	22.49	3.50	
Coronal plane (+ = inversion/ - =eversion)					
Angle at footstrike (°)	-2.75	8.02	-1.87	6.59	
Angle at toe-off (°)	4.64	6.06	5.48	5.16	
Peak eversion (°)	-9.37	6.64	-8.48	5.72	
ROM (°)	8.36	3.14	8.23	3.35	
Transverse plane (+ = external/ - =internal)					
Angle at footstrike (°)	-5.16	2.82	-5.32	2.91	
Angle at toe-off (°)	0.98	4.93	0.76	4.74	
Peak external rotation (°)	2.09	3.84	1.63	3.52	
ROM (°)	6.14	3.63	6.08	3.47	

Notes: * = significant difference ($p<0.05$)

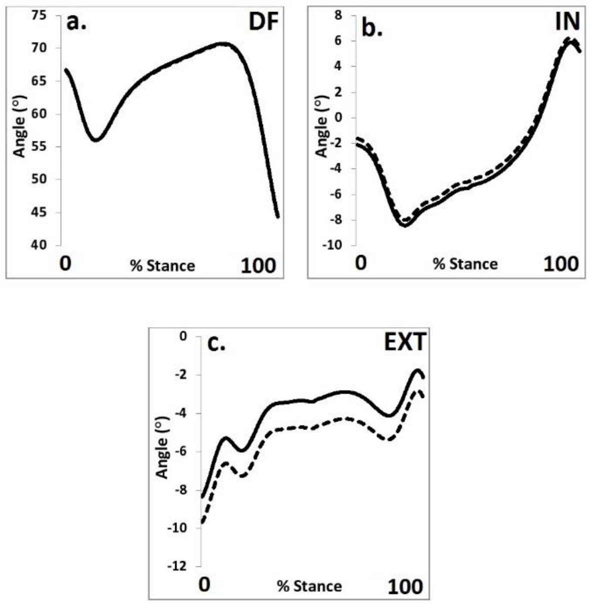

Figure 2.8. Ankle joint kinematics for the without condition in the a. sagittal, b. coronal and c. transverse planes (black = test and dash = retest) (DF = dorsiflexion, IN = inversion, EXT = external).

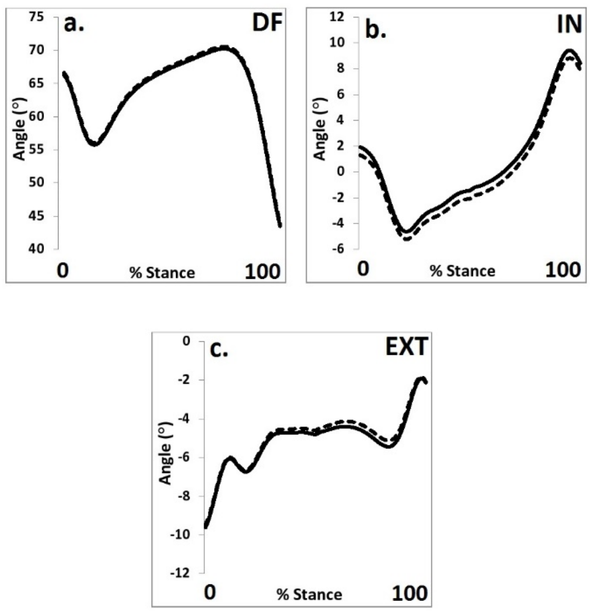

Figure 2.9. Ankle joint kinematics for the protector condition in the a. sagittal, b. coronal and c. transverse planes (black = test and dash = retest) (DF = dorsiflexion, IN = inversion, EXT = external).

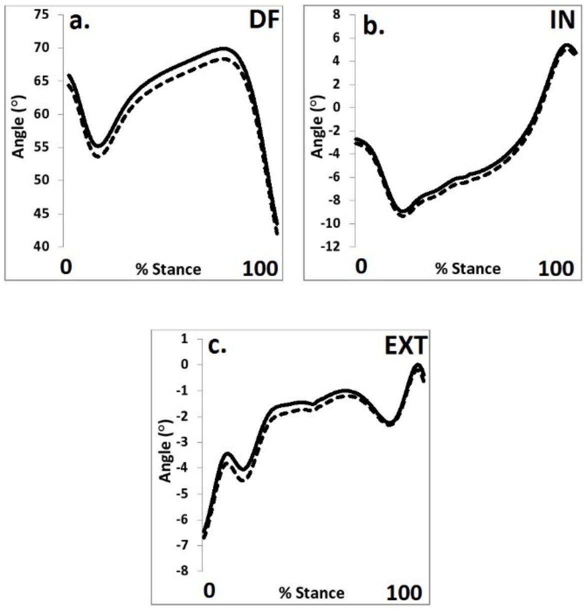

Figure 2.10. Ankle joint kinematics for the braced condition in the a. sagittal, b. coronal and c. transverse planes (black = test and dash = retest) (DF = dorsiflexion, IN = inversion, EXT = external).

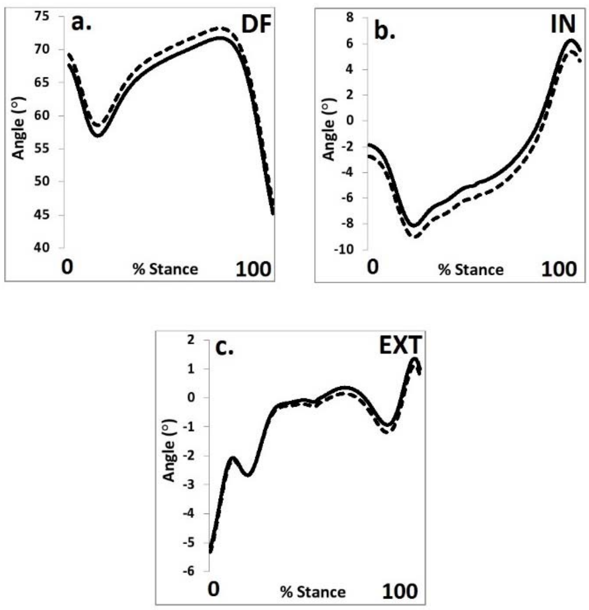

Figure 2.11. Ankle joint kinematics for the functional condition in the a. sagittal, b. coronal and c. transverse planes (black = test and dash = retest) (DF = dorsiflexion, IN = inversion, EXT = external).

2.6.1.4 Discussion

The aim of this pilot study was to assess the test-retest reliability of the TMM and FAM for defining ankle joint centre. To the authors knowledge this study represents the first study to assess the reliability of these two methods and to compare how wearing an ankle brace or ankle protector affects the reliability of the TMM. This study may provide important information to those looking to use 3D analysis to quantify reliable ankle joint kinematics.

It is important to note that all four conditions showed no significant test-retest differences in the coronal plane. The coronal plane waveforms also had the highest test-retest ICC's indicating a high level of reliability (ICC≥0.987). Therefore, the TMM and FAM methods can both be reliably utilised to assess inversion and eversion. This finding disagrees with findings by Besier, et al. (2003), Sinclair, et al. (2015), and Sinclair, et al. (2014) who proposed that sagittal plane kinematics are more reliable and less susceptible to alterations than the transverse and coronal planes. However, these studies looked at the hip and knee whereas the current study investigated the ankle.

Both the BRACE and FUNCTIONAL conditions showed significant differences in the sagittal plane for angle at footstrike, angle at toe off and peak dorsiflexion. Also the WITHOUT condition showed a significant difference for angle at toe off in the sagittal plane. These variations pose a problem for clinicians interested in the effects of ankle dorsiflexion and plantarflexion on injury aetiology (Donoghue, et al., 2008; Johanson, et al., 2006; Silbernagel, et al., 2012). However, it is important to acknowledge that all conditions exhibited a high level of reliability (ICC≥0.984) in the sagittal plane and the difference found was less than 2°. Therefore, it is recommended that clinicians use the TMM when interested in sagittal plane kinematics as it exhibited fewer significant differences between test and retest parameters. In the transverse plane only the angle at toe off in the without condition showed a significant difference between test-retest data. The transverse plane also had the lowest reliability when compared to the other two planes of motion, albeit still moderately reliable (ICC≥0.779).

Out of the four conditions the braced condition was the least reliable (ICC≥0.779) and exhibited a higher number of significant differences than the protector or without conditions. The error in the brace condition is most likely due to the hard outer shell making it difficult to palpate the malleoli. A proposed methodology to allow for more accurate data collection for a braced ankle could be to take a static using an unbraced ankle, making sure that the tracking markers on the footwear are secured using a strong adhesive, then removing the footwear and putting on the brace before putting the footwear back on. This methodology needs further investigation for test-retest reliability before being utilised by clinicians.

There are some limitations to the current investigation that should be acknowledged. Firstly, all participants were of a healthy BMI with no skeletal abnormalities. This made palpitation and identification of landmarks relatively easy whereas participants with a larger BMI and skeletal abnormalities may lead to difficulties with landmark identification. Secondly whilst the current study looked at the reliability of TMM and FAM it did not consider their accuracy in locating the true centre of the ankle joint. It is therefore recommended further work be undertaken to investigate which method, TMM or FAM, is more accurate and reliable at identifying the anatomical joint centre using radiographic techniques.

In conclusion, whilst research has considered the reliability of hip and knee joint centre locations techniques, information regarding the ankle joint centre is lacking. The present study adds to the current knowledge regarding the reliability of different ankle joint centre location techniques. The findings of the current investigation indicate that there are fewer errors using the TMM when the ankle is uncovered or when covered with a soft foam which is easy to palpate through. Therefore, the TMM is proposed as the best method to use by clinicians when examining participants with heathy BMI and no skeletal abnormalities.

2.7 Data filtering

When collecting kinematic data, the purpose of data filtering is to remove any undesirable noise from the digital signal and leave behind only what is considered to be the true signal (Winter, 1990). Noise can occur within a signal from improper digitization of the markers, electrical interference, or soft tissue artefacts (Winter, et al., 1974). Often in biomechanics to remove this noise from a signal, investigators will use low pass filters with a set cut off frequency dependent on the type of motion being recorded. The purpose of these low-pass filters is to keep the lower frequencies of the signal and attenuate the higher frequencies associated with noise. 4th order zero-lag Butterworth filters are a popular low pass filter within the field of biomechanics (Sinclair, et al., 2013). The reason for their popularity over other filtering methods is due to them being optimally flat in their pass band, have relatively high roll offs, and rapid response in the time domain (Robertson & Dowling, 2003). Therefore, all data collected during this thesis was processed using low pass 4th order zero-lag Butterworth filters. However, another important consideration is the cut off frequency used as a poor selection can be detrimental to the interpretation of findings. As the optimal cut off frequency can differ for different motions each study contained within this thesis will define the cut-off frequency used within its specific methodological section.

2.8 Segment co-ordinate system

When modelling segments, global or local co-ordinate systems can be utilised to establish the axes of rotations of each segment. Throughout this thesis each segments co-ordinate axes were defined using the segment co-ordinate system (SCS) which is a local co-ordinate system. The SCS was chosen over the GCS as the GCS has previously been found to produce considerable error when the segment is not aligned perpendicular to a plane of motion (Richards, et al., 2008). The SCS Z axis (internal/external rotation) is determined by the unit vector directed from the distal segment endpoint to the proximal segment endpoint. The SCS Y axis (abduction/adduction) is determined by the unit vector that is perpendicular to both the frontal plane and the Z axis. This axis is directed posterior to anterior. The SCS X axis (flexion/extension) is determined by the application of the right hand rule. This axis is medial-lateral in orientation.

2.9 3D kinematic calculations

All joint angles and joint velocities contained within this thesis were calculated following the Carden/Euler technique described by Grood & Suntay, (1983). These angles are calculated by establishing the orientation of the three axes of the SCS of the proximal segment relative to the three axes of SCS of the distal segment to form an agreeable axis between the two segments. To do this two key pieces of information are required; a rotation matrix and position vector. The rotation matrix describes the axis of the SCS (X, Y, and Z) and the position vector defines a pivot point between the two segments which then allows the orientation of the segments to be established and the joint angles to be calculated. The order in which the three axes are worked out relative to one another can affect the orientation of the segments axes. Therefore, when calculating multi-planar motions an important factor to consider is the order in which the planes should be calculated as the sequence could affect the angular kinematic outputs. Throughout this thesis The International Society of Biomechanics (ISB) guidelines were adhered to for defining the cardan sequence. ISB recommend that an order of X, Y, Z be followed when calculating lower extremity angular kinematics, where X is flexion/extension, Y is abduction/adduction, and Z is internal/external rotation (Wu & Cavanagh, 1995). Further work has validated this sequence as being the most reliable when investigating ankle kinematics (Sinclair, et al., 2012), knee and hip kinematics (Lees, et al., 2010).

2.10 Statistical analysis

2.10.1 Sample size

For each of the studies contained within this thesis a sample size of 12 for each gender was selected based on previous research finding meaningful findings using as few as seven participants (De Clercq, 1997), ten participants (Cloak, et al., 2010, Commons & Low, 2014, Greene, et al., 2014) and eleven participants (Tang, et al., 2010, Vanwanseele, et al., 2014) when comparing braced and unbraced dynamic movements.

2.10.2 Descriptive statistics

Throughout this thesis descriptive statistics are used to present the means and standard deviations of the outcome measures and also to present key characteristics of the participant populations used for each study.

2.10.3 Inferential statistics

All data analysis contained within this thesis was conducted using SPSS (SPSS Inc., Chicago, IL, USA) with statistical significance accepted at the $p \leq 0.05$ level. Each studies methodological section details the specific variables extracted for analysis, the specific analysis ran on the them, and the specific version of SPSS used for the analysis. It should be noted that the main aim of the thesis was to assess the effects of ankle protectors on lower-limb kinematics and not to investigate gender differences. Therefore, throughout this thesis the data for each movement investigated is grouped by gender and analysed/presented separately as not to detract from the focus of the thesis.

3. The effects of ankle protectors on the stance phase of running: a comparison to braced and unbraced ankles.

3.1 Introduction

Running is a fundamental movement in football and is utilised frequently when not in possession of the ball to move around the pitch. During a 90-minute football match professional male players will cover between 9 km and 12 km with the majority of this being walking (speeds between 0.19 and 2 m.s^{-1}) or jogging (speeds between 2 and 4 m.s^{-1}) (Rampinini, et al., 2007). Professional female players will cover between 9.7 and 11.3 km with between 8.4 and 9.8 km of this being either walking, jogging, or low-speed running (speeds between 1.67 and 3.33 m.s^{-1}) (Krustrup, et al., 2005). Football players at the highest standard perform more intervals of high-intensity running than those at a lower level (Mohr, et al., 2008). During jogging without the football contusion injuries are infrequent but could possibly occur during an accidental collision with another player. An ankle-inversion injury is more likely to occur during jogging and is the third most frequent mechanism of non-contact ankle-inversion injuries sustained by football players (Woods, et al., 2003).

Ankle braces have previously been found to be effective at reducing ankle kinematics in passive ROM tests and during dynamic tilt table tests (Eils, et al., 2002) and have been established as being effective at reducing the risk of inversion injuries (Surve, et al., 1994). Furthermore, ankle braces effect on stance limb kinematics during running has had some attention (De Clercq, 1997; Martin & Harter, 1993; Tamura, et al., 2017). However, there are currently no research papers investigating the effects of ankle protectors on stance limb kinematics during running.

Previous research investigating the effects of ankle braces has found that when running on a treadmill tilted at 8˚ semi-rigid ankle braces and lace-up ankle braces significantly reduce peak inversion angles compared to not wearing either (Martin & Harter, 1993). Another study found during overground running ankle braces significantly reduced ankle eversion, and peak eversion velocity when compared to not wearing any (De Clercq, 1997). More recently a study comparing semi-rigid ankle braces and lace-up ankle braces to a control of not wearing either found that during a continuous 30 minute run semi-rigid ankle braces significantly reduced peak inversion, inversion/eversion range of motion (ROM), and peak eversion velocity, whereas lace-up ankle braces significantly reduced inversion/eversion ROM, and peak eversion velocity but not peak inversion (Tamura, et al., 2017). Additionally, the type of ankle brace has also been found to effect sagittal plane motion. Tamura, et al., (2017) found that a lace-up ankle brace significantly reduced peak plantarflexion, plantarflexion angle at toe off, and peak plantarflexion velocity during running whereas a semi-rigid ankle brace did not exhibit any restrictions in this plane of motion.

During walking it has been found that by restricting dorsiflexion of the ankle by as little as 8˚ from normal gait can affect knee joint kinematics and kinetics in both the sagittal and coronal planes which could worsen knee osteoarthritis or posterior knee laxity in individuals with pre-existing conditions (Ota, et al., 2014). However, another study using a commercially available ankle brace found that ankle braces do not effect knee kinematics during running (West & Campbell, 2014). Tamura, et al., (2017) found that both a lace-up and semi-rigid ankle brace did not significantly alter knee flexion but did significantly increase hip adduction at footstrike. Studies investigating the effects of ankle braces on ground reaction forces during running have found no significant difference in peak force or time to peak force when compared to not wearing any (De Clercq, 1997; West & Campbell, 2014).

There are currently no research papers available on the effects of ankle protectors on running performance but there are several studies that have looked at the effect of ankle braces. The majority of studies have found that wearing ankle braces do not effect running performance and do not negatively affect sprint times (Locke, et al., 1997; Gross, et al., 1997; Bocchinfuso, et al., 1994). Although it has recently been found that ankle braces increase energy expenditure, but the effect is less than < 1 kcal/min, when compared to unbraced ankles (Tamura, et al., 2017).

Ankle braces are proficient at reducing the risk of ankle-inversion injuries and appear to have little effect on running performance, however it is unknown if ankle protectors effect the ankle in the same way as a brace. Therefore, when comparing ankle protectors effectiveness at reducing the risk of ankle-inversion injuries during running, a reduction in peak inversion, and ankle ROM in the coronal plane are key parameters of interest. However, the effects further up the kinematic chain should not be ignored due to the possible effects on the knee and hip joints. Therefore, the current study aims to investigate the effects of ankle protectors on ankle kinematics during the stance phase of running, compare the effects of ankle protectors with braced and unbraced ankles to establish which it more closely resembles, investigate the effects of ankle protectors on knee and hip kinematics, and investigate the effects on both male and female populations. It is hypothesised that ankle protectors will reduce ankle kinematics and produce similar kinematics to a braced ankle.

3.2 Methodology

3.2.1 Participants

Twelve male (age 24.75±4.81 years, height 174.77±5.83 cm, body mass 73.43±10.52 kg and BMI 23.98±2.73) and twelve female (age 24.25±6.58 years, height 165.17±4.93cm, body mass 64.07±8.23 kg and BMI 23.47±2.71) participants took part in this study. Participants were recruited from local and university football teams via opportunity sampling using poster adverts. The inclusion criteria for the study was that the participants were aged between 18 and 35, regularly partake in sport and were injury free at the time of testing. All participants provided written consent in line with the University of Central Lancashire's ethical panel (STEMH 309).

3.2.2 Procedure

Participants performed running trials across an 18m by 7.5m biomechanics laboratory in three test conditions; wearing ankle braces (BRACE), wearing ankle protectors (PROTECTOR) and with uncovered ankles (WITHOUT). The order the participants performed the test conditions in was randomised and five successful trials were recorded for each test condition. A successful trial was determined as one in which the participant landed with the whole of their right foot on an embedded force plate (Kistler Instruments Ltd., Alton, Hampshire) located in the centre of the laboratory, did not focus on the force plate as to alter their natural gait pattern (Sinclair, et al., 2014), and kept within a speed tolerance of 3.4 $m.s^{-1}$ ± 5%. This speed was selected due to the majority of jogging during a 90minute match being at low speeds (Rampinini, et al., 2007). The force plate sampled at 1000 Hz and was used to determine the start and end of the stance phase during the running trials. These points were determined as the point where the force plate first recorded a vertical ground reaction force (VGRF) that exceeded 20N and ended when the VGRF dropped back down below 20N (Sinclair, et al., 2011).

Kinematic data were recorded using an eight camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) tracking retro-reflective markers at a sampling rate of 250 Hz. Using the calibrated anatomical system technique (CAST) (Cappozzo, et al., 1995) the retro-reflective markers were attached to the 1st and 5th metatarsal heads, calcaneus, medial and lateral malleoli, the medial and lateral femoral epicondyles, the greater trochanter, left and right anterior superior iliac spine, and left and right posterior superior iliac spine. These markers were used to model the right foot, shank, thigh, and pelvis segments in six degrees of freedom. Rigid plastic mounts with four markers on each were also attached to the shank and thigh and were secured using elasticated bandage. These were used as tracking markers for the shank and thigh segments. To track the foot the 1st and 5th metatarsal heads and the calcaneus were used and to track the pelvis the left and right anterior superior iliac spine and left and right posterior superior iliac spine were used. In the BRACE condition the medial and lateral malleoli locations were found by placing the index finger under the rigid construct of the brace to locate the anatomical landmark then matching the location to the exterior of the Brace where the marker was then fixed to. In the PROTECTOR condition the medial and lateral malleoli locations were located by palpating the soft foam construct to find the underlying anatomical landmarks. To assess the speed of the participant a single marker was attached to the xiphoid process and was checked for velocity using the QTM software after each trial was recorded. Before dynamic trials were captured a static trial of the participant stood in the anatomical position was captured which was used to identify the location of the tracking makers with reference to the anatomical markers. To define each plane of motion firstly the Z (transverse) axis follows the segment from distal to proximal and denotes internal/external rotation, secondly the Y (coronal) axis is orientated from anterior to posterior of the segment and denotes adduction/abduction, and thirdly the X (sagittal) axis is orientated from medial to lateral of the segment and denotes flexion/extension.

3.2.3 Ankle Braces and Ankle Protectors

The ankle protectors used for the current investigation were a pair of Nike ankle shield 10 (Nike Inc, Washington County, Oregon, USA) and the ankle braces used were a pair of Aircast A60 (DJO, Vista, CA, USA).

Figure 3.1. On the left a pair of Nike ankle shield 10 ankle protectors and on the right an Aircast A60 ankle brace.

3.2.4 Data Processing

Anatomical and tracking markers were identified within the Qualisys Track Manager software and then exported as C3D files to be analysed using Visual 3D software (C-Motion, Germantown, MD, USA). To define the centre points of the ankle and knee segments the two marker methods were utilised for both. These methods calculate the centre of the joint using the positioning of the malleoli markers for the ankle centre and the femoral epicondyle markers for the knee centre (Graydon, et al., 2015; Sinclair, et al., 2015). To calculate the hip joint centre a regression equation which uses the position of the ASIS markers was utilised (Sinclair, et al., 2014). The running trials were filtered at 12 Hz using a low pass 4th order zero-lag Butterworth filter. A cut off frequency of 12 Hz was selected based on work done by Sinclair, et al. (2013) who used a residual analysis to establish that a cut off frequency range of 10-15Hz is optimal for filtering running data. Data were normalized to 100% of the stance phase then processed trials were used to produce means of the five trials for each test condition for each participant. 3D kinematics of the ankle, knee and hip joints of the right leg were calculated using an XYZ cardan sequence of rotations. The 3D joint kinematic measures which were extracted for further analysis were 1) angle at footstrike, 2) angle at toe-off, 3) peak angle during the stance phase, 4) absolute range of motion (absolute ROM) calculated by taking the maximum angle from the minimum angle during stance, 5) relative range of motion (relative ROM) calculated using the angle at footstrike and the first peak value after footstrike, 6) peak angular velocities for the ankle during the stance phase. Measures taken from the force plate to be analysed were 1) peak forces 2) instantaneous loading rate calculated as the maximum increase in vertical force between frequency intervals, 3) average loading rate calculated by dividing the peak vertical impact force by the time to the impacts peak 4) stance time. The force data were normalised to bodyweights (BWs) for each participant to allow comparisons across the data set to be investigated.

3.2.5 Statistical analyses

Data analysis was conducted using SPSS v22.0 (SPSS Inc., Chicago, IL, USA). The data were grouped by gender and analysed separately from one another. The means of the five trials for each of the three test conditions for each output measure were compared using one-way repeated measures ANOVA with significant findings, accepted at $P \leq 0.05$ level, being further explored using Bonferroni post-hoc pairwise comparisons. The Bonferroni method was selected due to it being a more conservative post-hoc test and better at reducing Type 1 errors than other methods (McHugh, 2011). Effect sizes were determined using partial Eta^2 (pη^2).

3.3 Male Results

Tables 3.1-3.3 and figures 3.2 & 3.3 present the key parameters of interest obtained during the stance phase of locomotion.

3.3.1 Kinetic and temporal parameters

Table 3.1. Kinetic and temporal variables (means and standard deviations) obtained during stance phase of the running gait from male participants.

	WITHOUT		PROTECTOR		BRACE	
Peak Vertical Impact Force (BW)	2.20 ± 0.35		2.29 ± 0.26		2.40 ± 0.40	
Peak Braking Force (BW)	0.46 ± 0.12		0.48 ± 0.12		0.48 ± 0.16	
Peak Propulsive force (BW)	0.33 ± 0.05		0.31 ± 0.06		0.31 ± 0.05	
Peak Medial Force (BW)	0.12 ± 0.06		0.12 ± 0.06		0.12 ± 0.07	
Peak Lateral Force (BW)	0.19 ± 0.07		0.20 ± 0.08		0.20 ± 0.06	
Instantaneous Loading Rate (BW.s)	177.25 ± 46.89		181.02 ± 45.95		209.28 ± 66.51	
Average Loading Rate (BW.s)	76.68 ± 22.20		78.19 ± 24.69		93.88 ± 30.60	
Stance Time (s)	0.22 ± 0.02		0.23 ± 0.02		0.22 ± 0.02	

The kinetic and temporal variables exhibited no significant differences ($P > 0.05$) between the WITHOUT, PROTECTOR, and BRACE conditions.

3.3.2 3D Kinematic Parameters

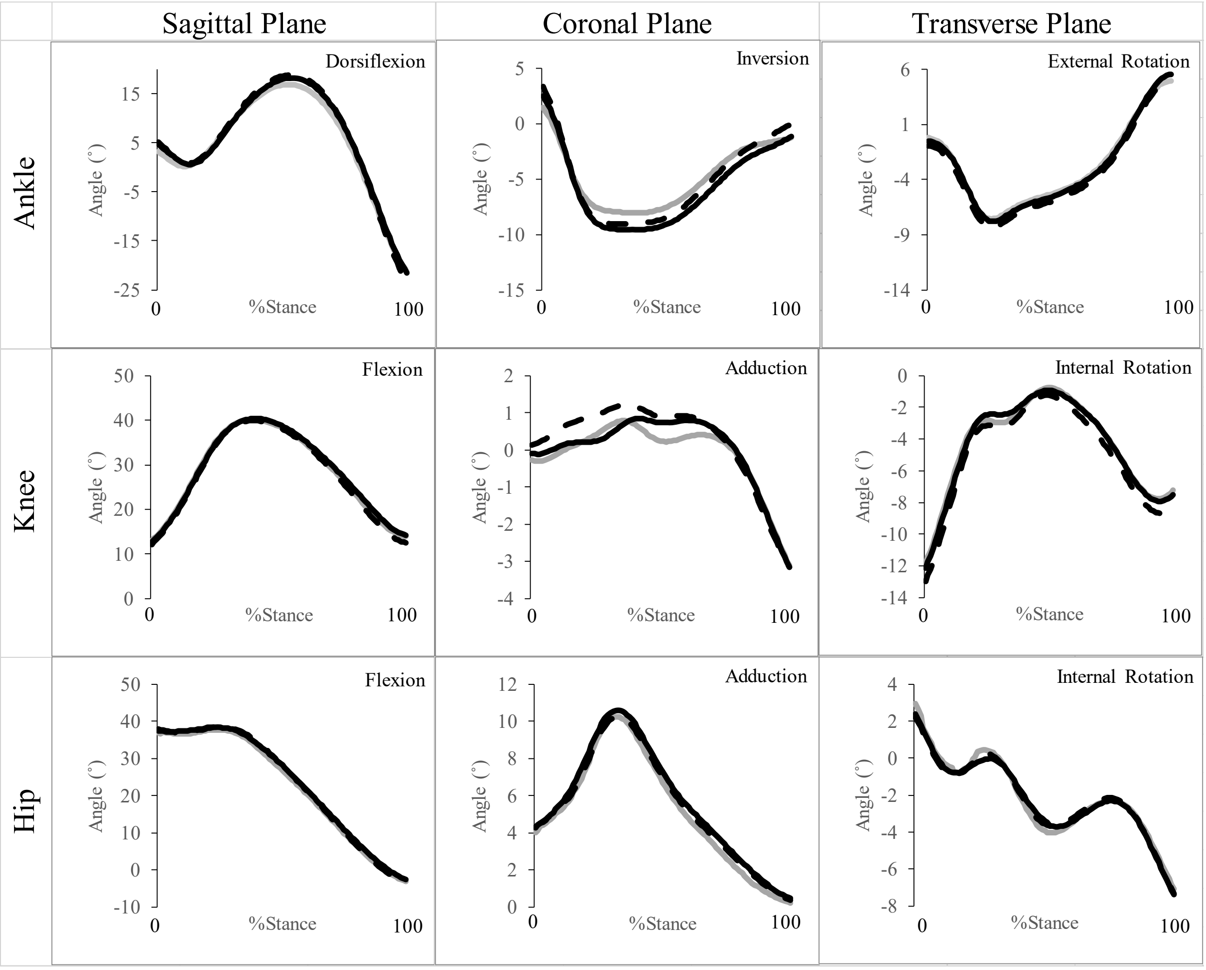

Figure 3.2. Mean ankle, knee, and hip kinematics during the stance phase of locomotion for the sagittal, coronal, and transverse planes from male participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey)

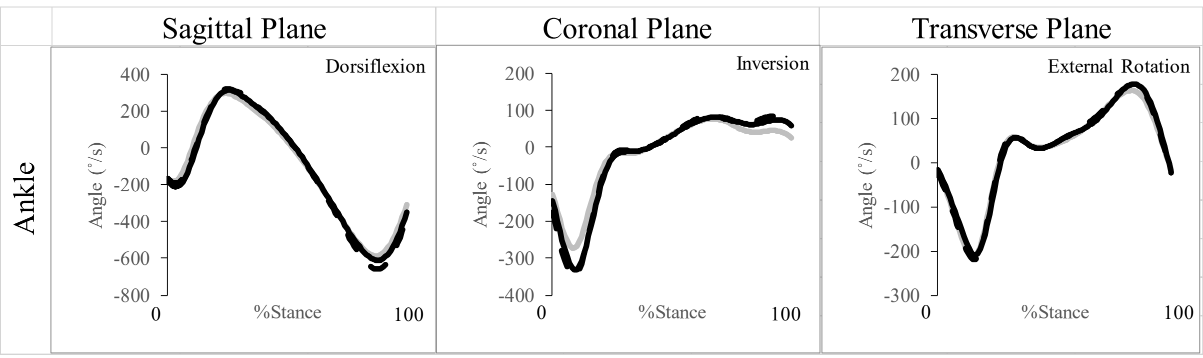

Figure 3.3. Mean ankle velocity during the stance phase of locomotion for the sagittal, coronal, and transverse planes from male participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

Table 3.2. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained during stance phase of the running gait from male participants.

		WITHOUT		PROTECTOR		BRACE	
ANKLE	**Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)**						
	Angle at footstrike	6.20 ± 7.42		6.05 ± 6.82		4.15 ± 5.64	B
	Angle at toe-off	-23.65 ± 4.13		-21.69 ± 3.85	A	-21.32 ± 3.22	A
	Peak dorsiflexion	19.20 ± 3.21		18.46 ± 2.41		17.02 ± 2.09	AB
	Absolute ROM	42.66 ± 3.29		40.15 ± 3.73	A	38.34 ± 2.99	AB
	Relative ROM	13.00 ± 6.45		12.41 ± 5.96		12.87 ± 5.41	
	Peak dorsifelxion velocity (˚/s)	357.16 ± 55.92		342.46 ± 52.27		316.85 ± 46.74	AB
	Peak plantarflexion velocity (˚/s)	-665.97 ± 82.45		-619.33 ± 88.36	A	-595.27 ± 75.35	A
	Coronal plane (+ = Inversion / - = Eversion)						
	Angle at footstrike	3.32 ± 2.86		2.54 ± 3.07		1.46 ± 2.55	A
	Angle at toe-off	0.02 ± 3.41		-1.06 ± 3.59		-1.24 ± 3.05	A
	Peak inversion	3.87 ± 2.79		3.16 ± 3.07		1.92 ± 2.74	A
	Peak eversion	-9.78 ± 3.70		-10.28 ± 3.78		-8.80 ± 3.74	B
	Absolute ROM	13.64 ± 3.23		13.44 ± 3.20		10.72 ± 2.30	AB
	Relative ROM	13.10 ± 3.94		12.82 ± 3.69		10.26 ± 2.87	AB
	Peak inversion velocity (˚/s)	140.04 ± 67.85		134.53 ± 68.14		108.96 ± 49.71	AB
	Peak eversion velocity (˚/s)	-363.35 ± 88.53		-350.48 ± 98.62		-284.47 ± 77.48	AB
	Transverse plane (+ = External / - = Internal)						
	Angle at footstrike	-1.15 ± 2.10		-0.56 ± 2.66		-0.43 ± 2.91	
	Angle at toe-off	5.06 ± 3.87		5.61 ± 3.95		4.87 ± 4.42	
	Peak rotation	-8.82 ± 4.44		-8.33 ± 4.53		-8.06 ± 4.38	
	Absolute ROM	13.94 ± 4.18		14.02 ± 4.02		13.12 ± 3.43	
	Relative ROM	7.67 ± 3.13		7.78 ± 2.83		7.63 ± 2.47	
	Peak external rotation velocity (˚/s)	195.11 ± 53.73		195.48 ± 51.68		181.75 ± 51.06	
	Peak internal rotation velocity (˚/s)	-254.25 ± 87.68		-257.38 ± 84.52		-248.07 ± 82.90	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

For the ankle joint, in the sagittal plane, significant main effects were found for the angle at footstrike $F_{(2, 22)} = 5.04$, $P \leq 0.05$, $p\eta^2 = 0.31$, angle at toe-off $F_{(2, 22)} = 11.95$, $P \leq 0.01$, $p\eta^2=0.52$, peak dorsiflexion angle $F_{(2, 22)} = 23.27$, $P \leq 0.01$, $p\eta^2=0.68$, absolute ROM $F_{(2, 22)} = 31.12$, $P \leq 0.01$, $p\eta^2=0.74$, peak dorsiflexion velocity $F_{(2, 22)} = 14.71$, $P \leq 0.01$, $p\eta^2=0.57$, and peak plantarflexion velocity $F_{(2, 22)} = 20.92$, $P \leq 0.01$, $p\eta^2=0.66$. Post-hoc analysis revealed that the BRACE condition exhibited significantly ($P \leq 0.05$, 95%CI: 0.11-3.69%) lower angle at footstrike than the PROTECTOR condition. It also revealed the BRACE ($P \leq 0.01$, 95%CI: 0.58-4.09%) and PROTECTOR ($P \leq 0.001$, 95%CI: 0.89-3.05%) conditions had a significant reduction in angle at toe off than the WITHOUT condition. The BRACE condition significantly reduced peak dorsiflexion when compared to the WITHOUT ($P \leq 0.001$, 95%CI: 1.03–3.33%) and PROTECTOR ($P \leq 0.001$, 95%CI: 0.84-2.05%) conditions. All three conditions were significantly different from each other for absolute range of motion with the WITHOUT condition having the most ROM and BRACE condition having the least ROM (WITHOUT & PROTECTOR $P \leq 0.001$, 95%CI: 1.62-3.40%, WITHOUT & BRACE $P \leq 0.001$, 95%CI: 2.53-6.11%, and PROTECTOR & BRACE $P \leq 0.05$, 95%CI: 0.20-3.61%). For the ankle velocities the BRACE condition significantly reduced the dorsiflexion velocity compared to the PRTOECTOR ($P \leq 0.01$, 95%CI: 9.14-42.07%) and WITHOUT ($P \leq 0.001$, 95%CI: 18.16-62.45%) conditions and the WITHOUT condition had significantly more plantarflexion velocity compared to the PROTECTOR ($P \leq 0.01$, 95%CI: 15.71-77.56%) and BRACE ($P \leq 0.001$, 95%CI: 40.66-100.74%) conditions.

For the ankle joint, in the coronal plane, significant main effects were found for the angle at footstrike $F_{(2, 22)} = 7.34$, $P \leq 0.01$, $p\eta^2=0.40$, angle at toe-off $F_{(2, 22)} = 6.02$, $P \leq 0.01$, $p\eta^2=0.35$, peak inversion angle $F_{(2, 22)} = 10.22$, $P \leq 0.01$, $p\eta^2=0.48$, peak eversion angle $F_{(1.19, 13.14)} = 6.80$, $P \leq 0.05$, $p\eta^2=0.38$, absolute ROM $F_{(2, 22)} = 25.19$, $P \leq 0.01$, $p\eta^2=0.70$, relative ROM $F_{(2, 22)} = 18.40$, $P \leq 0.01$, $p\eta^2=0.63$, peak inversion velocity $F_{(2, 22)} = 9.06$, $P \leq 0.01$, $p\eta^2=0.45$, and peak eversion velocity $F_{(2, 22)} = 20.90$, $P \leq 0.01$, $p\eta^2=0.66$. Post-hoc analysis revealed that the BRACE condition significantly reduced angle at footstrike ($P \leq 0.004$, 95%CI: 0.62-3.10%), angle at toe off ($P \leq 0.05$, 95%CI: 0.01-2.51%), and peak inversion angle ($P \leq 0.001$, 95%CI: 0.93-2.96%) when compared with the WITHOUT condition. The BRACE condition also exhibited significantly ($P \leq 0.001$, 95%CI: 0.87-2.08%) lower peak eversion angle when compared to the PROTECTOR condition. It was also revealed that the BRACE condition had significantly lower absolute and relative ROM's when compared to both the WITHOUT (ABS $P \leq 0.001$, 95%CI: 1.42-4.43% & REL $P \leq 0.001$, 95%CI: 1.21-4.46%) and PROTECTOR

(ABS $P \leq 0.001$, 95%CI: 1.36-4.08% & REL $P \leq 0.01$, 95%CI: 1.06-4.06%) conditions. For the ankle velocities the BRACE condition significantly reduced both inversion velocity and eversion velocity compared to the WITHOUT (INV $P \leq 0.05$, 95%CI: 5.15-57.01% & EVE $P \leq 0.001$, 95%CI: 38.73-119.04%) and PROTECTOR (INV $P \leq 0.01$, 95%CI: 8.64-42.50% & EVE $P \leq 0.001$, 95%CI: 31.62-100.41%) conditions. No significant differences ($P > 0.05$) were found in the transverse plane for the ankle

Table 3.3. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained during stance phase of the running gait from male participants.

		WITHOUT	PROTECTOR	BRACE
KNEE	**Sagittal plane (+ = Flexion / - = Extension)**			
	Angle at footstrike	11.99 ± 4.35	12.58 ± 4.36	12.83 ± 3.81
	Angle at toe-off	12.49 ± 4.62	14.32 ± 6.05	14.12 ± 5.50
	Peak flexion	40.09 ± 3.97	40.55 ± 3.70	40.17 ± 3.98
	Absolute ROM	30.56 ± 4.43	30.31 ± 3.42	29.54 ± 3.54
	Relative ROM	28.10 ± 4.96	27.97 ± 4.96	27.34 ± 4.08
	Coronal plane (+ = Adduction / - = Abduction)			
	Angle at footstrike	0.14 ± 4.18	-0.6 ± 4.24	-0.43 ± 4.50
	Angle at toe-off	-3.16 ± 2.78	-3.14 ± 2.92	-3.15 ± 3.00
	Peak adduction	2.92 ± 4.66	2.73 ± 4.66	2.56 ± 4.38
	Absolute ROM	6.52 ± 2.40	6.65 ± 2.30	6.42 ± 1.76
	Relative ROM	2.79 ± 2.65	2.79 ± 2.76	2.99 ± 2.60
	Transverse plane (+ = Internal / - = External)			
	Angle at footstrike	-12.96 ± 6.03	-12.18 ± 7.46	-11.94 ± 7.23
	Angle at toe-off	-8.37 ± 4.39	-7.52 ± 4.98	-7.17 ± 5.00
	Peak rotation	0.20 ± 6.72	0.62 ± 7.67	0.31 ± 7.22
	Absolute ROM	14.07 ± 5.89	13.84 ± 6.32	13.12 ± 6.30
	Relative ROM	13.16 ± 6.49	12.25 ± 6.90	12.25 ± 6.69
HIP		**WITHOUT**	**PROTECTOR**	**BRACE**
	Sagittal plane (+ = Flexion / - = Extension)			
	Angle at footstrike	36.72 ± 9.56	37.78 ± 8.34	36.82 ± 8.95
	Angle at toe-off	-3.61 ± 8.28	-2.72 ± 7.14	-3.11 ± 7.23
	Peak flexion	39.64 ± 9.24	39.81 ± 9.10	38.70 ± 9.38
	Absolute ROM	43.27 ± 9.48	42.45 ± 9.76	41.81 ± 9.64
	Relative ROM	40.35 ± 10.18	40.41 ± 9.86	39.93 ± 9.90
	Coronal plane (+ = Adduction / - = Abduction)			
	Angle at footstrike	4.41 ± 4.87	3.99 ± 4.70	4.55 ± 5.30
	Angle at toe-off	0.37 ± 2.36	0.38 ± 3.33	0.46 ± 3.63
	Peak adduction	10.51 ± 5.10	10.75 ± 5.30	10.79 ± 5.81
	Absolute ROM	10.86 ± 2.63	11.07 ± 2.53	11.09 ± 2.38
	Relative ROM	6.10 ± 3.28	6.76 ± 3.56	6.24 ± 3.76
	Transverse plane (+ = Internal / - = External)			
	Angle at footstrike	2.48 ± 7.76	2.45 ± 7.50	2.61 ± 8.57
	Angle at toe-off	-7.32 ± 6.56	-7.47 ± 7.21	-6.91 ± 6.74
	Peak rotation	-8.20 ± 6.71	-8.18 ± 7.01	-7.61 ± 6.59
	Absolute ROM	11.48 ± 4.24	11.56 ± 4.57	11.14 ± 4.59
	Relative ROM	10.68 ± 4.52	10.63 ± 4.83	10.22 ± 4.57

No significant differences ($P > 0.05$) were found in in any of the planes of motion for both the knee joint or the hip joint.

3.4 Female Results

Tables 3.4-3.6 and figures 3.4 & 3.5 present the key parameters of interest obtained during the stance phase of locomotion.

3.4.1 Kinetic and temporal parameters

Table 3.4. Kinetic and temporal variables (means and standard deviations) obtained during stance phase of the running gait from female participants.

	WITHOUT		PROTECTOR		BRACE	
Peak Vertical Impact Force (BW)	2.19 ± 0.22		2.31 ± 0.31		2.30 ± 0.28	
Peak Braking Force (BW)	0.39 ± 0.07		0.39 ± 0.08		0.39 ± 0.09	
Peak Propulsive force (BW)	0.34 ± 0.08		0.32 ± 0.08		0.32 ± 0.07	
Peak Medial Force (BW)	0.09 ± 0.04		0.07 ± 0.03		0.07 ± 0.03	
Peak Lateral Force (BW)	0.14 ± 0.06		0.14 ± 0.06		0.14 ± 0.05	
Instentaneous Loading Rate (BW.s)	125.88 ± 40.70		126.28 ± 37.67		139.92 ± 40.92	
Average Loading Rate (BW.s)	53.37 ± 21.62		59.85 ± 20.44		64.27 ± 23.79	
Stance Time (s)	0.24 ± 0.03		0.25 ± 0.03		0.24 ± 0.03	

The kinetic and temporal variables exhibited no significant differences ($P > 0.05$) between the WITHOUT, PROTECTOR, and BRACE conditions.

3.4.2 3D Kinematic Parameters

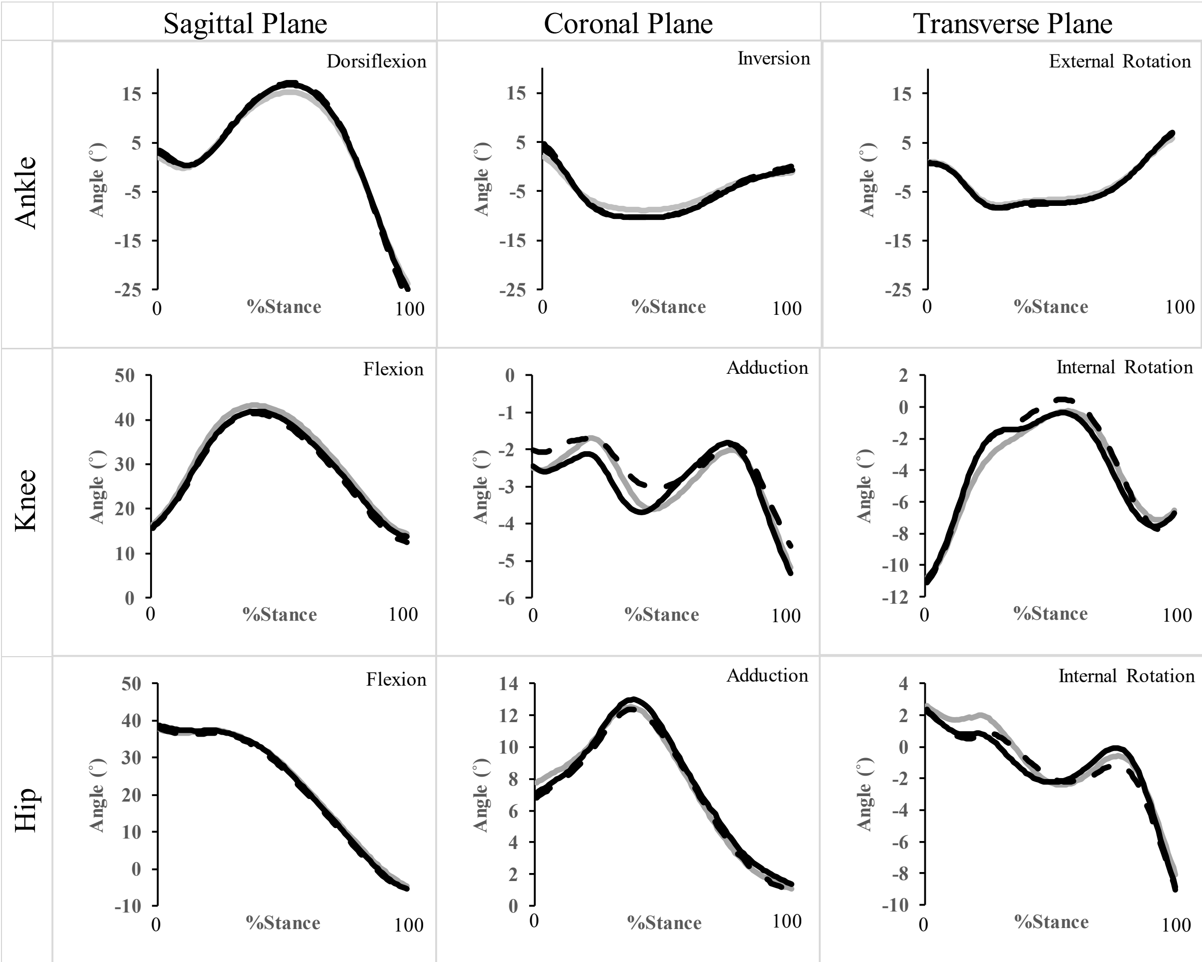

Figure 3.4. Mean ankle, knee, and hip kinematics during the stance phase of locomotion for the sagittal, coronal, and transverse planes from female participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

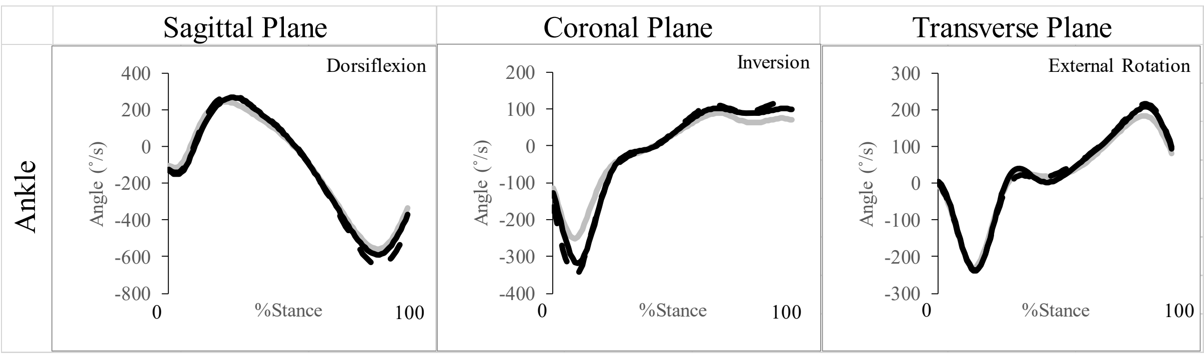

Figure 3.5. Mean ankle velocity during the stance phase of locomotion for the sagittal, coronal, and transverse planes from female participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

Table 3.5. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained during stance phase of the running gait from female participants.

		WITHOUT		PROTECTOR		BRACE	
ANKLE	**Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)**						
	Angle at footstrike (˚)	2.89 ± 10.07		4.11 ± 9.69		1.89 ± 8.96	B
	Angle at toe-off	-27.35 ± 4.04		-25.07 ± 5.05	A	-23.86 ± 4.62	AB
	Peak dorsiflexion	17.47 ± 3.11		16.98 ± 2.99		15.47 ± 3.06	AB
	Absolute ROM	44.83 ± 3.05		41.99 ± 4.52	A	39.33 ± 3.70	AB
	Relative ROM	14.58 ± 8.82		12.87 ± 8.57		13.58 ± 7.26	
	Peak dorsifelxion velocity (˚/s)	348.29 ± 123.05		339.88 ± 119.96	A	313.78 ± 117.76	A
	Peak plantarflexion velocity (˚/s)	-651.55 ± 113.69		-593.63 ± 111.23	A	-563.13 ± 93.06	AB
	Coronal plane (+ = Inversion / - = Eversion)						
	Angle at footstrike	4.54 ± 3.48		3.48 ± 3.59		1.87 ± 2.87	AB
	Angle at toe-off	0.09 ± 4.30		-0.70 ± 4.50		-1.19 ± 3.66	
	Peak inversion	5.01 ± 2.98		4.13 ± 3.12		2.30 ± 2.62	AB
	Peak eversion	-10.96 ± 1.90		-10.90 ± 2.18		-9.26 ± 1.72	AB
	Absolute ROM	15.97 ± 2.92		15.03 ± 3.33		11.57 ± 2.43	AB
	Relative ROM	15.50 ± 3.51		14.37 ± 3.85	A	11.13 ± 2.75	AB
	Peak inversion velocity (˚/s)	159.90 ± 35.99		140.67 ± 31.61	A	122.58 ± 31.08	A
	Peak eversion velocity (˚/s)	-402.02 ± 108.78		-369.25 ± 116.82		-292.14 ± 81.80	AB
	Transverse plane (+ = External / - = Internal)						
	Angle at footstrike	0.83 ± 3.99		0.70 ± 4.40		1.10 ± 3.53	
	Angle at toe-off	7.12 ± 3.40		6.83 ± 4.25		5.80 ± 3.51	A
	Peak rotation	-9.77 ± 3.79		-9.58 ± 4.05		-8.68 ± 3.79	A
	Absolute ROM	16.96 ± 4.91		16.53 ± 4.40		14.58 ± 3.95	AB
	Relative ROM	10.59 ± 5.22		10.27 ± 4.94		9.78 ± 4.39	
	Peak external rotation velocity (˚/s)	246.05 ± 45.40		236.96 ± 41.38		213.21 ± 36.14	AB
	Peak internal rotation velocity (˚/s)	-286.52 ± 117.44		-274.89 ± 85.53		-267.11 ± 98.70	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

For the ankle joint, in the sagittal plane, significant main effects were found for the angle at footstrike $F_{(2, 22)} = 5.69$, $P \leq 0.01$, $p\eta^2 = 0.34$, angle at toe-off $F_{(2, 22)} = 27.94$, $P \leq 0.01$, $p\eta^2=0.72$, peak dorsiflexion angle $F_{(2, 22)} = 20.45$, $P \leq 0.01$, $p\eta^2=0.65$, absolute ROM $F_{(2, 22)} = 40.32$, $P \leq 0.01$, $p\eta^2=0.80$, peak dorsiflexion velocity $F_{(2, 22)} = 9.95$, $P \leq 0.01$, $p\eta^2=0.48$, and peak plantarflexion velocity $F_{(2, 22)} = 58.25$, $P \leq 0.01$, $p\eta^2=0.84$. Post-hoc analysis revealed that the BRACE condition exhibited significantly ($P \leq 0.01$, 95%CI: 0.73-3.71%) lower angle at footstrike than the PROTECTOR condition. All three conditions were significantly different from each other for angle at toe off with the WITHOUT condition having the largest angle at toe off and the BRACE condition having the smallest angle at toe off (WITHOUT &

PROTECTOR $P \leq 0.01$, 95%CI: 0.70-3.87%, WITHOUT & BRACE $P \leq 0.001$, 95%CI: 2.22-4.76%, and PROTECTOR & BRACE $P \leq 0.05$, 95%CI: 0.09-2.32%). The BRACE condition significantly reduced peak dorsiflexion when compared to the WITHOUT ($P \leq 0.001$, 95%CI: 0.93-3.09%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.52-2.51%) conditions. Again all three conditions were significantly different from each other for absolute range of motion with the WITHOUT condition having the most ROM and BRACE condition having the least ROM (WITHOUT & PROTECTOR $P \leq 0.01$, 95%CI: 1.15-4.52%, WITHOUT & BRACE $P \leq 0.001$, 95%CI: 3.89-7.10%, and PROTECTOR & BRACE $P \leq 0.01$, 95%CI: 0.95-4.37%). For the ankle velocities the BRACE ($P \leq 0.001$, 95%CI: 15.52-53.49%) and PROTECTOR ($P \leq 0.01$, 95%CI: 7.58-44.62%) condition significantly reduced the dorsiflexion velocity compared to the WITHOUT condition. The WITHOUT condition had significantly more plantarflexion velocity compared to the PROTECTOR ($P \leq 0.001$, 95%CI: 35.35-80.50%) and BRACE ($P \leq 0.001$, 95%CI: 63.46-113.39%) conditions. Also the PROTECTOR condition had significantly more plantarflexion velocity than the BRACE ($P \leq 0.01$, 95%CI: 7.70-53.30%) condition.

For the ankle joint, in the coronal plane, significant main effects were found for the angle at footstrike $F_{(2, 22)} = 15.73$, $P \leq 0.01$, $p\eta^2$=0.59, peak inversion angle $F_{(2, 22)}$ =18.16, $P \leq 0.01$, $p\eta^2$=0.62, peak eversion angle $F_{(2, 22)} = 22.77$, $P \leq 0.01$, $p\eta^2$=0.67, absolute ROM $F_{(2, 22)}$ =43.65, $P \leq 0.01$, $p\eta^2$=0.80, relative ROM $F_{(2, 22)} = 38.45$, $P \leq 0.01$, $p\eta^2$=0.78, peak inversion velocity $F_{(1.10, 12.13)} = 8.33$, $P \leq 0.01$, $p\eta^2$=0.43, and peak eversion velocity $F_{(2, 22)} = 24.63$, $P \leq 0.01$, $p\eta^2$=0.69. Post-hoc analysis revealed that the BRACE condition significantly reduced angle at footstrike when compared to the PROTECTOR ($P \leq 0.05$, 95%CI: 0.25-2.97%) and WITHOUT ($P \leq 0.001$, 95%CI: 1.33-4.01%) conditions. The BRACE condition also significantly reduced both peak inversion and peak eversion when compared to both the PROTECTOR (INV $P \leq 0.01$, 95%CI: 0.66-3.00% & EVE $P \leq 0.001$, 95%CI: 0.77-2.50%)

and WITHOUT (INV $P \leq 0.001$, 95%CI: 1.40-4.01% & EVE $P \leq 0.001$, 95%CI: 0.94-2.45%) conditions. It was also revealed that the BRACE condition had significantly lower absolute and relative ROM's when compared to both the WITHOUT (ABS $P \leq 0.001$, 95%CI: 2.78-6.02% & REL $P \leq 0.001$, 95%CI: 2.69-6.04%) and PROTECTOR (ABS $P \leq 0.001$, 95%CI: 2.09-4.83% & REL $P \leq 0.001$, 95%CI: 1.67-4.81%) conditions. The PROTECTOR condition was also significantly ($P \leq 0.05$, 95%CI: 0.07-2.18%) lower than the WITHOUT condition for relative ROM. For the ankle velocities the BRACE ($P \leq 0.05$, 95%CI: 4.14-70.49%) and PROTECTOR ($P \leq 0.001$, 95%CI: 9.74-28.72%) conditions significantly reduced inversion velocity when compared to the WITHOUT condition. The BRACE condition significantly reduced eversion velocity compared to the WITHOUT ($P \leq 0.001$, 95%CI: 59.08-160.69%) and PROTECTOR ($P \leq 0.001$, 95%CI: 35.77-118.46%) conditions.

For the ankle joint, in the transverse plane, significant main effects were found for the angle at toe-off $F_{(2, 22)} = 5.47$, $P \leq 0.01$, $p\eta^2=0.33$, peak rotation $F_{(2, 22)} = 4.76$, $P \leq 0.01$, $p\eta^2=0.30$, absolute ROM $F_{(2, 22)} =20.01$, $P \leq 0.01$, $p\eta^2=0.65$, and peak external rotation velocity $F_{(2, 22)} = 9.31$, $P \leq 0.01$, $p\eta^2=0.46$. Post-hoc analysis revealed that the BRACE condition exhibited significantly lower angle at toe off ($P \leq 0.01$, 95%CI: 0.32-2.33%) and peak rotation ($P \leq 0.05$, 95%CI: 0.21-1.95%) than the WITHOUT condition. The BRACE condition also significantly reduced Absolute ROM when compared to both PROTECTOR ($P \leq 0.001$, 95%CI: 0.90-2.99%) and WITHOUT ($P \leq 0.001$, 95%CI: 1.09-3.66%) conditions. For the Peak external rotation velocity, the BRACE again was significantly lower than both the PROTECTOR ($P \leq 0.05$, 95%CI: 3.20-44.31%) and WITHOUT ($P \leq 0.01$, 95%CI: 8.74-56.94%) conditions.

Table 3.6. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained during stance phase of the running gait from female participants.

		WITHOUT		PROTECTOR		BRACE	
KNEE	**Sagittal plane (+ = Flexion / - = Extension)**						
	Angle at footstrike	15.74 ± 4.73		15.92 ± 4.39		16.47 ± 5.30	
	Angle at toe-off	12.67 ± 4.41		13.82 ± 3.89		14.46 ± 4.97	
	Peak flexion	41.84 ± 3.52		42.38 ± 3.36		43.53 ± 3.82	AB
	Absolute ROM	30.29 ± 3.46		30.00 ± 2.29		30.83 ± 3.36	
	Relative ROM	26.10 ± 4.92		26.46 ± 4.42		27.06 ± 5.00	
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at footstrike	-2.02 ± 3.75		-2.44 ± 4.18		-2.49 ± 4.17	
	Angle at toe-off	-4.51 ± 2.82		-5.34 ± 3.08		-5.18 ± 2.91	
	Peak adduction	0.97 ± 4.57		0.43 ± 4.76		0.55 ± 4.87	
	Absolute ROM	6.85 ± 3.33		6.92 ± 2.53		6.85 ± 2.50	
	Relative ROM	2.99 ± 2.54		2.87 ± 1.78		3.04 ± 2.02	
	Transverse plane (+ = Internal / - = External)						
	Angle at footstrike	-11.12 ± 5.60		-10.93 ± 5.15		-10.82 ± 6.10	
	Angle at toe-off	-7.41 ± 6.26		-6.69 ± 6.42		-6.53 ± 6.82	
	Peak rotation	2.51 ± 4.37		1.56 ± 4.24		1.43 ± 5.18	
	Absolute ROM	14.65 ± 4.74		13.49 ± 4.45		13.48 ± 4.39	
	Relative ROM	13.62 ± 4.53		12.49 ± 4.58		12.25 ± 4.42	
HIP		**WITHOUT**		**PROTECTOR**		**BRACE**	
	Sagittal plane (+ = Flexion / - = Extension)						
	Angle at footstrike	37.79 ± 9.18		38.13 ± 7.98		37.51 ± 7.81	
	Angle at toe-off	-6.03 ± 5.66		-5.78 ± 5.81		-4.73 ± 5.94	
	Peak flexion	39.91 ± 8.36		40.03 ± 7.07		40.18 ± 7.19	
	Absolute ROM	45.94 ± 5.03		45.72 ± 5.04		44.90 ± 4.76	
	Relative ROM	43.82 ± 5.81		43.81 ± 6.39		42.23 ± 6.13	
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at footstrike	6.84 ± 4.04		7.00 ± 5.12		7.58 ± 5.65	
	Angle at toe-off	0.71 ± 5.44		1.36 ± 5.80		1.07 ± 5.78	
	Peak adduction	13.01 ± 4.23		13.50 ± 4.31		13.33 ± 4.99	
	Absolute ROM	12.79 ± 3.98		12.53 ± 4.19		12.88 ± 4.35	
	Relative ROM	6.17 ± 3.32		6.50 ± 3.57		5.74 ± 3.69	
	Transverse plane (+ = Internal / - = External)						
	Angle at footstrike	2.37 ± 4.74		2.25 ± 4.36		2.46 ± 4.41	
	Angle at toe-off	-8.79 ± 8.56		-8.66 ± 7.35		-8.02 ± 7.68	
	Peak rotation	-9.55 ± 8.07		-9.26 ± 7.20		-8.63 ± 7.46	
	Absolute ROM	15.16 ± 3.88		15.04 ± 4.49		14.52 ± 4.36	
	Relative ROM	11.91 ± 5.15		11.51 ± 5.93		11.09 ± 4.93	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

For the knee a significant main effect was found in the sagittal plane for peak flexion $F_{(2, 22)} = 10.03$, $P \leq 0.01$, $p\eta^2=0.48$. Post-hoc analysis revealed a greater knee flexion in the BRACE condition than both the PROTECTOR (P=0.00911, 95%CI: %) and WITHOUT (P=0.006462, 95%CI: %) conditions. No significant differences (P > 0.05) were found in the coronal or transverse planes for the knee joint or in any of the planes of motion for the hip joint.

3.5 Discussion

The aim of the current study was to investigate the effects of ankle protectors on ankle kinematics during the stance phase during running, compare the effects of ankle protectors with braced and unbraced ankles to establish which it more closely resembles, investigate the effects of ankle protectors on knee and hip kinematics, and investigate the effects on both male and female populations. Previous research reviewing the effectiveness of ankle braces has found them to reduce the risk of inversion injury (Farwell, et al., 2013) and it is a reduction in coronal plane kinematics which is likely the main contributor to the reduction in risk of inversion injuries (Tang, et al., 2010). Ankle protectors aim to reduce contusion injuries and have previously been found to be effective at this (Ankrah & Mills, 2004). However, it was previously unknown whether ankle protectors inadvertently restrict the ankle during running, due to its location, which may cause restrictions similar to ankle braces.

3.5.1 Discussion of male results

The ankle braces used in the current study showed similar results to previous studies by reducing the motion of the ankle in the coronal plane (Martin & Harter, 1993; Tamura, et al., 2017). In particular, the peak inversion and peak inversion velocity, which are key contributors to inversion injuries were reduced (Kristianslund, et al., 2011). This makes them a good reference frame to compare the restrictive properties of ankle protectors to when investigating the effects on running. When comparing the coronal plane ankle kinematics using ankle protectors to braced and unbraced ankles it can be concluded that ankle protectors do not significantly restrict the ankle in the coronal plane and replicate similar movement to that of an ankle free of orthotic support in male populations. The lack of restriction is likely due to the soft foam construct of the ankle protector which is far less rigid than the plastic polymer contained within the ankle brace. This rigidness that is the main contributor to the ankle braces efficiency at restricting the ankle. Therefore, ankle protectors do not offer the benefits of protecting against ankle-inversion injuries like ankle braces during running for male populations.

However, the sagittal plane results produced some interesting observations. The angle at toe off was significantly reduced in the BRACED & PROTECTOR conditions when compared to the WITHOUT condition. Also absolute ROM was reduced in these conditions too, these results suggest that there is an impedance on the ankle when wearing ankle protectors and ankle braces. These findings are interesting because previous research has not found any reductions in sagittal plane ankle kinematics when using a semi-rigid ankle brace like the one used in the current study but did find a reduction in lace-up ankle braces (Tamura, et al., 2017). The reduction in movement in this plane might be due to the way both the ankle braces and ankle protectors sit on the ankle. The ankle braces have a support strap that runs around the front and rear of the ankle which allows the brace to be tightened. The tightening of this strap is likely to reduce the movement of the ankle by restricting the ankle in the sagittal plane. As for the ankle protector, although the soft foam is designed not to come all the way over the front of the foot, on many of the participants the foam did encroach on the front of the foot due to its “one size fits all” design. The location of the foam at the front of the ankle joint could possibly explain the reduction of sagittal plane movement when wearing the ankle protector. Reductions in ankle motion in the sagittal plane has previously been shown to increase energy expenditure (Huang, et al., 2015) and so these reductions in ankle ROM seen in the current study could suggest that ankle protectors and ankle braces could cause earlier onset of fatigue for a wearer during prolong use such as during competitive match play. Additionally, there could also be an effect on performance of movements such as vertical jumps in which sagittal plane motion of the ankle is a contributor to the performance of the height attained (Smith, et al., 2016). This is beyond the scope of the current study but should be investigated further.

Previous research has shown some ankle devices alter knee and hip kinematics, which could increase the likelihood of sustaining an injury higher up the kinematic chain (Ota, et al., 2014). The results of the current study found that knee and hip kinematics were not significantly different between the test conditions. The implementation of the ankle braces and ankle protectors used in the current study do not increase the risk of injuring the knee or hip by altering the kinematics of these locations. Additionally, similar to previous research there were no significant differences found for the ground reaction forces between any of the conditions in the current study (De Clercq, 1997; West & Campbell, 2014).

The current study has established that ankle protectors provide very little restriction to the ankle when running and do not restrict the ankle like ankle braces. Therefore, ankle protectors should only be used as a means to reduce risk of ankle-contusion injuries and not implemented as a method to reduce the risk of ankle-inversion injuries during running in male populations. It must be noted that although no restrictions were seen in the coronal plane there were reductions in sagittal plane motion for the ankle. These reductions could possibly increase energy demand needed for locomotion and affect performance of other football related movements. Therefore, further research on different football related movements is necessary to fully understand how ankle protectors affect male wearers.

3.5.2 Discussion of female results

Similar to the results of the males the results for the females found that ankle braces significantly reduced coronal plane motion when running similar to previous studies (Martin & Harter, 1993; Tamura, et al., 2017). These significant findings make them a good reference frame for which to assess the restrictive properties of ankle protectors to during running when used by females. Interestingly the ankle protector significantly reduced the peak inversion velocity and relative ROM in the female population compared to not wearing any which was not seen in the male population. This finding is interesting because previous research has shown ankle-inversion velocity is a key characteristic of ankle-inversion injury (Kristianslund, et al., 2011). Therefore, reducing this could reduce the risk of inversion injuries. It is doubtful that the rigidness of the ankle protector is the reason for this reduction as it would have likely also reduced other kinematic measures in the coronal plane. One possible explanation for this reduction could be that due to the ankle protector being designed to be 'one size fits all' and the average height of the females used is lower than the males has led to a larger portion of the lower shank and foot being covered by the ankle protector in the female population. This larger coverage might provide proprioceptive cues to the wearer, which may be beneficial to reduce the overall risk of inversion injury. This has been seen with ankle taping where the effectiveness of the tape does not exceed more than approximately fifteen minutes of use (Lohkamp, et al., 2009) but has been found to significantly reduce the risk of ankle injury when compared to not wearing any tape (Verhagen, et al., 2000). This is beyond the scope of the current investigation but one that should be researched in the future to compare inversion injury rates of players wearing ankle protector's verses players who do not wear ankle protector's. This reduction suggests that ankle protectors could reduce the risk of inversion injuries however, when comparing the ankle protectors to ankle braces it can be seen that the ankle braces significantly reduce almost all other kinematic variables in the coronal plane when compared to both wearing ankle protectors and not wearing any ankle support. Therefore, it must be concluded that the ankle protectors more closely resemble an ankle free of ankle support than one that is braced in the coronal plane when used by a female population.

Again similar to the male population some unexpected results were found in the sagittal plane where there were significant reductions in the angle at toe off, absolute ROM, peak dorsiflexion velocity, and peak plantarflexion velocity when wearing ankle protectors compared to not wearing any. Previous research has found reductions in this plane can effect performance of some sporting tasks (Smith, et al., 2016) and can also increase energy expenditure (Huang, et al., 2015) which suggests these reductions might have a negative impact on female players who use them. This reduction is likely due to the location of the foam covering the rear of the ankle as well as encroaching on the front of the ankle due to the 'one size fits all' design of the ankle protectors. On nearly all of the female participants used in the current study the foam padding came across the front of the ankle whereas the intended design is for the ankle protector foam to finish before coming around the front of the ankle which is probably one of the main contributors to the reductions in the sagittal plane motion. These findings suggest that the current design of ankle protectors require some attention as the current 'one size fits all' design may need to be reconsidered to accommodate variations in height of users. Additionally, the ankle braces used in the current study reduced sagittal plane motion during running which has not been found before when using a semi-rigid ankle brace during running (Tamura, et al., 2017). This is also likely due to the design of the ankle braces used as the braces have a support strap that runs around the front and rear of the ankle which allows the brace to be tightened. The tightening of this strap is likely to reduce the movement of the ankle by restricting the ankle in the sagittal plane. This strap is likely the reason why the ankle braces were significantly more restrictive than the ankle protectors in the sagittal plane.

Ankle protectors do not significantly alter the kinematics of the knee or hip, however, the semi-rigid ankle braces used by the current study did cause a significant increase in knee flexion when wearing them. This change in knee flexion could be compensating for the restriction in the sagittal plane motion around the ankle when using the ankle brace and has the possibility to increase the likelihood of knee injury (Ota, et al., 2014). Similar to previous research and the data recorded from the male population there were no significant differences found for the ground reaction force data between any of the conditions in the current study (West & Campbell, 2014).

The current study has established that ankle protectors provide very little restriction to the ankle when running and do not restrict the ankle like ankle braces. It must be noted that there were some restrictions in the coronal plane that could possibly contribute to the reduction in ankle-inversion injuries when wearing ankle protectors. However, these restrictions were far superior in the braced condition. Additionally, without further exploration these findings must be taken with caution as there were no restrictions found for many of the other coronal kinematic measures. Therefore, ankle protectors should only be used as a means to reduce risk of ankle-contusion injuries and not implemented as a method to reduce the risk of ankle-inversion injuries during running in female populations unless further research finds these reductions in motion for other football related motions. Also the reductions in sagittal plane motion found for the ankle could possibly increase energy demand needed for locomotion and affect performance of other football related movements when used by female wearers and so further research on different football related movements is necessary.

3.5.3 Limitations of the study

The current study has limited applicability due to the relatively comfortable jogging pace the participants ran at and further research is required to investigate the effects of ankle protectors during jumping, nonlinear motion, and kicking a football for both male and female populations before more comprehensive conclusions and recommendations can be formulated. It must be noted that the current study only investigated the kinematics of the right stance foot but symmetry between stance feet should not be assumed. Furthermore, some of the kinematic data show large standard deviations. These large deviations may be due to differing running styles exhibited by the participants, and in some cases such as the hip, due to the movement of the tightly fitted sports shorts worn by participants. Although markers affixed to the malleoli were not used to track the dynamic movement there is still a possibility that error in their application may cause errors within the data collected as they were used for defining segments in the static model. Another limitation with the marker set used is that markers were attached to the trainers of the participant and not to skin, however by removing the trainers and applying to the skin has the potential to alter the running gait of the participants and reduces the ecological validity of the test results. It must be noted though that the markers attached to the trainer might not accurately resemble the true movement of the foot contained within the trainer. Finally, the foot was considered a rigid segment which means the effects on the differing joints that make up the ankle complex cannot be individually investigated.

3.5.4 Considerations for next study

The focus of this first study has been to establish the effects of ankle protectors on a fundamental movement within football as a starting point for the thesis. Due to the study finding restrictions in the sagittal plane for both males and females there is a possibility that ankle protectors might significantly affect the performance of movements that require large sagittal plane motions such as vertical jumps. Therefore, based on this finding the next study will investigate the effects of ankle protectors on vertical jumps.

4. The effects of ankle protectors on a countermovement vertical jump (CMVJ): a comparison to braced and unbraced ankles.

4.1 Introduction

Countermovement vertical jumps (CMVJ) occur very frequently during football matches and are performed to gain an advantage over an opposing player when attempting to head a ball in flight. Ankle-inversion injuries often occur during the landing phase of a vertical jump (Andersen, et al., 2004, Bjørneboe, et al., 2014) and is the most frequent mechanism of non-contact ankle-inversion injuries (Woods, et al., 2003). Ankle-contusion injuries are infrequent during CMVJs but may occur if players come into contact with one another. The commonality of ankle-inversion injuries during the landing phase has led to it being a popular area of exploration when assessing ankle braces during a dynamic movement.

There are currently no research papers on the effects of ankle protectors on lower-limb kinematics during a CMVJ however the effects of ankle braces on ankle kinematics during landing has been well researched. The majority of studies focus on the effects of ankle braces on sagittal plane motion of the ankle during landing (Cordova, et al., 2010; DiStefano, et al., 2008; Hodgson, et al., 2005; West & Campbell, 2014) or on ground reaction forces (Riemann, et al., 2002; Williams & Riemann, 2009) whilst few have presented coronal plane motion as well (Hopper, et al., 1999; Simpson, et al., 2013; Vanwanseele, et al., 2014). Of the studies that have presented coronal plane motion two studies have found ankle braces have no effect when compared to not wearing any (Hopper, et al., 1999; Simpson, et al., 2013) whereas one study did find an effect (Vanwanseele, et al., 2014). Vanwanseele, et al. (2014) found ankle braces significantly reduce peak inversion, peak eversion, ankle ROM in the coronal plane, and peak eversion velocity when compared to not wearing any.

Ankle braces have been found to significantly decrease ankle plantarflexion at initial ground contact and reduce maximum dorsiflexion (DiStefano, et al., 2008, McCaw & Cerullo, 1999) as well as reducing the overall ROM of the ankle during the landing phase of the jump (Cordova, et al., 2010). Reducing ankle dorsiflexion and plantarflexion could affect knee and hip kinematics during the landing phase. West & Campbell (2014) found there to be no effect on knee kinematics when wearing an ankle brace however, DiStefano, et al. (2008) found that during landing a reduction in sagittal plane motion of the ankle does increase knee flexion. Cordova, et al. (2010) also found ankle braces to alter knee displacement but do not effect hip displacement. Therefore, when comparing the effects of ankle protectors to ankle braces and not wearing any during a vertical jump the knee and hip kinematics need to be explored to investigate if their implementation adversely alters the kinematics at these locations.

Sagittal plane motion of the ankle is one of the main contributors to dissipating GRFs during the landing phase of a jump (Devita & Skelly, 1992). Any restrictions in this plane of motion caused by ankle braces or ankle protectors might increase GRFs by decreasing the body's ability to deal with the forces. This change might, over time, be detrimental to the musculoskeletal structures. Both DiStefano, et al. (2008) and West & Campbell (2014) found no significant difference in peak GRF or time to peak when wearing ankle braces compared to not wearing any. Hodgson, et al. (2005) on the other hand found ankle braces to significantly increase peak GRF and loading rate and Cordova, et al. (2010) found them to significantly reduce the time to peak force. Again when comparing ankle protectors to braced and unbraced ankles considerations must be made to the effects on GRFs as a reduction in sagittal plane motion of the ankle could inadvertently increase the loading rate and peak vertical GRF which could increase the risk of overuse injuries in the lower extremities.

The research on the effects of ankle braces on vertical jump performance has been split. There are multiple studies showing that they do not significantly reduce jump height (Ambegaonkar, et al., 2011; Yaggie & Kinzey, 2001; Locke, et al., 1997) whilst others have found there to be a significant reduction in jump height (Smith, et al., 2016). Studies finding a reduction in jump height have found that the restrictions found in the sagittal plane during take-off appear to be one of the main contributing factors for reducing jump height (Smith, et al., 2016). Not all studies have found a decrease in jump height and this might be due to differing types of ankle brace used.

Ankle braces are proficient at reducing the risk of ankle-inversion injuries however it is unknown if ankle protectors' effect the ankle in the same way as a brace. Chapter 3 established that during running when wearing ankle protectors, the ankle kinematics are similar to an unbraced ankle in the coronal plane for both males and females. However, there was a significant reduction in sagittal plane motion which might affect performance of movements such as a CMVJ. Therefore, the current study aims to investigate the effects of ankle protectors on ankle kinematics during the take-off and landing phase of a CMVJ, compare the effects of ankle protectors with braced and unbraced ankles to establish which it more closely resembles, investigate the effects of ankle protectors on knee and hip kinematics, and investigate the effects of ankle protectors on jump height. Furthermore, the effects on both males and females will be investigated. It is hypothesised that ankle protectors will reduce sagittal plane ankle kinematics and reduce vertical jump height.

4.2 Methodology

4.2.1 Participants

Twelve male (aged 25.87±6.86 years, height 175.23±5.31 cm, body mass; 75.43±10.44 kg and BMI; 24.50±2.71) and twelve female (aged 21.92±2.11 years, height 165.73±4.88 cm, body mass; 62.19±7.48 kg and BMI; 22.62±2.27) participants took part in this study. Participants were recruited from local and university football teams via opportunity sampling using poster adverts. The inclusion criteria for the study was that the participants were aged between 18 and 35, regularly partake in sport and injury free at the time of testing. All participants provided written consent in line with the University of Central Lancashire's ethical panel (BuSH 183)

4.2.2 Procedure

Participants completed five maximal effort CMVJ's in three conditions; wearing ankle braces (BRACE), wearing ankle protectors (PROTECTOR) and with an uncovered ankle (WITHOUT). The order the participants performed the test conditions in was randomised. The participant started and finishing each jump with their dominant foot, the right foot for all participants, in contact with a force platform (Kistler Instruments Ltd., Alton, Hampshire) sampling at 1000 Hz, which was embedded into the floor of the biomechanics laboratory. The jump was defined as two phases; the take-off phase and landing phase. The jump phase was determined as the point at which the centre of gravity velocity (COGV) of the pelvis dropped below 0 in the Z axis (Negative value indicates the COGV is moving towards the floor) until the vertical ground reaction force fell below 20N (Sinclair, et al., 2011) which determined the point at which the participant left the force plate. The landing phase was determined as the first point after the jump phase at which the vertical ground reaction force exceeded 20N (Sinclair, et al., 2011) until maximum knee flexion (Yeow, et al., 2011). Kinematic data were recorded using an eight camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) tracking retro-reflective markers at a sampling rate of 250 Hz. Using the calibrated anatomical system technique (CAST) (Cappozzo, et al., 1995) the retro-reflective markers were attached to the 1st and 5th metatarsal heads, calcaneus, medial and lateral malleoli, the medial and lateral femoral epicondyles, the greater trochanter, Left and right anterior superior iliac spine, and left and right posterior superior iliac spine. These markers were used to model the right foot, shank, thigh, and pelvis segments in six degrees of freedom. Rigid plastic mounts with four markers on each were also attached to the shank and thigh and were secured using elasticated bandage. These were used as tracking markers for the shank and thigh segments. To track the foot the 1st and 5th metatarsal heads and the calcaneus were used and to track the pelvis the left and right anterior superior iliac spine, and left and right posterior superior iliac spine were used. Before dynamic trials were captured in each condition a static trial of the participant stood in the anatomical position was captured.

4.2.3 Ankle braces and ankle protectors

The ankle protectors used for the current investigation were a pair of Nike ankle shield 10 (Nike Inc, Washington County, Oregon, USA) and the ankle braces used were a pair of Aircast A60 (DJO, Vista, CA, USA).

4.2.4 Data processing

Anatomical and tracking markers were identified within the Qualisys Track Manager software and then exported as C3D files to be analysed using Visual 3D software (C-Motion, Germantown, MD, USA). To define the centre points of the ankle and knee segments the two marker methods were utilised for both. These methods calculate the centre of the joint using the positioning of the malleoli markers for the ankle centre and the femoral epicondyle markers for the knee centre (Graydon, et al., 2015; Sinclair, et al., 2015). To calculate the hip joint centre a regression equation which uses the position of the ASIS markers was utilised (Sinclair, et al., 2014). The countermovement vertical jumps were filtered at 12Hz using a low pass 4th order zero-lag filter Butterworth filter. A cut off frequency of 12Hz was selected based on work done by Ford, et al. (2010) who used residual analysis to establish that this cut off is optimal for filtering vertical jump data. Data were split into two phases; the jump phase and landing phase and each data set was normalised to 100% of the phase. Processed trials were used to produce means of the five trials for each test condition for each participant. 3D kinematics of the ankle, knee and hip joints of the right leg were calculated using an XYZ cardan sequence of rotations. The 3D joint kinematic measures which were extracted for further analysis were 1) angle at the start of each phase, 2) angle at the end of each phase, 3) peak angles during each phase, 4) absolute range of motion (absolute ROM) for each phase which was calculated by taking the maximum angle from the minimum angle during the phase, 5), relative range of motion (relative ROM) for each phase which was calculated using the angle at the start of the phase and the first peak value after the start of the phase, 6), peak angular velocities for the ankle in each phase. Measures taken from the force plate to be analysed were 1) peak forces during the take-off phase and the landing phase 2) impulse during the take-off phase, calculated as the area under the force-time curve 3) instantaneous loading rate during the landing phase, calculated as the maximum increase in vertical force between frequency intervals, 4) average loading rate during the landing phase, calculated by dividing the peak vertical impact force by the time to the impacts peak 5) ground contact time during the take-off phase and the landing phase. The force data were normalised to bodyweights (BWs) for each participant by dividing values obtained by the participant's mass to allow comparisons across the data set to be investigated. Additionally, the height attained during each jump was calculated by subtracting the position of the pelvis in the vertical plane at the point of initiation from the maximum height of the pelvis in the vertical plane.

4.2.5 Statistical analyses

Data analysis was conducted using SPSS v22.0 (SPSS Inc., Chicago, IL, USA). The three test conditions were compared using repeated measures ANOVA's with significant findings, accepted at $P<0.05$ level, being further explored using post-hoc pairwise comparisons. Effect sizes were determined using partial Eta^2 ($p\eta^2$).

4.3 Male Results

Tables 4.1-4.6 and figures 4.1-4.4 present the key parameters of interest obtained during a countermovement vertical jump from the male participants.

4.3.1.1 Vertical jump height

Table 4.1. Maximum jump height attained in metres for the male participants.

Without		Protector		Brace	
0.46 ± 0.10		0.45 ± 0.11		0.44 ± 0.10	A

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

A significant main effect was found between jump heights $F_{(2, 22)} = 6.66$, $P \leq 0.01$, $p\eta^2 = 0.52$. Post-hoc analysis found that the BRACE condition significantly lowered jump height when compared to the WITHOUT condition ($P \leq 0.05$, 95%CI: 0.00-0.05%).

4.3.1.2 Kinetic and temporal parameters during take-off phase of the countermovement vertical jump

Table 4.2. Kinetic and temporal variables (means and standard deviations) obtained during the take-off phase of the countermovement vertical jump for the male participants.

	WITHOUT		PROTECTOR		BRACE	
Peak Vertical Force (BW)	1.23 ± 0.12		1.30 ± 0.13		1.31 ± 0.14	
Peak Anterior Force (BW)	0.16 ± 0.04		0.16 ± 0.04		0.16 ± 0.04	
Peak Posterior Force (BW)	0.06 ± 0.02		0.06 ± 0.03		0.06 ± 0.03	
Peak Medial Force (BW)	0.01 ± 0.01		0.01 ± 0.01		0.01 ± 0.01	
Peak Lateral Force (BW)	0.16 ± 0.04		0.16 ± 0.05		0.16 ± 0.04	
Impulse (BW.s)	0.53 ± 0.07		0.51 ± 0.06		0.52 ± 0.08	
Ground Contact Time (s)	0.75 ± 0.08		0.72 ± 0.08		0.73 ± 0.13	

The kinetic and temporal variables exhibited no significant differences ($P > 0.05$) between the WITHOUT, PROTECTOR, and BRACE conditions during the take-off phase.

4.3.1.3 3D Kinematic parameters during the take-off phase

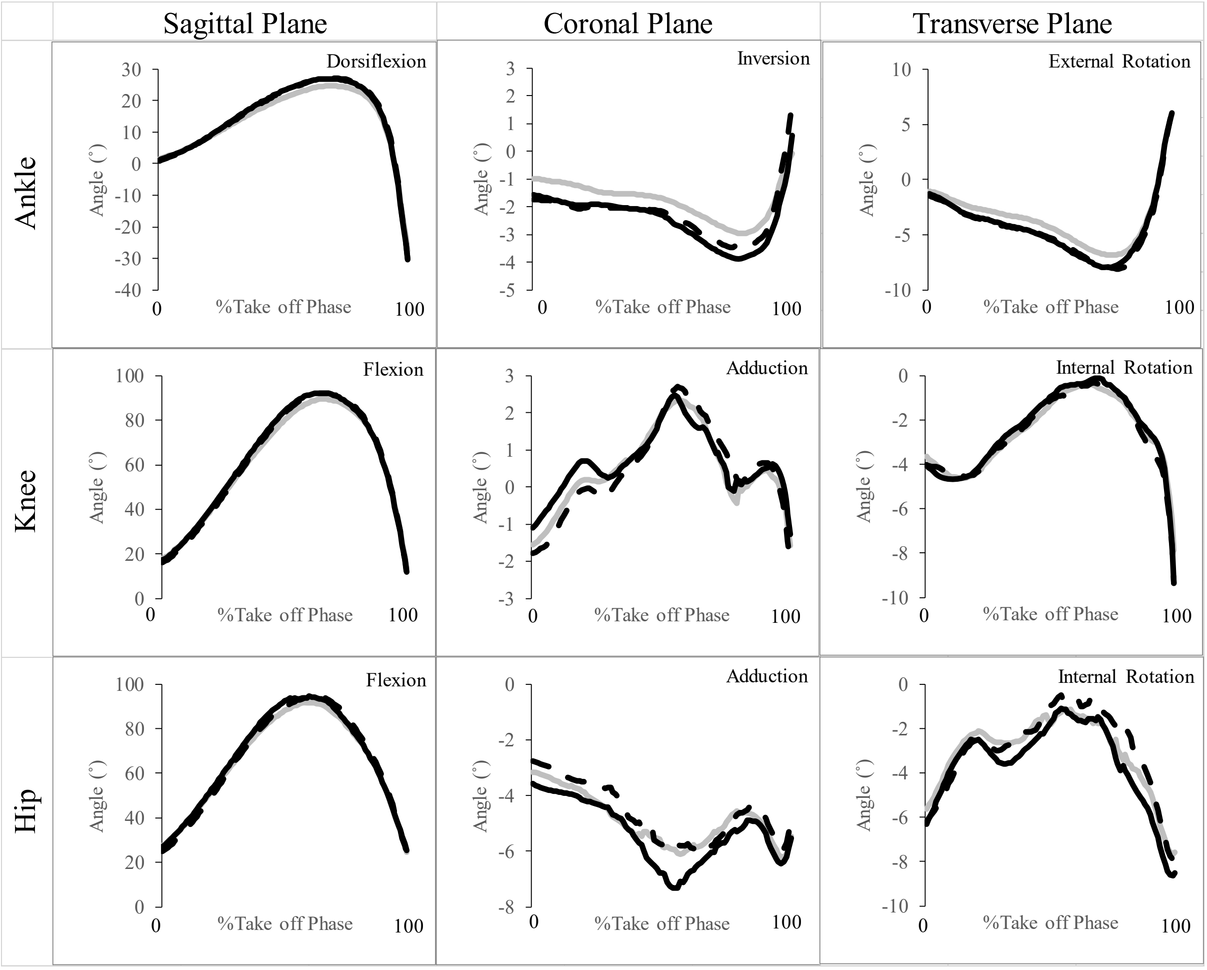

Figure 4.1. Mean ankle, knee, and hip kinematics during the take-off phase for the sagittal, coronal, and transverse planes for the male participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

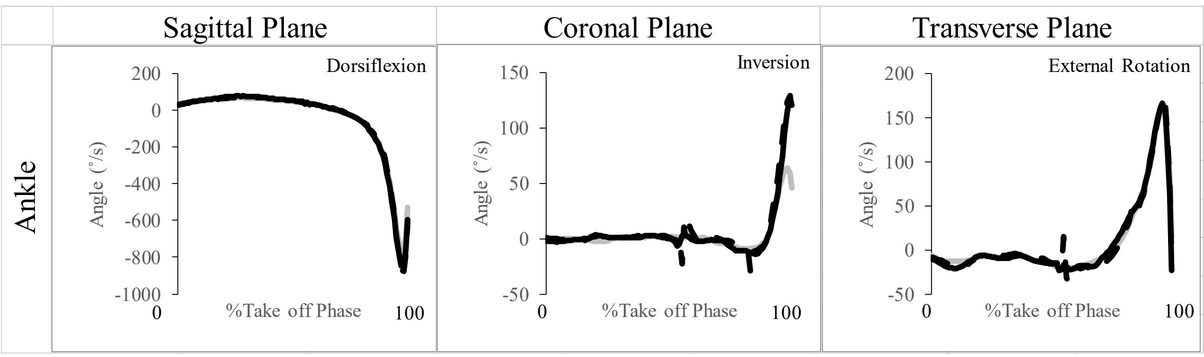

Figure 4.2. Mean ankle velocity during the take-off phase for the sagittal, coronal, and transverse planes for the male participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

Table 4.3. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained during the take-off phase for the male participants.

		WITHOUT		PROTECTOR		BRACE	
Ankle Angles	**Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)**						
	Angle at initiation	1.00 ± 2.52		1.31 ± 2.99		1.54 ± 3.07	
	Angle at take off	-30.56 ± 4.56		-30.11 ± 5.23		-27.43 ± 5.50	AB
	Peak dorsiflexion	27.89 ± 4.61		27.35 ± 3.71		25.33 ± 3.75	AB
	Absolute ROM	58.45 ± 4.58		57.46 ± 5.35		52.76 ± 6.18	AB
	Relative ROM	26.89 ± 3.27		26.05 ± 2.61		23.78 ± 2.31	AB
	Peak dorsifelxion velocity (˚/s)	96.15 ± 32.74		96.10 ± 29.28		84.86 ± 16.50	
	Peak plantarflexion velocity (˚/s)	-900.57 ± 138.27		-868.67 ± 150.90		-800.05 ± 155.27	AB
	Coronal plane (+ = Inversion / - = Eversion)						
	Angle at initiation	-1.75 ± 3.28		-1.57 ± 2.87		-1.00 ± 2.61	
	Angle at take off	1.74 ± 4.60		0.57 ± 4.60		-0.08 ± 3.95	A
	Peak inversion	2.95 ± 4.11		2.04 ± 3.89		1.63 ± 3.04	
	Peak eversion	-5.16 ± 2.95		-5.53 ± 2.56		-4.50 ± 2.74	
	Absolute ROM	8.11 ± 3.95		7.57 ± 3.85		6.13 ± 2.82	A
	Relative ROM	3.41 ± 1.99		3.96 ± 2.41		3.50 ± 1.74	
	Peak inversion velocity (˚/s)	176.70 ± 115.68		163.53 ± 112.36		107.94 ± 76.31	AB
	Peak eversion velocity (˚/s)	-70.54 ± 87.56		-53.35 ± 35.08		-51.66 ± 29.89	
	Transverse plane (+ = External / - = Internal)						
	Angle at initiation	-1.48 ± 2.58		-1.28 ± 2.80		-1.04 ± 2.34	
	Angle at take off	6.12 ± 2.93		6.04 ± 3.30		5.91 ± 3.51	
	Peak rotation	-8.72 ± 3.66		-8.57 ± 3.43		-7.76 ± 3.45	AB
	Absolute ROM	15.06 ± 2.98		14.84 ± 2.73		13.90 ± 2.26	
	Relative ROM	7.23 ± 2.21		7.28 ± 2.63		6.72 ± 2.37	
	Peak external rotation velocity (˚/s)	226.42 ± 101.11		194.36 ± 37.74		187.47 ± 45.54	
	Peak internal rotation velocity (˚/s)	-109.39 ± 91.06		-95.64 ± 67.76		-75.79 ± 61.22	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

For the ankle joint, in the sagittal plane, significant main effects were found for the angle at take-off $F_{(2, 22)} = 8.31$, $P \leq 0.01$, $p\eta^2 = 0.43$, peak dorsiflexion $F_{(1.37, 14.69)} = 9.11$, $P \leq 0.01$, $p\eta^2 = 0.45$, absolute ROM $F_{(1.31, 14.41)} = 12.85$, $P \leq 0.01$, $p\eta^2 = 0.54$, relative ROM $F_{(2, 22)} = 10.17$, $P \leq 0.01$, $p\eta^2 = 0.48$, and peak plantarflexion velocity $F_{(1.24, 13.66)} = 17.16$, $P \leq 0.01$, $p\eta^2 = 0.61$. Post-hoc analysis found that the angle at take-off was significantly lower in the BRACE condition when compared to both the WITHOUT ($P \leq 0.01$, 95%CI: 0.63-5.61%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.96-4.39%) conditions. Peak dorsiflexion was also significantly lower in the BRACE condition when compared to both the WITHOUT ($P \leq 0.05$, 95%CI: 0.23-4.89%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.52-3.54%) conditions. Again the

BRACE condition exhibited significantly less absolute ROM and relative ROM when compared to the WITHOUT (ABS $P \leq 0.01$, 95%CI: 1.28-10.09% & REL $P \leq 0.01$, 95%CI: 0.65-5.55%) and PROTECTOR (ABS $P \leq 0.001$, 95%CI: 2.36-7.04% & REL $P \leq 0.05$, 95%CI: 0.27-4.25%) conditions. For the peak plantarflexion velocity again the BRACE condition was significantly lower than the WITHOUT ($P \leq 0.01$, 95%CI: 37.79-163.25%) and PROTECTOR ($P \leq 0.001$, 95%CI: 42.86-94.38%) conditions.

For the ankle joint, in the coronal plane, significant main effects were found for angle at take-off $F_{(2, 22)} = 8.40$, $P \leq 0.01$, $p\eta^2 = 0.43$, absolute ROM $F_{(2, 22)} = 7.97$, $P \leq 0.01$, $p\eta^2 = 0.42$, and peak inversion velocity $F_{(2, 22)} = 18.65$, $P \leq 0.01$, $p\eta^2 = 0.63$. Post-hoc analysis revealed the BRACE condition significantly reduced angle at take-off ($P \leq 0.05$, 95%CI: 0.25-3.38%) and absolute ROM ($P \leq 0.05$, 95%CI: 0.18-3.77%) when compared to the WITHOUT condition. The BRACE condition also significantly reduced peak inversion velocity when compared to both the WITHOUT ($P \leq 0.001$, 95%CI: 28.99-108.53%) and PROTECTOR ($P \leq 0.001$, 95%CI: 23.61-87.57%) conditions.

For the ankle joint, in the transverse plane, a significant main effect was found for peak rotation $F_{(2, 22)} = 8.20$, $P \leq 0.01$, $p\eta^2 = 0.43$. Post-hoc analysis revealed that the BRACE condition significantly reduced rotation when compared to the WITHOUT ($P \leq 0.05$, 95%CI: 0.08-1.84%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.33-1.29%) conditions.

Table 4.4. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained during the take-off phase for the male participants.

		WITHOUT	PROTECTOR	BRACE
Knee Angles	**Sagittal plane (+ = Flexion / - = Extension)**			
	Angle at initiation	16.03 ± 7.68	17.18 ± 8.59	17.81 ± 8.27
	Angle at take off	11.07 ± 8.05	12.07 ± 8.53	13.41 ± 9.11
	Peak flexion	93.56 ± 8.04	93.23 ± 7.44	91.61 ± 8.79
	Absolute ROM	84.39 ± 5.43	83.60 ± 4.48	81.01 ± 7.65
	Relative ROM	77.54 ± 9.73	76.06 ± 9.97	73.80 ± 10.61
	Coronal plane (+ = Adduction / - = Abduction)			
	Angle at initiation	-1.79 ± 3.34	-1.09 ± 3.34	-1.55 ± 3.51
	Angle at take off	-2.00 ± 2.64	-1.28 ± 2.42	-1.59 ± 2.64
	Peak adduction	5.47 ± 5.05	5.36 ± 4.98	5.13 ± 5.63
	Absolute ROM	10.01 ± 3.76	9.54 ± 3.71	9.62 ± 4.76
	Relative ROM	7.26 ± 4.78	6.45 ± 4.93	6.68 ± 5.51
	Transverse plane (+ = Internal / - = External)			
	Angle at initiation	-4.05 ± 5.14	-4.03 ± 5.99	-3.65 ± 5.46
	Angle at take off	-9.35 ± 5.31	-9.37 ± 5.74	-7.86 ± 6.09
	Peak rotation	2.36 ± 5.50	2.50 ± 5.80	2.52 ± 5.72
	Absolute ROM	12.67 ± 6.34	12.71 ± 6.28	11.72 ± 6.06
	Relative ROM	6.41 ± 5.81	6.53 ± 6.11	6.17 ± 6.19
Hip Angles		**WITHOUT**	**PROTECTOR**	**BRACE**
	Sagittal plane (+ = Flexion / - = Extension)			
	Angle at initiation	24.78 ± 13.55	27.12 ± 16.37	27.61 ± 13.92
	Angle at take off	23.92 ± 11.09	25.30 ± 10.73	24.42 ± 12.21
	Peak flexion	96.02 ± 12.52	94.13 ± 14.54	93.80 ± 15.95
	Absolute ROM	76.62 ± 12.31	73.46 ± 12.77	73.78 ± 15.28
	Relative ROM	71.24 ± 16.74	67.01 ± 18.54	66.19 ± 20.04
	Coronal plane (+ = Adduction / - = Abduction)			
	Angle at initiation	-2.75 ± 3.06	-3.58 ± 3.21	-3.16 ± 3.43
	Angle at take off	-4.94 ± 3.25	-5.54 ± 3.45	-5.46 ± 3.66
	Peak adduction	-0.87 ± 3.74	-1.60 ± 2.97	-1.16 ± 3.51
	Absolute ROM	7.74 ± 2.76	7.45 ± 2.65	7.38 ± 2.25
	Relative ROM	5.87 ± 3.57	5.47 ± 3.21	5.38 ± 2.77
	Transverse plane (+ = Internal / - = External)			
	Angle at initiation	-6.23 ± 7.89	-6.32 ± 8.26	-5.64 ± 8.08
	Angle at take off	-7.96 ± 6.98	-8.50 ± 8.14	-7.58 ± 7.28
	Peak rotation	-9.94 ± 7.65	-10.84 ± 7.86	-10.22 ± 7.32
	Absolute ROM	12.06 ± 3.71	12.56 ± 3.75	11.99 ± 3.82
	Relative ROM	8.34 ± 3.91	8.04 ± 4.10	7.40 ± 3.44

No significant differences (P > 0.05) were found in in any of the planes of motion for both the knee joint or the hip joint.

4.3.2.1 Kinetic and temporal parameters during landing phase of the countermovement vertical jump

Table 4.5. Kinetic and temporal variables (means and standard deviations) obtained during the landing phase of the countermovement vertical jump for the male participants.

	WITHOUT		**PROTECTOR**		**BRACE**	
Peak Vertical Force (BW)	2.12 ± 0.58		2.14 ± 0.62		2.10 ± 0.59	
Peak Anterior Force (BW)	0.51 ± 0.14		0.51 ± 0.16		0.51 ± 0.14	
Peak Posterior Force (BW)	0.55 ± 0.19		0.55 ± 0.18		0.52 ± 0.18	
Peak Medial Force (BW)	0.06 ± 0.04		0.07 ± 0.07		0.05 ± 0.05	
Peak Lateral Force (BW)	0.30 ± 0.17		0.30 ± 0.16		0.25 ± 0.11	
Instantaneous Loading Rate (BW.s)	244.60 ± 149.27		260.84 ± 152.30		231.99 ± 106.87	
Average Loading Rate (BW.s)	57.67 ± 62.22		64.09 ± 62.99		46.28 ± 26.67	
Ground Contact Time (s)	0.18 ± 0.06		0.17 ± 0.05		0.17 ± 0.06	

The kinetic and temporal variables exhibited no significant differences (P > 0.05) between the WITHOUT, PROTECTOR, and BRACE conditions during the landing phase.

4.3.2.2 3D Kinematic parameters during the landing phase

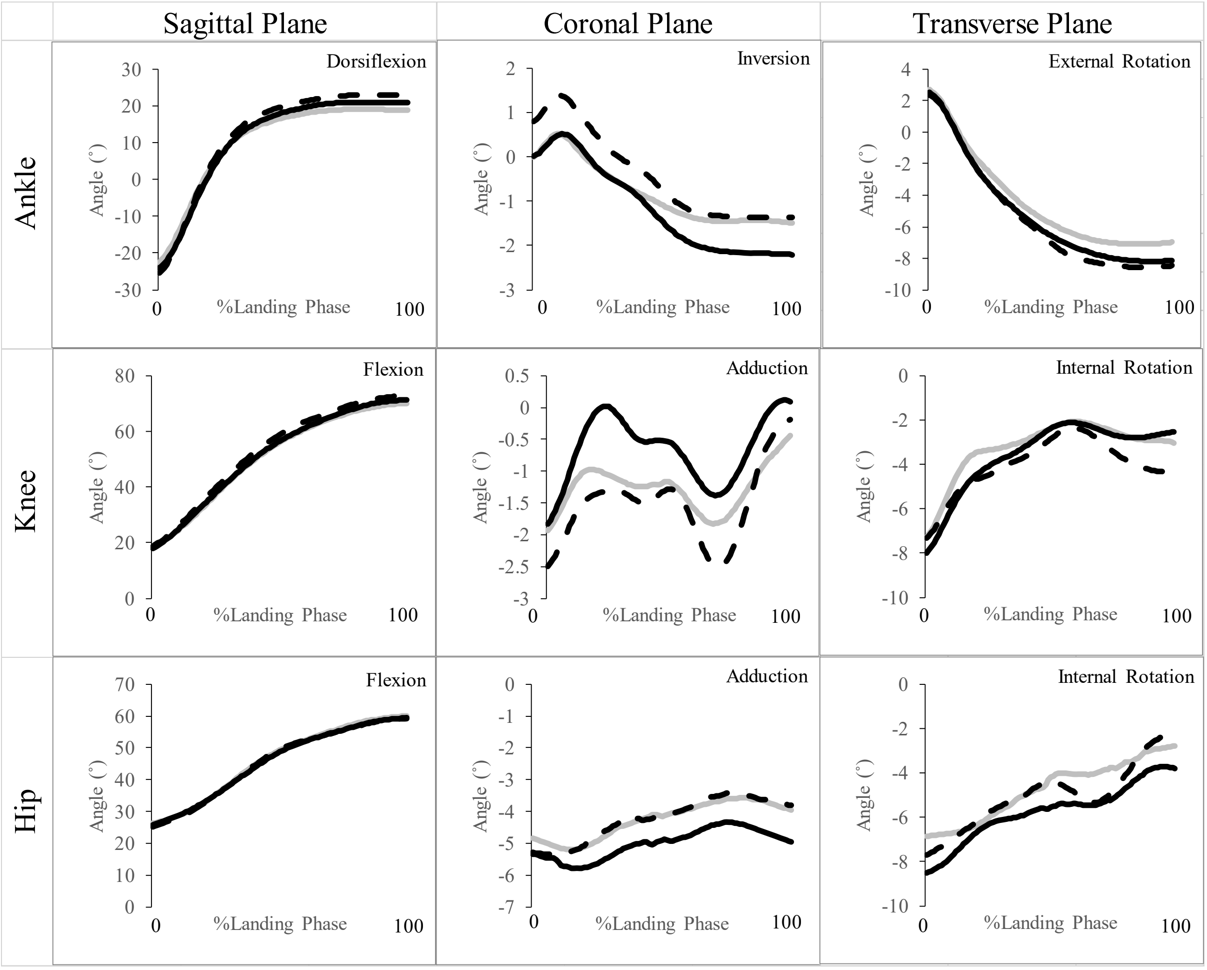

Figure 4.3. Mean ankle, knee, and hip kinematics during the landing phase for the sagittal, coronal, and transverse planes for the male participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

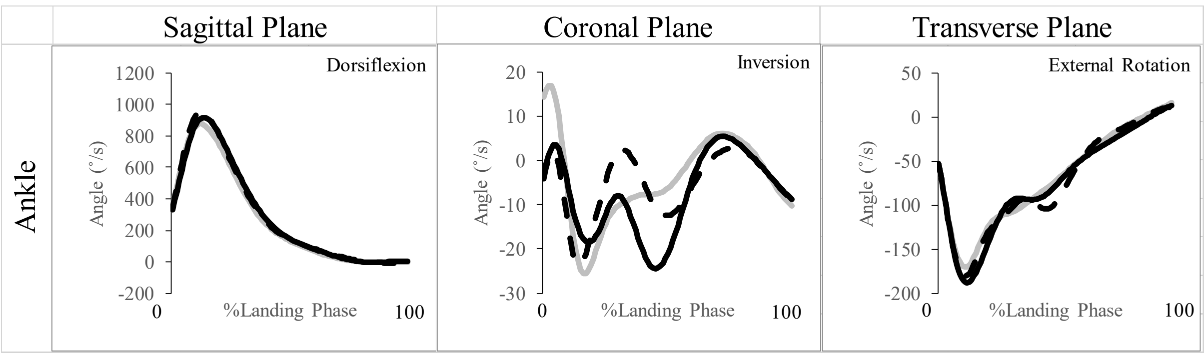

Figure 4.4. Mean ankle velocity during the landing phase for the sagittal, coronal, and transverse planes for the male participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

Table 4.6. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained during landing phase for the male participants.

		WITHOUT		PROTECTOR		BRACE	
Ankle Angles	**Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)**						
	Angle at impact	-25.47 ± 10.44		-24.11 ± 12.25		-22.49 ± 8.67	
	Angle at max knee flexion	22.85 ± 5.54		20.94 ± 5.19		18.90 ± 3.79	A
	Peak dorsiflexion	23.62 ± 4.85		21.76 ± 4.67		19.78 ± 3.31	A
	Absolute ROM	49.13 ± 9.63		45.88 ± 10.93		42.27 ± 7.12	A
	Relative ROM	49.10 ± 9.73		45.88 ± 10.93		42.27 ± 7.12	A
	Peak dorsifelxion velocity (˚/s)	1065.25 ± 212.48		1005.96 ± 248.53		988.04 ± 144.53	
	Peak plantarflexion velocity (˚/s)	-37.53 ± 50.78		-36.31 ± 50.45		49.75 ± 45.21	
	Coronal plane (+ = Inversion / - = Eversion)						
	Angle at impact	0.80 ± 4.86		0.02 ± 4.42		-0.01 ± 3.50	
	Angle at max knee flexion	-1.36 ± 2.83		-2.20 ± 3.17		-1.49 ± 3.34	
	Peak inversion	3.46 ± 3.34		2.41 ± 3.26		1.78 ± 3.19	A
	Peak eversion	-3.23 ± 3.62		-3.86 ± 3.51		-2.82 ± 3.10	
	Absolute ROM	6.69 ± 2.57		6.26 ± 1.72		4.60 ± 0.82	AB
	Relative ROM	4.03 ± 3.52		3.88 ± 2.78		2.82 ± 1.89	
	Peak inversion velocity (˚/s)	116.50 ± 90.16		112.47 ± 70.56		89.83 ± 47.51	
	Peak eversion velocity (˚/s)	-123.67 ± 60.12		-121.99 ± 50.93		-102.94 ± 37.42	
	Transverse plane (+ = External / - = Internal)						
	Angle at impact	2.50 ± 4.22		2.36 ± 3.75		2.69 ± 3.23	
	Angle at max knee flexion	-8.47 ± 3.80		-8.14 ± 4.19		-6.97 ± 4.02	AB
	Peak rotation	-9.55 ± 4.15		-9.14 ± 4.14		-8.02 ± 4.14	A
	Absolute ROM	12.35 ± 2.85		11.82 ± 2.64		10.87 ± 2.36	
	Relative ROM	12.05 ± 3.57		11.50 ± 3.46		10.71 ± 2.77	
	Peak external rotation velocity (˚/s)	65.94 ± 27.63		68.22 ± 40.40		62.57 ± 25.62	
	Peak internal rotation velocity (˚/s)	-250.88 ± 68.08		-248.19 ± 62.37		-232.61 ± 69.90	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

For the ankle joint, in the sagittal plane, significant main effects were found for angle at max knee flexion $F_{(2, 22)} = 9.94$, $P \leq 0.01$, $p\eta^2 = 0.48$, peak dorsiflexion $F_{(2, 22)} = 11.30$, $P \leq 0.01$, $p\eta^2 = 0.51$, absolute ROM $F_{(2, 22)} = 10.20$, $P \leq 0.01$, $p\eta^2 = 0.48$, and relative ROM $F_{(2, 22)} = 10.02$, $P \leq 0.01$, $p\eta^2 = 0.48$. Post-hoc analysis found that the BRACE condition was significantly lower than the WITHOUT condition for the angle at max knee flexion ($P \leq 0.01$, 95%CI: 1.56-6.33%), peak dorsiflexion ($P \leq 0.001$, 95%CI: 1.76-5.92%), absolute ROM ($P \leq 0.01$, 95%CI: 2.47-11.25%), and relative ROM ($P \leq 0.01$, 95%CI: 2.37-11.28%).

For the ankle joint, in the coronal plane, significant main effects were found for peak inversion $F_{(2, 22)} = 10.22$, $P \leq 0.01$, $p\eta^2 = 0.48$, and absolute ROM $F_{(2, 22)} = 9.81$, $P \leq 0.01$, $p\eta^2 = 0.47$. Post-hoc analysis revealed the BRACE condition significantly reduced peak inversion when compared to the WITHOUT ($P \leq 0.01$, 95%CI: 0.45-2.91%) condition. Also the BRACE condition significantly reduced absolute ROM when compared to both the WITHOUT ($P \leq 0.05$, 95%CI: 0.29-3.88%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.58-2.74%) conditions.

For the ankle joint, in the transverse plane, significant main effects were found for angle at max knee flexion $F_{(2, 22)} = 10.04$, $P \leq 0.01$, $p\eta^2 = 0.48$, and peak rotation $F_{(2, 22)} = 8.68$, $P \leq 0.01$, $p\eta^2 = 0.44$. Post-hoc analysis found the BRACE condition had a significantly lower angle at max knee flexion when compared to the WITHOUT ($P \leq 0.001$, 95%CI: 0.68-2.31%) and PROTECTOR ($P \leq 0.05$, 95%CI: 0.11-2.23%) conditions. Also the BRACE significantly reduced peak rotation when compared to the WITHOUT ($P \leq 0.01$, 95%CI: 0.60-2.45%) condition.

Table 4.7. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained during the landing phase for the male participants.

		WITHOUT		PROTECTOR		BRACE	
Knee Angles	**Sagittal plane (+ = Flexion / - = Extension)**						
	Angle at impact	18.78 ± 7.16		17.91 ± 6.77		18.25 ± 7.10	
	Angle at max knee flexion	72.68 ± 16.91		71.25 ± 15.89		70.08 ± 17.32	
	Absolute ROM	53.91 ± 11.18		53.34 ± 13.12		51.83 ± 13.89	
	Relative ROM	53.91 ± 11.18		53.34 ± 13.12		51.83 ± 13.89	
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at impact	-2.49 ± 3.05		-1.82 ± 3.43		-1.94 ± 3.50	
	Angle at max knee flexion	-0.20 ± 5.33		-0.10 ± 5.56		-0.43 ± 5.13	
	Peak adduction	2.08 ± 4.13		2.62 ± 4.58		1.92 ± 4.31	
	Absolute ROM	7.11 ± 2.77		6.81 ± 2.56		6.41 ± 2.35	
	Relative ROM	4.58 ± 3.15		4.44 ± 3.38		3.86 ± 3.08	
	Transverse plane (+ = Internal / - = External)						
	Angle at impact	-7.34 ± 5.84		-8.02 ± 5.86		-7.32 ± 5.24	
	Angle at max knee flexion	-4.32 ± 8.34		-2.53 ± 8.41		-3.02 ± 8.45	
	Peak rotation	-9.14 ± 6.23		-9.25 ± 6.28		-8.48 ± 5.93	
	Absolute ROM	8.42 ± 3.23		8.91 ± 4.17		8.42 ± 3.89	
	Relative ROM	6.63 ± 4.42		7.68 ± 5.15		7.26 ± 4.92	
Hip Angles		WITHOUT		PROTECTOR		BRACE	
	Sagittal plane (+ = Flexion / - = Extension)						
	Angle at impact	25.49 ± 9.49		25.83 ± 7.39		26.23 ± 10.75	
	Angle at max knee flexion	60.01 ± 23.01		59.50 ± 23.38		60.07 ± 26.65	
	Peak flexion	59.58 ± 22.52		59.90 ± 23.18		60.36 ± 26.60	
	Absolute ROM	34.29 ± 16.69		34.07 ± 19.42		34.18 ± 20.22	
	Relative ROM	34.09 ± 16.21		34.07 ± 19.42		34.14 ± 20.25	
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at impact	-5.30 ± 4.30		-5.33 ± 4.45		-4.83 ± 4.53	
	Angle at max knee flexion	-3.76 ± 5.23		-4.94 ± 6.53		-3.94 ± 5.35	
	Peak adduction	-1.99 ± 4.80		-2.41 ± 5.48		-1.58 ± 4.54	
	Absolute ROM	4.68 ± 1.72		4.80 ± 1.96		4.76 ± 1.71	
	Relative ROM	1.37 ± 1.05		1.88 ± 2.34		1.51 ± 1.48	
	Transverse plane (+ = Internal / - = External)						
	Angle at impact	-7.65 ± 6.13		-8.53 ± 7.39		-6.86 ± 6.72	
	Angle at max knee flexion	-2.84 ± 9.91		-3.84 ± 9.80		-2.78 ± 9.33	
	Peak rotation	-9.13 ± 7.06		-10.04 ± 7.83		-8.82 ± 7.30	
	Absolute ROM	9.00 ± 3.22		8.79 ± 2.64		8.63 ± 2.07	
	Relative ROM	7.52 ± 4.47		7.28 ± 3.15		6.68 ± 2.46	

No significant differences ($P > 0.05$) were found in in any of the planes of motion for both the knee joint or the hip joint during the landing phase.

4.4 Female Results

Tables 4.7-4.12 and figures 4.5-4.8 present the key parameters of interest obtained during a countermovement vertical jump from the female participants.

4.4.1.1 Vertical jump height

Table 4.8. Maximum jump height attained in metres for the female participants.

Without		Protector		Brace	
0.35 ± 0.06		0.34 ± 0.06	A	0.33 ± 0.06	A

Note. A = significant difference from WITHOUT condition, B = significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

A significant main effect was found between jump heights $F_{(2, 22)} = 9.80$, $P \leq 0.01$, $p\eta^2 = 0.47$. Post-hoc analysis found that the BRACE ($P \leq 0.01$, 95%CI: 0.01-0.03%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.00-0.02%) conditions significantly lowered jump height when compared to the WITHOUT condition.

4.4.1.2 Kinetic and temporal parameters during take-off phase of the countermovement vertical jump

Table 4.9. Kinetic and temporal variables (means and standard deviations) obtained during the take-off phase of the countermovement vertical jump for the female participants.

	WITHOUT	PROTECTOR	BRACE
Peak Vertical Force (BW)	1.15 ± 0.16	1.18 ± 0.16	1.17 ± 0.16
Peak Anterior Force (BW)	0.13 ± 0.05	0.13 ± 0.04	0.13 ± 0.05
Peak Posterior Force (BW)	0.05 ± 0.02	0.04 ± 0.02	0.04 ± 0.02
Peak Medial Force (BW)	0.01 ± 0.01	0.01 ± 0.01	0.01 ± 0.01
Peak Lateral Force (BW)	0.16 ± 0.06	0.16 ± 0.06	0.16 ± 0.06
Impulse (BW.s)	0.52 ± 0.08	0.51 ± 0.07	0.52 ± 0.08
Ground Contact Time (s)	0.77 ± 0.17	0.74 ± 0.15	0.77 ± 0.18

The kinetic and temporal variables exhibited no significant differences ($P > 0.05$) between the WITHOUT, PROTECTOR, and BRACE conditions during the take-off phase.

4.4.1.3 3D Kinematic parameters during the take-off phase

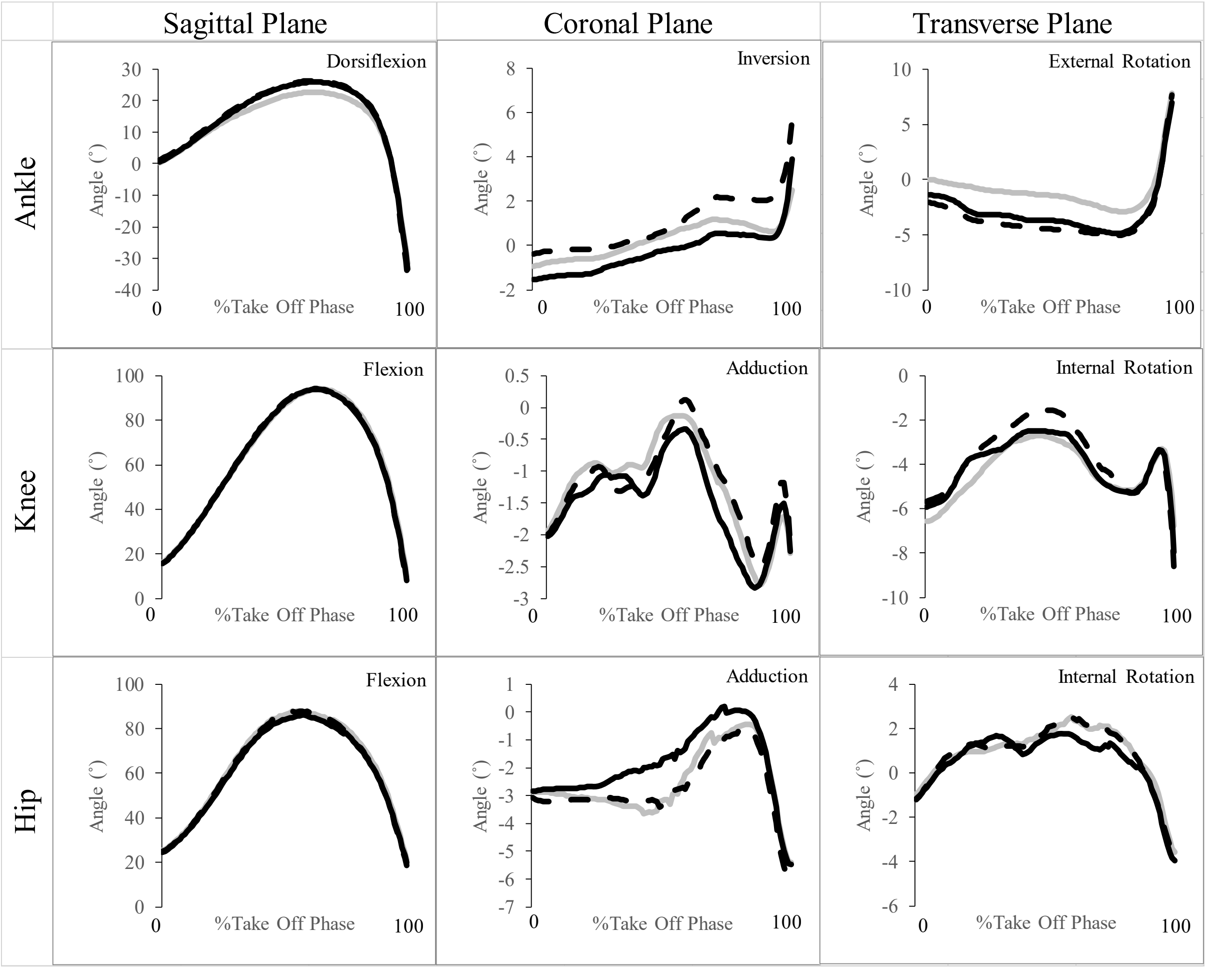

Figure 4.5. Mean ankle, knee, and hip kinematics during the take-off phase for the sagittal, coronal, and transverse planes for the female participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

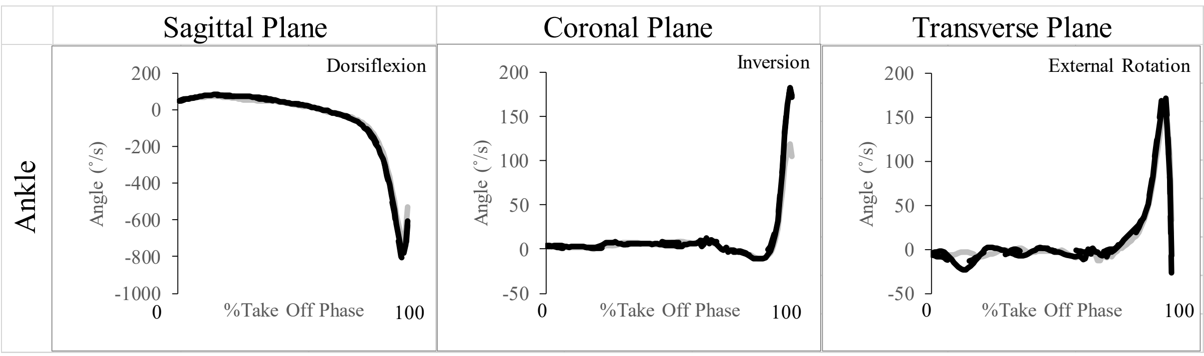

Figure 4.6. Mean ankle velocity during the take-off phase for the sagittal, coronal, and transverse planes for the female participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

Table 4.10. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained during the take-off phase for the female participants.

		WITHOUT		PROTECTOR		BRACE	
ANKLE	**Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)**						
	Angle at initiation	1.16 ± 2.79		0.49 ± 2.97		0.34 ± 3.48	
	Angle at take off	-36.38 ± 4.52		-33.44 ± 4.34	A	-31.50 ± 4.07	A
	Peak dorsiflexion	26.98 ± 6.11		26.41 ± 5.56		23.49 ± 5.25	AB
	Absolute ROM	63.36 ± 7.79		59.85 ± 7.98	A	54.99 ± 6.58	AB
	Relative ROM	25.82 ± 4.41		25.92 ± 4.04		23.15 ± 4.41	AB
	Peak dorsifelxion velocity (˚/s)	90.84 ± 15.87		99.83 ± 22.20		89.58 ± 21.00	
	Peak plantarflexion velocity (˚/s)	-839.34 ± 117.91		-794.05 ± 124.43	A	-733.10 ± 114.11	AB
	Coronal plane (+ = Inversion / - = Eversion)						
	Angle at initiation	-0.39 ± 2.57		-1.52 ± 3.07		-0.94 ± 2.72	
	Angle at take off	5.68 ± 4.05		3.96 ± 4.31	A	2.49 ± 3.29	AB
	Peak inversion	6.64 ± 3.57		4.80 ± 3.57	A	3.74 ± 2.80	A
	Peak eversion	-1.57 ± 2.94		-2.80 ± 3.45	A	-1.99 ± 3.28	B
	Absolute ROM	8.21 ± 2.40		7.60 ± 2.53		5.73 ± 1.94	AB
	Relative ROM	1.18 ± 1.22		1.28 ± 1.58		1.05 ± 1.21	
	Peak inversion velocity (˚/s)	196.15 ± 75.23		197.35 ± 78.51		132.70 ± 78.97	AB
	Peak eversion velocity (˚/s)	-42.72 ± 25.28		-45.53 ± 31.29		-41.94 ± 29.44	
	Transverse plane (+ = External / - = Internal)						
	Angle at initiation	-2.08 ± 1.58		-1.38 ± 1.57	A	0.00 ± 1.69	AB
	Angle at take off	7.62 ± 4.04		7.03 ± 4.42		7.82 ± 4.23	
	Peak rotation	-6.91 ± 2.29		-6.46 ± 2.78		-4.55 ± 2.28	AB
	Absolute ROM	14.75 ± 3.95		13.70 ± 3.70		12.73 ± 3.80	A
	Relative ROM	4.82 ± 2.50		5.08 ± 2.67		4.55 ± 1.82	
	Peak external rotation velocity (˚/s)	207.92 ± 55.40		191.01 ± 46.45		189.26 ± 56.39	A
	Peak internal rotation velocity (˚/s)	-106.76 ± 65.24		-91.68 ± 41.70		-75.59 ± 40.89	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

For the ankle joint, in the sagittal plane, significant main effects were found for angle at take-off $F_{(2, 22)} = 19.15$, $P \leq 0.01$, $p\eta^2 = 0.64$, peak dorsiflexion $F_{(2, 22)} = 28.58$, $P \leq 0.01$, $p\eta^2 = 0.72$, absolute ROM $F_{(2, 22)} = 72.18$, $P \leq 0.01$, $p\eta^2 = 0.87$, relative ROM $F_{(1.20, 13.23)} = 10.73$, $P \leq 0.01$, $p\eta^2 = 0.49$, and peak plantarflexion velocity $F_{(2, 22)} = 72.66$, $P \leq 0.01$, $p\eta^2 = 0.87$. Post-hoc analysis found that the WITHOUT condition had a significantly greater angle at take-off than the PROTECTOR ($P \leq 0.05$, 95%CI: 0.46-5.42%) and BRACE ($P \leq 0.001$, 95%CI: 2.74-7.02%) conditions. The BRACE condition significantly reduced peak dorsiflexion compared to the WITHOUT ($P \leq 0.001$, 95%CI: 1.79-5.20%) and PROTECTOR ($P \leq 0.001$, 95%CI: 0.48-1.63%) conditions. All three conditions were significantly different from each other for absolute ROM with the WITHOUT condition having the largest ROM and the BRACE condition having the least ROM (WITHOUT & PROTECTOR $P \leq 0.001$, 95%CI: 1.53-5.50%, WITHOUT & BRACE $P \leq 0.001$, 95%CI: 6.38-10.37%, and PROTECTOR & BRACE $P \leq 0.001$, 95%CI: 2.92-6.80%). The BRACE condition significantly reduced relative ROM when compared to both the WITHOUT ($P \leq 0.05$, 95%CI: 0.17-5.18%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.88-4.65%) conditions. For peak plantarflexion velocity all three conditions were significantly different from each other with the WITHOUT condition having the highest velocity and the BRACE condition having the least velocity (WITHOUT & PROTECTOR $P \leq 0.001$, 95%CI: 20.46-70.11%, WITHOUT & BRACE $P \leq 0.001$, 95%CI: 79.08-133.39%, and PROTECTOR & BRACE $P \leq 0.001$, 95%CI: 38.31-83.58%).

For the ankle joint, in the coronal plane, significant main effects were found for angle at take-off $F_{(2, 22)} = 22.11$, $P \leq 0.01$, $p\eta^2 = 0.67$, peak inversion $F_{(2, 22)} = 18.45$, $P \leq 0.01$, $p\eta^2 = 0.63$, peak eversion $F_{(2, 22)} = 6.67$, $P \leq 0.01$, $p\eta^2 = 0.38$, absolute ROM $F_{(2, 22)} = 19.76$, $P \leq 0.01$, $p\eta^2 = 0.64$, and peak inversion velocity $F_{(2, 22)} = 48.63$, $P \leq 0.01$, $p\eta^2 = 0.82$. Post-hoc analysis found all three conditions were significantly different from each other for angle at take-off with the WITHOUT condition having the largest angle and the BRACE condition having the smallest angle (WITHOUT & PROTECTOR $P \leq 0.001$, 95%CI: 0.81-2.62%, WITHOUT & BRACE $P \leq 0.001$, 95%CI: 1.54-4.84%, and PROTECTOR & BRACE $P \leq 0.05$, 95%CI: 0.08-2.87%). The WITHOUT condition had a significantly larger peak inversion than the PROTECTOR ($P \leq 0.001$, 95%CI: 0.86-2.82%) and BRACE ($P \leq 0.001$, 95%CI: 1.31-4.48%) conditions. The PROTECTOR condition exhibited a greater peak eversion angle when compared to both the WITHOUT ($P \leq 0.05$, 95%CI: 0.21-2.25%) and BRACE ($P \leq 0.05$, 95%CI: 0.02-1.60%) conditions. For the absolute ROM the BRACE condition significantly reduced ROM when compared to both the WITHOUT ($P \leq 0.001$, 95%CI: 1.07-3.89%) and

PROTECTOR ($P \leq 0.01$, 95%CI: 0.62-3.13%) conditions. For peak inversion velocity the BRACE condition had a significantly lower velocity than both the WITHOUT ($P \leq 0.001$, 95%CI: 43.78-83.12%) and PROTECTOR ($P \leq 0.001$, 95%CI: 39.17-90.11%) conditions.

For the ankle joint, in the transverse plane, significant main effects were found for angle at initiation $F_{(1.24, 13.60)} = 28.63$, $P \leq 0.01$, $p\eta^2 = 0.72$, peak rotation $F_{(2, 22)} = 26.57$, $P \leq 0.01$, $p\eta^2 = 0.71$, absolute ROM $F_{(2, 22)} = 7.54$, $P \leq 0.01$, $p\eta^2 = 0.41$, and peak external rotation velocity $F_{(2, 22)} = 4.27$, $P < 0.05$, $p\eta^2 = 0.28$. Post-hoc analysis found all three conditions were significantly different from each other for angle at initiation with the WITHOUT condition having the more internal rotation and the BRACE condition having the least rotation (WITHOUT & PROTECTOR $P \leq 0.01$, 95%CI: 0.18-1.23%, WITHOUT & BRACE $P \leq 0.001$, 95%CI: 1.04-3.13%, and PROTECTOR & BRACE $P \leq 0.001$, 95%CI: 0.67-2.09%). The BRACE condition had a significantly lower peak rotation than the WITHOUT ($P \leq 0.001$, 95%CI: 1.42-3.29%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.72-3.10%) conditions. For the absolute ROM the BRACE condition significantly reduced ROM when compared to the WITHOUT ($P \leq 0.01$, 95%CI: 0.58-3.45%) condition. For peak external rotation velocity, the BRACE condition had a significantly lower velocity than the WITHOUT ($P \leq 0.05$, 95%CI: 2.98-34.34%) condition.

Table 4.11. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained during the take-off phase for the female participants.

		WITHOUT		PROTECTOR		BRACE	
KNEE	**Sagittal plane (+ = Flexion / - = Extension)**						
	Angle at initiation	15.74 ± 6.27		15.73 ± 7.11		16.25 ± 7.12	
	Angle at take off	6.09 ± 4.48		8.21 ± 4.51	A	9.94 ± 4.99	AB
	Peak flexion	95.76 ± 13.04		94.76 ± 13.64		95.13 ± 12.47	
	Absolute ROM	89.59 ± 15.63		87.47 ± 15.39		86.25 ± 14.09	
	Relative ROM	79.45 ± 12.81		79.05 ± 13.32		78.89 ± 12.20	
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at initiation	-1.99 ± 2.78		-2.09 ± 3.13		-1.93 ± 3.27	
	Angle at take off	-2.17 ± 2.70		-2.33 ± 2.92		-2.29 ± 2.86	
	Peak adduction	2.67 ± 3.52		2.14 ± 4.20		2.42 ± 3.86	
	Absolute ROM	7.94 ± 2.34		7.76 ± 1.79		7.77 ± 1.89	
	Relative ROM	4.67 ± 2.68		4.23 ± 2.12		4.35 ± 2.24	
	Transverse plane (+ = Internal / - = External)						
	Angle at initiation	-5.66 ± 3.83		-5.96 ± 3.75		-6.56 ± 4.19	
	Angle at take off	-8.60 ± 3.25		-7.90 ± 4.11		-6.76 ± 3.27	
	Peak rotation	0.67 ± 4.17		0.61 ± 4.05		0.34 ± 3.43	
	Absolute ROM	11.67 ± 5.02		11.05 ± 4.30		10.76 ± 3.48	
	Relative ROM	6.33 ± 3.81		6.56 ± 2.90		6.90 ± 3.64	
HIP		**WITHOUT**		**PROTECTOR**		**BRACE**	
	Sagittal plane (+ = Flexion / - = Extension)						
	Angle at initiation	24.31 ± 8.45		24.65 ± 8.39		25.29 ± 8.90	
	Angle at take off	18.57 ± 4.84		19.94 ± 5.12		21.04 ± 5.04	
	Peak flexion	89.40 ± 13.84		87.58 ± 14.65		89.73 ± 13.73	
	Absolute ROM	71.84 ± 13.01		69.25 ± 13.19		70.32 ± 12.42	
	Relative ROM	65.09 ± 15.69		62.92 ± 16.69		64.44 ± 16.49	
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at initiation	-3.10 ± 2.79		-2.82 ± 2.71		-2.87 ± 3.20	
	Angle at take off	-5.95 ± 3.59		-5.42 ± 3.55		-5.42 ± 4.28	
	Peak adduction	0.72 ± 3.71		1.56 ± 3.76		0.83 ± 3.53	
	Absolute ROM	8.08 ± 2.24		8.07 ± 2.43		8.05 ± 3.02	
	Relative ROM	4.27 ± 2.11		3.70 ± 2.14		4.36 ± 3.26	
	Transverse plane (+ = Internal / - = External)						
	Angle at initiation	-1.20 ± 5.72		-1.16 ± 4.33		-0.96 ± 4.77	
	Angle at take off	-3.71 ± 5.90		-4.04 ± 4.77		-3.59 ± 5.18	
	Peak rotation	-5.71 ± 5.11		-5.83 ± 3.90		-5.39 ± 4.25	
	Absolute ROM	11.49 ± 3.15		10.78 ± 3.42		11.16 ± 4.53	
	Relative ROM	6.98 ± 4.31		6.11 ± 2.99		6.72 ± 4.71	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

For the knee a significant main effect was found in the sagittal plane for angle at take-off $F_{(2, 22)} = 31.50$, $P \leq 0.01$, $p\eta^2=0.74$. Post-hoc analysis revealed all three conditions were significantly different from each other with BRACE condition exhibiting the largest angle and the WITHOUT condition exhibiting the smallest angle (WITHOUT & PROTECTOR $P \leq 0.01$,

95%CI: 0.79-3.45%, WITHOUT & BRACE P ≤ 0.001, 95%CI: 2.55-5.15%, and PROTECTOR & BRACE P ≤ 0.05, 95%CI: 0.25-3.21%). No significant differences (P > 0.05) were found in the coronal or transverse planes. No significant differences (P > 0.05) were found in any of the planes of motion for hip joint.

4.4.2.1 Kinetic and temporal parameters during landing phase of the countermovement vertical jump

Table 4.12. Kinetic and temporal variables (means and standard deviations) obtained during the landing phase of the countermovement vertical jump for the female participants.

	WITHOUT		**PROTECTOR**		**BRACE**	
Peak Vertical Force (BW)	1.92 ± 0.47		1.94 ± 0.46		1.91 ± 0.40	
Peak Anterior Force (BW)	0.37 ± 0.12		0.37 ± 0.13		0.37 ± 0.12	
Peak Posterior Force (BW)	0.38 ± 0.24		0.42 ± 0.23		0.41 ± 0.22	
Peak Medial Force (BW)	0.00 ± 0.02		0.00 ± 0.02		0.01 ± 0.03	
Peak Lateral Force (BW)	0.26 ± 0.11		0.25 ± 0.97		0.25 ± 0.10	
Instantaneous Loading Rate (BW.s)	159.38 ± 107.47		175.99 ± 126.84		152.21 ± 99.19	
Average Loading Rate (BW.s)	49.58 ± 61.71		43.76 ± 32.61		37.14 ± 21.10	
Ground Contact Time (s)	0.22 ± 0.06		0.22 ± 0.06		0.22 ± 0.05	

The kinetic and temporal variables exhibited no significant differences (P > 0.05) between the WITHOUT, PROTECTOR, and BRACE conditions.

4.4.2.2 3D Kinematic parameters during the landing phase

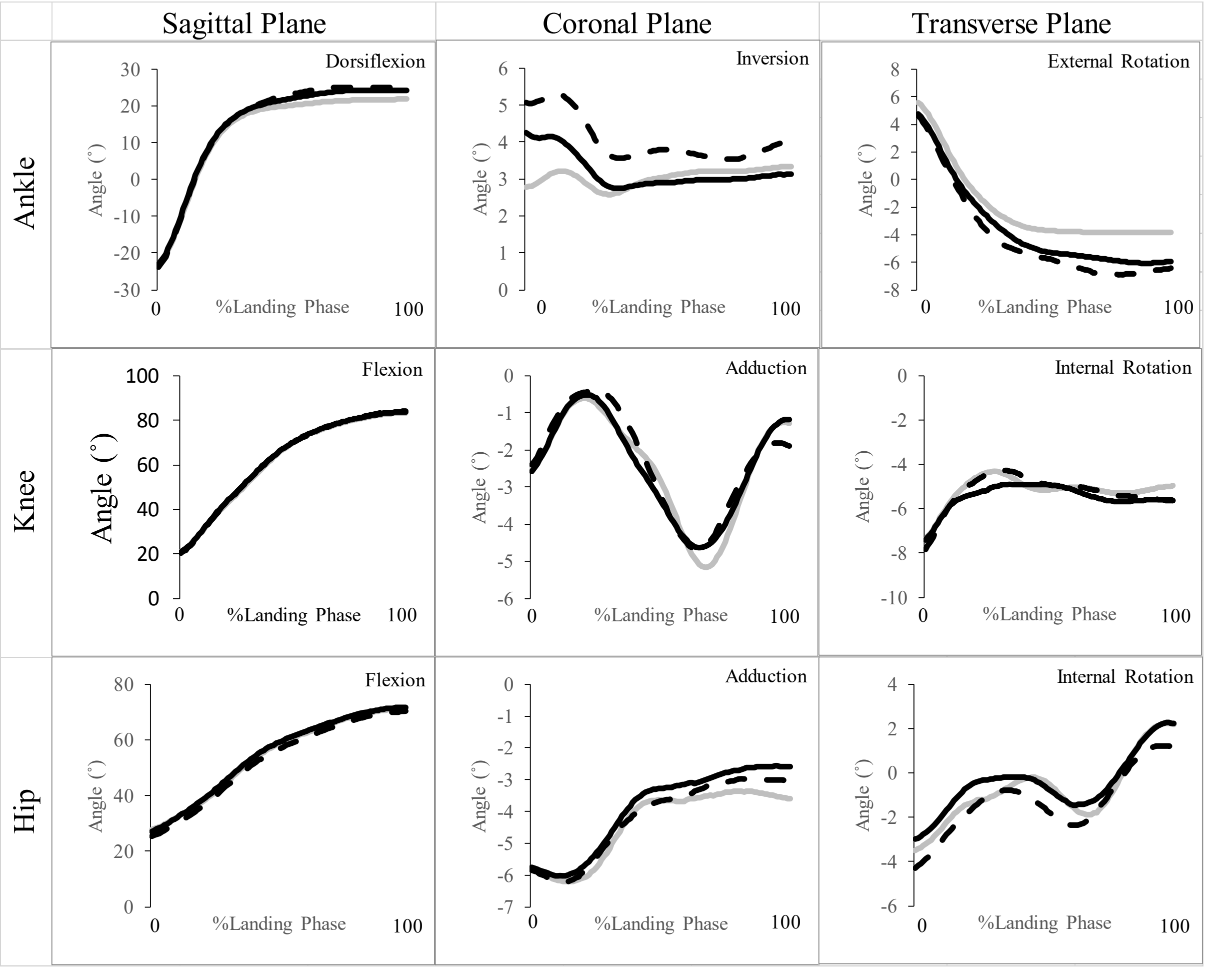

Figure 4.7. Mean ankle, knee, and hip kinematics during the landing phase for the sagittal, coronal, and transverse planes for the female participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

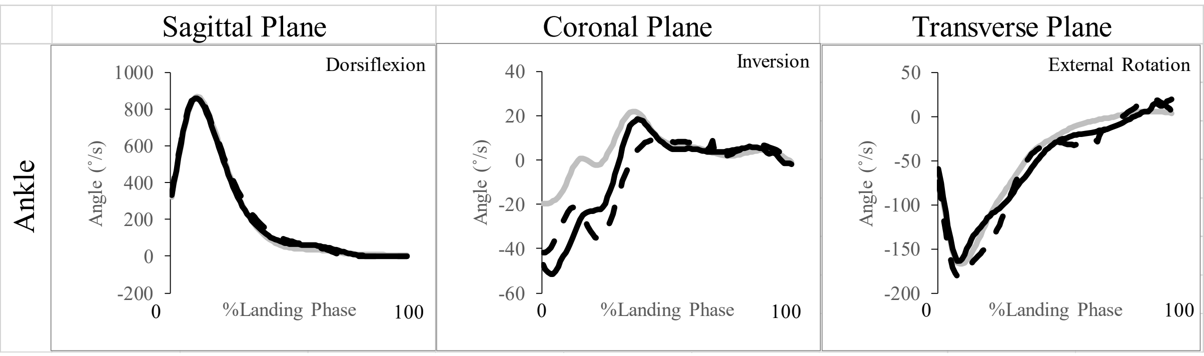

Figure 4.8. Mean ankle velocity during the landing phase for the sagittal, coronal, and transverse planes for the female participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

Table 4.13. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained during landing phase for the female participants.

		WITHOUT		PROTECTOR		BRACE	
ANKLE	**Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)**						
	Angle at impact	-23.90 ± 9.57		-22.97 ± 9.60		-23.13 ± 8.12	
	Angle at max knee flexion	25.02 ± 4.77		24.21 ± 4.23		21.80 ± 3.49	AB
	Peak dorsiflexion	25.73 ± 4.70		25.02 ± 4.19		22.57 ± 3.74	AB
	Absolute ROM	49.64 ± 9.83		47.99 ± 10.44		45.70 ± 8.90	
	Relative ROM	49.64 ± 9.83		47.99 ± 10.44		45.70 ± 8.90	
	Peak dorsifelxion velocity (˚/s)	932.60 ± 159.60		915.58 ± 151.26		913.28 ± 117.26	
	Peak plantarflexion velocity (˚/s)	-35.16 ± 19.38		-34.33 ± 24.91		-37.33 ± 30.73	
	Coronal plane (+ = Inversion / - = Eversion)						
	Angle at impact	5.08 ± 5.18		4.33 ± 4.91		2.78 ± 3.97	A
	Angle at max knee flexion	4.03 ± 4.00		3.19 ± 4.81		3.34 ± 3.90	
	Peak inversion	7.43 ± 4.05		6.54 ± 4.23		5.23 ± 3.05	A
	Peak eversion	1.08 ± 4.21		0.54 ± 4.17		0.76 ± 3.84	
	Absolute ROM	6.35 ± 1.69		5.99 ± 1.06		4.47 ± 1.83	AB
	Relative ROM	3.99 ± 2.38		3.79 ± 2.28		2.03 ± 1.93	AB
	Peak inversion velocity (˚/s)	82.17 ± 43.88		76.53 ± 36.78		80.71 ± 41.11	
	Peak eversion velocity (˚/s)	-112.23 ± 35.00		-111.07 ± 31.84		-73.84 ± 40.96	AB
	Transverse plane (+ = External / - = Internal)						
	Angle at impact	4.62 ± 2.89		4.77 ± 2.44		5.56 ± 2.34	
	Angle at max knee flexion	-6.44 ± 4.32		-5.97 ± 3.72		-3.84 ± 3.80	AB
	Peak rotation	-8.43 ± 3.58		-7.71 ± 3.36		-5.64 ± 2.95	AB
	Absolute ROM	13.07 ± 3.60		12.51 ± 3.01		11.23 ± 2.11	
	Relative ROM	13.06 ± 3.62		12.48 ± 3.07		11.20 ± 2.09	
	Peak external rotation velocity (˚/s)	95.94 ± 68.88		106.28 ± 64.66		90.77 ± 56.82	
	Peak internal rotation velocity (˚/s)	-251.86 ± 57.22		-237.93 ± 50.20		-230.57 ± 33.54	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions.

For the ankle joint, in the sagittal plane, significant main effects were found for angle at max knee flexion $F_{(2, 22)} = 9.36$, $P \leq 0.01$, $p\eta^2 = 0.46$, and peak dorsiflexion $F_{(2, 22)} = 11.75$, $P \leq 0.01$, $p\eta^2 = 0.52$. Post-hoc analysis revealed that the BRACE condition significantly reduced the angle at max knee flexion when compared to the WITHOUT ($P \leq 0.05$, 95%CI: 0.52-5.91%) and PROTECTOR ($P \leq 0.05$, 95%CI: 0.46-4.36%) conditions. The BRACE condition also significantly reduced peak dorsiflexion when compared to the WITHOUT ($P \leq 0.01$, 95%CI: 0.74-5.58%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.79-4.012%) conditions.

For the ankle joint, in the coronal plane, significant main effects were found for angle at impact $F_{(2, 22)} = 7.51$, $P \leq 0.01$, $p\eta^2 = 0.41$, peak inversion $F_{(2, 22)} = 8.99$, $P \leq 0.01$, $p\eta^2 = 0.45$, absolute ROM $F_{(2, 22)} = 7.29$, $P \leq 0.01$, $p\eta^2 = 0.40$, relative ROM $F_{(2, 22)} = 5.47$, $P \leq 0.01$, $p\eta^2 = 0.33$, and peak eversion velocity $F_{(2, 22)} = 7.76$, $P \leq 0.01$, $p\eta^2 = 0.41$. Post-hoc analysis found that the BRACE condition significantly reduced the angle at impact when compared to the WITHOUT ($P \leq 0.05$, 95%CI: 0.27-4.32%) condition. The BRACE condition significantly reduced peak inversion when compared to the WITHOUT ($P \leq 0.01$, 95%CI: 0.58-3.81%) condition. Also the BRACE condition significantly reduced absolute ROM and relative ROM when compared to both the WITHOUT (ABS $P \leq 0.05$, 95%CI: 0.18-3.57% & REL $P \leq 0.05$, 95%CI: 0.26-3.2%) and PROTECTOR (ABS $P \leq 0.05$, 95%CI: 0.16-2.88% & REL $P \leq 0.05$, 95%CI: 0.06-3.46%) conditions. For peak eversion velocity it was found that the BRACE condition significantly reduced the velocity compared to both the WITHOUT ($P \leq 0.05$, 95%CI: 1.84-74.96%) and PROTECTOR ($P \leq 0.01$, 95%CI: 9.17-65.29%) conditions.

For the ankle joint, in the transverse plane, significant main effects were found for angle at max knee flexion $F_{(2, 22)} = 10.58$, $P \leq 0.01$, $p\eta^2 = 0.49$, and peak rotation $F_{(2, 22)} = 14.64$, $P \leq 0.01$, $p\eta^2 = 0.57$. Post-hoc analysis revealed the BRACE condition significantly reduced the angle at max knee flexion when compared to both the WITHOUT ($P \leq 0.01$, 95%CI: 0.70-4.52%) and PROTECTOR ($P \leq 0.01$, 95%CI: 0.44-3.83%) conditions. The BRACE condition also significantly reduced peak rotation when compared to both the WITHOUT ($P \leq 0.01$, 95%CI: 1.13-4.45%) and PROTECTOR ($P \leq 0.05$, 95%CI: 0.40-3.74%) conditions.

Table 4.14. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained during the landing phase for the female participants.

		WITHOUT	PROTECTOR	BRACE
Knee Angles	**Sagittal plane (+ = Flexion / - = Extension)**			
	Angle at impact	20.19 ± 7.37	20.88 ± 7.64	20.98 ± 7.21
	Angle at max knee flexion	83.98 ± 14.08	83.94 ± 14.49	83.33 ± 13.65
	Absolute ROM	63.79 ± 11.80	63.42 ± 11.12	62.35 ± 11.22
	Relative ROM	63.79 ± 11.80	63.06 ± 11.01	62.35 ± 11.22
	Coronal plane (+ = Adduction / - = Abduction)			
	Angle at impact	-2.41 ± 2.84	-2.67 ± 3.28	-2.51 ± 3.17
	Angle at max knee flexion	-1.89 ± 4.01	-1.27 ± 4.16	-1.29 ± 4.49
	Peak adduction	1.82 ± 3.87	2.01 ± 4.28	1.65 ± 4.15
	Absolute ROM	8.89 ± 3.32	9.70 ± 3.15	9.30 ± 2.50
	Relative ROM	4.23 ± 1.77	4.69 ± 2.34	4.16 ± 1.99
	Transverse plane (+ = Internal / - = External)			
	Angle at impact	-7.84 ± 3.20	-7.39 ± 3.81	-7.39 ± 3.44
	Angle at max knee flexion	-5.63 ± 4.57	-5.71 ± 4.58	-4.97 ± 5.17
	Peak rotation	-1.78 ± 3.32	-1.44 ± 3.41	-1.34 ± 3.34
	Absolute ROM	7.75 ± 2.77	8.26 ± 2.37	7.92 ± 2.47
	Relative ROM	6.05 ± 3.66	5.95 ± 3.63	6.05 ± 3.29
Hip Angles		**WITHOUT**	**PROTECTOR**	**BRACE**
	Sagittal plane (+ = Flexion / - = Extension)			
	Angle at impact	25.29 ± 6.34	26.99 ± 6.10	27.58 ± 7.57
	Angle at max knee flexion	70.33 ± 12.00	71.89 ± 16.39	71.33 ± 12.62
	Peak flexion	70.58 ± 11.77	72.16 ± 16.22	71.61 ± 12.53
	Absolute ROM	45.29 ± 12.14	45.19 ± 13.75	44.03 ± 11.48
	Relative ROM	45.28 ± 12.14	45.18 ± 13.76	44.03 ± 11.48
	Coronal plane (+ = Adduction / - = Abduction)			
	Angle at impact	-5.84 ± 2.24	-5.75 ± 2.37	-5.86 ± 3.22
	Angle at max knee flexion	-3.03 ± 5.10	-2.48 ± 4.51	-3.60 ± 5.05
	Peak adduction	-1.16 ± 4.41	-0.50 ± 3.67	-1.40 ± 4.11
	Absolute ROM	6.37 ± 1.99	6.89 ± 2.30	6.17 ± 1.79
	Relative ROM	1.69 ± 1.62	1.64 ± 1.43	1.71 ± 2.08
	Transverse plane (+ = Internal / - = External)			
	Angle at impact	-4.28 ± 5.49	-3.03 ± 4.79	-3.48 ± 4.73
	Angle at max knee flexion	1.15 ± 4.22	2.17 ± 2.94	2.19 ± 3.93
	Peak rotation	-6.78 ± 4.45	-6.33 ± 3.85	-6.62 ± 3.68
	Absolute ROM	10.95 ± 3.89	11.62 ± 3.95	11.42 ± 3.57
	Relative ROM	8.44 ± 3.63	8.33 ± 3.65	8.28 ± 4.53

No significant differences ($P > 0.05$) were found in in any of the planes of motion for both the knee joint or the hip joint during the landing phase.

4.5 Discussion

The aim of the current study was to investigate the effects of ankle protectors on ankle kinematics during the take-off and landing phase of a countermovement vertical jump (CMVJ), compare the effects of ankle protectors with braced and unbraced ankles to establish which it more closely resembles, investigate the effects of ankle protectors on knee and hip kinematics, investigate the effects of ankle protectors on jump height, and investigate the effects on male and female populations. Previous research reviewing the effectiveness of ankle braces has found them to reduce the risk of inversion injury (Farwell, et al., 2013) and it is a reduction in coronal plane kinematics which is likely the main contributor to the reduction in risk of inversion injuries (Tang, et al., 2010). Ankle protectors aim to reduce contusion injuries and have previously been found to be effective at this (Ankrah & Mills, 2004). However, it was previously unknown whether an ankle protector inadvertently restrict the ankle during a CMVJ, due to its location, which may cause restrictions similar to ankle braces and possibly affect jump height.

4.5.1 Landing phase results

The most important phase to consider when investigating the effects of ankle protectors on ankle kinematics is the landing phase of a CMVJ as this phase is the most likely to cause an ankle-inversion injury (Bjørneboe, et al., 2014). With the kinematics in the coronal plane being the most likely to cause ankle-inversion injuries (Kristianslund, et al., 2011). Therefore, the landing phase will be the first to be discussed.

4.5.1.1 Discussion of male landing phase results

The ankle braces used by the current study produced similar results as previous studies during the landing phase of a CMVJ as they significantly reduced peak inversion and absolute ROM in the coronal plane (Vanwanseele, et al., 2014) which have been established as key parameters attributed to ankle-inversion injuries (Kristianslund, et al., 2011). Therefore, the ankle braces used by the current study produce an adequate kinematic reference frame to compare the ankle protectors to, to establish if they can reduce the risk of inversion injuries during the landing phase of a CMVJ. During the landing phase, the ankle protectors did not significantly reduce any ankle kinematics in any of the planes of motion when compared to the without condition. Similar to the findings in chapter 3 on the stance limb during running, ankle protectors do not restrict the ankle like an ankle brace and cannot protect against inversion injuries during landing from a CMVJ in male populations. The difference in restrictive properties between the ankle protectors and ankle braces is likely due to the construct of each. The soft EVA foam the ankle protectors are made from is not as rigid as the plastic polymer strips found in the medial and lateral aspects of the ankle braces. These strips are specifically designed to reduce inversion of the ankle whilst the foam in the ankle protectors is not thick enough or rigid enough to replicate these restrictions. Additionally, the use of ankle protectors by male football players does not significantly affect the kinematics of the knee or hip meaning their utilisation will not increase the risk of injury further up the lower kinematic chain during the landing phase of a CMVJ. Although the ankle braces showed significant reductions in dorsiflexion and ankle ROM in the sagittal plane these reductions did not significantly affect knee and hip kinematics either. These results mirror previous research looking at the effects of ankle orthotics on knee and hip kinematics which have also found significant reductions in sagittal plane ankle kinematics do not affect knee and hip kinematics (Cordova, et al., 2010; West & Campbell, 2014). The use of both ankle protectors and ankle braces did not significantly affect peak ground reaction forces, length of time between initial ground contact during landing and the point of maximum knee flexion during landing, or loading rate again mirroring previous research (DiStefano, et al., 2008; West & Campbell, 2014). Therefore, the use of both ankle protectors and ankle braces does not increase the risk of injury further up the kinetic chain by adversely affecting the knee and hip kinematics. It can be concluded that for male populations ankle protectors do not perform like an ankle brace and are only effective at reducing the risk of contusion injuries around the ankle during the landing phase of a CMVJ and cannot protect against ankle-inversion injuries during this manoeuvre. Furthermore, the use of ankle protectors does not significantly affect knee kinematics, hip kinematics, or GRFs and therefore their use does not increase the likelihood of injuries further up the kinematic chain during landing when used by males.

4.5.1.2 Discussion of female landing phase results

Similar to the male results it was found that the ankle braces reduced the key parameters attributed to the risk of an ankle-inversion injury (Kristianslund, et al., 2011) when used by a female population which again makes them a good reference frame to compare the ankle protectors to. During the landing phase the ankle protectors did not significantly reduce any ankle kinematics in any of the planes of motion when compared to not wearing any. Therefore, ankle protectors do not restrict the ankle like an ankle brace and cannot protect against inversion injuries during landing from a CMVJ in female populations. Again, similar to the males, it is likely the difference in constructs of the ankle protectors and ankle braces which has caused the braces to significantly reduce ankle motion but the ankle protectors not to. The use of ankle protectors by female football players does not significantly affect the kinematics of the knee or hip meaning the usage of them will not increase the risk of injury further up the lower kinematic chain during the landing phase of a CMVJ. Additionally, the use of ankle braces did not significantly affect knee and hip kinematics either. These results mirror the findings in the male population and also previous research by both DiStefano, et al. (2008) and West & Campbell (2014) who both found no effect of ankle orthotics on knee and hip kinematics during the landing phase of a drop jump. Again, the use of both ankle protectors and ankle braces did not significantly affect peak ground reaction forces, length of time between initial ground contact during landing and the point of maximum knee flexion during landing, or loading rate. Therefore, the use of either does not increase the risk of injury further up the kinematic chain by adversely affecting the knee and hip kinematics.

Similar to the male population the ankle protectors do not perform like an ankle brace and are only effective at reducing the risk of contusion injuries around the ankle during the landing phase of a CMVJ when used by female populations and cannot protect against ankle-inversion injuries during this manoeuvre. Furthermore, the use of ankle protectors does not significantly affect knee kinematics, hip kinematics, or GRFs and therefore their use does not increase the likelihood of injuries further up the kinematic chain during landing when used by females.

4.5.2 Discussion of take-off phase and jump height results

It is important to understand the effects of ankle protectors on ankle kinematics during the take-off phase as impediments in this phase might affect jump performance. Previous studies have found reductions in jump height when using ankle braces and have suggested a reduction in sagittal plane ankle kinematics to be one of the main contributors to this reduction in performance (Smith, et al., 2016). Therefore, investigating the take-off phase can offer insights into any differences found in jump height.

4.5.2.1 Discussion of male take-off phase and jump height results

The results from chapter 3 found a reduction in sagittal plane motion when wearing ankle protectors in a male population which suggested a possible adverse effect when wearing them to perform a CMVJ. The current study found that the use of ankle protectors by male football players did not significantly affect the performance of a CMVJ as there was no significant reduction in jump height when compared to not wearing any. The ankle braces used by the current study had a moderate effect on jump height attained as they significantly reduce jump height by 0.02m. During take-off the ankle protectors did not significantly change sagittal, coronal, or transverse kinematics of the ankle. Whereas the ankle braces significantly reduced sagittal plane ankle kinematics in particular plantarflexion at take-off, ankle ROM, and peak plantarflexion velocity. Previous research has found similar reductions in jump height when using ankle braces and have also found reductions in sagittal plane motion when reductions in jump height occurs (Smith, et al., 2016). These findings suggest that reductions in sagittal plane kinematics of the ankle are likely the main contributors to the reduction in vertical jump height. This could be due to the ankle not being able to travel through its full range of motion, reducing the momentum of the body being propelled upwards and into the air, ultimately reducing the total height attained. The difference in results is possibly down to the constructs of the ankle protectors and ankle braces. The ankle protectors soft foam construct appears to allow freedom of motion in the sagittal plane during a CMVJ whereas the ankle braces reduction in sagittal plane motion is likely down to the supportive strip that wraps around the ankle. The tightening of this strap appears to reduce the ankles ability to dorsiflex and plantarflex affecting jump performance. Ankle protectors and ankle braces do not significantly alter knee and hip biomechanics and therefore do not increase the risk of injury at these location during the take-off phase of a CMVJ in a male population. Additionally, the use of ankle protectors and ankle braces do not significantly affect peak ground reaction forces, length of time between initiation of a CMVJ and point of take-off, or impulse. Therefore, it can be concluded that during the take-off phase of a CMVJ ankle protectors do not adversely affect male populations who utilise them. However, the use of semi-rigid ankle braces, such as the one used in the current study, do adversely affect performance of a CMVJ.

4.5.2.2 Discussion of female take-off phase and jump height results

Again the results from chapter 3 found a reduction in sagittal plane motion when wearing ankle protectors in a female population which suggested a possible adverse effect when wearing them to perform a CMVJ. Unlike the males' the ankle protectors had a moderate effect on jump height and significantly reduced jump height for the female population when compared to the without condition. The height attained in the protector condition was similar to the braced condition. Looking at the ankle kinematics it's likely the significant reductions in the sagittal plane motion of the ankle in these conditions, in particular the range of motion of the ankle, and peak plantarflexion velocity which affected the jump height. This is supported by Smith, et al. (2016) who also found that reductions in sagittal plane motion during jumping significantly reduces jump height. The reductions in this plane were not seen for the males for the ankle protectors and so it could be speculated that the 'one size fits all' design of the ankle protectors may impede female uses. Similar to the observations in the discussion of the female running data in chapter 3 it was also observed that the 'one size fits all' design meant that the ankle protectors covered more of the ankle and lower shank of the shorter participants and did not fit as snuggly as the taller participants. On average the female participants used in the current study were shorter than the male participants, as seen in the participant information presented in section 4.2.1, and therefore more likely to be affected by the larger coverage area and less snug fit of the ankle protector. This larger coverage and less snug fit appears to reduce the ankles ability to plantarflex, as shown by the significant reduction of plantarflexion angle at take-off, possibly due to the excess foam bunching around the back of the ankle and being wedged against the rear of the trainers worn by the participant. This reduction in the ankles ability to plantarflex has reduced the absolute ROM of the ankle in this plane leading to a reduction in peak plantarflexion velocity and overall leading to a significant reduction in jump height. This affect could possibly be negated if ankle protectors of varied sizes were produced to accommodate variations in the height of football players instead of a 'one size fits all' design.

There are also significant reductions in the coronal plane during the take-off phase for both the ankle protector and ankle brace however, reductions in this plane during the take-off phase is unlikely to provide much benefit for the wearer as ankle-inversion injuries during take-off are infrequent (Bjørneboe, et al., 2014; Woods, et al., 2003). One interesting finding though is that ankle protectors and ankle braces significantly increase flexion of the knee at take-off. This increase could be compensating for the reductions in motion around the ankle joint and could suggest that more strain is put on the knee during take-off. Previous research has established that female footballers have a four times higher risk of knee and anterior cruciate ligament (ACL) injuries than males (Giza, et al., 2005) and the use of both ankle protectors and ankle braces could possibly increase the risk of these types of injuries in a female population. This is beyond the scope of this study but future research should investigate frequency of knee injuries of players who use ankle protectors and ankle braces. It should be also noted that the changes in ankle and knee kinematics when using ankle protectors and ankle braces does not significantly affect peak ground reaction forces, length of time between initiation of a CMVJ and point of take-off, or impulse.

Therefore, it can be concluded that during the take-off phase of a CMVJ ankle protectors and semi-rigid ankle braces, such as the one used in the current study, adversely affect the performance of a CMVJ and this reduction in performance is likely due to the reductions in sagittal plane motion of the ankle. Additionally, the use of ankle protectors and ankle braces significantly change knee kinematics in the sagittal plane during take-off which could possibly increase risk of knee injuries when used by a female population.

4.5.3 Limitations of the study

It must be noted that the current study only investigated the kinematics of the right stance foot but symmetry between stance feet should not be assumed. Furthermore, some of the kinematic data show large standard deviations. These large deviations may be due to differing jumping styles exhibited by the participants, and in some cases such as the hip, due to the movement of the tightly fitted sports shorts worn by participants. Although markers affixed to the malleoli were not used to track the dynamic movement there is still a possibility that error in their application may cause errors within the data collected as they were used for defining segments in the static model. Another limitation with the marker set used is that markers were attached to the trainers of the participant and not to skin, therefore the true movement of the foot contained within the trainer has not been collected, only an estimation of it. However, by removing the trainers and applying to the skin has the potential to alter the jumping and landing mechanics of the participants and reduce the ecological validity of the test results. Finally, the foot was considered a rigid segment which means the effects on the differing joints that make up the ankle complex cannot be individually investigated.

4.5.4 Considerations for next study

Both the first and second studies in this thesis have concentrated on straight line movements and have focused on the dominant limb of participants. No considerations have thus far been given to whether or not ankle protectors effect the dominant and non-dominant limb differently, the effects of ankle protectors on a change of direction, or the effects of non-standardised footwear on ankle kinematics. Therefore, the next study will investigate the effects of ankle protectors on both the dominant and non-dominant limb during a manoeuvre that requires a change of direction. Additionally, for the next study the footwear for the participants will be standardised so that the effects of external variables on the kinematic data are further limited.

5. The effects of ankle protectors on the dominant and non-dominant limb during a 45° cutting manoeuvre: a comparison to braced and unbraced ankles.

5.1 Introduction

The previous two chapters have established that ankle protectors have a small effect on coronal kinematics during running and also reduce performance of CMVJs by reducing sagittal plane motion when used by females. Additionally, it was found that there were significant effects on knee kinematics during take-off of a CMVJ when used by females. For males they appear to have very little effect on lower-limb kinematics and only significant differences in the sagittal plane during running have been found. However, football does not just consist of linear movements, it is a multi-directional sport which involves fast twists and turns with and without being in possession of the football, and so these small alterations found in the previous chapters may have greater effects during changes in direction. During sudden changes in direction ankle-inversion injuries can occur and have been found to be the second most common cause of non-contact ankle-inversion injuries to football players (Woods, et al., 2003). Additionally, ankle-contusion injuries can occur during these manoeuvres by a mistimed tackle from an opponent or a collision with another player. To assess the effects of ankle devices on lower-limb kinetics and kinematics during a sudden change of direction researchers often use cutting manoeuvres (Commons & Low, 2014; Greene, et al., 2014; Gudibanda & Wang, 2005; Klem, et al., 2017). Similar to the chapters on running and countermovement vertical jumping there are currently no research papers on the effects of ankle protectors on lower-limb kinematics during a cutting manoeuvre however ankle braces have had some attention.

During cutting manoeuvres studies have found ankle braces reduce the ankles coronal plane motion when compared to not wearing any (Commons & Low, 2014; Klem, et al., 2017) however, some have found no effect (Greene, et al., 2014). Using a male population Commons & Low (2014) found that ASO lace-up ankle braces reduce ankle-inversion angle. However, Klem, et al. (2017) using the same ankle brace with a female population found that the ASO ankle braces did not reduce peak inversion whereas using an Active T2 hinged ankle brace did significantly reduce peak inversion when compared to not wearing any. Greene, et al. (2014) found similar results when using lace-up ankle braces with a female population as it was found there was no reduction in coronal plane ROM of the ankle or inversion eversion angle at initial contact. These findings suggest that during cutting manoeuvres females require a more rigid ankle brace such as a semi-rigid or hinged ankle brace to reduce coronal plane kinematics. In the sagittal plane Commons & Low (2014) found that when using ankle braces during a 45˚ V-cut did not affect peak plantar flexion and Klem, et al. (2017) found no significant reductions in dorsiflexion when using lace-up or hinged ankle braces but did not report values for plantarflexion. However, Greene, et al. (2014) found that ankle ROM was significantly reduced in the sagittal plane when wearing a lace-up ankle brace by a female population.

Looking at the knee and hip during cutting has produced some interesting and contradictory findings. Greene, et al. (2014) found that ankle braces do not alter knee kinematics in a female population and West & Campbell (2014) found ankle braces reduce the magnitude of medial and lateral shear forces at the knee. These findings suggest that wearing ankle braces might be beneficial for reducing knee loading during cutting manoeuvres. However these findings must be taken with caution because Klem, et al. (2017) found when using a female population both ASO lace-up ankle braces and Active T2 hinged ankle braces significantly increased internal knee rotation and knee abduction angles which could increase risk of knee injuries. It has been found that ankle braces do not significantly affect ground reaction forces in either male or female populations when compared to unbraced ankles during a cutting manoeuvre (Bezalel, 2009). These findings have been mirrored by both Greene, et al. (2014) and West & Campbell (2014) who also found no significant difference in time to peak GRF. However, Cloak, et al.

(2010) found a reduction in peak mediolateral GRF when wearing an Aircast AirSport ankle brace which could possibly reduce the risk of ankle-inversion injuries. Ankle braces have also been found not to significantly affect stance time (Gudibanda & Wang, 2005).

Previous research has established that ankle braces reduce the risk of ankle-inversion injuries (McGuine, et al., 2011; McGuine, et al., 2012; Pedowitz, et al., 2008; Surve, et al., 1994) and the previous chapters have established that ankle protectors have a small effect on sagittal plane ankle kinematics in males during running and also effect sagittal and coronal ankle kinematics in females during running. Also there have been effects found using ankle protectors during the take-off phase of a CMVJ when used by females which also have an effect on knee kinematics. However, to the authors knowledge, there are no studies looking at the effects of ankle protectors on non-linear movements. Therefore, the current study aims to investigate the effects of ankle protectors on ankle kinematics during a 45˚ cutting manoeuvre, compare the effects of ankle protectors with braced and unbraced ankles to establish which it more closely resembles, investigate the effects of ankle protectors on knee and hip kinematics, investigate the effects on dominant and non-dominant limb, and investigate the effects on both male and female populations. It is hypothesised that ankle protectors will not reduce ankle kinematics and produce similar kinematics to an unbraced ankle.

5.2 Methodology

5.2.1 Participants

Twelve male (aged; 23.17±3.88 years, height; 178.25±5.16 cm, body mass; 73.32±9.40 kg and BMI; 23.12±3.18) and twelve female (aged; 26.83±5.02 years, height; 163.98±5.09 cm, body mass; 58.98±4.25 kg and BMI; 21.98±1.87) participants took part in this study. Participants were recruited from local and university football teams via opportunity sampling using poster adverts. The inclusion criteria for the study was that the participant were aged between 18 and 35, currently playing for a football team, were injury free at the time of testing, and were right foot dominant. The dominant foot was defined as the foot the participant would use to kick a football with. All participants provided written consent in line with the University of Central Lancashire's ethical panel (STEMH 309).

5.2.2 Procedure

Participants ran towards an embedded force plate (Kistler Instruments Ltd., Alton, Hampshire) which was located in the centre of an 18m by 7.5m biomechanics laboratory. Upon arrival at the force platform the participant was required to either plant their left foot (LEFT foot strike) on the force plate and propel themselves right at a 45˚ angle or plant their right foot (RIGHT foot strike) on the force plate and propel themselves left at a 45˚ angle to simulate a change of direction when approaching an opponent in a sporting context. Participants performed the cutting manoeuvre in three test conditions; wearing ankle braces (BRACE), wearing ankle protectors (PROTECTOR) and with uncovered ankles (WITHOUT). The order of foot strike and condition was randomised for the participants. To make the manoeuvre more realistic a skeleton manikin was placed on the far side of the force plate, from the starting point of the participant, to obstruct forward travel and simulate an opponent. Five successful trials were recorded for each test condition and each stance foot. A successful trial was determined as one in which the participant landed with the whole of their stance foot (left foot for a right cut or right foot for a left cut) on the force platform and kept within a speed tolerance of 3.6 $m.s^{-1}$ ± 5%. The force platform sampled at 1000 Hz and was used to determine the start and end of the stance phase during the cutting trials. These points were determined as the point where the force plate first recorded a vertical ground reaction force (VGRF) that exceeded 20N and ended when the VGRF dropped back down below 20N (Sinclair, et al., 2011).

Kinematic data were recorded using an eight camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) tracking retro-reflective markers at a sampling rate of 250 Hz. Using the calibrated anatomical system technique (CAST) (Cappozzo, et al., 1995) the retro-reflective markers were attached to the 1st and 5th metatarsal heads, calcaneus, medial and lateral malleoli, the medial and lateral femoral epicondyles, the greater trochanter, Left and right anterior superior iliac spine, and left and right posterior superior iliac spine. These markers were used to model the right and left foot, shank, thigh, and pelvis segments in six degrees of freedom. Rigid plastic mounts with four markers on each were also attached to the shanks and thighs and were secured using elasticated bandage. These were used as tracking markers for the shank and thigh segments. To track the feet the 1st and 5th metatarsal heads and the calcaneus were used and to track the pelvis the left and right anterior superior iliac spine and left and right posterior superior iliac spine were used. In the BRACE condition the medial and lateral malleoli locations were found by placing the index finger under the rigid construct of the brace to locate the anatomical landmark then matching the location to the exterior of the Brace where the marker was then fixed to. In the PROTECTOR condition the medial and lateral malleoli locations were located by palpating the soft foam construct to find the underlying anatomical landmarks. To assess the speed of the participant a single marker was attached to the xiphoid process and was checked for velocity using the QTM software after each trial was recorded. Before dynamic trials were captured a static trial of the participant stood in the anatomical position was captured which was used to identify the location of the tracking makers with reference to the anatomical markers. To define each plane of motion firstly the Z (transverse) axis follows the segment from distal to proximal and denotes internal/external rotation, secondly the Y (coronal) axis is orientated from anterior to posterior of the segment and denotes adduction/abduction, and thirdly the X (sagittal) axis is orientated from medial to lateral of the segment and denotes flexion/extension.

5.2.3 Ankle braces, ankle protectors, and footwear

The ankle protectors and ankle braces used for the current study are outlined in section 2.2. All males completed the testing wearing a pair of Adidas F10 TRX TF football trainers and all females completed the testing wearing a pair of Umbro Speciali cup TF football trainers.

5.2.4 Data processing

Anatomical and tracking markers were identified within the Qualisys Track Manager software and then exported as C3D files to be analysed using Visual 3D software (C-Motion, Germantown, MD, USA). To define the centre points of the ankle and knee segments the two marker methods were utilised for both. These methods calculate the centre of the joint using the positioning of the malleoli markers for the ankle centre and the femoral epicondyle markers for the knee centre (Graydon, et al., 2015; Sinclair, et al., 2015). To calculate the hip joint centre a regression equation which uses the position of the ASIS markers was utilised (Sinclair, et al., 2014). The cutting trials were filtered at 15Hz using a low pass 4th order zero-lag filter Butterworth filter (Malinzak, et al., 2001; Savage, et al., 2018). Data were normalized to 100%

of the stance phase for each foot strike then processed trials were used to produce means of the five trials for each test condition for each stance foot for each participant. 3D kinematics of the ankle, knee and hip joints of the right and left leg were calculated using an XYZ cardan sequence of rotations. The 3D joint kinematic measures which were extracted for further analysis were 1) angle at footstrike, 2) angle at toe-off, 3) peak angle during the stance phase, 4) absolute range of motion (absolute ROM) calculated by taking the maximum angle from the minimum angle during stance, 5) relative range of motion (relative ROM) calculated using the angle at footstrike and the first peak value after footstrike, 6) peak angular velocities for the ankle during the stance phase. Measures taken from the force plate to be analysed were 1) peak forces 2) instantaneous loading rate calculated as the maximum increase in vertical force between frequency intervals, 3) average loading rate calculated by dividing the peak vertical impact force by the time to the impacts peak 4) stance time. The force data were normalised to bodyweights (BWs) for each participant to allow comparisons across the data set to be investigated.

5.2.5 Statistical analyses

Data analysis was conducted using SPSS v22.0 (SPSS Inc., Chicago, IL, USA). Descriptive statistics were generated using means and standard deviations for each of the outcome measures and were grouped by gender and analysed separately from one another. Differences between Conditions (WITHOUT, PROTECTOR, and BRACE) and Stance limbs (LEFT foot strike and RIGHT foot strike) were examined using two-way repeated measures factorial ANOVA in a 3×2 design. Statistical significance was accepted at the $p \leq 0.05$ level. Significant main effects were investigated using pairwise comparisons after Bonferroni adjustment to control for type I error. Effect sizes were calculated using partial Eta^2 ($p\eta^2$).

5.3 Male Results

Tables 5.1-5.3 and figures 5.1 & 5.2 present the key parameters of interest obtained from the left and right limb during stance phase of a V-Cut manoeuvre.

5.3.1 Kinetic and temporal parameters

Table 5.1. Kinetic and temporal variables (means and standard deviations) obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the male participants.

	LEFT Foot Strike			RIGHT Foot Strike		
	WITHOUT	**PROTECTOR**	**BRACE**	**WITHOUT**	**PROTECTOR**	**BRACE**
Peak Vertical Impact Force (BW)	2.06 ± 0.20	2.13 ± 0.16	2.16 ± 0.20	2.24 ± 0.29	2.30 ± 0.19	2.15 ± 0.22
Peak Braking Force (BW)	0.78 ± 0.16	0.82 ± 0.17	0.80 ± 0.14	0.82 ± 0.15	0.82 ± 0.16	0.74 ± 0.18
Peak Propulsive force (BW)	0.20 ± 0.07	0.19 ± 0.05	0.18 ± 0.06	0.19 ± 0.05	0.18 ± 0.06	0.19 ± 0.06
Peak Medial Force (BW)	0.01 ± 0.01	0.01 ± 0.01	0.02 ± 0.01	0.01 ± 0.02	0.02 ± 0.02	0.02 ± 0.02
Peak Lateral Force (BW)	0.78 ± 0.07	0.77 ± 0.10	0.77 ± 0.08	0.81 ± 0.07	0.83 ± 0.11	0.82 ± 0.07
Instantanteous Loading Rate (BW.s)	167.44 ± 88.65	200.84 ± 130.24	185.84 ± 116.44	200.38 ± 111.46	226.13 ± 121.57	168.73 ± 96.04
Average Loading Rate (BW.s)	71.90 ± 28.97	71.39 ± 25.33	71.85 ± 24.91	72.88 ± 23.82	80.38 ± 35.43	66.13 ± 26.32
Stance Time (s)	0.28 ± 0.02	0.28 ± 0.02	0.28 ± 0.03	0.27 ± 0.02	0.27 ± 0.02	0.26 ± 0.02

The kinetic and temporal variables exhibited no significant differences ($P > 0.05$) between conditions or stance limbs.

5.3.2 3D Kinematic Parameters

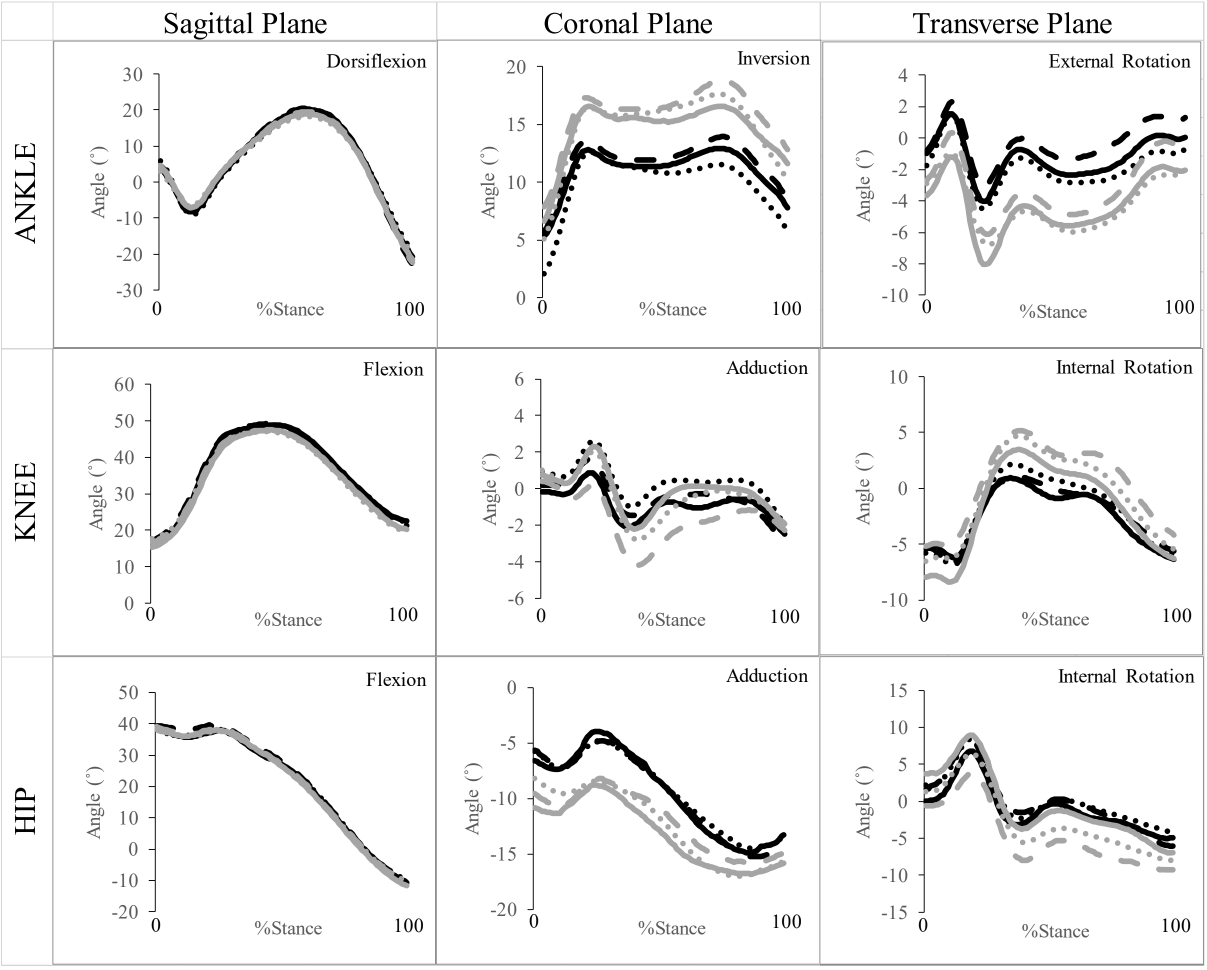

Figure 5.1. Mean ankle, knee, and hip kinematics obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the sagittal, coronal, and transverse planes for the male participants. (WITHOUT left leg = black dash, PROTECTOR left leg = black solid, BRACE left leg = black dot, WITHOUT right leg = grey dash, PROTECTOR right leg = grey solid, BRACE right leg = grey dot).

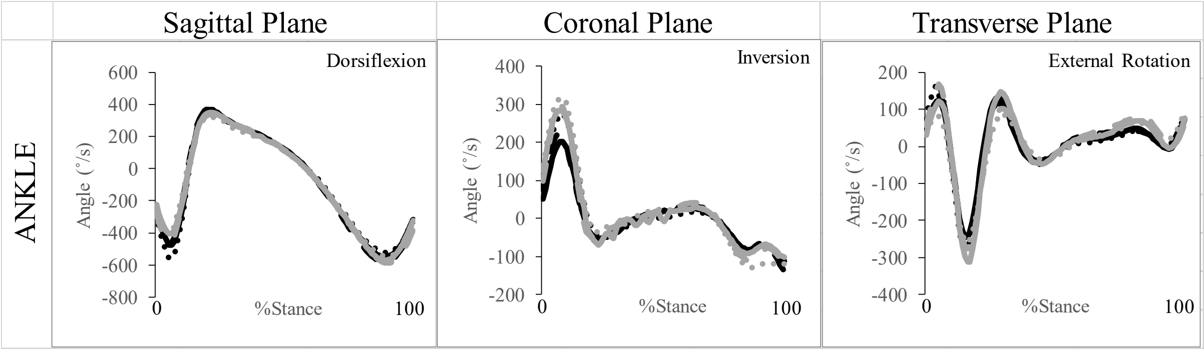

Figure 5.2. Mean ankle velocity obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the sagittal, coronal, and transverse planes for the male participants. (WITHOUT left leg = black dash, PROTECTOR left leg = black solid, BRACE left leg = black dot, WITHOUT right leg = grey dash, PROTECTOR right leg = grey solid, BRACE right leg = grey dot).

Table 5.2. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the male participants.

		LEFT Foot Strike						RIGHT Foot Strike					
		WITHOUT		PROTECTOR		BRACE		WITHOUT		PROTECTOR		BRACE	
ANKLE	**Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)**												
	Angle at footstrike	3.86 ± 12.31		4.64 ± 8.44		5.84 ± 5.60		4.45 ± 6.88		4.18 ± 9.12		3.72 ± 8.03	
	Angle at toe-off	-22.74 ± 6.83		-21.02 ± 7.31		-19.51 ± 6.15		-22.30 ± 5.42		-21.43 ± 5.82		-19.96 ± 4.40	
	Peak dorsiflexion	21.02 ± 5.60		20.48 ± 6.28		19.38 ± 6.03		19.86 ± 4.44		19.27 ± 5.33		18.37 ± 3.89	
	Absolute ROM	44.29 ± 8.44		42.39 ± 7.57		39.85 ± 6.57	AB	42.50 ± 5.80		41.01 ± 6.42		38.56 ± 4.39	AB
	Relative ROM	27.13 ± 11.34		26.56 ± 10.36		26.31 ± 6.61		27.09 ± 8.51		25.92 ± 10.67		23.91 ± 9.11	
	Peak dorsifelxion velocity (˚/s)	439.47 ± 114.42		427.30 ± 82.98		403.10 ± 57.74		423.64 ± 66.88		454.43 ± 107.13		388.49 ± 73.29	
	Peak plantarflexion velocity (˚/s)	-746.68 ± 150.87		-712.91 ± 166.23		-692.66 ± 118.45		-695.93 ± 129.62		-703.92 ± 102.02		-654.66 ± 121.15	
	Coronal plane (+ = Inversion / - = Eversion)												
	Angle at footstrike	5.74 ± 7.60		5.46 ± 5.97		2.05 ± 4.39	AB	7.71 ± 7.17	*	6.32 ± 5.62	*	5.42 ± 4.43	AB*
	Angle at toe-off	8.30 ± 5.35		7.68 ± 5.70		5.78 ± 5.08	AB	12.80 ± 5.53	*	11.58 ± 5.39	*	10.07 ± 3.39	AB *
	Peak inversion	16.85 ± 4.40		15.65 ± 3.14		13.99 ± 4.51	A	21.31 ± 4.77	*	19.62 ± 2.93	*	19.49 ± 4.47	A*
	Peak eversion	3.07 ± 5.90		3.15 ± 5.06		0.85 ± 3.83		5.98 ± 6.35	*	5.61 ± 4.51	*	5.00 ± 4.04	*
	Absolute ROM	13.78 ± 4.85		12.50 ± 4.84		13.14 ± 5.52		15.33 ± 7.30		14.01 ± 4.04		14.49 ± 7.63	
	Relative ROM	11.11 ± 7.46		10.19 ± 6.73		11.94 ± 6.74		13.61 ± 8.70		13.30 ± 5.49		14.07 ± 7.97	
	Peak inversion velocity (˚/s)	283.30 ± 172.28		264.25 ± 145.79		305.47 ± 146.30		356.18 ± 198.51	*	394.09 ± 221.39	*	347.74 ± 155.44	*
	Peak eversion velocity (˚/s)	-208.35 ± 72.30		-187.98 ± 73.67		-154.62 ± 49.38		-230.82 ± 88.66	*	-254.64 ± 209.22	*	-197.83 ± 81.75	*
	Transverse plane (+ = External / - = Internal)												
	Angle at footstrike	-0.83 ± 4.23		-0.85 ± 3.76		-1.74 ± 3.15		-2.85 ± 4.22		-3.81 ± 4.01		-2.61 ± 3.45	
	Angle at toe-off	1.27 ± 4.76		0.32 ± 3.93		-0.71 ± 3.67		-0.46 ± 4.53		-2.04 ± 4.16		-2.04 ± 3.31	
	Peak rotation	-5.46 ± 3.03		-6.10 ± 3.15		-6.53 ± 2.48		-7.79 ± 4.55		-9.41 ± 4.69		-8.28 ± 3.75	
	Absolute ROM	10.38 ± 2.61		10.29 ± 2.07		9.84 ± 2.68		10.25 ± 2.59		10.98 ± 2.33		9.60 ± 2.53	
	Relative ROM	5.75 ± 3.65		5.04 ± 3.11		5.04 ± 2.55		5.31 ± 2.49		5.38 ± 2.69		3.92 ± 2.06	
	Peak external rotation velocity (˚/s)	262.92 ± 91.10		264.07 ± 91.10		246.77 ± 79.36		248.76 ± 85.76		271.92 ± 66.35		218.69 ± 66.59	
	Peak internal rotation velocity (˚/s)	-312.63 ± 120.43		-332.07 ± 132.31		-347.07 ± 131.80		-350.65 ± 123.31		-380.12 ± 133.91		-324.81 ± 98.68	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition., AB = significant difference from WITHOUT & PROTECTOR conditions. * = Significant difference between stance foot.

For the ankle in the sagittal plane there was a significant main effect between conditions $F_{(2, 22)} = 13.46$, $P \leq 0.01$, $p\eta^2=0.55$ but not between stance foot for the absolute ROM. Further analysis revealed that the BRACE condition was significantly lower than the WITHOUT ($P \leq 0.01$, 95%CI: 1.66-6.71%) and the PROTECTOR ($P \leq 0.05$, 95%CI: 0.33-4.66%) conditions. In the coronal plane significant main effects were found for the stance foot and between the conditions for angle at footstrike (Stance foot; $F_{(1, 11)} = 6.77$, $P \leq 0.05$, $p\eta^2=0.38$, Condition; $F_{(2, 22)} = 6.88$, $P \leq 0.01$, $p\eta^2=0.39$), angle at toe-off (Stance foot; $F_{(1, 11)} = 11.92$, $P \leq 0.01$, $p\eta^2=0.52$, Condition; $F_{(2, 22)} = 9.64$, $P \leq 0.01$, $p\eta^2=0.47$), and peak inversion (Stance foot; $F_{(1, 11)} = 23.57$, $P \leq 0.01$, $p\eta^2=0.68$, Condition; $F_{(2, 22)} = 3.70$, $P \leq 0.05$, $p\eta^2=0.25$). Whilst significant main effects were found for stance foot but not for conditions for peak eversion ($F_{(1, 11)} = 13.52$, $P \leq 0.01$, $p\eta^2=0.55$), peak inversion velocity ($F_{(1, 11)} = 17.56$, $P \leq 0.01$, $p\eta^2=0.62$) and for peak eversion velocity ($F_{(1, 11)} = 5.38$, $P \leq 0.05$, $p\eta^2=0.33$). Further analysis found that the BRACE condition significantly reduced the angle at footstrike (WITHOUT; $P \leq 0.05$, 95%CI: 0.16-5.82%, PROTECTOR; $P \leq 0.05$, 95%CI: 0.35-3.95%) and angle at toe off (WITHOUT; $P \leq 0.01$, 95%CI: 0.91-4.34%, PROTECTOR; $P \leq 0.05$, 95%CI: 0.18-3.24%) when compared to both the WITHOUT and PROTECTOR conditions. As well as significantly reducing peak inversion when compared to the WITHOUT ($P \leq 0.05$, 95%CI: 0.38-5.05%) condition. The Left foot strike was significantly lower than the Right foot strike for angle at footstrike ($P \leq 0.001$, 95%CI: 0.32-3.81%), angle at toe off ($P \leq 0.01$, 95%CI: 1.53-6.92%), peak inversion ($P \leq 0.001$, 95%CI: 2.54-6.75%), peak eversion ($P \leq 0.05$, 95%CI: 1.27-5.07%), peak inversion velocity ($P \leq 0.01$, 95%CI: 38.78-124.57%) and peak eversion velocity ($P \leq 0.05$, 95%CI: 2.09-86.13%).

No significant differences ($P > 0.05$) were found in the transverse pane for the ankle or any of the planes of motion for both the knee joint or the hip joint. No significant differences ($P > 0.05$) were found between LEFT foot strike and RIGHT foot strike either.

Table 5.3. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the male participants.

		LEFT Foot Strike			**RIGHT Foot Strike**		
		WITHOUT	**PROTECTOR**	**BRACE**	**WITHOUT**	**PROTECTOR**	**BRACE**
KNEE	**Sagittal plane (+ = Flexion / - = Extension)**						
	Angle at footstrike	17.26 ± 7.78	16.96 ± 6.70	15.82 ± 5.80	16.68 ± 6.15	15.28 ± 5.64	17.57 ± 5.54
	Angle at toe-off	21.39 ± 10.40	22.68 ± 10.98	22.64 ± 10.49	20.67 ± 9.68	20.28 ± 7.99	20.04 ± 6.46
	Peak flexion	49.96 ± 5.99	49.51 ± 6.83	49.18 ± 7.60	48.10 ± 5.96	47.95 ± 5.86	47.56 ± 5.81
	Absolute ROM	35.94 ± 4.03	34.95 ± 4.57	35.28 ± 4.64	33.28 ± 4.39	34.93 ± 4.12	32.75 ± 4.07
	Relative ROM	32.71 ± 5.60	32.55 ± 2.78	33.36 ± 4.25	31.42 ± 5.11	32.66 ± 4.94	29.99 ± 4.39
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at footstrike	0.20 ± 3.40	-0.15 ± 2.69	0.92 ± 3.18	0.40 ± 1.84	0.75 ± 2.61	1.07 ± 1.87
	Angle at toe-off	-2.86 ± 3.55	-2.30 ± 3.63	-2.04 ± 3.78	-2.29 ± 3.20	-1.94 ± 2.93	-1.78 ± 2.99
	Peak adduction	3.07 ± 3.39	2.42 ± 2.36	3.64 ± 3.06	2.80 ± 3.48	4.29 ± 5.44	3.65 ± 4.49
	Absolute ROM	7.29 ± 2.36	6.48 ± 1.73	6.84 ± 1.61	8.68 ± 2.15	8.98 ± 2.82	8.40 ± 3.10
	Relative ROM	2.87 ± 2.17	2.57 ± 1.57	2.72 ± 1.81	2.40 ± 2.92	3.54 ± 4.09	2.58 ± 3.59
	Transverse plane (+ = Internal / - = External)						
	Angle at footstrike	-5.40 ± 3.34	-5.40 ± 3.74	-5.84 ± 4.62	-5.19 ± 3.48	-8.01 ± 3.65	-6.58 ± 4.44
	Angle at toe-off	-5.64 ± 4.39	-6.42 ± 3.21	-6.03 ± 4.41	-4.17 ± 5.14	-6.31 ± 4.02	-5.34 ± 4.83
	Peak rotation	3.58 ± 3.92	3.74 ± 4.01	4.35 ± 3.79	7.10 ± 4.48	5.07 ± 4.28	6.25 ± 3.63
	Absolute ROM	12.86 ± 2.70	13.07 ± 3.99	13.77 ± 3.61	15.30 ± 3.35	16.37 ± 3.69	15.67 ± 2.83
	Relative ROM	8.98 ± 3.24	9.14 ± 3.57	10.19 ± 4.37	12.29 ± 3.87	13.08 ± 4.51	12.83 ± 3.04
HIP		**LEFT Foot Strike**			**RIGHT Foot Strike**		
		WITHOUT	**PROTECTOR**	**BRACE**	**WITHOUT**	**PROTECTOR**	**BRACE**
	Sagittal plane (+ = Flexion / - = Extension)						
	Angle at footstrike	39.69 ± 7.29	38.40 ± 7.40	38.33 ± 8.19	39.18 ± 7.03	38.31 ± 6.99	38.42 ± 8.19
	Angle at toe-off	-9.69 ± 7.20	-10.74 ± 7.89	-9.81 ± 7.35	-10.35 ± 6.72	-11.59 ± 6.77	-10.12 ± 6.97
	Peak flexion	42.18 ± 7.16	40.15 ± 7.43	40.60 ± 8.39	40.92 ± 7.59	40.51 ± 6.72	40.27 ± 7.95
	Absolute ROM	52.01 ± 6.81	50.91 ± 7.35	50.42 ± 5.64	51.38 ± 7.20	51.83 ± 7.28	50.52 ± 7.26
	Relative ROM	49.52 ± 8.00	49.15 ± 7.95	48.16 ± 6.56	49.63 ± 7.28	49.64 ± 8.44	48.67 ± 7.90
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at footstrike	-6.03 ± 6.31	-6.56 ± 7.36	-6.04 ± 6.76	-9.58 ± 7.57	-10.34 ± 8.20	-7.92 ± 7.68
	Angle at toe-off	-14.57 ± 5.54	-13.37 ± 5.01	-13.38 ± 3.77	-14.83 ± 3.85	-15.82 ± 2.91	-15.51 ± 3.97
	Peak adduction	-3.30 ± 5.38	-2.21 ± 6.48	-3.24 ± 5.40	-5.58 ± 6.04	-6.74 ± 6.05	-5.54 ± 6.69
	Absolute ROM	13.65 ± 5.64	14.39 ± 5.18	12.95 ± 4.53	12.94 ± 4.86	12.33 ± 3.87	13.80 ± 5.06
	Relative ROM	10.92 ± 7.09	10.04 ± 6.09	10.16 ± 5.89	8.95 ± 5.95	8.73 ± 5.38	11.41 ± 5.77
	Transverse plane (+ = Internal / - = External)						
	Angle at footstrike	2.07 ± 9.75	0.37 ± 10.03	2.06 ± 8.99	0.77 ± 7.71	3.62 ± 12.43	1.42 ± 7.48
	Angle at toe-off	-6.11 ± 8.58	-5.00 ± 7.44	-3.86 ± 8.48	-9.19 ± 6.56	-7.06 ± 9.28	-7.80 ± 5.40
	Peak rotation	-9.06 ± 7.64	-9.14 ± 7.33	-7.74 ± 7.80	-13.27 ± 5.40	-9.71 ± 9.86	-11.38 ± 5.10
	Absolute ROM	18.93 ± 5.30	18.28 ± 6.06	18.00 ± 5.39	19.96 ± 4.97	20.73 ± 4.92	19.06 ± 4.54
	Relative ROM	7.80 ± 6.63	8.77 ± 5.98	8.20 ± 6.00	7.46 ± 6.21	7.41 ± 4.71	6.26 ± 4.54

5.4 Female Results

Tables 5.4-5.6 and figures 5.3 & 5.4 present the key parameters of interest obtained from the left and right limb during stance phase of a V-Cut manoeuvre.

5.4.1 Kinetic and temporal parameters

Table 5.4. Kinetic and temporal variables (means and standard deviations) obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the female participants.

	LEFT Foot Strike						RIGHT Foot Strike					
	WITHOUT		PROTECTOR		BRACE		WITHOUT		PROTECTOR		BRACE	
Peak Vertical Impact Force (BW)	2.43 ± 0.43		2.34 ± 0.36		2.50 ± 0.43		2.40 ± 0.25		2.38 ± 0.26		2.51 ± 0.27	
Peak Braking Force (BW)	0.90 ± 0.21		0.90 ± 0.20		0.94 ± 0.23		0.89 ± 0.19		0.87 ± 0.15		0.91 ± 0.18	
Peak Propulsive force (BW)	0.14 ± 0.04		0.14 ± 0.04		0.15 ± 0.05		0.15 ± 0.04		0.15 ± 0.05		0.14 ± 0.06	
Peak Medial Force (BW)	0.02 ± 0.02		0.03 ± 0.04		0.04 ± 0.06		0.02 ± 0.02		0.02 ± 0.01		0.03 ± 0.04	
Peak Lateral Force (BW)	0.71 ± 0.10		0.70 ± 0.11		0.71 ± 0.10		0.71 ± 0.07		0.72 ± 0.09		0.73 ± 0.10	
Instantanteous Loading Rate (BW.s)	255.77 ± 150.46		253.41 ± 160.22		266.26 ± 165.52		188.54 ± 88.03		218.45 ± 106.38		237.89 ± 125.79	
Average Loading Rate (BW.s)	88.86 ± 32.77		85.15 ± 25.64		87.82 ± 20.19		76.67 ± 24.88		85.56 ± 31.96		87.67 ± 22.47	
Stance Time (s)	0.25 ± 0.04		0.26 ± 0.04		0.26 ± 0.05		0.25 ± 0.03		0.25 ± 0.03		0.25 ± 0.04	

The kinetic and temporal variables exhibited no significant differences ($P > 0.05$) between conditions or stance limbs.

5.4.2 3D Kinematic Parameters

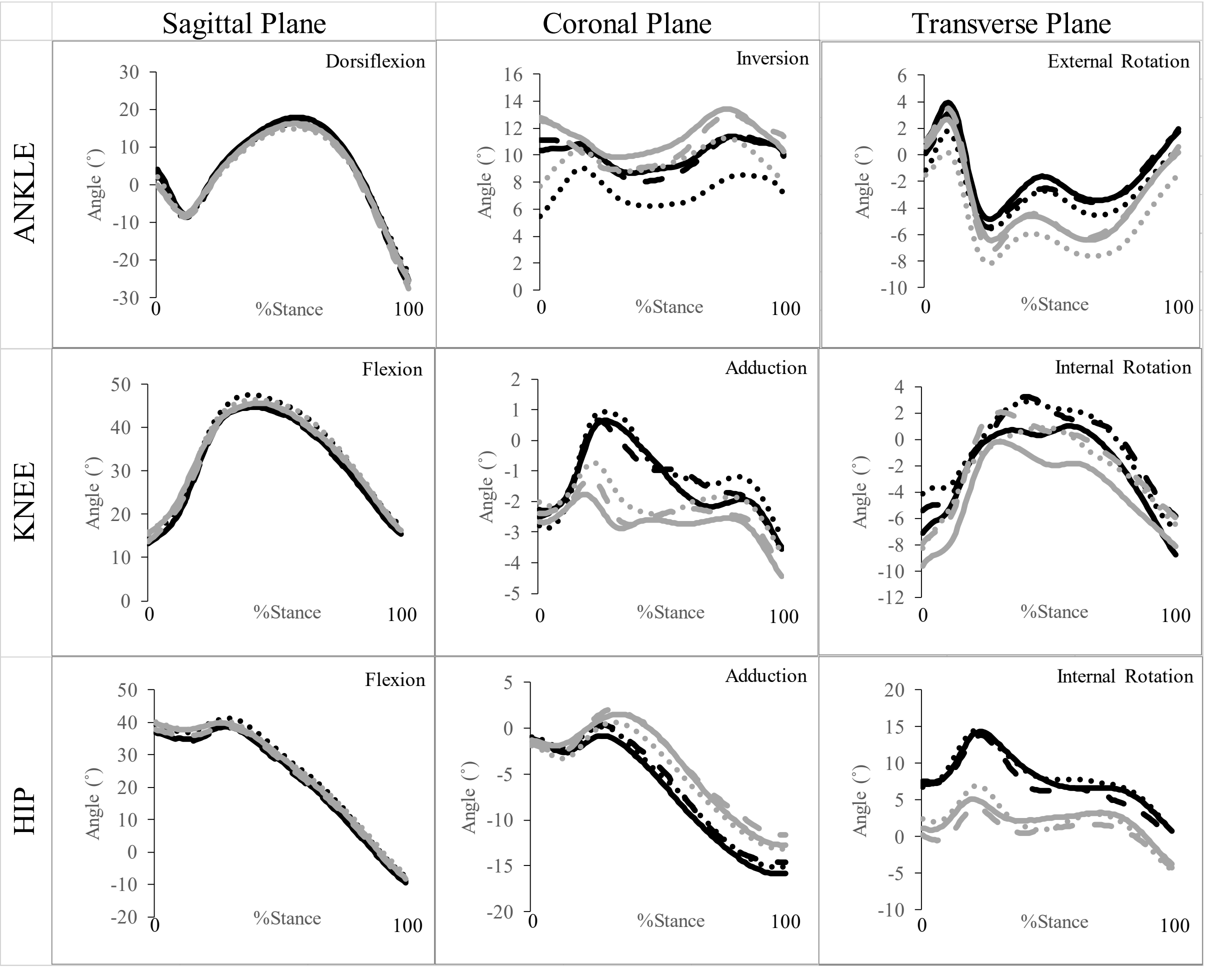

Figure 5.3. Mean ankle, knee, and hip kinematics obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the sagittal, coronal, and transverse planes for the female participants. (WITHOUT left leg = black dash, PROTECTOR left leg = black solid, BRACE left leg = black dot, WITHOUT right leg = grey dash, PROTECTOR right leg = grey solid, BRACE right leg = grey dot).

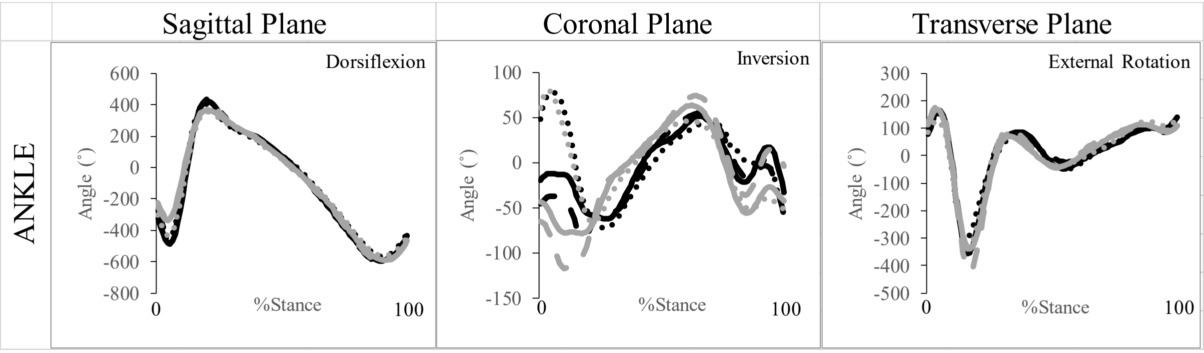

Figure 5.4. Mean ankle velocity obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the sagittal, coronal, and transverse planes for the female participants. (WITHOUT left leg = black dash, PROTECTOR left leg = black solid, BRACE left leg = black dot, WITHOUT right leg = grey dash, PROTECTOR right leg = grey solid, BRACE right leg = grey dot).

Table 5.5. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the female participants.

		LEFT Foot Strike						**RIGHT Foot Strike**					
		WITHOUT		**PROTECTOR**		**BRACE**		**WITHOUT**		**PROTECTOR**		**BRACE**	
ANKLE	**Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)**												
	Angle at footstrike	2.91 ± 16.42		4.08 ± 15.79		2.07 ± 17.11		0.26 ± 17.15		-0.10 ± 15.51		2.16 ± 14.05	
	Angle at toe-off	-26.78 ± 6.95		-25.37 ± 7.46		-23.87 ± 6.44		-26.80 ± 5.30		-25.68 ± 6.23		-24.49 ± 5.43	
	Peak dorsiflexion	17.59 ± 5.06		18.55 ± 4.94		16.68 ± 5.16		16.35 ± 5.75		16.72 ± 5.92		15.54 ± 4.84	
	Absolute ROM	44.45 ± 6.30		44.10 ± 7.35		41.50 ± 5.54	AB	44.70 ± 4.57		42.95 ± 6.00		40.25 ± 3.38	AB
	Relative ROM	29.77 ± 14.10		29.63 ± 13.41		26.89 ± 12.83		28.61 ± 16.59		26.13 ± 15.20		26.87 ± 12.32	
	Peak dorsifelxion velocity (˚/s)	514.93 ± 208.65		496.51 ± 192.92		489.19 ± 157.97		507.25 ± 167.02		468.47 ± 139.84		433.20 ± 111.24	
	Peak plantarflexion velocity (˚/s)	-740.30 ± 162.31		-709.50 ± 139.97		-718.94 ± 162.81		-711.14 ± 148.88		-680.76 ± 136.05		-668.01 ± 147.12	
	Coronal plane (+ = Inversion / - = Eversion)												
	Angle at footstrike	11.12 ± 5.88		10.35 ± 5.30		5.55 ± 5.06	AB	12.70 ± 6.73		12.57 ± 7.32		7.73 ± 5.57	AB
	Angle at toe-off	9.90 ± 4.81		10.09 ± 4.43		7.03 ± 4.07	AB	11.44 ± 3.60		10.26 ± 2.76		7.83 ± 3.49	AB
	Peak inversion	16.03 ± 4.25		15.52 ± 2.95		11.61 ± 4.16	AB	16.45 ± 4.45		16.75 ± 4.33		13.60 ± 4.45	AB
	Peak eversion	4.79 ± 3.35		5.12 ± 2.32		2.31 ± 3.47	AB	6.69 ± 4.10		6.43 ± 4.32		4.98 ± 4.29	AB
	Absolute ROM	11.24 ± 1.73		10.40 ± 2.19		9.29 ± 3.35		9.76 ± 3.91		10.32 ± 4.17		8.63 ± 3.39	
	Relative ROM	4.91 ± 4.62		5.17 ± 4.26		6.05 ± 5.24		3.75 ± 3.48		4.18 ± 4.28		5.87 ± 3.62	
	Peak inversion velocity (˚/s)	163.26 ± 75.59		153.14 ± 72.46		168.21 ± 107.31		139.87 ± 54.14		146.21 ± 69.21		170.15 ± 87.25	
	Peak eversion velocity (˚/s)	-240.20 ± 73.07		-198.22 ± 61.41		-162.89 ± 43.56	A	-279.64 ± 177.76		-251.12 ± 188.26		-200.11 ± 138.37	A
	Transverse plane (+ = External / - = Internal)												
	Angle at footstrike	0.09 ± 4.26		0.97 ± 4.38		-1.00 ± 4.13		0.76 ± 6.89		0.58 ± 6.75		-1.52 ± 5.19	
	Angle at toe-off	1.73 ± 4.55		1.94 ± 4.91		0.80 ± 4.88		0.34 ± 5.31		0.22 ± 4.88		-1.39 ± 4.60	
	Peak rotation	-7.89 ± 4.87		-7.26 ± 6.33		-8.10 ± 5.51		-9.32 ± 4.17		-8.89 ± 5.51		-9.96 ± 4.20	
	Absolute ROM	13.43 ± 1.89		13.40 ± 2.62		12.31 ± 2.21		14.44 ± 3.92		13.89 ± 3.47		12.05 ± 3.56	
	Relative ROM	5.45 ± 3.86		5.17 ± 3.26		5.20 ± 3.55		4.36 ± 3.64		4.41 ± 3.57		3.62 ± 2.58	
	Peak external rotation velocity (˚/s)	163.26 ± 75.59		153.14 ± 72.46		168.21 ± 107.31		321.67 ± 83.39		317.51 ± 74.45		277.29 ± 52.91	
	Peak internal rotation velocity (˚/s)	-424.96 ± 91.91		-435.79 ± 138.18		-403.52 ± 130.43		-475.62 ± 79.34		-445.89 ± 95.11		-410.90 ± 70.63	

Note. A = significant difference from WITHOUT condition, B = Significant difference from PROTECTOR condition, AB = significant difference from WITHOUT & PROTECTOR conditions. * = Significant difference between stance foot.

For the ankle in the sagittal plane there was a significant main effect between conditions $F_{(2, 22)} = 7.54$, $P \leq 0.01$, $p\eta^2=0.41$ but not between stance foot for the absolute ROM. Further analysis revealed that the BRACE condition was significantly lower than the WITHOUT ($P \leq 0.01$, 95%CI: 0.98-6.41%) and the PROTECTOR ($P \leq 0.05$, 95%CI: 0.02-5.28%) conditions.

In the coronal plane no significant main effects were found between stance foot. However, significant main effects were found between conditions for angle at footstrike $F_{(2, 22)} = 29.28$, $P \leq 0.01$, $p\eta^2=0.73$, angle at toe-off $F_{(2, 22)} = 13.10$, $P \leq 0.01$, $p\eta^2=0.54$, peak inversion $F_{(2, 22)} = 18.85$, $P \leq 0.01$, $p\eta^2=0.63$, peak eversion $F_{(2, 22)} = 7.12$, $P \leq 0.01$, $p\eta^2=0.39$, and peak eversion velocity $F_{(2, 22)} = 11.92$, $P \leq 0.01$, $p\eta^2=0.52$. Further analysis found the BRACE condition significantly reduced angle at footstrike (WITHOUT; $P \leq 0.001$, 95%CI: 3.30-7.24%, PROTECTOR; $P \leq 0.001$, 95%CI: 2.58-7.06%), angle at toe off (WITHOUT; $P \leq 0.01$, 95%CI: 1.17-5.31%, PROTECTOR; $P \leq 0.01$, 95%CI: 1.15-4.35%), peak inversion (WITHOUT; $P \leq 0.001$, 95%CI: 1.81-5.46%, PROTECTOR; $P \leq 0.001$, 95%CI: 1.51-5.55%), and peak eversion (WITHOUT; $P \leq 0.05$, 95%CI: 0.36-3.83%, PROTECTOR; $P \leq 0.05$, 95%CI: 0.37-3.89%), compared to both the WITHOUT and PROTECTOR conditions. The BRACE condition also significantly ($P \leq 0.01$, 95%CI: 24.87-131.98%) reduced peak eversion velocity compared to the WITHOUT condition.

No significant differences ($P > 0.05$) were found in any of the planes of motion for both the knee joint or the hip joint. No significant differences ($P > 0.05$) were found between LEFT foot strike and RIGHT foot strike.

Table 5.6. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained from the left and right limb during stance phase of a V-Cut manoeuvre for the female participants.

		LEFT Foot Strike			RIGHT Foot Strike		
		WITHOUT	PROTECTOR	BRACE	WITHOUT	PROTECTOR	BRACE
KNEE	**Sagittal plane (+ = Flexion / - = Extension)**						
	Angle at footstrike	13.46 ± 5.04	13.34 ± 6.88	14.88 ± 6.26	13.74 ± 6.37	15.49 ± 8.54	15.76 ± 6.42
	Angle at toe-off	16.79 ± 7.95	15.48 ± 6.73	17.50 ± 9.46	16.71 ± 5.76	16.23 ± 8.52	16.84 ± 7.69
	Peak flexion	45.66 ± 7.06	46.02 ± 8.14	48.48 ± 9.17	46.57 ± 8.00	46.48 ± 10.07	47.59 ± 7.18
	Absolute ROM	34.44 ± 4.94	35.34 ± 7.69	36.16 ± 7.28	34.79 ± 5.74	34.76 ± 7.78	35.17 ± 5.86
	Relative ROM	32.19 ± 5.70	32.68 ± 6.93	33.59 ± 8.28	32.83 ± 5.48	30.99 ± 5.59	31.83 ± 6.32
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at footstrike	-2.50 ± 2.48	-2.28 ± 3.03	-2.73 ± 2.70	-2.32 ± 3.38	-2.66 ± 2.90	-2.03 ± 2.99
	Angle at toe-off	-3.56 ± 2.41	-3.51 ± 2.83	-3.13 ± 1.99	-4.14 ± 3.31	-4.45 ± 3.27	-3.66 ± 2.98
	Peak adduction	2.48 ± 4.14	2.33 ± 3.83	2.92 ± 3.72	1.36 ± 4.50	0.67 ± 4.37	1.45 ± 4.01
	Absolute ROM	7.30 ± 3.52	7.17 ± 3.35	7.76 ± 3.83	7.15 ± 2.89	6.66 ± 2.42	7.01 ± 2.26
	Relative ROM	4.98 ± 3.65	4.60 ± 3.01	5.65 ± 3.31	3.68 ± 2.68	3.33 ± 2.80	3.47 ± 2.37
	Transverse plane (+ = Internal / - = External)						
	Angle at footstrike	-5.39 ± 6.46	-7.11 ± 6.38	-4.33 ± 7.27	-7.80 ± 7.24	-9.60 ± 7.17	-8.29 ± 6.99
	Angle at toe-off	-5.80 ± 4.65	-8.78 ± 5.09	-6.84 ± 3.36	-5.92 ± 4.21	-8.11 ± 4.87	-6.49 ± 5.44
	Peak rotation	6.37 ± 6.62	4.16 ± 5.51	5.90 ± 6.09	5.06 ± 5.04	2.22 ± 4.63	3.97 ± 5.07
	Absolute ROM	15.80 ± 6.04	15.39 ± 3.63	15.18 ± 5.18	15.59 ± 5.50	14.76 ± 4.43	14.54 ± 4.52
	Relative ROM	11.76 ± 6.78	11.27 ± 3.38	10.23 ± 4.86	12.86 ± 6.06	11.82 ± 4.91	12.26 ± 5.00
		LEFT Foot Strike			**RIGHT Foot Strike**		
		WITHOUT	**PROTECTOR**	**BRACE**	**WITHOUT**	**PROTECTOR**	**BRACE**
HIP	**Sagittal plane (+ = Flexion / - = Extension)**						
	Angle at footstrike	36.92 ± 9.62	37.29 ± 11.40	39.24 ± 11.77	37.70 ± 10.78	39.83 ± 13.93	39.96 ± 12.06
	Angle at toe-off	-8.49 ± 9.40	-9.45 ± 8.55	-7.60 ± 12.60	-7.83 ± 10.43	-8.26 ± 10.76	-6.91 ± 11.02
	Peak flexion	40.35 ± 11.63	40.34 ± 13.18	43.34 ± 14.85	40.82 ± 12.84	42.20 ± 16.03	43.25 ± 13.54
	Absolute ROM	48.82 ± 6.96	50.12 ± 8.31	50.95 ± 7.73	49.35 ± 7.24	50.47 ± 9.52	50.16 ± 8.06
	Relative ROM	45.40 ± 7.93	47.06 ± 7.14	46.85 ± 8.97	46.23 ± 6.20	48.10 ± 7.84	46.87 ± 6.97
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at footstrike	-1.36 ± 6.90	-1.43 ± 6.81	-1.09 ± 7.26	-1.84 ± 3.05	-1.33 ± 3.40	-1.92 ± 3.76
	Angle at toe-off	-14.58 ± 6.20	-15.69 ± 7.70	-14.33 ± 8.69	-11.61 ± 7.09	-12.75 ± 7.64	-13.21 ± 7.10
	Peak adduction	2.40 ± 6.68	1.23 ± 7.47	2.26 ± 7.18	3.27 ± 4.27	3.13 ± 5.48	2.21 ± 4.62
	Absolute ROM	18.05 ± 5.14	18.06 ± 6.62	18.78 ± 7.64	15.86 ± 7.53	16.91 ± 9.15	16.30 ± 8.36
	Relative ROM	14.28 ± 4.84	15.40 ± 5.88	15.43 ± 7.33	10.75 ± 6.79	12.45 ± 7.74	12.17 ± 6.76
	Transverse plane (+ = Internal / - = External)						
	Angle at footstrike	7.28 ± 7.16	7.72 ± 7.39	6.74 ± 7.30	0.27 ± 7.05	1.27 ± 6.57	2.44 ± 7.36
	Angle at toe-off	0.11 ± 7.49	0.93 ± 7.99	1.16 ± 8.1	-3.92 ± 5.47	-3.67 ± 5.83	-4.57 ± 6.73
	Peak rotation	-2.42 ± 7.48	-0.99 ± 7.01	-1.68 ± 6.06	-7.56 ± 5.99	-7.33 ± 4.00	-8.30 ± 5.62
	Absolute ROM	18.48 ± 5.05	17.66 ± 4.97	18.63 ± 4.43	15.28 ± 4.54	16.89 ± 4.58	19.63 ± 4.57
	Relative ROM	8.78 ± 6.41	8.95 ± 7.22	10.21 ± 6.73	7.46 ± 6.28	8.29 ± 7.32	8.89 ± 6.80

5.5 Discussion

The aims of the current study were to investigate the effects of ankle protectors on ankle kinematics during a 45˚ cutting manoeuvre, compare the effects of ankle protectors with braced and unbraced ankles to establish which it more closely resembles, investigate the effects of ankle protectors on knee and hip kinematics, investigate the effects on dominant and non-dominant limb, and investigate the effects on both male and female populations. Previous research reviewing the effectiveness of ankle braces has found them to reduce the risk of inversion injury (Farwell, et al., 2013) and it is a reduction in coronal plane kinematics which is likely the main contributor to the reduction in risk of inversion injuries (Tang, et al., 2010). Ankle protectors aim to reduce contusion injuries and have previously been found to be effective at this (Ankrah & Mills, 2004). The previous chapters have established that ankle protectors have a small effect on sagittal plane ankle kinematics in males during running and also effect sagittal and coronal ankle kinematics in females during running. Also there have been effects found using ankle protectors during the take-off phase of a CMVJ when used by females which also have an effect on knee kinematics. However, it was previously unknown whether ankle protectors inadvertently restrict the ankle during a 45˚ cutting manoeuvre, due to their location, which may cause restrictions similar to ankle braces.

5.5.1 Discussion of male results

The ankle brace used for the current study produced similar results to previous studies for the ankle kinematics as it significantly reduced peak inversion (Commons & Low, 2014; Klem, et al., 2017). This means that the ankle brace used by the current study are a good reference frame to compare the ankle protectors to, to assess restrictive properties. However, the results of the current study found no significant restrictions for any ankle kinematic variables for the left or right ankle between wearing and not wearing ankle protectors. Therefore, using the ankle brace as the reference frame was not necessary. Based on this finding ankle protectors cannot reduce the risk of ankle-inversion injuries during a 45˚ cutting manoeuvre when used by a male population and can only protect against contusion injuries.

Although not the focus of the current study one interesting finding is that there were significant differences in coronal kinematics found between right stance foot and the left stance foot for ankle angles. All participants were right foot dominant, defined as the foot they used to kick the ball. It can then be speculated that the participants were more accustomed to cutting to the right using their left foot as to take the football onto their favoured kicking foot. This could possibly account for the significant differences between stance foot for the coronal measures. Studies looking at the difference between dominant and non-dominant limbs have mainly focused on the effects of leg dominance on ACL injury, and have found no significant difference for ankle kinematics (Weinhandl, et al., 2017), or knee and hip kinematics in males when cutting (Greska, et al., 2017; Pollard, et al., 2018). However, studies looking at the frequency of injuries to dominant and non-dominant limbs of male football players have found that more injuries are sustained by the dominant ankle compared with the non-dominant ankle (Hawkins, et al., 2001; Woods, et al., 2003). The significant difference in coronal plane ankle kinematics between the dominant and non-dominant limb found by the current study could offer an insight into one of the mechanics behind this increased risk of sustaining an ankle injury to the dominant limb.

Although there were significant differences between stance limbs for ankle kinematics and significant reductions found when using the ankle brace compared to the ankle protectors and not wearing any ankle device these restrictions did not significantly alter knee or hip kinematics of either of the legs. Previous research has also found when using ankle orthotics during cutting manoeuvres that there are no significant changes to knee or hip kinematics (Greene, et al., 2014). This means that the use of ankle protectors and ankle braces does not increase the likelihood of an injury occurring further up the kinematic chain during a 45˚ cut. Additionally, no significant differences were found between any of the conditions or stance limbs for ground reaction forces and stance times which mirrors previous research (Bezalel, 2009; Greene, et al., 2014; West & Campbell, 2014). These findings indicate that ankle protectors do not increase the risk of injuries associated with attenuating ground reaction forces.

The current study has established that ankle protectors do not restrict either the dominant or non-dominant ankle when performing a 45˚ cutting manoeuvre and do not restrict the ankle like ankle braces when used by males. Therefore, ankle protectors can only reduce the risk of ankle-contusion injuries and not ankle-inversion injuries during cutting manoeuvres in male populations. Additionally, ankle protectors do not significantly affect knee or hip kinematics or ground reaction forces. It must be noted however that there appears to be significant differences between ankle kinematics when comparing the dominant limb to the non-dominant limb in males. These difference could increase the risk of ankle-inversion injuries to the dominant limb of male football players.

5.5.2 Discussion of female results

The female results show that the ankle braces significantly reduce coronal plane motion, in particular angle at footstrike and peak inversion angle, for both the dominant and non-dominant stance foot. These findings are consistent with previous research investigating the effects of ankle braces on cutting manoeuvres which have also found reductions in peak inversion (Commons & Low, 2014; Klem, et al., 2017). Previous research has also established excessive ankle-inversion to be a key component in ankle-inversion injuries (Kristianslund, et al., 2011) which makes the semi-rigid ankle braces used in the current study a good reference frame to assess the ankle protectors effectiveness at reducing the risk of inversion injuries. Unlike the previous chapters the ankle protectors used in the current study did not significantly reduce any ankle kinematics in any plane of motion for either the dominant or non-dominant stance foot and replicated ankle kinematics similar to an unbraced ankle when used by females. Also there were no significant effects on kinematics found at the knee or hip of either stance limb which is in line with previous research which has found the same during cutting manoeuvres (Greene, et al., 2014). This means that the use of ankle protectors and ankle braces do not increase the likelihood of an injury occurring further up the kinematic chain during a 45° cut. Additionally, no significant differences were found between any of the conditions or stance limbs for ground reaction forces and stance times which mirrors previous research (Bezalel, 2009; Greene, et al., 2014; West & Campbell, 2014). These findings indicate that ankle protectors do not increase the risk of injuries associated with attenuating ground reaction forces.

Unlike the males' the females' dominant and non-dominant ankle kinematics did not significantly differ from one another meaning that neither limb has a greater risk of injury than the other. Additionally, there were no significant differences between dominant and non-dominant knee and hip kinematics. These findings are interesting because limb dominance theory suggests that females exhibit greater kinematic leg asymmetries than males (Hewett, et al., 2010; Weinhandl, et al., 2017) whereas the results of the current study has found the reverse. The discrepancy between the findings of the current study and previous research is not clear, one possible explanation could be the playing experience or playing level of the participants used. However, this is beyond the scope of the current study and further research should investigate the effects of playing experience and level on asymmetries between the dominant and non-dominant limbs.

The current study has established that ankle protectors do not restrict either the dominant or non-dominant ankle when performing a 45° cutting manoeuvre and do not restrict the ankle like ankle braces when used by females. Therefore, ankle protectors can only reduce the risk of ankle-contusion injuries and not ankle-inversion injuries during cutting manoeuvres in female populations. Additionally, ankle protectors do not significantly affect knee kinematics, hip kinematics or ground reaction forces.

5.5.3 Limitations of the study

One of the main limitations of the current study is markers were affixed to the malleoli and used for defining segments in the static model however although these markers were not used to track the dynamic movement there is still a possibility that error in their application may cause errors within the data collected. Another limitation with the marker set used is that markers were attached to the trainers of the participant and not to skin, but by removing the trainers and applying to the skin has the potential to alter the running and cutting motion of the participants and reduces the ecological validity of the test results. However, it must be noted that the markers attached to the trainer might not accurately resemble the true movement of the foot contained within the trainer. The foot was considered a rigid segment which means the effects on the differing joints that make up the ankle complex cannot be individually investigated. Finally, some of the kinematic data show large standard deviations. These large deviations may be due to differing running and cutting styles exhibited by the participants, and in some cases such as the hip, due to the movement of the tightly fitted sports shorts worn by participants.

5.5.4 Considerations for next study

This study and the previous two have all considered movements common within football but none have included a fundamental piece of equipment frequently utilised in the sport. Therefore, the fourth and final study will investigate the effects of ankle protectors on a movement that includes a football.

6. The effects of ankle protectors on the stance limb during kicking a football: a comparison to braced and unbraced ankles.

6.1 Introduction

The previous chapters have focused on motions common in football but ones that can be performed without a ball. This chapter will consider the stance limb of a player during kicking a ball. Planting the foot during kicking a ball leaves the ankle of the stance limb in a vulnerable position. At this point in time a poorly timed tackle from an opponent that impacts the stance limb can cause both an ankle-inversion injury and/or an ankle-contusion injury (Andersen, et al., 2004). Additionally, ankle-inversion injuries can occur by the misplacement of the stance foot (Andersen, et al., 2004). These injuries account for a smaller portion of total ankle-inversion injuries when compared to other ankle injury mechanics but still make up a sizable portion (Woods, et al., 2003).

Studies investigating kicking a football have mainly concentrated on the kicking limb (Barfield, 1998; De Witt & Hinrichs, 2012; Kellis & Katis, 2007; Lees & Nolan, 1998; Sinclair, et al., 2014b; Sinclair, et al., 2014c) whilst some have considered the stance limb as well (Lees & Nolan, 1998; Kellis, et al., 2004; Lees, et al., 2009; Sinclair, et al., 2014b; Sinclair, et al., 2014c) however, none have considered the effects of ankle protectors or ankle braces on the stance limb kinematics. Previous chapters have established that the Aircast A60 ankle braces reduce coronal and sagittal plane motion, in particular ankle inversion which is considered the main mechanism behind ankle-inversion injuries (Tang, et al., 2010), and so it is possible that ankle braces will reduce these motions of the stance limb during kicking a ball. Additionally, the use of ankle braces during kicking a ball has been found not to significantly affect kicking accuracy (Putnam, et al., 2012). Lees, et al. (2009) found that for the stance limb, at the point of ground contact, the ankle exhibits a small amount of plantarflexion, followed by dorsiflexion as the shank moves over the point of contact with the floor, this continues until the ball is kicked where the ankle then goes back into plantarflexion during the follow through. The knee of the stance limb is flexed at initial ground contact and remains flexed throughout the duration of the kick reaching a peak of around 42˚ and the hip is flexed and subsequently only extends, most rapidly prior to ball contact (Lees, et al., 2009). Unlike previous studies Sinclair, et al. (2014b) reported more detailed kinematics, including coronal plane kinematics, on the stance limb when kicking with the dominant and non-dominant limb. The study found the stance limb exhibited a small amount of ankle eversion at the point of initial footstrike and continued to evert until the end of the phase, determined as the point at which the kicking limb impacted the ball. These findings were replicated again by Sinclair, et al. (2014c) in another study this time investigating ball velocity.

Ankle protectors have been found to be effective at reducing the risk of contusion injuries (Ankrah & Mills, 2004) and ankle braces have been found to be effective at reducing the risk of ankle-inversion injuries for both male (Surve, et al., 1994) and female (Sharpe, et al., 1997) football players. However, it is currently unknown how either effect the stance limb during kicking. The previous chapters have established that ankle protectors have a small effect on sagittal plane ankle kinematics in males during running and also effect sagittal and coronal ankle kinematics in females during running. There have been effects found using ankle protectors during the take-off phase of a CMVJ when used by females which also have an effect on knee kinematics. However, during a 45˚ cutting manoeuvre ankle protectors were found not to restrict either the dominant or non-dominant ankle when used by either gender whereas ankle braces have reduced coronal plane ankle kinematics in all of the previous chapters. Based on these findings there is a possibility that ankle protectors might have a small effect on stance limb of football players when kicking a ball and ankle braces might significantly effect coronal plane ankle kinematics. Therefore, the current study aims to investigate the effects of ankle protectors on ankle kinematics of the stance limb of a player during kicking a football, compare the effects of ankle protectors with braced and unbraced ankles to establish which it more closely resembles, investigate the effects of ankle protectors on knee and hip kinematics, and investigate the effects on both males and female populations. It is hypothesised that ankle protectors will not reduce ankle kinematics and produce similar kinematics to an unbraced ankle.

6.2 Methodology

6.2.1 Participants

Twelve male (aged; 25.83±6.37 years, height; 177.78±7.17 cm, body mass; 74.59±10.95 kg and BMI; 23.57±2.82) and twelve female (aged; 26.75±9.66 years, height; 164.19±4.11 cm, body mass; 64.44±6.69 kg and BMI; 23.95±2.72) participants took part in this study. Participants were recruited from local and university football teams via opportunity sampling using poster adverts. The inclusion criteria for the study was that the participants were aged between 18 and 35, currently playing for a football team, were injury free at the time of testing, and were right foot dominant. Foot dominance was determined as the foot the participant favoured to kick a football with. All participants provided written consent in line with the University of Central Lancashire's ethical panel (STEMH 391).

6.2.2 Procedure

Participants kicked a stationary football at a goal in three test conditions; wearing ankle braces (BRACE), wearing ankle protectors (PROTECTOR) and with uncovered ankles (WITHOUT). The order the participants performed the test conditions in was randomised and five successful trials were recorded for each test condition. A successful trial was determined as one in which the participant followed a 30˚ approach angle to the football, planted their stance limb on an embedded force plate (Kistler Instruments Ltd., Alton, Hampshire) and struck the ball as hard as they could into the centre of a goal. The force plate sampled at 1000 Hz and was used to determine the start point of the kicking manoeuvre. This point was determined as the point where the force plate first recorded a vertical ground reaction force (VGRF) that exceeded 20N (Sinclair, et al., 2011). The end point was determined as the point in which the kicking foot impacted with the football. This point was determined as the point where the balls velocity passed a threshold of 0.1. Kinematic data were recorded using an eight camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) tracking retro-reflective markers at a sampling rate of 250 Hz. Using the calibrated anatomical system technique (CAST) (Cappozzo, et al., 1995) the retro-reflective markers were attached to the 1st and 5th metatarsal heads, calcaneus, medial and lateral malleoli, the medial and lateral femoral epicondyles, the greater trochanter, Left and right anterior superior iliac spine, and left and right posterior superior iliac spine. These markers were used to model the stance foot, shank, thigh, and pelvis segments in six degrees of freedom. Rigid plastic mounts with four markers on each were also attached to the shank and thigh and were secured using elasticated bandage. These were used as tracking markers for the shank and thigh segments. To track the foot the 1st and 5th metatarsal heads and the calcaneus were used and to track the pelvis the left and right anterior superior iliac spine, and left and right posterior superior iliac spine were used. To model and track the football five strips of retroreflective tape were strategically placed around the ball so that all five could be viewed by the cameras when placed next to the force plate. Before dynamic trials were captured in each condition a static trial of the participant stood in the anatomical position was captured with the football placed in front of them.

6.2.3 Ankle braces, ankle protectors, football, and footwear.

The ankle protectors used for the current investigation were a pair of Nike ankle shield 10 (Nike Inc, Washington County, Oregon, USA) and the ankle braces used were a pair of Aircast A60 (DJO, Vista, CA, USA). The football used for the study was an Addidas UEFA champions league final MILANO 2016 match replica ball. All males completed the testing wearing a pair of Adidas F10 TRX TF football trainers and all female participants completed the testing wearing a pair of Umbro Speciali cup TF football trainers.

6.2.4 Data processing

Anatomical and tracking markers were identified within the Qualisys Track Manager software and then exported as C3D files to be analysed using Visual 3D software (C-Motion, Germantown, MD, USA). To define the centre points of the ankle and knee segments the two marker methods were utilised for both. These methods calculate the centre of the joint using the positioning of the malleoli markers for the ankle centre and the femoral epicondyle markers for the knee centre (Graydon, et al., 2015; Sinclair, et al., 2015). To calculate the hip joint centre a regression equation which uses the position of the ASIS markers was utilised (Sinclair, et al., 2014a). The stance phase of the kicking manoeuvres was filtered at 15Hz using a low pass 4th order zero-lag filter Butterworth filter (Sinclair, et al., 2014b; Sinclair, et al., 2014c). Data were normalized to 100% of the stance phase then processed trials were used to produce means of the five trials for each test condition for each participant. 3D kinematics of the ankle, knee and hip joints of the stance limb were calculated using an XYZ cardan sequence of rotations. The 3D joint kinematic measures which were extracted for further analysis were 1) angle at footstrike, 2) angle at ball impact, 3) peak angle during the kicking phase, 4) absolute range of motion (absolute ROM) calculated by taking the maximum angle from the minimum angle during stance, 5) relative range of motion (relative ROM) calculated using the angle at footstrike and the first peak value after footstrike, 6) peak angular velocities for the ankle during the stance phase. Measures taken from the force plate to be analysed were 1) peak forces 2) instantaneous loading rate calculated as the maximum increase in vertical force between frequency intervals, 3) average loading rate calculated by dividing the peak vertical impact force by the time to the impacts peak 4) stance time. The force data were normalised to bodyweights (BWs) for each participant to allow comparisons across the data set to be investigated.

6.2.5 Statistical analyses

Data analysis was conducted using SPSS v22.0 (SPSS Inc., Chicago, IL, USA). The means of the five trials for each of the three test conditions were compared using one-way repeated measures ANOVA with significant findings, accepted at $P \leq 0.05$ level, being further explored using post-hoc pairwise comparisons. Effect sizes were determined using partial Eta^2 (pη^2).

6.3 Male Results

Tables 6.1-6.3 and figures 6.1 & 6.2 present the key parameters of interest obtained from the stance limb during kicking a stationary football.

6.3.1 Kinetic and temporal parameters

Table 6.1. Kinetic and temporal variables (means and standard deviations) obtained from the stance limb during kicking a stationary football for the male participants.

	WITHOUT		**PROTECTOR**		**BRACE**	
Peak Vertical Impact Force (BW)	2.30 ± 0.28		2.27 ± 0.23		2.30 ± 0.36	
Peak Braking Force (BW)	0.83 ± 0.11		0.79 ± 0.15		0.80 ± 0.14	
Peak Medial Force (BW)	0.72 ± 0.19		0.70 ± 0.20		0.70 ± 0.18	
Peak Lateral Force (BW)	0.02 ± 0.03		0.01 ± 0.03		0.01 ± 0.03	
Instantaneous Loading Rate (BW.s)	202.44 ± 58.31		199.37 ± 63.19		213.07 ± 77.19	
Average Loading Rate (BW.s)	114.72 ± 46.31		106.52 ± 36.58		107.86 ± 51.45	
Stance Time (s)	0.14 ± 0.02		0.14 ± 0.02		0.14 ± 0.02	

The kinetic and temporal variables exhibited no significant differences ($P > 0.05$) between the WITHOUT, PROTECTOR, and BRACE conditions.

6.3.2 3D Kinematic Parameters

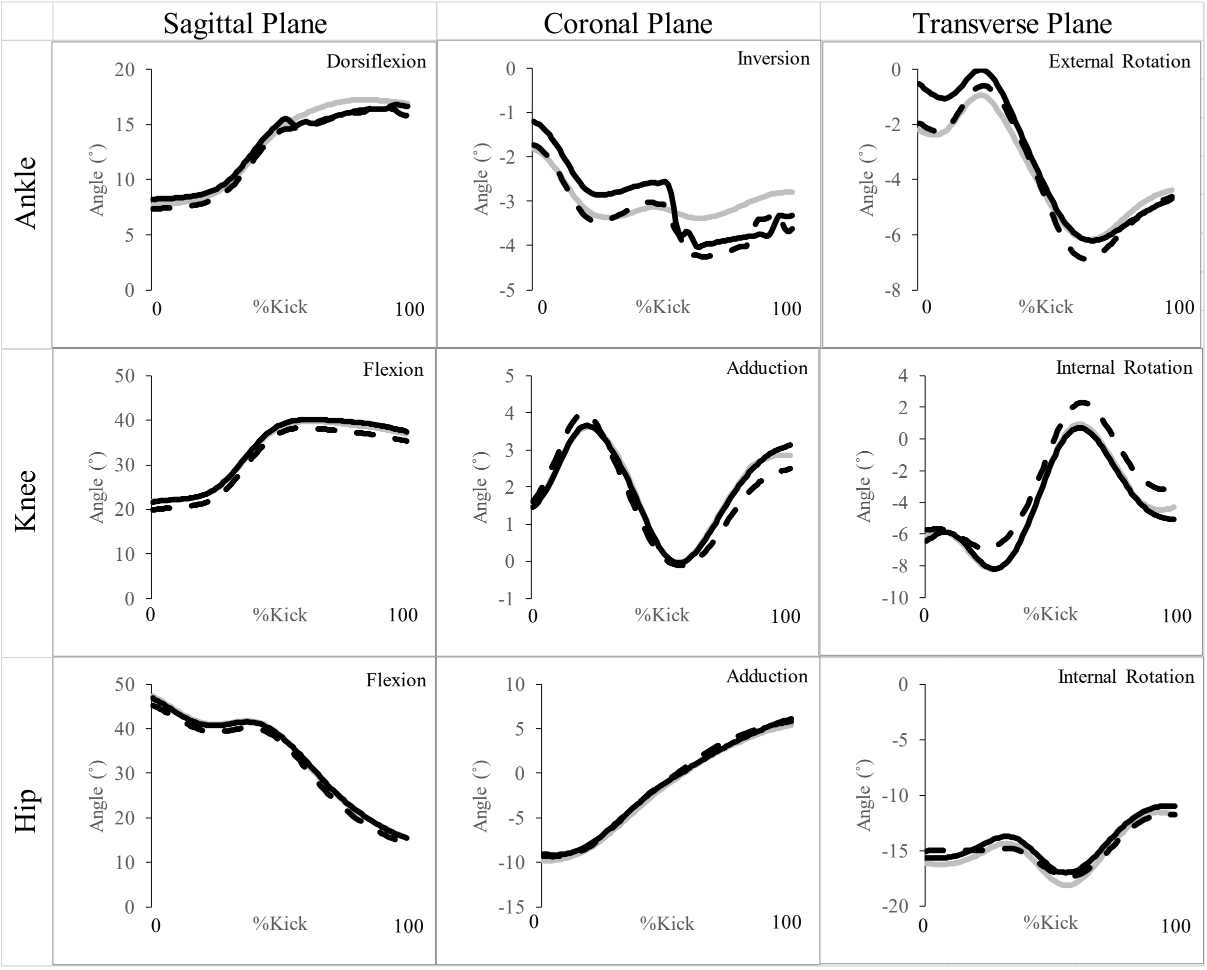

Figure 6.1. Mean ankle, knee, and hip kinematics obtained from the stance limb during kicking a stationary football for the sagittal, coronal, and transverse planes for the male participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

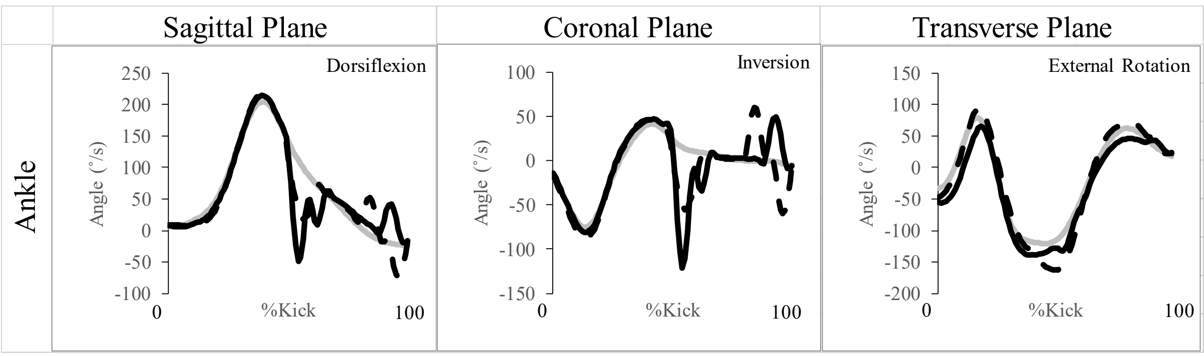

Figure 6.2. Mean ankle velocity obtained from the stance limb during kicking a stationary football for the sagittal, coronal, and transverse planes for the male participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

Table 6.2. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained from the stance limb during kicking a stationary football for the male participants.

		WITHOUT	PROTECTOR	BRACE
ANKLE	**Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)**			
	Angle at footstrike	7.13 ± 4.29	8.13 ± 3.34	7.47 ± 4.09
	Angle at ball impact	14.39 ± 7.26	15.32 ± 7.99	15.68 ± 6.71
	Peak dorsiflexion	16.40 ± 5.92	17.34 ± 5.92	16.77 ± 5.83
	Absolute ROM	11.62 ± 3.56	11.74 ± 4.24	10.68 ± 3.13
	Relative ROM	10.11 ± 3.68	10.04 ± 3.69	10.15 ± 3.56
	Peak dorsifelxion velocity (˚/s)	266.23 ± 146.33	271.48 ± 154.68	222.24 ± 63.39
	Peak plantarflexion velocity (˚/s)	-102.55 ± 110.13	-97.94 ± 113.40	-87.25 ± 97.09
	Coronal plane (+ = Inversion / - = Eversion)			
	Angle at footstrike	-1.78 ± 1.98	-1.41 ± 2.48	-1.87 ± 2.35
	Angle at ball impact	-4.94 ± 5.60	-4.61 ± 6.27	-4.07 ± 5.08
	Peak inversion	-1.33 ± 1.72	-0.89 ± 2.40	-1.40 ± 2.00
	Peak eversion	-6.81 ± 6.01	-6.57 ± 6.67	-5.63 ± 5.07
	Absolute ROM	5.49 ± 5.34	5.67 ± 5.65	4.23 ± 4.28
	Relative ROM	5.04 ± 5.61	5.15 ± 5.93	3.76 ± 4.53
	Peak inversion velocity (˚/s)	119.78 ± 147.64	121.48 ± 139.83	73.45 ± 32.34
	Peak eversion velocity (˚/s)	-139.74 ± 150.65	-135.47 ± 150.77	-122.44 ± 134.10
	Transverse plane (+ = External / - = Internal)			
	Angle at footstrike	-2.16 ± 3.01	-0.86 ± 3.61	-2.36 ± 3.39
	Angle at ball impact	-4.44 ± 3.61	-4.65 ± 4.66	-4.61 ± 3.21
	Peak rotation	-7.97 ± 3.15	-7.72 ± 3.93	-7.64 ± 2.83
	Absolute ROM	8.75 ± 3.58	9.30 ± 3.26	7.77 ± 2.78
	Relative ROM	5.82 ± 3.02	6.86 ± 3.39	5.28 ± 2.48
	Peak external rotation velocity (˚/s)	202.36 ± 122.61	185.27 ± 128.24	174.28 ± 96.82
	Peak internal rotation velocity (˚/s)	-272.50 ± 93.57	-260.97 ± 71.96	-242.52 ± 71.11

No significant differences (P > 0.05) were found in in any of the planes of motion for the ankle joint.

Table 6.3. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained from the stance limb during kicking a stationary football for the male participants.

		WITHOUT	PROTECTOR	BRACE
KNEE	**Sagittal plane (+ = Flexion / - = Extension)**			
	Angle at footstrike	20.44 ± 6.12	22.31 ± 5.64	21.99 ± 6.50
	Angle at ball impact	34.90 ± 8.99	37.46 ± 8.43	37.70 ± 7.77
	Peak flexion	40.56 ± 5.78	41.87 ± 5.97	41.90 ± 6.00
	Absolute ROM	21.51 ± 5.41	20.51 ± 5.24	20.73 ± 5.52
	Relative ROM	20.13 ± 6.19	19.56 ± 5.76	19.90 ± 6.14
	Coronal plane (+ = Adduction / - = Abduction)			
	Angle at footstrike	1.32 ± 3.04	1.25 ± 2.07	1.38 ± 2.20
	Angle at ball impact	2.16 ± 3.06	2.79 ± 3.42	2.39 ± 3.11
	Peak adduction	4.17 ± 3.03	4.57 ± 3.14	4.28 ± 2.51
	Absolute ROM	5.42 ± 3.38	6.01 ± 2.13	5.85 ± 1.83
	Relative ROM	2.85 ± 1.60	3.33 ± 1.99	2.90 ± 1.71
	Transverse plane (+ = Internal / - = External)			
	Angle at footstrike	-5.99 ± 2.08	-6.13 ± 3.15	-5.87 ± 1.80
	Angle at ball impact	-3.43 ± 3.10	-4.54 ± 3.41	-3.46 ± 3.75
	Peak rotation	2.39 ± 3.71	2.38 ± 3.58	2.57 ± 3.20
	Absolute ROM	10.93 ± 3.63	12.05 ± 2.48	11.58 ± 2.28
	Relative ROM	8.38 ± 2.57	8.52 ± 2.22	8.43 ± 2.66
HIP		**WITHOUT**	**PROTECTOR**	**BRACE**
	Sagittal plane (+ = Flexion / - = Extension)			
	Angle at footstrike	46.64 ± 10.22	47.53 ± 10.28	47.60 ± 9.70
	Angle at ball impact	14.81 ± 12.46	15.76 ± 12.01	16.09 ± 10.93
	Peak flexion	46.79 ± 10.13	47.61 ± 10.24	47.61 ± 9.71
	Absolute ROM	32.00 ± 9.44	31.93 ± 8.45	31.62 ± 8.22
	Relative ROM	31.85 ± 9.66	31.84 ± 8.62	31.61 ± 8.25
	Coronal plane (+ = Adduction / - = Abduction)			
	Angle at footstrike	-8.82 ± 6.89	-8.93 ± 6.04	-9.25 ± 6.33
	Angle at ball impact	6.09 ± 3.23	5.91 ± 5.01	5.33 ± 4.33
	Peak adduction	6.39 ± 3.42	5.95 ± 5.01	5.38 ± 4.35
	Absolute ROM	15.90 ± 4.55	15.66 ± 5.39	15.36 ± 5.34
	Relative ROM	15.21 ± 5.18	14.88 ± 6.10	14.64 ± 6.04
	Transverse plane (+ = Internal / - = External)			
	Angle at footstrike	-15.17 ± 8.43	-15.47 ± 8.67	-15.82 ± 9.36
	Angle at ball impact	-12.39 ± 8.62	-12.06 ± 8.76	-12.83 ± 8.99
	Peak rotation	-19.99 ± 8.33	-19.65 ± 8.50	-20.64 ± 9.09
	Absolute ROM	10.33 ± 3.84	10.20 ± 3.21	10.87 ± 3.53
	Relative ROM	5.52 ± 3.39	6.02 ± 5.02	6.05 ± 4.33

No significant differences ($P > 0.05$) were found in in any of the planes of motion for the knee joint or the hip joint.

6.4 Female Results

Tables 6.4-6.6 and figures 6.3 & 6.4 present the key parameters of interest obtained from the stance limb during kicking a stationary football.

6.4.1 Kinetic and temporal parameters

Table 6.4. Kinetic and temporal variables (means and standard deviations) obtained from the stance limb during kicking a stationary football for the female participants.

	WITHOUT		**PROTECTOR**		**BRACE**	
Peak Vertical Impact Force (BW)	2.31 ± 0.36		2.33 ± 0.37		2.30 ± 0.28	
Peak Braking Force (BW)	0.77 ± 0.24		0.75 ± 0.22		0.76 ± 0.21	
Peak Medial Force (BW)	0.77 ± 0.21		0.76 ± 0.20		0.77 ± 0.18	
Peak Lateral Force (BW)	0.02 ± 0.02		0.02 ± 0.03		0.02 ± 0.02	
Instantaneous Loading Rate (BW.s)	187.03 ± 64.67		186.29 ± 80.14		173.85 ± 44.33	
Average Loading Rate (BW.s)	84.44 ± 29.70		85.33 ± 41.83		88.00 ± 30.78	
Stance Time (s)	0.15 ± 0.01		0.15 ± 0.01		0.15 ± 0.02	

The kinetic and temporal variables exhibited no significant differences ($P > 0.05$) between the WITHOUT, PROTECTOR, and BRACE conditions.

6.4.2 3D Kinematic Parameters

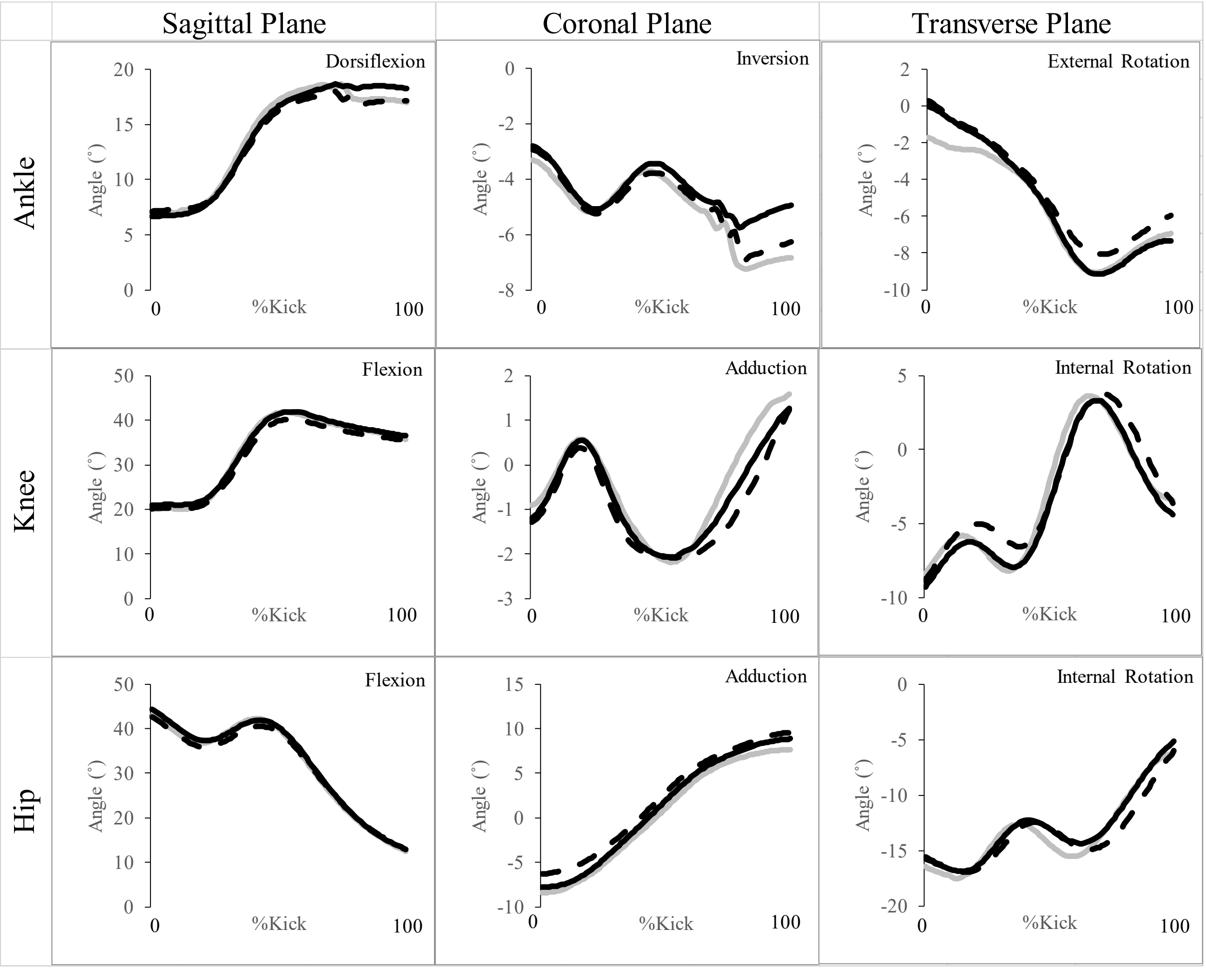

Figure 6.3. Mean ankle, knee, and hip kinematics obtained from the stance limb during kicking a stationary football for the sagittal, coronal, and transverse planes for the female participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

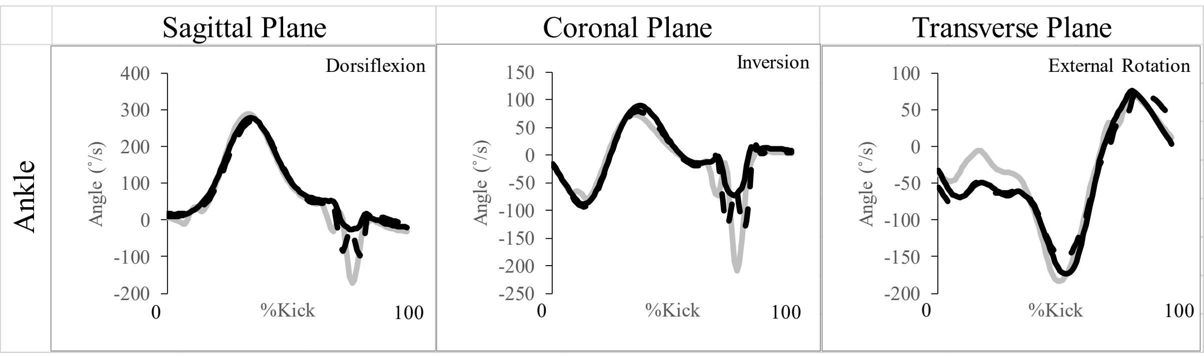

Figure 6.4. Mean ankle velocity obtained from the stance limb during kicking a stationary football for the sagittal, coronal, and transverse planes for the female participants. (WITHOUT = dash, PROTECTOR = black, BRACE = grey).

Table 6.5. Kinematic data (means and stand deviations measured in degrees) for the ankle obtained from the stance limb during kicking a stationary football for the female participants.

		WITHOUT	PROTECTOR	BRACE
	Sagittal plane (+ = Dorsiflexion / - = Plantarflexion)			
	Angle at footstrike	6.86 ± 2.49	6.53 ± 2.60	7.03 ± 1.95
	Angle at ball impact	15.94 ± 8.72	17.10 ± 7.36	16.08 ± 8.70
	Peak dorsiflexion	18.29 ± 5.88	18.61 ± 5.72	18.70 ± 5.60
	Absolute ROM	14.48 ± 4.80	14.32 ± 4.58	14.65 ± 4.26
	Relative ROM	11.44 ± 5.49	12.08 ± 5.96	11.67 ± 5.46
	Peak dorsifelxion velocity (˚/s)	316.26 ± 60.70	310.94 ± 41.53	339.02 ± 63.03
	Peak plantarflexion velocity (˚/s)	-274.95 ± 661.69	-164.10 ± 312.46	-332.09 ± 754.30
	Coronal plane (+ = Inversion / - = Eversion)			
	Angle at footstrike	-2.90 ± 2.09	-2.90 ± 2.42	-3.34 ± 2.33
	Angle at ball impact	-6.95 ± 6.98	-6.08 ± 5.33	-7.71 ± 7.81
ANKLE	Peak inversion	-2.36 ± 1.87	-2.24 ± 2.64	-2.76 ± 2.10
	Peak eversion	-8.54 ± 6.99	-7.82 ± 5.41	-9.41 ± 7.91
	Absolute ROM	6.18 ± 5.82	5.57 ± 4.39	6.65 ± 6.86
	Relative ROM	5.64 ± 6.04	4.92 ± 4.35	6.07 ± 7.07
	Peak inversion velocity (˚/s)	97.89 ± 34.59	107.69 ± 38.77	106.14 ± 43.84
	Peak eversion velocity (˚/s)	-324.05 ± 644.00	-240.04 ± 365.34	-366.35 ± 769.29
	Transverse plane (+ = External / - = Internal)			
	Angle at footstrike	0.11 ± 4.32	-0.38 ± 4.51	-2.10 ± 2.82
	Angle at ball impact	-6.60 ± 4.80	-8.12 ± 5.59	-7.37 ± 4.34
	Peak rotation	-10.12 ± 3.54	-10.84 ± 4.97	-10.13 ± 4.24
	Absolute ROM	12.33 ± 4.34	12.43 ± 5.30	10.09 ± 3.39
	Relative ROM	10.23 ± 5.50	10.46 ± 5.97	8.03 ± 3.91
	Peak external rotation velocity (˚/s)	217.92 ± 87.46	193.43 ± 91.65	194.98 ± 79.64
	Peak internal rotation velocity (˚/s)	-346.44 ± 94.32	-341.34 ± 114.81	-298.71 ± 66.89

No significant differences ($P > 0.05$) were found in in any of the planes of motion for the ankle joint.

Table 6.6. Kinematic data (means and stand deviations measured in degrees) for the knee and hip obtained from the stance limb during kicking a stationary football for the female participants.

		WITHOUT		PROTECTOR		BRACE	
KNEE	**Sagittal plane (+ = Flexion / - = Extension)**						
	Angle at footstrike	19.34 ± 4.44		19.91 ± 5.06		19.85 ± 3.41	
	Angle at ball impact	34.82 ± 9.01		36.04 ± 8.75		35.23 ± 8.45	
	Peak flexion	41.46 ± 5.82		42.93 ± 6.27		423.32 ± 5.63	
	Absolute ROM	23.05 ± 4.93		23.96 ± 5.69		23.70 ± 6.25	
	Relative ROM	22.12 ± 5.29		23.03 ± 5.96		22.47 ± 6.70	
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at footstrike	-1.06 ± 3.27		-1.02 ± 3.09		-0.75 ± 2.85	
	Angle at ball impact	1.83 ± 4.66		2.10 ± 4.86		1.85 ± 4.06	
	Peak adduction	2.68 ± 4.13		2.69 ± 3.66		2.63 ± 3.72	
	Absolute ROM	5.89 ± 1.52		5.79 ± 1.57		5.67 ± 1.33	
	Relative ROM	3.74 ± 2.17		3.71 ± 1.24		3.38 ± 1.71	
	Transverse plane (+ = Internal / - = External)						
	Angle at footstrike	-8.15 ± 5.49		-9.13 ± 5.52		-7.89 ± 4.86	
	Angle at ball impact	-3.85 ± 4.41		-4.74 ± 5.54		-3.70 ± 4.35	
	Peak rotation	4.79 ± 4.03		4.36 ± 4.50		4.87 ± 3.25	
	Absolute ROM	15.49 ± 3.79		15.90 ± 3.13		15.32 ± 2.95	
	Relative ROM	12.94 ± 5.13		13.49 ± 4.28		12.77 ± 4.01	
HIP		**WITHOUT**		**PROTECTOR**		**BRACE**	
	Sagittal plane (+ = Flexion / - = Extension)						
	Angle at footstrike	42.06 ± 6.19		43.19 ± 6.90		42.13 ± 7.54	
	Angle at ball impact	12.28 ± 14.43		12.94 ± 14.24		12.09 ± 15.12	
	Peak flexion	43.45 ± 6.15		44.98 ± 6.73		44.45 ± 7.92	
	Absolute ROM	32.25 ± 10.88		33.35 ± 11.21		33.45 ± 10.83	
	Relative ROM	30.86 ± 12.22		31.57 ± 13.54		31.13 ± 12.77	
	Coronal plane (+ = Adduction / - = Abduction)						
	Angle at footstrike	-6.10 ± 5.20		-7.29 ± 5.32		-7.83 ± 6.01	
	Angle at ball impact	9.50 ± 5.88		8.56 ± 5.14		7.86 ± 7.60	
	Peak adduction	9.98 ± 5.85		9.35 ± 4.68		8.80 ± 7.04	
	Absolute ROM	16.57 ± 5.45		17.19 ± 5.36		17.08 ± 4.66	
	Relative ROM	16.08 ± 5.83		16.64 ± 5.69		16.63 ± 4.87	
	Transverse plane (+ = Internal / - = External)						
	Angle at footstrike	-16.21 ± 11.26		-16.14 ± 12.34		-17.24 ± 10.65	
	Angle at ball impact	-6.40 ± 11.75		-5.36 ± 11.44		-6.46 ± 11.74	
	Peak rotation	-20.09 ± 9.72		-19.90 ± 11.47		-20.61 ± 9.70	
	Absolute ROM	15.05 ± 4.57		15.59 ± 3.92		16.18 ± 4.33	
	Relative ROM	11.17 ± 5.81		11.83 ± 6.41		12.81 ± 6.50	

No significant differences ($P > 0.05$) were found in in any of the planes of motion for the knee joint or the hip joint.

6.5 Discussion

The aims of the current study were; to investigate the effects of ankle protectors on ankle kinematics of the stance limb during kicking a football, compare the effects of ankle protectors with braced and unbraced ankles to establish which it more closely resembles, investigate the effects of ankle protectors on knee and hip kinematics, and investigate the effects on both males and female populations. Previous research reviewing the effectiveness of ankle braces has found them to reduce the risk of inversion injury (Farwell, et al., 2013) and it is a reduction in coronal plane kinematics which is likely the main contributor to the reduction in risk of inversion injuries (Tang, et al., 2010). Ankle protectors aim to reduce contusion injuries and have previously been found to be effective at this (Ankrah & Mills, 2004). The previous chapters have established that ankle protectors have a small effect on sagittal plane ankle kinematics in males during running and also effect sagittal and coronal ankle kinematics in females during running. There have been effects found using ankle protectors during the take-off phase of a CMVJ when used by females which also have an effect on knee kinematics. However, during a 45˚ cutting manoeuvre ankle protectors were found not to restrict either the dominant or non-dominant ankle when used by either gender. It was previously unknown whether ankle protectors or ankle braces restrict the ankle of the stance limb during kicking a football.

6.5.1 Discussion of male and female results

The results of the current study have found that the use of both ankle protectors and ankle braces did not significantly affect any plane of motion of the ankle of the stance limb during kicking a ball when used by males or females. Additionally, both ankle protectors and ankle braces did not significantly affect any plane of motion for the knee or hip, or adversely affect any ground reaction forces. The findings for the ankle protector mirror similar findings to the previous chapters where no differences were also found. However, the findings for the ankle braces produced some surprising results as all of the previous chapters have found the Aircast A60 braces to significantly reduce motion of the ankle. There are a few possible explanations for not finding any significant differences between conditions. Firstly, previous research has found that the stance limb exhibits a small amount of eversion during initial ground contact and continues to evert until the ball is kicked (Sinclair, et al., 2014b) therefore little, if any, inversion occurs during kicking. Secondly, the sagittal plane motion of the ankle exhibits a lot less ROM than the other sporting movements investigated in previous chapters which suggests that the position of the body is over the stance limb for most of the movement to aid with balance and allow the pendulum motion of the kicking limb. This is supported by Lees, et al. (2009) who found the ankle to exhibit a small amount of plantarflexion at initial ground contact, followed by dorsiflexion as the shank moved over the point of contact with the floor, until the ball is kicked where the ankle then goes back into plantarflexion. Therefore, due to this the point at which the restrictive properties of the ankle brace take affect might not be reached. Thirdly, even though there were restrictions in place to try to generate a repeatable kicking technique between participants there were still differences present in the kicking styles of the individuals which are shown by the large standard deviations present in the kinematic data. These differences between kicking techniques might mask any difference between conditions. It is possible that the 30˚ approach angle was not sufficient to elicit a restriction on the ankle from the ankle braces and so future work should look at using greater approach angles to investigate if a change in approach angle causes significant restrictions when using a brace.

The current study has established that ankle protectors and ankle braces do not restrict the ankle of the stance limb when kicking a stationary football when used by males or females. The lack of restriction is likely due to the position of the body relative to the stance limb location during the kick which means little motion of the stance limb occurs. Additionally, ankle protectors do not significantly affect knee kinematics, hip kinematics or ground reaction forces. Therefore, based on these findings ankle protectors and ankle braces do not restrict the ankle and do not adversely affect the placement or position of the stance limb during kicking a ball. However, investigation into the restrictive properties of ankle braces, and comparisons between ankle protectors and braces, cannot be investigated during this movement as the position of the stance limb is as such as to not elicit the restrictive properties of the ankle brace. This makes it difficult to conclude if ankle protectors and ankle braces can reduce the risk of inversion injuries to the stance limb during kicking.

6.5.2 Limitations of the study

Firstly, one of the main limitations of the current study was that no performance measure of final ball velocity or accuracy of shot were measured and therefore it is a possible that there was significant differences in overall performance of the kick which might have had an effect on the 3D kinematic parameters measured. Secondly, the approach angle used was 30˚ which restricted the participants to a prescribed approach route. This might have altered their preferred kicking technique by forcing them to adopt a different approach. Thirdly, only the stance limb was investigated and there may be effects to the kicking limb. Unfortunately, due to the motion of the foot impacting the ball the data collected on the kicking limb was poor and so the decision was made to exclude this data from the current study. Further investigation into methods of tracking the kicking limb should be investigated as the current investigation found that at the point where the foot impacted with the ball caused disruptions in the cameras capturing the markers of the kicking limb. Fourthly, although markers affixed to the malleoli were not used to track the dynamic movement there is still a possibility that error in their application may cause errors within the data collected as they were used for defining segments in the static model. Fifthly, markers were attached to the trainers of the participant and not to skin, however by removing the trainers and applying to the skin has the potential to alter the position of the stance limb and effect the performance of the kick. However, it must be noted that the markers attached to the trainer might not accurately resemble the true movement of the foot contained within the trainer. Finally, the foot was considered a rigid segment which means the effects on the differing joints that make up the ankle complex cannot be individually investigated.

7. Summary of Conclusions.

7.1 Summary of conclusions for running study

When ankle protectors were used by male participants during running it was found that ankle protectors provided very little restriction to the ankle and did not restrict the ankle like ankle braces. It must be noted that although no restrictions were seen in the coronal plane there were reductions in sagittal plane motion for the ankle. These reductions could possibly increase energy demand needed for locomotion and affect performance of other football related movements. Therefore, ankle protectors should only be used as a means to reduce the risk of ankle-contusion injuries and not implemented as a method to reduce the risk of ankle-inversion injuries during running in male populations.

When ankle protectors were used by female participants during running it was found that ankle protectors provided very little restriction to the ankle and did not restrict the ankle like ankle braces. It must be noted that there were some restrictions in the coronal plane that could possibly contribute to the reduction in ankle-inversion injuries when wearing ankle protectors. However, these restrictions were far superior in the braced condition. Additionally, without further exploration these findings must be taken with caution as there were no restrictions found for many of the other coronal kinematic measures. Therefore, ankle protectors should only be used as a means to reduce the risk of ankle-contusion injuries and not implemented as a method to reduce the risk of ankle-inversion injuries unless further research finds these reductions in motion for other football related movements. Also the reductions in sagittal plane motion found for the ankle could possibly increase energy demand needed for locomotion and affect performance of other football related movements when used by female wearers and so further research on different football related movements is necessary.

7.2 Summary of conclusions for countermovement vertical jump study

When ankle protectors were used during the take-off phase of a countermovement vertical jump (CMVJ) they did not adversely affect male populations who utilised them. However, the use of semi-rigid ankle braces, such as the one used in the current study, did adversely affect performance of a CMVJ. During the landing phase of a CMVJ the ankle protectors did not perform like the ankle braces and are only effective at reducing the risk of contusion injuries and cannot protect against ankle-inversion injuries during this manoeuvre. Furthermore, the use of ankle protectors did not significantly affect knee kinematics, hip kinematics, or GRFs and therefore their use does not increase the likelihood of injuries further up the kinematic chain.

When ankle protectors and semi-rigid ankle braces, such as the one used in the current study, were used by female participants during the take-off phase of a CMVJ they reduced total jump height ascertained. This reduction in performance is likely due to the reductions found during take-off in sagittal plane motion of the ankle. Additionally, the use of ankle protectors and ankle braces significantly changed knee kinematics in the sagittal plane during take-off which could possibly increase the risk of knee injuries when used by a female population. During the landing phase of a CMVJ ankle protectors did not perform like an ankle brace and are only effective at reducing the risk of contusion injuries around the ankle and cannot protect against ankle-inversion injuries during this manoeuvre. Furthermore, the use of ankle protectors did not significantly affect knee kinematics, hip kinematics, or GRFs and therefore their use does not increase the likelihood of injuries further up the kinematic chain during landing.

7.3 Summary of conclusions for 45˚ cutting manoeuvre study

When ankle protectors were used by male participants during a 45˚ cutting manoeuvre they were found not to restrict either the dominant or non-dominant ankle and did not restrict the ankle like ankle braces. Therefore, ankle protectors can only reduce the risk of ankle-contusion injuries and not ankle-inversion injuries during cutting manoeuvres. Additionally, ankle protectors did not significantly affect knee or hip kinematics or ground reaction forces. It must be noted however that there appears to be significant differences between ankle kinematics when comparing the dominant limb to the non-dominant limb in males. These difference could increase the risk of ankle-inversion injuries to the dominant limb of male football players.

When ankle protectors were used by female participants during a 45˚ cutting manoeuvre they did not restrict either the dominant or non-dominant ankle and did not restrict the ankle like ankle braces. Therefore, ankle protectors can only reduce the risk of ankle-contusion injuries and not ankle-inversion injuries during cutting manoeuvres. Additionally, ankle protectors did not significantly affect knee kinematics, hip kinematics or ground reaction forces

7.4 Summary of conclusions for the kicking study

Ankle protectors and ankle braces did not restrict the ankle of the stance limb when kicking a stationary football when used by males or females. The lack of restriction was likely due to the position of the body relative to the stance limb location during the kick which meant little motion of the stance limb occurred. Additionally, ankle protectors did not significantly affect knee kinematics, hip kinematics or ground reaction forces. Therefore, based on these findings ankle protectors and ankle braces do not restrict the ankle and do not adversely affect the placement or position of the stance limb during kicking a ball. However, investigation into the restrictive properties of ankle braces, and comparisons between ankle protectors and braces, could not be investigated during this movement as the position of the stance limb was as such as to not elicit the restrictive properties of the ankle brace. This makes it difficult to conclude if ankle protectors and ankle braces can reduce the risk of inversion injuries to the stance limb during kicking.

8. Synthesis of research.

8.1 Contribution to knowledge

Overall the key finding of this thesis is that ankle protectors can only protect against contusion injuries and cannot protect against ankle-inversion injuries. This finding is important as now football players can be better informed when selecting a device to reduce their risk of ankle-contusion injuries or ankle-inversion injuries Ankle injuries are a common occurrence in football (Junge & Dvorak, 2013, Peterson, et al., 2000) and can result in lengthy periods of absence for key players (Waldén, et al., 2013) resulting in reduced chances of teams winning competitive matches (Hägglund, et al., 2013). Ankle injuries cost premier league clubs on average £253,000 a season (JLT Specialty, 2017) with an average time loss of 16±27 days with the more severe ankle ligament injuries leading to 43±33 days lost (Waldén, et al., 2013). Therefore, it is of paramount importance for clubs to reduce the frequency of ankle injuries by investigating and implementing the best methods for reducing ankle injury risk. Previous research has found that the frequency of ankle-inversion injuries can be reduced by using ankle braces (Pedowitz et al 2008) and ankle-contusion injuries can be reduced through the use of ankle protectors (Ankrah & Mills, 2002, Ankrah & Mills, 2004). However, prior to this thesis, no research had investigated if ankle protectors can reduce the risk of ankle-inversion injuries by reducing ankle kinematics.

Another important finding is that the current “one size fits all” design of ankle protectors should be re-evaluated as it can cause significant alterations to sagittal plane kinematics of the ankle for certain footballing related movements. These restrictions can cause reductions in performance of vertical jumps and may increase energy demand during competitive matches. Ideally ankle protectors should offer a range of sizes to accommodate individuals of differing heights to reduce the encroachment of the EVA foam around the front of the foot. This would likely aid in reducing the significant affects found in the sagittal plane for some of the studies found in this thesis. The current design of ankle protectors, like the ones used by the current study, could benefit from changes in material construct to either make them better at dissipating forces, by using newer materials, or by the introduction of firmer materials which are integrated into the foam to protect against both contusion and inversion injuries.

8.2 Areas for further research

This thesis has concentrated on the effects of ankle protectors on kinetic and kinematic parameters but has not accounted for the effects of their use during training and competitive match play on injury frequencies. Although there were few restrictions found when using the ankle protectors there remains a possibility that they do still reduce the frequency of ankle-inversion injuries when utilised by players in competition. The use of ankle protectors could provide the wearer with proprioceptive cues which could reduce the risk of an inversion injury. A future research project should implement a longitudinal study investigating the use of ankle protectors on ankle-inversion injury frequency in football.

During the undertaking of this thesis there have been developments made by manufactures to offer products of which contain ankle protector like structures into football trainers and football boots. Some such products like the Nike phantom series and Adidas x series football boots currently contain structures that partially cover the ankle, and although not specifically designed as ankle protectors, could be further developed to incorporate foams to reduce the risk of contusion injuries. The style of boot is similar to that of high top trainers found in basketball and could offer similar reductions in inversion injuries as seen when high top trainers are utilised in basketball (Taylor, et al., 2015). This would be an interesting avenue for exploration by future research and could be another variable to consider when constructing the study suggested at the end of the paragraph above. Additionally, some companies are now offering ankle protectors of various sizes to accommodate users of different heights. A further research project should investigate the effects of using ankle protectors that are selected based on height verse the “one size fits all” design to assess if the newer protectors reduce the significant reductions in sagittal plane motion that was found during this thesis.

An interesting finding during this thesis was that during the 45° cutting study it was found that for the males there were significant differences between ankle kinematics when comparing the dominant limb to the non-dominant limb. These difference could increase the risk of ankle-inversion injuries to the dominant limb of male football players. A future study should investigate the frequency of ankle-inversion injuries to the dominant and non-dominant legs to ascertain if either is at a greater risk of the injury.

Another development of note is that of the technology to produce materials with better shock absorbent properties than the currently used EVA foams found in ankle protectors. Foams such as that developed by D3O are offering greater impact protection at less thicknesses than that of its EVA counterparts. The use of this foam could be beneficial when incorporated into ankle protectors so that a thinner, more lightweight protector could be developed which offered similar or better protection from contusion injuries than the ankle protectors using EVA foam. A worthy avenue of exploration for a material technologist would be investigating the incorporation of D3O foam into an ankle protector for the use within football. The experiments should focus on making the D3O based ankle protector as thin as possible whilst not detrimentally affecting shock absorption properties.

8.3 General limitations

Although every precaution was taken to make sure that the kinematic data collected was accurate and reliable there still remains a possibility that small amounts of error occurred causing small but significant changes in the angles recorded. The accuracy of the Qualysis system study in section 2.4 found that the maximum error when recording in the centre of the calibrated area was 1.24 mm. Although the maximum error was still relatively small, when looking at small changes in ankle rotations this small error across multiple markers has a small chance of causing significant differences where there were none. However, when looking at the mean error from the same location during the study it was found to be 0.06 mm which indicates the likelihood of finding differences due to the accuracy of the system was very low. Another possible cause of significant differences where there were none could have been due to marker placement. The study in section 2.6.1 which was investigating the ankle joint centre location techniques showed that significant differences could be produced by misplacement of the markers as opposed to the effects of the ankle braces and ankle protectors. This study also showed that although the two marker method was the most reliable for creating the ankle joint centre it could still produce small but significant differences in the kinematic data. However, there were very few significant difference found and none were not found in the coronal plane which was the plane of most interest when investigating ankle-inversion injuries. One other way the kinematic data could have been effected was the cut-offs frequencies used for the Butterworth filters. As a residual analysis was not conducted on the data for each movement there is a possibility that the data could have been over smoothed, resulting in data points of interest being removed, or under smoothed, resulting in white noise skewing the data. This error could have resulted in misinterpretations of the data obtained. However, although a residual analysis was not conducted on the data for each movement the cut-off frequencies were selected for each study based on residual analysis conducted and contained within publications on the same movements as those under investigation.

Sample-size calculations are often seen as an ethical issue if the study is seen as not having sufficient power to produce meaningful results (Bacchetti, et al., 2005). As a priori power calculations were not conducted during this thesis it could be accused of being unethical under this premise if no significant differences were found. However, there was a process in place to determine adequate sample sizes which drew upon previous research investigating similar movements to those contained within this thesis to help determine the sample sizes used. Additionally, as significant differences were found in this thesis the selection of the sample size used were therefore ethically adequate.

Although the above limitations could have possibly effected the data contained within this thesis every precaution has been taken to try to minimise the possibility of this occurring. Due to the likelihood of these occurring being low it should be concluded that the data contained herein this thesis is as accurate as it can possibly be. Therefore, the conclusions drawn from the data should be interpreted as accurate and reliable allowing these findings to help football players make better informed decisions when choosing to use either an ankle brace or ankle protectors.

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10. Appendices

10.1 Peer reviewed publications

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The test-retest reliability of different ankle joint center location techniques

by Graydon, R[1], Fewtrell, D[1], Atkins, S[1] and Sinclair, J [1]

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Accurate and reliable joint identification is imperative for the collection of meaningful kinetic and kinematic data. Of the lower kinetic chain both the hip and knee joints have received a considerable amount of attention in 3D modelling. However, the reliability of methods to define the ankle joint center have received very little attention. This study investigated the reliability of the two marker method (TMM) and the functional ankle method (FAM) on estimating the ankle joint center. Furthermore, the effects of the two-marker method reliability for defining the ankle joint center when the ankle was covered with a brace or protector was investigated. 3D kinematic data was collected from ten participants (8 female and 2 male) whilst walking. The ankle joint center was defined twice using each test condition; TMM (WITHOUT), FAM (FUNCTIONAL), TMM when the ankle was covered with a brace (BRACE), and TMM when the ankle was covered with a protector (PROTECTOR). Intraclass correlations (ICC) were utilised to compare test and retest waveforms and paired samples t-tests were used to compare angular parameters. Significant differences were found in the test-retest angular parameters in the transverse and sagittal planes for the WITHOUT, BRACE, and FUNCTIONAL conditions. The strongest test-retest ICC's were observed in the WITHOUT and PROTECTOR conditions. The findings of the current investigation indicate that there are fewer errors using the TMM when the ankle is uncovered or when covered with soft foam that is easy to palpate through.

Keywords: Biomechanics, motion analysis, gait analysis, ankle, reliability

Accurate joint center identification is imperative for the collection of reliable and accurate lower limb kinetic and kinematic data for gait analysis [1,2]. Advances in three-dimensional (3D) motion capture systems are allowing for more detailed analysis of human movement than ever before but the efficacy and clinical interpretation of the data is only as good as the application of markers used to track the movement [3]. Unfortunately there can be inherent flaws in the application of surface markers when attempting to quantify movement of the underlying bones.

Address correspondence to: Robert Graydon, University of Central Lancashire, Preston, Lancashire
E-mail: RGraydon@uclan.ac.uk
[1] Division of Sport Exercise and Nutritional Sciences, School of Sport Tourism and Outdoors, University of Central Lancashire

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Skin artefacts [4] and experience level of the investigator applying the marker [5] both contribute to errors in data collection accuracy.

One of the key sources of measurement ambiguity in 3D kinematic analyses using surface marker placement is the definition of the joint center about which segmental rotations are considered to occur. Methods to accurately identify the hip joint center have been extensively researched [6,7,8] and to a lesser extent methods of accurately identifying the knee joint center have been researched [9,10,11]. However, ankle joint center identification techniques have received very little attention. The three main methods of identifying the ankle joint center are the two-marker-model (TMM), plug-in-gait model (PGM), and functional ankle model (FAM). Each method

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The effects of ankle protectors on lower limb kinematics in male football players: a comparison to braced and unbraced ankles

R. Graydon[1*], D. Fewtrell[1], S. Atkins[2] and J. Sinclair[1]

[1]Centre for Applied Sport and Exercise Sciences, School of Sport and Wellbeing, University of Central Lancashire, Fylde road, Preston, PR1 2HE Lancashire, United Kingdom; [2]School of Health Sciences, University of Salford, M5 4WT Manchester, United Kingdom; rwgraydon@uclan.ac.uk

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RESEARCH ARTICLE

Abstract

Football (soccer) players have a high risk of injuring the lower extremities. To reduce the risk of ankle inversion injuries ankle braces can be worn. To reduce the risk of ankle contusion injuries ankle protectors can be utilised. However, athletes can only wear one of these devices at a time. The effects of ankle braces on stance limb kinematics has been extensively researched, however ankle protectors have had little attention. Therefore, the current study aimed to investigate the effects of ankle protectors on lower extremity kinematics during the stance phase of jogging and compare them with braced and uncovered ankles. Twelve male participants ran at 3.4 m/s in three test conditions; ankle braces (BRACE), ankle protectors (PROTECTOR) and with uncovered ankles (WITHOUT). Stance phase kinematics were collected using an eight-camera motion capture system. Kinematic data between conditions were analysed using one-way repeated measures ANOVA. The results showed that BRACE (absolute range of motion (ROM) = 10.72° and relative ROM = 10.26°) significantly (P<0.05) restricted the ankle in the coronal plane when compared to PROTECTOR (absolute ROM=13.44° and relative ROM =12.82°) and WITHOUT (absolute ROM=13.64° and relative ROM=13.10°). It was also found that both BRACE (peak dorsiflexion=17.02° and absolute ROM=38.34°) and PROTECTOR (peak dorsiflexion =18.46° and absolute ROM =40.15°) significantly (P<0.05) reduced sagittal plane motion when compared to WITHOUT (peak dorsiflexion =19.20° and absolute ROM =42.66°). Ankle protectors' effects on lower limb kinematics closely resemble that of an unbraced ankle. Therefore, ankle protectors should only be used as a means to reduce risk of ankle contusion injuries and not implemented as a method to reduce the risk of ankle inversion injuries. Furthermore, the reductions found in sagittal plane motion of the ankle could possibly increase the bodies energy demand needed for locomotion when ankle protectors are utilised.

Keywords: biomechanics, motion analysis, ankle braces, ankle protectors, football, soccer

1. Introduction

Football (soccer) is an immensely popular sport with an estimated 265 million participants worldwide (FIFA Communications Division, 2007). Unfortunately, as with any sport, there is an inherent risk of injury to participants and football is no exception. Figures for injury incidences vary among studies due to differing methodologies, time frames observed, ability of participants and competitions observed, but conclude there are approximately 25 to 43.53 injuries per 1000 h of competitive match play (Andersen *et al.*, 2004; Hägglund *et al.*, 2013; Hawkins and Fuller, 1999; Salces *et al.*, 2014). Losing an integral team member can lead to a reduced chance of winning competitive matches and further more lead to loss of major trophies (Hägglund *et al.*, 2013). Therefore, an understanding of the common types of injury sustained by players and also methods to reduce the occurrence of injury is a high priority for football clubs.

Footballing injuries mainly occur to the lower extremities (Ekstrand *et al.*, 2011) with the ankle being one of the most commonly injured sites amongst players (Junge and Dvorak, 2013). Ankle inversion injuries and contusion injuries account for a large proportion of the total amount of ankle injuries (Waldén *et al.*, 2013). Once a player has suffered an ankle inversion injury they have an increased

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10.2 Ethical approval

10.2.1 Running study

uclan
University of Central Lancashire

20th May 2015

David John Fewtrell/Robert Graydon
School of Sport, Tourism and the Outdoors
University of Central Lancashire

Dear David/Robert,

Re: STEMH Ethics Committee Application
Unique Reference Number: STEMH 309

The STEMH ethics committee has granted approval of your proposal application 'The effects of ankle protectors and braces on the kinetics and kinematics of association football players'. Approval is granted up to the end of project date* or for 5 years from the date of this letter, whichever is the longer.

It is your responsibility to ensure that

- the project is carried out in line with the information provided in the forms you have submitted
- you regularly re-consider the ethical issues that may be raised in generating and analysing your data
- any proposed amendments/changes to the project are raised with, and approved, by Committee
- you notify roffice@uclan.ac.uk if the end date changes or the project does not start
- serious adverse events that occur from the project are reported to Committee
- a closure report is submitted to complete the ethics governance procedures (Existing paperwork can be used for this purposes e.g. funder's end of grant report; abstract for student award or NRES final report. If none of these are available use e-Ethics Closure Report Proforma).

Yours sincerely,

Ambreen Chohan
Deputy Vice Chair
STEMH Ethics Committee

* for research degree students this will be the final lapse date

NB - Ethical approval is contingent on any health and safety checklists having been completed, and necessary approvals as a result of gained.

10.2.2 countermovement vertical jump study

uclan
University of Central Lancashire

8th August 2013

Dave Fewtrell and Robert Graydon
School of Sports Tourism & the Outdoors
University of Central Lancashire

Dear Dave & Robert

Re: BuSH Ethics Committee Application
Unique Reference Number: BuSH 183

The BuSH ethics committee has granted approval of your proposal application 'The ankle protector's effects on performance optimisation in association football'.

Please note that approval is granted up to the end of project date or for 5 years, whichever is the longer. This is on the assumption that the project does not significantly change, in which case, you should check whether further ethical clearance is required

We shall e-mail you a copy of the end-of-project report form to complete within a month of the anticipated date of project completion you specified on your application form. This should be completed, within 3 months, to complete the ethics governance procedures or, alternatively, an amended end-of-project date forwarded to roffice@uclan.ac.uk quoting your unique reference number.

Yours sincerely

Gill Thomson
Vice Chair
BuSH Ethics Committee

NB - Ethical approval is contingent on any health and safety checklists having been completed, and necessary approvals as a result of gained.

10.2.3 45° cutting manoeuvre study

20th May 2015

David John Fewtrell/Robert Graydon
School of Sport, Tourism and the Outdoors
University of Central Lancashire

Dear David/Robert,

Re: STEMH Ethics Committee Application
Unique Reference Number: STEMH 309

The STEMH ethics committee has granted approval of your proposal application 'The effects of ankle protectors and braces on the kinetics and kinematics of association football players'. Approval is granted up to the end of project date* or for 5 years from the date of this letter, whichever is the longer.

It is your responsibility to ensure that

- the project is carried out in line with the information provided in the forms you have submitted
- you regularly re-consider the ethical issues that may be raised in generating and analysing your data
- any proposed amendments/changes to the project are raised with, and approved, by Committee
- you notify roffice@uclan.ac.uk if the end date changes or the project does not start
- serious adverse events that occur from the project are reported to Committee
- a closure report is submitted to complete the ethics governance procedures (Existing paperwork can be used for this purposes e.g. funder's end of grant report; abstract for student award or NRES final report. If none of these are available use e-Ethics Closure Report Proforma).

Yours sincerely,

A Chohan

Ambreen Chohan
Deputy Vice Chair
STEMH Ethics Committee

* for research degree students this will be the final lapse date

NB - Ethical approval is contingent on any health and safety checklists having been completed, and necessary approvals as a result of gained.

10.2.4 Stance limb during kicking study

20 November 2015

Dave Fewtrell / Robert Graydon
School of Sport and Wellbeing
University of Central Lancashire

Dear Dave / Robert

Re: STEMH Ethics Committee Application
Unique Reference Number: STEMH 391

The STEMH ethics committee has granted approval of your proposal application '**The effects of ankle protectors and braces on the kinetics and kinematics of association football players**'. Approval is granted up to the end of project date* or for 5 years from the date of this letter, whichever is the longer.

It is your responsibility to ensure that

- the project is carried out in line with the information provided in the forms you have submitted
- you regularly re-consider the ethical issues that may be raised in generating and analysing your data
- any proposed amendments/changes to the project are raised with, and approved, by Committee
- you notify roffice@uclan.ac.uk if the end date changes or the project does not start
- serious adverse events that occur from the project are reported to Committee
- a closure report is submitted to complete the ethics governance procedures (Existing paperwork can be used for this purposes e.g. funder's end of grant report; abstract for student award or NRES final report. If none of these are available use e-Ethics Closure Report Proforma).

Yours sincerely

Arati Iyengar

Arati Iyengar
Vice Chair
STEMH Ethics Committee

* for research degree students this will be the final lapse date

NB - Ethical approval is contingent on any health and safety checklists having been completed, and necessary approvals as a result of gained.

10.3 Example participant information sheet

University of Central Lancashire

The ankle protector's effects on performance optimisation in association football
Participant information sheet

Introduction

You are being invited to take part in a research study at the University of Central Lancashire. Before you decide whether or not to take part it is important for you to understand why the research is being done and what it will involve. Please take time to read the following information carefully.

The purpose of this study is to assess how ankle protectors and ankle braces effect the performance of sporting tasks and is being conducted as part of an MPhil/PhD project. Previous research has established the protective capabilities of ankle protectors and braces but to date there has been no research investigating the effects of ankle protectors on movement.

You have been invited to participate in this investigation because you regularly partake in sporting activities which involve regular intervals of running. It is up to you to decide whether or not to take part.

Do I Have To Take Part?

Participation is entirely voluntary. Even if you decide to take part in the study, you can still leave/end the testing session at any point. Please note that your data can be withdrawn up until the point at which data analysis has been undertaken. For more information on data withdrawal please see the section titled "withdrawal".

Location and Duration

Data collection will take place at the University of Central Lancashire in the biomechanics laboratory (Darwin building room DB018). If you choose to partake in the study the data collection should take no longer than one hour per participant.

Procedure

To participate in this study you are required to be in good health and injury free. A date and time will be agreed with you to attend a data collection session at the biomechanics laboratory. On the day of testing, before any data is collected, you will be required to complete a PAR-Q form to confirm you are physically able to participate. Please see the PAR-Q form which accompanies this information sheet. If you have any questions regarding the PAR-Q form please contact the lead researcher on the details found in the last section of this information sheet.

You will be required to complete five trials of jogging at a comfortable pace over ten metres. There will be three test conditions; with ankle protectors, with ankle braces and without any ankle protector or braces. You will perform a total of 15 jogging trials and will be allowed sufficient rest periods between each test. The investigator will demonstrate what you are required to do before testing commences and you will be required to perform a series of pre-test warm up movements.

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To record the movement data retro-reflective markers will be placed on your lower body using non-invasive tape to define various anatomical positions.

If you decide to participate in the study it is advised you wear ankle socks, athletic shorts, t-shirt and trainers on the day of testing.

Risk

The risk of injury to participants is low as you will be performing movements common in most athletic sports. An assessment of the risks associated with this study has been completed and where possible any potential risks have been minimised. If you wish to see a copy of the risk assessment then please contact the researcher using the details provided at the bottom of this sheet. If you do sustain an injury during the study the testing will be immediately stopped and first aid will be administered. If the pain continues in the following days after the testing it is strongly recommended that you consult a physiotherapist or a doctor.

Benefits

After taking part in this study the investigator can provide feedback on your running gait. The investigator has served as a running shoe specialist and is able to offer recommendations on training shoe selection based on the information obtained during testing. Please ask at the end of the end of testing if you would like this information.

Data protection/Confidentiality

All data obtained during this study will be saved to a password protected laptop computer and will be used to produce means and standard deviations. All information linking participants to the study will remain confidential and will be securely stored within a locked filing cabinet. This data will be retained for 5 years from the end of the study and then destroyed (in line with data protection requirements). Once the data is obtained the participant will be allocated a number so that anonymity will remain throughout any produced report. The investigator may ask to take a photo during testing to show experimental set up in the final report. To stop any identification of participants from any photos used in the final document the face of the participant will be covered. If you are happy for your photo to be taken and used then please provide consent.

Withdrawal

If you choose to withdraw your information please contact the lead researcher on the contact details found in the last section of this information sheet. . You can choose to withdraw your data up until the point at which data analysis has been undertaken.

Ethical Approval

This study is in the process of seeking ethical approval from Science, Technology, Engineering, Medicine and Health (STEMH) ethics committee at the University of Central Lancashire.

University of Central Lancashire

Issues and/or Complaints

If you have any issues or complaints with regards to the study or investigator then please contact the Director of Studies Dr. David Fewtrell. If your complaint is not satisfactorily dealt with by the Director of Studies you may also contact the University Officer for Ethics. The contact details for both are provided in the last section of this information sheet.

Participation

If you have any further questions then please contact the lead research using the details provided. If you are happy to participate in this study after reading the above information then please contact the lead researcher to arrange a date and time for data collection to take place.

Thank you for taking the time to read this information sheet.

Contact Details

Lead Researcher: Robert Graydon
E-mail. RGraydon@uclan.ac.uk

Director of Studies: Dr. David Fewtrell
E-mail. DJFewtrell@uclan.ac.uk

Officer for Ethics
E-Mail. OfficerforEthics@uclan.ac.uk

10.4 Example Consent form

uclan
University of Central Lancashire

CONSENT FORM

Project Title:
The effects of ankle protectors and braces on the kinetics and kinematics of association football players.

Researcher:
Robert Graydon
MPhil/PhD Research Student
RGraydon@uclan.ac.uk

	Please initial box
I confirm that I have read and understand the information sheet, dated ………….. for the above study and have had the opportunity to consider the information, ask questions and have had these answered satisfactorily.	☐
I understand that my participation is voluntary and that I am free to withdraw from the physical testing at any time, without giving reason.	☐
I agree to take part in the above study.	☐
I agree that my data gathered in this study may be stored (after it has been anonymised). Please Note: All information linking participants to the study will remain confidential and will be securely stored within a locked filing cabinet. This data will be retained for 5 years from the end of the study and then destroyed (in line with data protection requirements).	☐
I understand that I have 24 hours after completion of the data collection session to remove my data from the study if I so desire. After this period I understand that it will not be possible to withdraw my data from the study.	☐

	Please initial Yes	Please initial No
I agree to the use of anonymised images/photos in publications	☐	☐

Name of Participant	Date	Signature
Name of Researcher	Date	Signature

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