THE EFFECT OF ANKLE BRACING ON JOINT DYNAMICS IN THE LOWER LIMB DURING JUMPING TASKS IN ELITE FEMALE ATHLETES

A Thesis Submitted to the College of

Graduate Studies and Research

in Partial Fulfillment of the Requirements

for the Degree of Master of Science

in the College of Kinesiology

University of Saskatchewan

Saskatoon

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August 2011

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Abstract

Wearing ankle braces reduces the incidence of new and recurring ankle injuries in sports. Several studies have examined the effect of bracing on the mechanics of the ankle but little research has been done examining biomechanical changes at the knee and hip when ankle braces are used. The purpose of this study was to determine if the application of an ankle brace had an effect on the kinematics and kinetics of the ankle, knee and hip joints during two simulated athletic jumping manoeuvres. Eight members of the University of Saskatchewan Women's Huskie Basketball team were recruited for this study. Each subject performed a series of single leg jump landing/takeoff manoeuvres in forward and sideways directions while their movements and ground reaction forces (GRF) were recorded. The participants performed the movements both with and without wearing a lace-up style ankle brace. Dependent variables for this study included ground reaction forces (GRF) and ankle, knee and hip joint angles and joint moments as well as ankle and knee joint stiffness. Comparisons were made between the braced and non-braced conditions using paired t-tests. Using a conservative statistical approach, significant changes were only observed for ankle joint kinematics, with the braced condition exhibiting significant decreased overall sagittal range of motion, and a significant increase in ankle external rotation. A strong trend for increased ankle inversion was also observed during both the forward and sideways manoeuvres. There were no significant differences for GRFs, in ankle knee or hip joint moments or knee and hip kinematics at the p<0.001 level for any time point during contact. During the braced condition the GRFs displayed a strong trend for increasing in magnitude as well as decreasing in time to peak magnitude, with the largest differences observed in the breaking and vertical GRFs at or near the time of impact. Trends were observed in ankle moments with an increase in the eversion moment, plantar flexor moment and external rotation

moment at impact. Smaller kinematic changes were observed at the knee joint with trends indicating an increase in knee flexion at impact and a decrease in knee abduction angle. The hip did not display any difference with regards to kinematic changes however there was a trend for increased hip flexion moments at impact. There were no major differences observed for GRFs, ankle, knee or hip kinematics or kinetics during the propulsive phase of each movement. These results indicate that the largest ankle brace effect is primarily constrained to the time period surrounding impact with the ground and the largest change in joint mechanics occurs at the ankle.

Acknowledgements

I would like to acknowledge my supervisor Dr. Joel Lanovaz for his guidance, support and dedication he has provided to me throughout the completion of my Master's thesis. Dr. Lanovaz has supplied me with both the tools and information I required to complete my studies in the field of biomechanics. It has been a pleasure working with you throughout the past three years in all endeavours undertaken. Thank you for your time and dedication. I would also like to thank my thesis committee members Dr. Phil Chilibeck, Dr. Alison Oates and Mr. Bart Arnold. Thank you for all the input and guidance you have given me throughout my academic endeavours.

Dedication

I dedicate this Master's thesis to my parents, friends and family members. You have all had a tremendous role in supporting me through the entire process of completing this thesis.

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Chapter 1

Scientific Framework

1.1 Introduction

Injuries to the lateral ligaments of the ankle joint complex are among the most frequent injuries to occur during sports (Thonnard, Bragard, Willems, & Plaghki, 1996; Wright, Neptune, van den Bogert, & Nigg, 2000; Eils et al., 2002). Up to 86% of ankle injuries are sprains which can account for 10- 28% of all reported sports injuries (Dizon & Reyes, 2010). As many as 73% of recreational and competitive athletes experience reoccurring ankle sprains, accounting for the single largest absence from activity of any sports injury, however, those participants competing in basketball, soccer and handball are the most susceptible (Dizon et al., 2010). Although ankle injuries may be complex and multifaceted, the primary site for injury has typically been the lateral ankle, and more specifically the anterior talo-fibular ligament (Wright et al., 2000). With the high incidence of ankle joint injuries, many have studied the ability of bracing and taping to prevent and protect the ankle joint ligaments from sprain mechanisms (Mickel, et al, 2006; Verhagen, van der Beek, & van Mechelen, 2001).

In addition to ankle injuries, female athletes competing in sports are at a 4-6 times greater risk to sustain a knee injury than their male counterparts (Pollard, Sigward, & Powers, 2010). Specifically, females are particularly susceptible to non-contact anterior cruciate ligament (ACL) injuries during activities involving quick decelerations, changes in directions and jumping which also indicates high injury rates among basketball, soccer and handball players (Decker, Torry, Wyland, Sterett, and Stedman, 2003; Hewett et al., 2005). Numerous studies have examined gender differences in lower extremity mechanics during athletic movement to better understand why females are considered at risk for ACL injuries. In addition to the relatively non-modifiable differences in hormone levels and body morphology, females exhibit altered landing mechanics displaying decreased knee flexion, increased quadriceps activation, decreased hamstrings activation, and increased frontal plane knee muscle activation (Hewett, et al., 2005; McLean, Huang, & van den Bogert, 2005; Stoffel et al., 2010; Yu, &Garrett, 2007). Knee injuries can also be related to external environmental factors, which may include playing surface or shoe type. Due to their ability to modify ankle mechanics, ankle stabilizers have received attention as one external factor which may have the potential to modify knee and hip mechanisms (Cloak, Galloway, & Wyon, 2010; DiStefano, Padua, Brown, & Guskiewicz, 2008; Santos, Mclntire, Foecking, & Liu, 2004).

The lower limb joints work together as an interconnected system to absorb loads and optimize movement performance. Wearing an ankle brace, which may reduce ankle range of motion and increase the stiffness around the joint, may have the potential to alter the proximal joints of the leg during movements (McCaw, & Cerullo, 1999; Santos et al., 2004). For a population that seems to be susceptible to knee injuries, females may be increasingly prone to greater changes in knee and hip mechanical changes than their male counterparts (Fagenbaum, & Darling, 2003). It is relatively unknown what additional effects ankle stabilizers have on the proximal joints of the lower limb over the course of contact with the floor while performing an athletic maneuver, specifically during the timing of impact and in preparation for take-off. Analyzing females completing these movements would give a better overall understanding of how ankle stabilization affects ankle, knee and hip mechanics around each joint.

1.2 Review of the Literature

1.2.1Ankle Stabilization

The lateral ligament complex of the ankle is the most commonly injured anatomical structure in the body, with over one million reported injuries, per year, in the United States alone (Thonnard et al., 1996; Wright et al., 2000; Eils et al., 2002). For athletes, ankle injuries can account for 10 – 30 % of all reported sports injuries which equates to approximately 10-20 % of the time lost from athletic participation (Dizon et al., 2010; Shapiro, et al. 1994). The ligaments which are most commonly damaged or torn include the anterior talofibular ligaments (ATFL) as well as the calcaneofibular ligaments (CL) (Mickel et al., 2006). These ligaments lie on the lateral side of the ankle joint. Lateral ligaments are often injured due to an inversion motion occurring at the ankle joint. The primary cause of inversion motion occurs by an individual landing or rolling onto the outside of his or her foot (Shapiro, et al. 1994). Initial inversions are often amplified by the continued momentum of the player's body, forcing the medial malleolus to move closer to the calcaneous and causing excessive strain on the both the ATFL and CL ligaments (Mickel et al., 2006).

One of the functions of the ankle musculature is to prevent injuries to the ankle ligament structure. Muscles surrounding the ankle joint provide persistent stabilization to the joint during motion, also known as dynamic stability; muscles can contract to a greater degree if an excessive strain is placed on the ankle joint (Midgley et al., 2007). This dynamic protection occurs in response to neural input, and can stiffen up the ankle joint to various degrees to offer specific levels of protection. However, dynamic stabilization takes time and the latency period between the perturbation stimuli and the firing response of an ankle muscle can only occur after 50 to 68 milliseconds (Sefton, Hicks-Little, Koceja, & Cordova, 2007). This firing delay may be too slow for ankle musculature to adequately protect the ankle joint from a sudden inversion perturbation (Sefton et al., 2007). In addition to the neural delay, an individual competing in athletics or prolonged activity will undoubtedly fatigue ankle musculature. Fatigue may further delay this muscle firing response. Known as electromechanical delay, a slower firing muscle offers less protection for the joint ligaments which, depending on the inversion stimuli, may not adequately protect the ankle ligaments from injury (Mickel et al., 2006; Midgley et al., 2007). The application of an ankle brace provides additional mechanical stability to the ankle joint, which may reduce the need for an increased reflex response from ankle musculature by reducing the rate at which the ankle ligaments are loaded (Sefton, et al., 2007).

1.2.2 Ankle Braces

Ankle stabilizers are used to prevent the occurrence and reoccurrence of ankle ligament damage. Traditionally taping has been considered the gold standard for the prevention of ankle sprains (Dizon et al., 2010). However, the ankle brace has been developed as an alternative to ankle taping. Along with ankle taping procedures, the aim of an ankle brace is to restrict the ankle joint range of motion of the wearer, specifically to restrict excessive inversion at the ankle (Cordova, Takahashi, Kress, Brucker, & Finch, 2010; Popadopoulos et al., 2005). Ankle braces can come in a variety of different support levels and styles. For athletic populations the semi-rigid and lace-up styles are the most commonly prescribed and studied. The difference between the two brace styles is the material composition of the brace. A semi-rigid brace is composed of rigid plastic that surrounds the medial and lateral sides of the foot, ankle and lower shin. The lace-up style ankle braces are composed of nylon and other stretch resistant materials that provide constant support to the joint by surrounding the foot and ankle. Due to the full wrap of the foot and ankle joint the lace-up types more closely mimic ankle taping methods than the semi-rigid braces. Recent studies have shown that lace-up ankle braces possess a number of important advantages over ankle taping including cost effectiveness, reusability, decreased loosening effects over the same wear period and increased comfort and ease of application (Gudibanda & Wang, 2005; Meana, Alegre, Elvira, & Aguado, 2008; Mickel et al., 2006; Siegler, Liu, Sennett, Nobilini, & Dunbar, 1997; Verhagen et al., 2001). For these reasons, lace-up ankle braces have been commonly prescribed and purchased to reduce both the incidence and reoccurrence of lateral ligament sprains.

1.2.2.1 Ankle Brace Effects

Effectiveness of the ankle brace has often been assessed by comparing the absolute reduction in inversion range of motion (ROM) in a passive setting. Passive ROM testing examines the ankle ROM rotation limit by applying a standardized torque to the ankle joint. To determine the level of torque application, the joint is rotated to the limit of comfort without a brace applied. Subsequent measures match this level of torque with a range of ankle braces to determine the specific ankle brace ROM restriction (Eils et al., 2002). Using this technique, research has shown that ankle braces have the potential to restrict inversion range of motion by 18% to 53% (Alves, Alday, Ketcha, & Lentell, 2002; Eils et al., 2002). This percentage can correspond to a range of approximately 14.9° to 20° restriction to inversion motion (Cordova, Ingersoll & Palmieri, 2002).

Trap door measures are the second commonly used procedure to measure isolated inversion ROM restriction. The procedure simply involves having subjects stand upright on a platform which drops away suddenly, mimicking a situation that might lead to an inversion ankle sprain. The differences in ROM between braced and control conditions are much smaller than under passive ROM testing due to the additional body weight effect. Examining inversion restriction with a trap door with the same braces used in the passive tests, Elis et al. (2002) found braces to reduce inversion between 5.8° and 10.14 ° (a 15% to 26% reduction) compared to a no brace condition. These results are similar to Zhang, Wortley, Chen, & Freedman (2009) who found the Ankle Stabilizing Orthosis (ASO) brand ankle brace to reduce ankle inversion ROM by 7.4°.

Ankle inversion is not the only range of motion restricted by ankle bracing. In addition to their effectiveness in restricting the frontal plane of motion, ankle braces have also demonstrated an ability to modify sagittal plane ankle joint kinematics (Cordova et al., 2010; DiStefano et al., 2008; Gudibanda et al., 2005; Simpson, Craven, Theodorou, & DelRay, 1999). Numerous studies have examined the potential for ankle braces to limit the passive range of both plantar and dorsi flexion. Using a passive ROM testing mechanism, Elis et al. (2002) determined that plantar flexion was restricted between 8.6° to 15° and dorsi flexion was restricted between 7° and 14°. Cordova et al., (2002) also demonstrated a reduction of 9.7° of plantar flexion when an ankle brace ROM was measured, a significant difference from the no brace condition. In regards to dorsi flexion, Paris, Vardaxis, & Kokkaliariaet (1995) found that the lace up brace condition provided 5.6° restriction in ROM when compared to the control condition. Reporting 1.85° plantar ROM reduction and 7.52° dorsi flexion ROM reduction, Seigler et al. (1997) also confirms sagittal plane restriction.

Previous research has shown ankle bracing's ability to limit and change ankle motion. While there are advantages of reproducibility in using passive and trap door methods, they are limited in their ability to evoke brace and joint responses observed under game movement conditions (Duysens, & Levin, 2010). These limitations are due to the ROM stimulus being quite different between static experiments and real dynamic moments. Without the ability to fully replicate actual sprain stimuli, researchers have begun to move away from the artificial measures and towards using real sport simulations to determine the effectiveness of ankle bracing (Duysens et al., 2010). Simulated athletic movements such as lateral cutting, or landing from a height have been two commonly implicated protocols that examine dynamically the effects of ankle bracing.

During dynamic movement, injury focus has been on the landing or impact phase. Along with restriction in the frontal plane, one suggested injury prevention mechanism provided by bracing may be the role of maintaining the ankle in a proper anatomic position prior to landing (Ubell, Boylan, Ashton-Miller, & Wojtys, 2003; Thonnard et al., 1996). Wright et al. (2000) suggests ankle bracing may influence the position of the unloaded foot prior to impact by decreasing the ROM in the sagittal plane, specifically a tendency to be plantar flexed.

As the mechanism for ankle sprains is described as a combination of both ankle inversion and plantar flexion, the ankle brace’s ability to limit plantar flexion prior to impact may protect the ankle from a potentially injury prone position (Eils et al., 2002; DiStefano et al., 2008). Overall ankle joint range of motion in the sagittal plane has been shown to decrease with braces during drop landings. Drop landings measure the effects of a purely vertical impact on leg mechanics. DiStefano et al. (2008) demonstrated with ankle brace application the ankle joint displayed between 2.8° and 3.4° reduction in plantar flexion at initial ground contact during a landing from a 0.30 meter height. These results correspond to the 8.9 degree overall sagittal plane ROM restriction reported by Cordova et al. (2010) during the interval between ground contact and maximal ground reaction force during a drop landing also from a 0.30 meter height. From a 0.60 meter height, McCaw et al. (1999) reported a reduction of between 5 and 6 degree sagittal plane ROM when participants impacted the ground under a brace condition. Using 85 %

of maximum velocity side shuffle as opposed to a drop jump, ankle bracing displayed a 3 to 4 degree reduction in plantar flexion (Simpson et al., 1999). Also testing a lateral maneuver Zhang et al. (2009) reported a reduction of 1.1 degree in plantar flexion at impact. These results indicate that the type of brace used as well as the movement protocol may alter that absolute level of sagittal plane restriction at the ankle. The consistency of reduced plantar flexion between studies indicate that during both drop landings and lateral movement protocols ankle braces have the potential to reduce sagittal plane ankle motion.

1.2.3 Ankle Stabilization Effects

The lower leg is a kinetic chain, and as such each joint is only one single part of that chain. During movement, the hip, knee and ankle all work together to provide support to the body and act as an impact absorption mechanism. Practitioners have long recognized the importance of unimpeded hip, knee and ankle flexion to absorb the impact of landing (McCaw et al., 1999). During landing, motions begin distally at the ankle and progress proximally through the knee and hip joints (DiStefano et al., 2008). The results originally reported by McCaw et al. (1999) of ankle restriction in the sagittal plane, has prompted the speculation that ankle braces may have a significant effect on ground reaction force attenuation, leg joint kinematic changes and leg kinetic changes. The sagittal plane offers the largest range of motion of the ankle's three axes, and is a primary mechanism in which ground reaction forces are attenuated and energy is absorbed at the ankle during landing (Cordova et al., 2010). Unlike in the frontal plane, the ankle joint has large musculature to control the flexion and extension at the ankle joint. Isolated movement in the sagittal plane is not a primary cause for the typical ankle sprain injury. Therefore the application of an ankle brace may not overtly benefit the ankle joint stability in the sagittal plane, and may have a detrimental effect on the joint due to range of motion restricting normal ankle motion (Simpson et al., 1999). Not allowing for normal ankle dorsi flexion and plantar flexion range of motion may therefore alter the dynamics of the ankle as well as the proximal joints of the leg (McCaw et al., 1999).

1.2.3.1 Effects on Ground Reaction Forces

Ankle sagittal plane motion is a primary mechanism through which the ankle joint contributes to the performance of movement. For this reason a decreased ROM in the sagittal plane may be an undesirable feature for the attenuation of ground reaction forces (Siegler et al., 1997). Research examining impact forces during landings have determined that during landings an athlete’s body can experience ground reaction forces in excess of 6000 Newtons, with mean vertical ground reaction forces ranging between 3.33 and 5.39 times body weight during a landing between 30 and 90 cm respectively (Wallace et al., 2010). These values are similar to landing from heights between 32 -128 centimeters, which can cause ground reaction forces between 3.0 and 11.0 times bodyweight (Zhang, Bates, & Dufeek, 2000). During a landing from a maximal vertical jump for example, vertical ground reaction forces have been found to range from 2.58 to 9.92 times body weight (Ortega, Bíes, & de la Rosa, 2010). For these reasons unimpeded hip, knee, and ankle flexion during landings are critical to ensuring joint safety during impact absorption (McCaw et al., 1999).

Several studies have demonstrated the importance of ankle dorsi flexion for energy absorption during landing, with differing contributions of energy absorption occurring during stiff and soft landings (McCaw et al., 1999). In soft landings with more joint flexion during impact, the hip, knee, and ankle joints contributed 25, 37, and 37%, respectively, to the total energy absorbed by the lower extremity. During stiff landings with minimal joint flexion, the relative contributions of the hip, knee, and ankle joints were altered to 20, 31, and 50%, respectively, of the total energy absorbed (DeVita, & Skelly, 1992). This comparison outlines that leg joints are able to disperse and transfer the impact of landing quite equally when each joint is allowed to move through an unimpeded range of motion. These results are further supported by McCaw et al. (1999)showing that ankle taping and bracing may adversely influence impact absorption during landings by limiting sagittal range of motion by 5°, which may increase energy dissipation demands at the knee and hip. The results of McCaw et al. (1999) suggest that ankle stabilizing techniques may impinge on the normal ankle kinematics, which will act as a precursor to changing knee and hip dynamics.

If ankle plantar flexors play a large role in the absorption of landing forces, a smaller range of sagittal ROM during landing may result in greater peak landing forces. Studies examining the effect of ankle bracing on the generation of ground reaction forces have generally supported the theory that a reduction in sagittal ankle ROM would lead to an increase in vertical ground reaction force (Cordova et al., 2010; DiStefano et al., 2008; Hodgson ,Tis, Cobb, & Higbie, 2005; Riemann, Schmitz, Gale, & McCaw, 2002 and Sacco et al., 2006). All theorized ankle bracing would display an increase in vertical GRF at impact over a control condition. However, only the results of Hodgson et al. (2005) confirmed their theory of bracing increasing vertical GRF. Their results demonstrated a significant 12 % increase in magnitude over the first 10 milliseconds of impact, with a trend for increased magnitude (5% increase) observed during peak vertical GRF. Of the five other studies either no significant difference was determined between bracing and control conditions or the control condition displayed larger GRF magnitude. One possible explanation for this lack of change may be attributed to the way the ankle brace conforms to the foot and ankle complex, mediating the initial peak vertical force along the mediolateral and or the anterioposterior axes instead of the vertical axis (Cordova, et al., 2010).

More consistent within the ankle bracing literature has been the change in the timing of the GRF profiles, specifically the rate at which initial impact force and peak force were generated during landing. The timing of the first peak (forefoot impact) and second peak (maximum vertical force, heel contact) forces were reduced significantly under ankle brace conditions by an average of 185 milliseconds and 425 milliseconds respectively (Riemann et al., 2002). The results of Cordova et al. (2010) demonstrated a smaller mean decrease during the brace condition to the first (3 milliseconds) and second peak (4 milliseconds) but concluded these differences were a significant reduction compared to the control condition. Although not significant, the results of Hodgson et al. (2005) indicated a trend for the ankle brace to decrease the time to peak by an average of 3milliseconds and 17 milliseconds for peak one and two respectively. DiStefano et al. (2008) did not observe significant differences between braced and control conditions for the rate of force development between braced conditions. It has been speculated that the reduction in time to reach the initial peak reaction force may be related to the brace restricting forefoot and midfoot mobility, transforming the foot into a rigid segment and diminishing the allowable movement at initial impact (Cordova et al., 2010; Riemann et al., 2002). The reduction in time to reach the second peak impact force is thought to occur because the decrease in plantar flexion coupled with a decrease in dorsi flexion at impact would essentially create a flat footed landing strategy (Riemann et al., 2002). Overall, the decreased interval observed before peak force indicates that proximal musculoskeletal structures of the body may be subjected to loads within a shorter time interval (Sacco et al., 2006). This has implications in terms of how stress is applied to the lower extremity kinetic chain, as well as how stress is ultimately dissipated proximally through the knee and hip joints (Cordova et al., 2010).

1.2.3.2 Effects on Proximal Joint Kinematics

In addition to influencing the ground reaction force profiles during landings, ankle bracing has also been examined as to the extent in which alterations occur in proximal joint kinematics. It has been speculated that because of an ankle brace's ability to alter the normal ankle biomechanics during movement, ankle braces may have the ability to change knee and hip mechanics (Cordova, et al., 2010). Since braces can reduce the amount of plantar flexion angle at touchdown and therefore decrease available ROM, bracing may result in the ankle musculature absorbing less force. There is potential for the knee and hip joints to increase in flexion at impact to compensate for the reduced ankle motion and reduced force absorption, thereby potentially keeping the magnitude of rate of vertical GRF relatively constant (DiStefano, et al., 2008). With each joint of the lower extremity kinetic chain having its own role during landing, adapting to the inability of the ankle to rotate or absorb force may place the proximal joints at risk by trying to compensate (Venesky, Docherty, Dapena and Schrader, 2006). This change in knee and hip motion associated with the use of an ankle brace may depend on the specific adaptation strategy used by the subjects or the movement in which the participants are subjected (Santos et al., 2004).

Using a two footed drop landing protocol (i.e. landing on two feet versus one foot), DiStefano et al. (2008) found no change in vertical ground reaction force at impact between brace conditions but noticed a significant increase in knee flexion at the same time interval while the participants were wearing the ankle brace (12°) compared with the control (9°). They suggest the greater knee flexion angle when wearing the brace offset the restriction in ankle

ROM and allowed the vertical ground reaction force to remain constant. However, overall knee range of motion was smaller under the brace condition (79°) than the control (82°). Cordova et al. (2010) hypothesized that knee flexion would increase due to the application of an ankle brace to compensate for the kinematic changes at the ankle during single legged landing. Their results contradict this hypothesis and indicate that the braced condition displayed the least amount of knee flexion (42.6°) compared to the control condition (45.1°). This lack of change at the knee may have resulted in the decrease time to peak vertical force, confining the leg to a more upright position at impact during the braced condition. Landing with less flexion may place greater reliance on the knee joint articular surfaces (menisci, and articular cartilage) to absorb the compressive loads at impact and possibly lead to more stress at the knee joint (Cordova, et al., 2010). Decreased knee flexion at impact has also been observed to increase strain on the ligaments of the knee, specifically the anterior cruciate ligament due to increased quadriceps pull at low joint angles (i.e. closer to 0° flexion) (Fagenbaum et al., 2003).

Bracing can also significantly reduce inversion and eversion as well as internal and external rotation at the ankle. This inability to rotate the ankle may cause excessive knee joint varus/valgus and internal/external rotation motion during movement (Venesky, et al., 2006). Santos et al. (2004) used a study design in which subjects performed trunk rotation tasks while standing on one leg: turning sideways to catch a ball (open task) and turning sideways to touch a target with their shoulder (closed task). The results of these studies showed that the effect of ankle bracing on the axis of rotation on the knee depended on the context of the tasks performed. Under situations where the subject was required to make a forceful trunk rotation while on a single leg, the results showed the ankle braces causing an increase in knee axial rotation indicating a higher risk of knee injury (Santos, et al., 2004).

Less data is available regarding the change in hip kinematics due to ankle stabilization. Only Cordova et al. (2010) measured the change in hip kinematics during their investigation and concluded that no differences were observable between braced and non-braced conditions. It has been suggested that the hip joint ROM observed remained unchanged under the ankle brace condition because of the relatively low contribution of the hip joint to the total lower extremity force absorption during a drop landing (Cordova, et al., 2010). This is in accordance to previous work by DeVita et al. (1992) who have shown the hips' contribution to the energy absorbed during landing to be significantly less than at the knee or ankle during drop landings. Alterations in hip kinematics due to bracing effects may therefore be smaller than changes observed at the knee.

In recent years, investigators have also explored the relationship of ankle instability and lower limb joint kinematics. While ankle joint instability, referring to an excessive range of motion and decreased stabilization at the ankle joint, is on the opposite spectrum than ankle joint stabilization, it is another example of how the lower limb is a series of interconnected joints that work together (Gribble, & Robinson, 2009). Investigators have shown that during landings, knee and hip kinematic parameters are altered in the presence of ankle instability. Disruption to ankle joint stability during landing can often alter the degree of knee flexion/ extension at impact with greater knee extension (stiffer landing) at impact possibly allowing for a longer period of time to dissipate and control ground reaction forces in the presence of ankle instability (Gribble et.al, 2009). This response is opposite to the theorized response of knee flexion increasing during ankle brace application. Research applied to ankle stability and instability reference the extent in which leg joints to work together. It is apparent that the lower limb acts as a very structured and interconnected series of joints, where a change in one joint range of motion and stability during movement can potentially affect other joints in series.

1.2.3.3 Effects on Proximal Joint Kinetics

To further quantify the effect of ankle bracing on knee and hip joint mechanics, joint moments, calculated through inverse dynamics, can be used to assess changes in muscle action. If there is a change in the amount of force transferred to the knee and hip with the application of an ankle brace, it is likely that the musculature surrounding the joint will have to act differently in order to compensate. Literature examining how joint moments are altered due to ankle brace application are limited, however, recent studies have shown that knee kinematics can be altered with the application of ankle stabilizers (Stoffel et al., 2010; Venesky, et al., 2006)

By measuring knee moment variables with a drop landing on a slant board occurring under braced and control conditions, Venesky, et al. (2006) showed an increase in external knee rotation joint moment (i.e. moments in the transverse plane) at impact under the braced condition. The braced condition displayed a 7.55% increase over the control condition. The results did not display any significant frontal plane valgus (abduction) knee moment difference between brace and control conditions. Stofffel et al. (2010) examining the effects of ankle bracing on the knee joint moments during running forward and cutting at a 45° angle. Peak internal rotation moments in the transverse plane were significantly less under the braced condition for both forward and sideways maneuvers, a reduction of 18% compared to control trials. Knee varus (adduction) moments were also examined, with the braced condition displaying between a 4% and 18% reduction over the control condition for running and sideways cutting respectively. Stoffel et al. (2010) did report that both the knee internal rotation and valgus moments were significantly larger for side step cutting versus running for both conditions.

These results may have implications for possible knee injury, with the authors concluding that side stepping motions are more likely to alter knee joint moments. Both of these studies partially refuted their initial hypotheses that ankle bracing would increase knee joint loading. The results are in partial support of ankle stabilization altering knee joint kikinetics. Therefore it becomes logical to investigate factors that are associated with the specific knee injuries to determine what extent knee moment alteration may be explicit to specific injury mechanisms.

1.2.4 Knee Injury Mechanisms and Females

The knee joint, while typically less prone to injury than the ankle is also a common site for injury during sports involving running, landing, decelerating, and rapid lateral changes in direction (Hughes, Watkins, & Owen, 2008). Anterior cruciate ligament (ACL) injuries are arguably the most common and most serious knee injury with approximately 70% of these injuries occurring during sports participation (Hughes et al, 2008). Between 70% and 90% of ACL injuries have been reported to occur in non-contact situations and therefore sports such as basketball, soccer and handball have shown high incidence of ACL knee injury (Hughes et al, 2008; Quatman, Quatman-Yates, Hewett, 2010). Along with the high incidence rate, knee injuries are typically more serious than ankle injuries with return rate to sport ranging as low as 30%- 50% (Myklebust, et al, 2003). Of those same individuals who had returned from surgery as many as half reported significant problems with instability pain and loss of range of motion when examined 8-10 years after their injuries. The serious implications of sustaining a knee ACL injury has increased focus and research pertaining to understanding the underlying mechanisms of ACL injuries (Gilchrist, et al, 2008).

Mechanically, ACL injury occurs when excessive tension force is applied on the ACL ligament. Based on the various methods used to study ACL injury mechanisms it is apparent that the ACL can be subject to high forces under varying loading conditions (Decker et al., 2003; Kenozek, Torry, Van Hoof, Cowley, & Tanner, 2005; McLean et al., 2005; Quatman, et al, 2010; Yu et al., 2007). Based on previous research, it can be concluded that ACL injuries do not occur solely in the sagittal, frontal or transverse planes, however, research groups have typically focused on the sagittal and frontal planes in isolation when describing the mechanism of ACL injury.

In recent years there has been a disproportionate amount of female athletes damaging their ACL compared to males (Sanna & O' Connor, 2008). Recent studies have proposed that females are six to eight times more likely to suffer non-contact ACL injures than their male counterparts (Hughes et al. 2008). Because of these increased rates occurring with women there has been a considerable effort to understand the gender mechanisms that may predispose females to non-contact ACL injuries (Medina, et al., 2008). A number of these studies looking at female ACL injuries and risk factors have focused on physiological and anatomical measures such as hormone production, limb lengths, height and hip to knee angles (Hewett, et al., 2005). Although these factors may contribute to knee injuries they are essentially non-modifiable in nature. For this reason attention has shifted, focusing now on the differing neuromuscular control strategies and functional abilities of muscles controlling the knee and hip and ankle during movement to account for the discrepancy in non-contact ACL injuries between males and females (Houck, Duncan & De Haven, 2006; Medina, et al., 2008).

While males and females adopt similar dynamic body positioning during athletics, female athletes may use control strategies that emphasize and focus strain on the ACL to a greater extent than their male counterparts (Sigward & Powers, 2007). Studies have repeatedly shown that women compared with men appear to land from a jump, side cut or deceleration from run with less knee and hip flexion, increased knee valgus and associated abduction moments and high quadriceps activity relative to hamstrings activity (Griffin. et al, 2006). Often, females demonstrate insufficient neuromuscular control including; altered muscular strength ratios, insufficient recruitment or inappropriate timing of muscle firing patterns which all may contribute to increased ACL injuries (Myer, Ford, & Hewitt, 2005).

Neuromuscular control in females during sporting movements is viewed as a primary contributor to their increased risk of ACL injuries compared to their male counterparts (McLean, et al., 2005). Hewett et al (2005) have shown prospectively that larger knee abduction moments during the impact phase of landing are more commonly associated with females. Females completing side stepping trials were observed by McLean et al. (2005) to demonstrate significantly larger peak abduction knee moments than males and this trend was found to be dependent on initial contact valgus angle. These results are in accordance with Hewett et al. (2005) who demonstrated that female athletes landed with increased knee abduction (valgus) angles when ACL injuries occur. Borotikar, Newcomer, Koppes, & McLean (2008) further observed muscle fatigue to increase the peak knee abduction angle in females. These results demonstrate that females are at a combined risk of ACL injury by landing in an abducted knee position and increasing the muscle contraction with a valgus moment, further straining the ACL ligament.

Females have also demonstrated a larger discrepancy between their quadriceps to hamstring muscle strength ratio compared to males (Fagenbaum et al., 2003). Overly developed quadriceps strength with under developed hamstrings strength can increase risk of ACL strain, specifically when the knee is close to full extension during impact (Myer et al., 2005). Increased quadriceps contraction in this position can increase the anterior shear forces acting on the proximal tibia (Renstrom, et al., 2008). It has been speculated that due to this increased tibial shear, females are more likely to injure their ACL than their male counterparts (Yu et al., 2007). However recent investigations have found conflicting results regarding female knee flexion angle at impact. In their study Fagenbaum et al. (2003) reported that females consistently landed with increased knee flexion compared to males, and that increasing flexion immediately after impact appeared to be beneficial in preventing ACL injury. Decker et al. (2003) reported that females, who landed more erect at initial impact, demonstrate a larger knee flexion ROM throughout the landing phase compared to males. Landing performance differences between males and females require investigations beyond the kinematic level to fully understand the neuromuscular control strategies by which females differ than males.

Knee kinematic and kinetic variables have been examined in an attempt to determine the mechanism by which females are at risk for ACL injuries. While the ACL injury is a direct result of what occurs at the knee, it is important to consider the contribution of the entire kinetic chain to the knee loading. Alterations in the ankle's ROM have been previously described to affect knee, and to a lesser degree hip, kinematics and kinetics. Poor or abnormal neuromuscular control of the lower limb during athletic movements has been observed in females to a larger extent than males. Therefore the application of ankle stabilization may affect females to a larger degree and potentially further alter the risk of sustaining a knee injury. It may be possible to modify training programs to reduce ACL injuries for female athletes who wear ankle braces.

1.2.5 Theoretical Effects of Bracing on Performance

Athletic therapists have questioned ankle bracings ability to limit peak performance. This theory stems from the belief that the application of a brace can subsequently reduce ankle performance by not allowing for the optimal ankle motion due to the restriction of the brace

(Cordova et al., 2005). During all performance movements, such as sprinting, performing agility maneuvers and vertical jumping the ankle must be able to quickly plantar and dorsi flex the foot/ ankle complex, as this allows for push off and attainment of maximal velocity (Mackean, Bell, & Burnham, 1995). With the benefits of reduced ankle injuries clear, a compromise must be determined and one must consider the value of reduced injury versus a potential for decreased performance when choosing to apply an ankle brace. There have been several studies that have evaluated the effects of ankle stabilization on lower extremity functional performance tasks. Specifically the movements that have been looked at are vertical jumping, running speed and agility performance, all of which are very common and critical to high performing athletes (Cordova et al., 2005).

Sprint times are often measured over short distances between 40 to 80 yards (Cordova, et al., 2005). With the inherent designs of ankle braces it is possible that they will restrict the foot/ankle movements that are required to generate speed. Collectively, studies that have utilized sprint times have found on average, performance detriments of 20 milliseconds seconds while wearing an ankle brace over a 40 yard sprint. This increase in time translates into approximately a 1.0% decrease in running speed (Cordova, et al., 2005; Verbugge, 1996). In terms of athletics this difference may only be substantial for elite level athletes (Bot & van Mechelen, 1999). Vertical jump height, assessed by measuring the distance between the individual's maximum standing reach and their mark at the highest point in their jump, and agility, assessed by measuring the time required to complete a series of quick changes in direction do not seem to be significantly affected by ankle bracing (Bot et al., 1999; Verbrugge, 1996). Cordova, et al. (2005) found that, although there were slight limitations in vertical jump height, comprehensive analysis indicated that ankle stabilizers do not meaningfully or significantly restrict height obtained during a controlled vertical jump performance. Studies have also indicated that ankle brace conditions had little effect on agility performance (Bot et al., 1999; Verbugge, 1996). Verbugge (1996) reported that the average difference observed between agility run times when wearing ankle brace was not substantial. These findings are further supported in a review completed by Cordova, et al. (2005), who reported that ankle stabilization has the least effect on agility course timed performance.

While there may be conflicting results indicating larger discrepancies between braced and control conditions in regards to performance it is still unclear why braces my limit performance. In terms of performance measures, sprint and agility times along with vertical jump height measures are very broad measures that do not describe any underlying lower limb biomechanical differences. Evidence that bracing may or may not hinder performance has typically presented as a direct effect to reduced ankle ROM (Bot et al., 1999; Cordova, et al, 2005; Mackean, Bell, & Burnham, 1995; Verbugge, 1996) without examination into the effect ankle bracing may have on the proximal joint biomechanics, which may under represent how ankle bracing affects the entire lower limb.

1.2.6 Summary

With the increase in knowledge about ankle stabilization there are many reasons why an athlete would choose to wear an ankle brace to prevent initial or reoccurring ankle sprains from happening. Research has proven that ankle stabilization modalities prevent injury. With the lower limb joints working together as an interconnected series to reduce impacts and optimize landing performance along with the wide spread use of braces, there is value to knowing how these ankle braces affect the other joints of the leg as well as affect the ground reaction force profiles acting on the participants. An ankle brace can reduce ankle range of motion, which has the potential to alter the proximal joints of the leg during movements. For a population that seems to be susceptible to knee injuries, females may be more prone to greater changes in knee and hip mechanics than their male counterparts while wearing ankle braces. Ankle bracing may therefore have a greater ability to alter the joint dynamics during athletic movements compared to a non-braced condition. Analyzing females completing simulated athletic maneuvers under both braced and non-braced conditions would give a better overall understanding of how ankle stabilization affects ankle, knee and hip kinematics and kinetics around each joint.

1.3 Purpose Statement, Research Questions, and Hypotheses

1.3.1 Purpose Statement

The primary purpose of this study is to determine whether the application of a standard commercial lace-up style ankle brace (ASO, Medical Specialties Inc, Charlotte, NC, USA) has an effect on the ground reaction force profiles, joint kinematics or the joint moments occurring around the ankle, knee and hip joints during two different simulated athletic maneuvers. Subjects performed both a forward jumping maneuver as well as a sideways jumping maneuver with and without ankle braces. The jumping conditions aim to simulate a standard set of game movements. Values of the primary outcome variables will be compared between bracing conditions within a single movement; no comparisons will be made between movement types.

1.3.2 Research Questions

1) Does the ASO ankle brace modify the ground reaction force profiles at impact and throughout the ground contact phase?

2) Does the ASO ankle brace modify the kinematics and joint moments occurring around the ankle joint compared to a control condition?

3) Is there a change in the knee and hip kinematics and joint moments due to the application of the ASO ankle brace?

4) If the ASO ankle brace increases the stiffness around the ankle joint, will there be an associated decrease in stiffness around the knee joint to compensate?

1.3.3 Hypotheses

It is hypothesized that for the current investigation there will be a change in the ground reaction force profiles, as well as a change in the ankle knee and hip kinematics and joint moments when movements are performed under the ASO brace condition.

1) Previous research has determined that the application of an ankle brace can restrict the ankle joint range of motion of the wearer, (Cordova, et al., 2010; Popadopoulos et al., 2005) in the sagittal, frontal and transverse axes (Eils et al., 2002; DiStefano et al., 2008). Due to the ankle joint restriction, there is potential for the knee and hip joints to increase in flexion at impact to compensate for the reduced ankle motion (Cordova, et al., 2010). **Therefore it is hypothesized that brace application will decrease the ankle joint range of motion while simultaneously increase the knee and hip joint ranges of motion to compensate across contact.**
2) The ground reaction force profiles are important as they relate to the absorption and transmission of energy onto the different tissues comprising the musculoskeletal system of the lower limb (McCaw et al., 1999). The application of an ankle brace has been shown to increase vertical ground reaction forces by increasing the rigidity of the ankle and limiting ankle range of motion during drop landings (Cordova et al., 2010). **It is hypothesized that brace application will increase both the magnitude of and the rate of loading for the ground reaction forces.**
3) If there is a change in the amount of force transferred to the ankle, knee and hip with the application of an ankle brace, it is likely that the musculature surrounding the joint will have to act differently in order to compensate. **It is hypothesized that the application of**

an ankle brace will change the joint moments occurring around the ankle as well as in the proximal knee and hip joints

4) Joint stiffness properties are essential to controlling the dynamic stability of the leg, and recent evidence has revealed that there can be adjustments in the coordinating pattern of lower limb stiffness by modulating ankle stiffness (Farley & Morgenroth., 1999; Zinder, Granata, Shultz and Gansneder, 2009). The application of a brace adds-non stretch material around the ankle joint which could stiffen the joint. **It is hypothesized that the addition of an ankle brace will increase the stiffness at the ankle joint, and subsequently decrease the knee stiffness in response in order to regulate overall joint stiffness.**

Chapter 2

Methods

2.1 Study Overview

Eight participants, currently active members of the University of Saskatchewan Women's Huskie Basketball team, were recruited for this study. Data collection took place at the College of Kinesiology's Musculoskeletal Biomechanics Laboratory located within the Physical Activity Complex on the University of Saskatchewan Campus. Each subject performed a series of single legged jump landing/takeoff maneuvers in forward and sideways directions while their movements and ground reaction forces (GRF) were recorded. The participants performed the movements both with and without wearing lace-up ankle braces. Dependent variables included GRF and ankle, knee and hip joint angles and joint moments and ankle and knee joint stiffness. Comparisons were made between the braced and non-braced conditions.

2.2 Participants

A sample of 8 Canadian Interuniversity Sport (CIS) level female basketball players gave their written informed consent to participate. The participants were 21.38 ± 2.23 years (mean ± SD), 171.5 ± 6.1 cm tall, 67 ± 7.1 kg and had an average of 4.0 ± 1.3 years of CIS level experience playing basketball. Inclusion criteria included: 1) free of significant physical or neurological impairment; 2) free of any significant lower body injury (such as broken bones, torn or sprained ligaments) for the previous 6 months leading up to the study; 3) ability to perform a series of single leg jump landings/takeoffs. The study was approved by the University of Saskatchewan biomedical review board for research in human subjects (See Appendix A for a copy of the Ethics: Certificate of Approval).

All of the participants recruited were considered part of a trained, athletic population with familiarity with ankle brace application and wear. The familiarity with brace application and wear was due to an agreement between Huskie coaching and training staff along with the Huskie basketball players, requiring all members of the team to wear ankle braces on both the left and right ankles for all basketball games and practices as well as all training sessions excluding weight training. A population that had familiarity with ankle brace wear was selected to remove any learning aspects associated with ankle brace wear.

2.3 Instruments & Devices

2.3.1 Ground Reaction Forces

Two force platforms were used. The primary force platform (AMTI model OR6-7 strain gauge, AMTI, Watertown, MA, USA) was imbedded in the floor of the data collection area. It was used to record the six GRF components (three force components: Fx, Fy, Fz; three moment components: Mx, My, Mz) of the landing phase of the jump maneuvers (impact to take-off). A secondary force platform (Model 4060-10, Bertec, Columbus, OH, USA) was used as the take-off platform during the jumps. Vertical force data from this platform were used to identify the initiation of the airborne phase, defined as the time between contact between the two force paltormas, of the jumps. Analog signals from both force plates were collected using the data acquisition system built into the motion capture system (Vicon Motion Systems, CO, USA) used to collect the kinematic data (see next section). The force data were sampled at a rate of 2000 Hz.

2.3.2 Kinematics

A commercial motion capture system (Vicon Nexus, Vicon Motion Systems, CO, USA) was used to record the 3D kinematics of the lower limbs, pelvis and upper arms of each participant. The motion capture system consisted of eight specialized high speed video cameras (Model F-20, Vicon Motion Systems, CO, USA) which were used to track the 3D positions of 14 mm diameter spherical retro-reflective markers attached to the participants. The motion capture data were collected at a sampling rate of 200 Hz and synchronized with the force data.

The movement protocols used a full body bilateral marker set consisting of 35 required (tracking) markers and 12 calibration markers, used to track the position of the body moving through space. Each marker was covered in reflective tape and attached to a plastic base for mounting. Each marker was taped using double sided hypoallergenic wig tape and either mounted directly onto the participant's skin or applied onto a marker cluster composed of heat moldable thermoplastic sheets cut and molded to fit onto their associated anatomical position (Clusters can be observed attached to a participant's femur(lateral thigh) and shank (lateral calf) in Figure 2.1a & 2.1b). Details concerning the marker placements and calibration protocols are given in Appendix B.

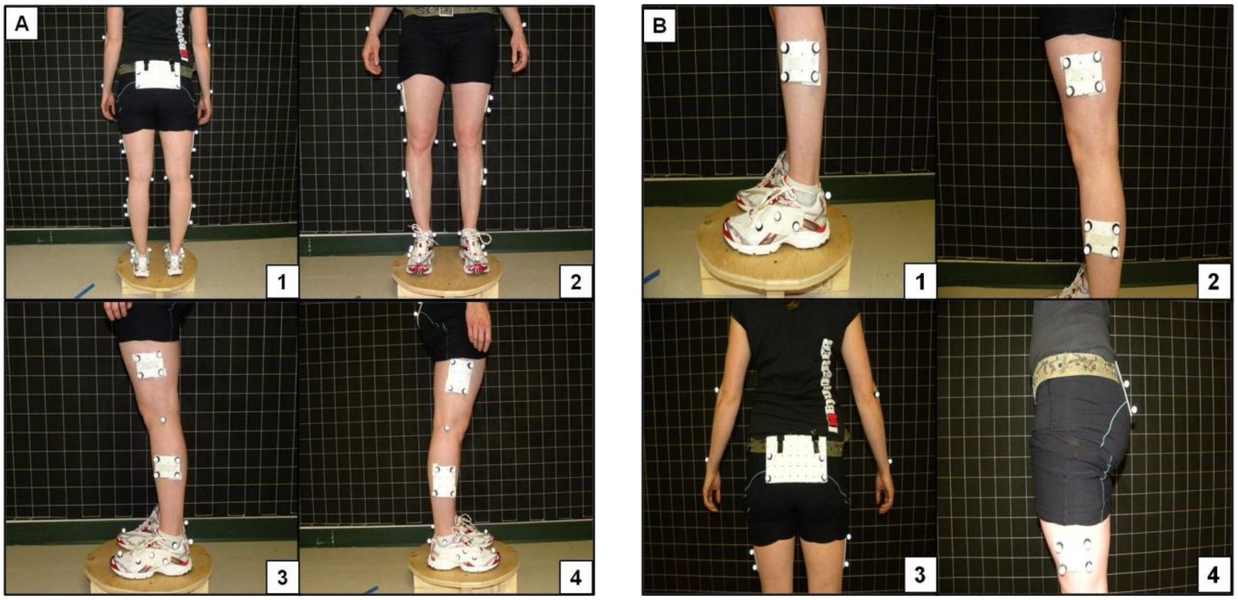

Figure 2.1a: The calibration markers displayed in 1) the posterior view; 2) anterior view;) Left side; 4) Right side. Also displayed are the mounting of the clusters
Figure 2.1b: The lower body required marker set displaying 1a) tibia cluster and foot markers; 2a) femur and tibia clusters 3a) pelvis and femur clusters and 4a) lateral view of pelvis and femur clusters.

2.3.3 Ankle Braces/ Shoes

The ankle braces chosen for the current study were the ASO brand ankle lace up brace (Medical Specialties, Inc., Charlotte, NC, USA). All participants were familiar with the ASO brace as it was the prescribed ankle brace provided to all players on the women's Huskie basketball team. The ASO braces are a lace up style ankle brace consisting of a non-stretch nylon shell, constructed of 840 denier nylon. The term denier refers to the mass of the fibers composing the material of interest and indicates the material's durability. A larger denier value indicates a heavier and more durable woven material. A typical ankle brace is composed of nylon ranging from 840 -1000 denier. The ASO brace includes built in non-stretch stabilizing support straps which provide added support to the ankle joint complex by replicating an ankle taping procedure. The ASO also includes removable stabilizing plastic inserts located on the medial and lateral side of the brace to further support the ankle joint. The ASO brace is completed by an elastic strap which encloses both the ankle laces and the stabilized position of the lateral support straps, ensuring nothing will come undone during wear. Images of all components of the ASO brace can be found in Figure 2.2

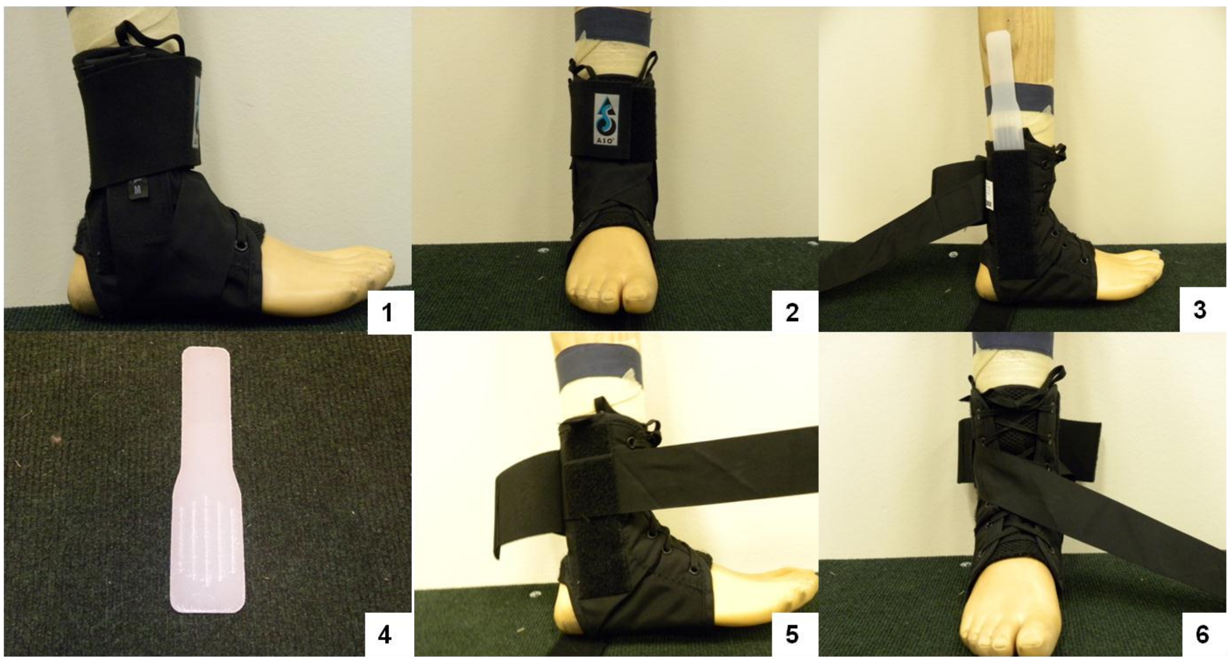

Figure 2.2: The ASO ankle brace on ankle model. The lateral and front view of the completed brace with elastic closure finished (1 and 2). The ASO brace is constructed with additional support structures, including lateral stabilizing plastic insert shown extended out of its sleeve and alone (3 and 4), and the non-stretch stabilizing support straps shown from both the side and anterior view (5 and 6).

All participants were required to wear braces on both ankles to mimic the same support provided to them during actual games and practices. Each participant was given a brand new, never worn ankle brace for their jumping leg to perform the jumping movement protocol with. The brace worn on their non-jumping leg was either brand new, or a brace that had been worn by a previous participant. Each participant was matched for brace size based on the size prescribed for them by the University of Saskatchewan Athletic Training department. The braces were assumed to be identical out of the box and were all obtained from one single order through the University of Saskatchewan Athletic Department.

All participants were allowed to lace up their own ASO ankle brace. Each participant was instructed that the braces be laced up to a tightness level matching what they would wear for games, practice, and training activities. The same researcher observed that the braces were tightened and that no one left the ankle braces loose and non-supportive.

2.4 Procedures

Participants reported to the University of Saskatchewan's Kinesiology department Musculoskeletal Biomechanics laboratory for one single testing session. Each participant was instructed to arrive changed, wearing shorts and a T-shirt, and her own Huskie basketball team shoes. The team shoes were a mid-height Nike Zoom Kobe V basketball shoe (Nike, Beaverton, OR, USA). One participant did not use the team shoe based on personal preference. This participant shoe was a mid- height shoe similar to the other seven participants but was a Nike Air Max (Nike, Beaverton, Or, USA). Upon arrival all volunteers completed an informed consent that described testing protocols (Appendix A). In addition to the informed consent participants were asked to identify which leg was the preferred leg for takeoff during a layup. All participants indicated that their left leg was preferred, and as such all participants were instructed to jump, land and push-off using their left leg.

Participants were systematically assigned to one of two braced groups in alternating fashion based on the order of recruitment, with all odd number participants (1,3,5,7) assigned to the control condition first and even numbered participants (2,4,6,8) assigned to the braced condition first.

2.4.1 Movement Protocol – Jumping

Participants completed a series of standardized single leg jump/take off movements. These were chosen to simulate movements typically encountered during athletic participation. The two movements used in this study were: 1) a forward single leg jump; and 2) a sideways single leg jump. Both movements required the participants to jump down from a 30 cm high box onto a force platform integrated into the biomechanics runway located approximately 30 cm from the front of box. This protocol allowed each participant to gain momentum prior to contacting the force platform, which ensured that each movement generated contact forces similar to those experienced in real athletic situations. The height was chosen so that the contact force magnitudes would be similar to those seen if the participants had been jogging up to the runway prior to initiating a forward or sideways movement (Kellis, & Kouvelioti, 2009; Yeow, Lee, & Goh, 2009).

Each movement began with a forward drop from the raised force platform down onto the force platform imbedded in the floor of the data collection area. From there the participants either performed a forward jump or a sideways jump (to the side) off the lower force platform. The criteria for an acceptable jump were as follows:

1) The criteria for the forward jump stipulated that as soon as the participants landed with their left foot, they were instructed to jump completely forward in one continuous motion without stopping, and take off with the same preferred foot as contacted the ground. The jump was plyometric in nature with the non-preferred foot never coming in contact with the ground but was not a fully explosive plyometric jump. The participants were instructed to jump forward off the raised platform and not 'jump up' in an attempt to control for variability in jump height, and to finish the motion by landing on their

preferred leg once again. A representative trial is presented in Figure. 2.3. & Figure. 2.3.b

2) The criteria for the sideways jump were to have the participant land on their preferred leg, and in one continuous motion without pausing, take-off laterally directing their body completely sideways with no further motion forward. The participant was instructed to land on their right foot after completing the take off to ensure a safe landing. A representative trial is presented in Figure. 2.4. & Figure. 2.4.b- Alternate view.

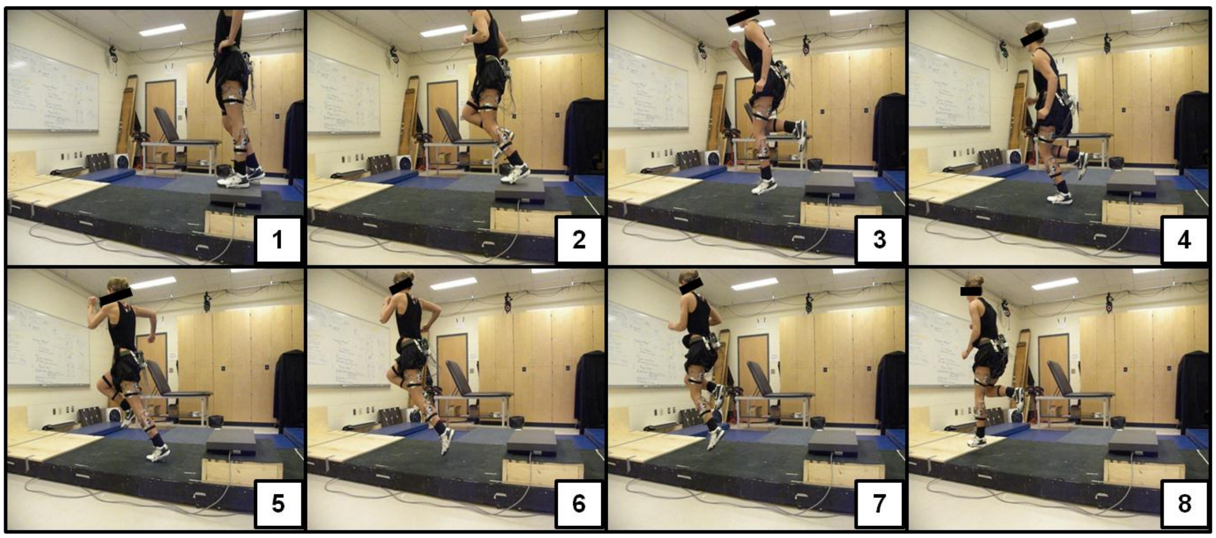

Figure 2.3: Representative trial of a Forward jump (*Numbers indicate progression in sequence of photos taken) outlining the key components of the movement including 1)Ready position; 2) Initial take-off ; 3)Flight phase prior to impact of 2^{nd} force platform; 4) 1^{st} impact phase; 5) Propulsive pushing phase; 6) 2^{nd} take-off phase; 7) 2^{nd} flight phase; 8) Final impact and completion of movement.

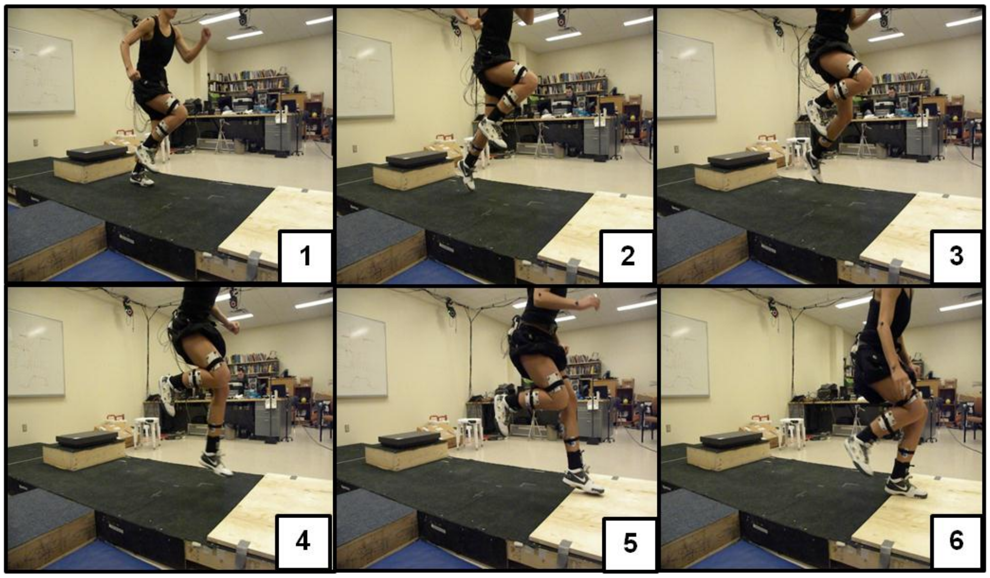

Figure 2.3.B: Alternative view of the Forward jumping protocol. (*Numbers indicate progression in sequence of photos taken)

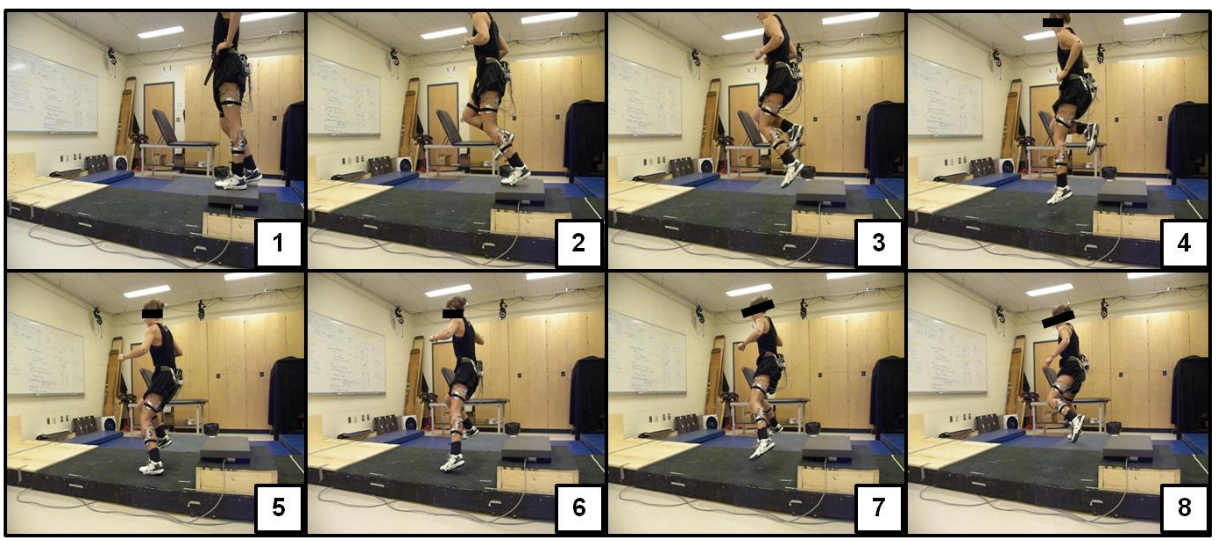

Figure 2.4: Representative trial of a Sideways jump (*Numbers indicate progression in sequence of photos taken) outlining the key components of the movement including 1)Ready position; 2) Initial push-off ; 3)Take-off; 4) Flight phase prior to impact of 2nd force platform; 5) 1st impact phase; 6) Propulsive pushing phase 2nd take-off phase; 7) 2nd take-off and flight phase; 8) Final impact and completion of movement.

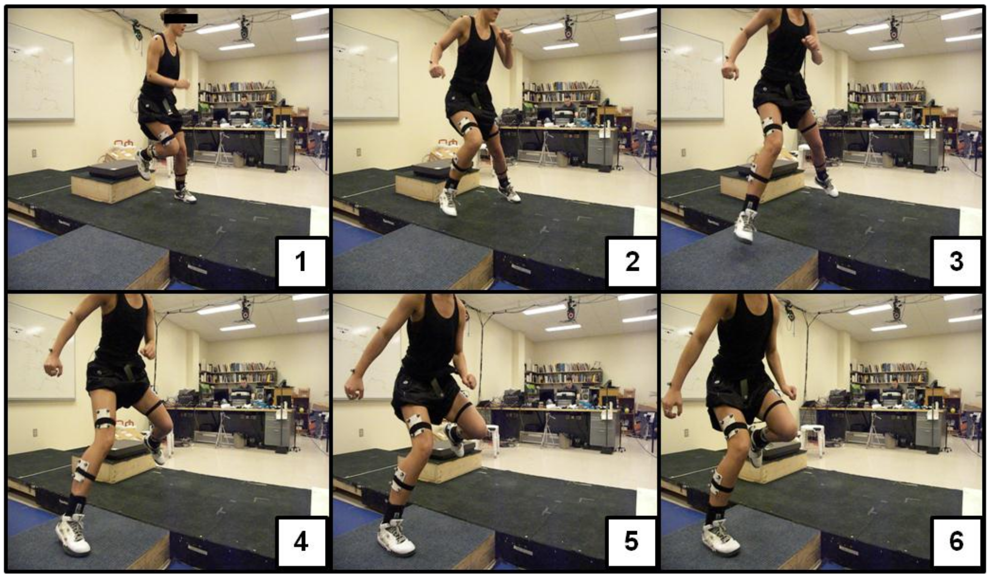

Figure. 2.4.b: Alternative Representation of Sideways jumping protocol. (*Numbers indicate progression in sequence of photos taken)

2.4.2 Instructions

Prior to each jump, the participant was instructed to get themselves in the ready position consisting of them balancing on their preferred (i.e. left) leg, with their non-preferred (i.e. right) leg bent at the knee at a 90° angle. Once the participant verbally indicated readiness the researcher activated a movement cue displayed on a 22 inch computer monitor located in front of the landing platform.   One of three movement cues ‘Forward’ or ‘Sideways’ or ‘Stop’ would appear in a randomized order that was the same for all participants(See Appendix B. Figure 2.3.2 for Visual Cues for the Forward, Sideways and Stop movements). The ‘Stop’ cue indicated that no action was to take place and was utilized to prevent anticipation of the actual movement cues. Movement cues appeared on the video monitor after a randomized amount of time ranging from 2- 4 seconds. All participants were instructed to begin movement as soon as the cue became observable on the video monitor. No measure of reaction time was included in this study, and this was also explained to the participants.   A total of 15 forward and15 sideways trials that matched the qualifying criteria of an acceptable trial were collected for each participant, for each brace condition. Due to data collection issues and participant movement error, this resulted in a range of 35 to 39 trials collected for each participant including 'STOP' cued trials.

Minimal instructions were given to the participants regarding the general nature of the jump landings and subsequent takeoff (drop down from box when cue is given, and land on force platform embedded in the landing surface, be sure not to pause during the landings and complete the movements in one continuous motion), with only one demonstration by the researcher per movement given to limit coaching effects. The participants were also instructed to perform the jumping task at their own pace and with their own preferred technique and were subsequently given one or two practice trials. Current research has determined that a standardized jogging warm-up protocol can reduce brace tightness (Dizon et al., 2010). Therefore a warm-up was not included in the study protocol due to potential loosening effects that may have altered braced dynamics for those in the initial braced condition. However each participant was verbally questioned as to their willingness to begin without further warm-up, and no participant was opposed to beginning the jumping protocol.

Following the completion of all jumping trials for the first brace condition the participant was instructed to complete a series of walking trials prior to the change in brace condition for the second set of jumping trials. Walking trials consisted of a set of standardized (6 meter) distance gait trials. Each walking trail was completed by having the participant start at one end of the lab runway and at a self-selected pace walk across the runway stopping at the end. Trials were accumulated for both directions across the runway to obtain data from both the left and right legs. The walking data was intended to determine alterations in gait caused by brace effects, however the scope of this thesis does not include the results of the walking analysis. After the change in braced conditions the participants repeated the jumping protocols**.**

2.5 Data Analysis

A minimum of 10 trials per condition and movement were processed within the Vicon Nexus software for each participant. Subsequently six trials were analyzed for each brace condition and movement direction of each participant for statistical analysis. These trials were systematically chosen to match for both the flight time between takeoff and impact, as well as the ground contact time, occurring between impact and takeoff, between conditions within a given participant. This systematic approach was utilized to select representative jumping trials between brace conditions which were not biased by any biomechanical variable.

2.5.1 Ground Reaction Force Profiles

The GRF profiles from the landing force platform were examined in the medial (-)/lateral (+) (**x**), anterior (-)/posterior(+) (**y**), and vertical(+) (**z**) directions. The vertical force was used to define the start and end of the foot contact phase. The contact phase was defined as the time period when the vertical force was above 15 N. Since the GRF data were collected at a higher sampling rate compared to the kinematic data, the GRF data were down-sampled (from 2000 Hz to 200 Hz) to match the time points of the kinematic data. The GRF data over the contact phase were then normalized to 101 points corresponding to 0 – 100% of contact. The GRF data were not filtered. The location of the center of pressure during foot contact was calculated from the GRF data and used in the subsequent joint moment calculations.

GRF data were used to define key points within the contact phase. The time of maximal vertical GRF ($GRFz^{max}$) and maximal braking (i.e. posterior directed) force ($GRFy^{max}$) for both the forward and sideways movements were identified. The time of peak propulsive force was also found. In the forward direction this was the peak force in the anterior direction ($GRFy^{min}$) and for the sideways jumps it was the peak force in the medial direction ($GRFx^{min}$). The GRF values and the times of occurrence were analyzed. GRF data was used to define the two time points at which the kinematics and joint moments were to be examined. The two time points were impact and maximal propulsive. The impact phase corresponded with the 0-5% of contact, while the maximal propulsive phase corresponded to the percentage of contact time which maximal anterior force (forward maneuver) and maximal medial force (sideways maneuver) were achieved and occurred at approximately 75% of stance.

2.5.2 Kinematics

Anatomical coordinate systems were defined for the pelvis and the left femur, tibia and foot using static calibration data as described in Appendix B. For all participants except one, raw kinematic data were filtered using a low pass 4th order Butterworth filter set with a cutoff frequency of 40 Hz. Due to noise in the raw data, one subject's data were filtered with a cutoff frequency of 20 Hz. These relatively high cutoff frequencies were chosen based on visual examination of the data and the desire to retain as much of the original data as apssible. First and second derivatives used in subsequent calculations were based on kinematic data that were filtered with cutoff frequencies of 20 Hz and 15 Hz respectively in order to reduce errors inherent in numerical differentiation techniques. This dual filtering approach was used to obtain the best data from both the first and second derivatives. Filtering for each derivative must be applied to the raw data depending on the variable analyzed in order to optimize the removal of the noise artifact (van den Bogert, de Koning, 1996). Three dimensional joint angles for each trial were calculated for the left hip, knee and ankle using the Cardan rotation sequence described by Grood and Suntay (1983). For each jumping trial the abduction/adduction (x-axis), flexion/extension (y-axis) and internal/external rotation (z-axis) ranges of motion over the contact phase for the ankle, knee and hip joints were calculated. In addition, joint angle data were extracted at the time points corresponding to initial impact and at the time of peak propulsive force.

2.5.3 Kinetics

Using an inverse dynamics approach (Winter, 2005) the three dimensional net joint moments at the ankle, knee and hip were determined. The net joint moments are the resultant torques that act around a joint at any given point in time. For most movements, these torques are primarily generated by the muscles that cross the joint but can also include contributions provided by soft tissues such as ligaments. Net joint moments give an indication as to which muscle groups are dominant around a joint at a given time point. For example, an extensor joint moment at the knee would indicate that knee extensor muscles (i.e. quadriceps) are active, over and above any co contractions.

Moment data were expressed along the three joint axes. The x-axis moments corresponded to inversion/eversion at the ankle and abduction/adduction at the knee and hip. The y-axis moments corresponded to the plantar/dorsi flexion at the ankle and flexion/extension at the knee and hip. The z-axis moments corresponded to the internal/external rotation at the ankle, knee and hip. Joint moments are calculated relative to the local coordinate system of the distal segment (Figure 2.5). The peak joint moments in each axis during the contact phase were identified. Joint moment values were also extracted at the time of impact and the time of peak propulsive force.

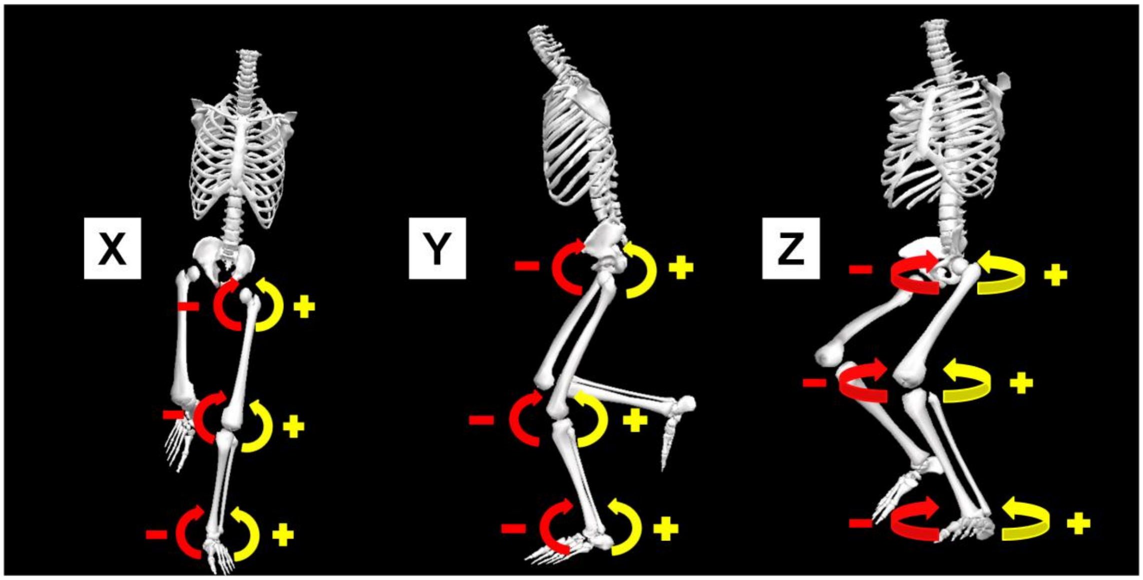

Figure 2.5: Joint moments in orientation used for analysis in X, Y & Z axes.
Ankle X-(+) eversion Knee X-(+) abduction Hip X- (+) abduction
Ankle Y-(+) plantar flexion Knee Y-(+) flexion Hip Y- (+) extension
Ankle Z-(+) external rotation Knee Z -(+) external rotation Hip Z - (+) external rotation

2.5.4 Stiffness

Sagittal plane joint stiffness for the ankle and knee was calculated during the early contact phase. Joint stiffness is defined as the torque required to move a joint through a given angular displacement. This can be calculated by plotting the joint moment against the joint angle and taking the slope of the curve. This was done mathematically by calculating the first derivative of the joint moment-angle curve to generate an instantaneous stiffness curve. The average sagittal plane joint stiffness at the ankle and knee was calculated by taking the average joint stiffness from impact until peak vertical force. Additionally, impact stiffness was examined by taking the instantaneous stiffness value occurring at the time corresponding to 0 % of contact.

2.6 Statistical Analysis:

Due to the exploratory nature of the research design, separate two tailed paired t-tests were used to examine the main effect of brace (braced vs. non-braced). Analyses were run separately for each jumping maneuver (forward and sideways). This statistical model was used to assess the effect of ankle stabilizers on ankle knee and hip joint kinematic and kinetic variables as well as to assess the difference in ground reaction force variables and the locations (timing-percentage of contact) of the changes. Comparisons, on either the peak magnitude of each variable or the timing (percentage of contact) of each variable, were made using paired *t*-test with Bonferroni adjustment for the total number of variables examined within each maneuver. For each maneuver 45 variables were examined, this adjusted our level of significance to $p < 0.001$ (0.05/45).

Bonferroni adjustment was used specifically to protect against Type I error. This adjustment however may cause Type II error and subsequently the results may not reach the level of significance between conditions where there really was a change. To help address the issue of Type II error, trends were examined in addition to any significant findings. Within this study, trends were defined as variables with an alpha level of $p<0.05$. The decision to include the trends was to explore the biomechanical differences that occurred during the jumping movements which did not meet the adjusted $p<0.001$ level of significance but were still deemed relevant, and allow for the discussion of variables that were deemed important to the study.

Chapter 3

Results

3.0 Summary

The series of paired t-test revealed that there were significant differences in joint kinematics at the ankle, along with trends observed at the ankle, knee with no changes in hip kinematics. There were also strong trends observed associated with changes in the joint moments produced at the ankle, with smaller trends at the knee and hip. Trends were also observed between brace conditions for the GRF's generated for both the forward and sideway maneuvers.

The statistical data is presented in the form of tables outlining each group of variables *p-value.* As well, certain variables have been graphed in a manner to better explain and clarify the measured data occurring for the participant's movement under both stabilization conditions. As outlined above, the statistics (paired *t-tests*) were run for either the average of the peak magnitudes or for the percentage of contact at which the variable occurs. Effect size is also presented in the tables pertaining to ankle, knee and hip angle and moment data as well as ankle and knee stiffness. Effect size is included as a secondary description of the magnitude of the difference observed between braced and control conditions.

Some measures are presented in graphs displaying a variable during both the forward and sideways manoeuvres. When presenting a time-series data that has been averaged across a number of participants, overall peak values tend to be under-represented. This "smoothing" effect is due to peak magnitudes occurring at different time points within the various time-series being averaged. The final effect is that the mean time-series curves will typically underestimate the individual peak values and often does not offer a true representation of the differences reported in the tables. Subsequently, as a way of presenting the graphs to represent the statistical information, curves from single representative subjects will often be used to indicate the position and magnitude changes within each variable presented.

3.1 Flight Time and Contact Time

There were no differences between either brace condition for the time each participant spent in the flight phase prior to impact, or the time spent in contact with the force platform prior to making their directional jump in either the sideways or forward direction (Table 3.1). This is an indication that the overall momentum for each condition was similar and allowed for comparisons between the two conditions.

Table 3.1 Flight and Contact Time:
The Time in Flight corresponds to the interval between the first instance of toe off from the 30cm box and touchdown onto force platform within the landing surface. Time in Contact corresponds to time spent on landing surface force plate until time of takeoff. Flight and contact time are recorded in seconds (s).

Time in Flight	Brace	Mean (±S.D.)	*t*	*p*
Forward	ASO	0.240 s (0.03)	0.828	0.435
	Control	0.236 s (0.03)		
Sideways	ASO	0.247 s (0.02)	0.961	0.368
	Control	0.244 s (0.02)		
Time in Contact				
Forward	ASO	0.405 s (0.07)	0.741	0.483
	Control	0.402 s (0.07)		
Sideways	ASO	0.467 s (0.09)	1.613	0.151
	Control	0.457 s (0.08)		

3.2 Ground Reaction Force

3.2.1 Vertical GRF

There was a trend observed for the magnitude and timing of the ground reaction forces for both the forward and sideways jump between braced conditions. The average vertical GRF profiles for the forward and sideways jumps are shown in Figure 3.1. The ankle braced condition demonstrated greater a maximal vertical ground reaction force ($GRFz^{max}$) and a faster time to $GRFz^{max}$ ($TGRFz^{max}$) for both the forward and sideways direction. (Table 3.2)

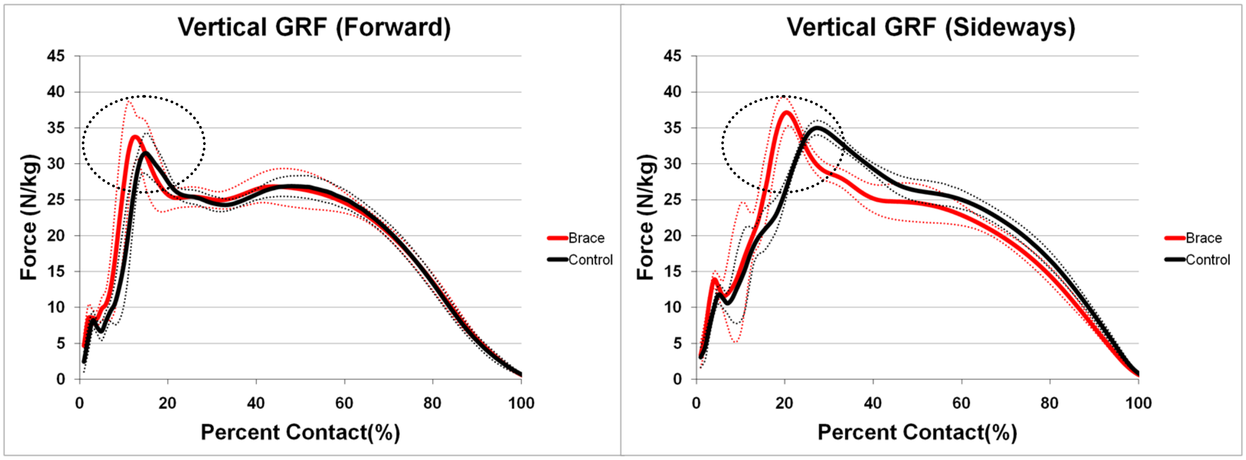

Figure 3.1 Maximal $GRFz^{max}$ magnitude and location:
Vertical GRF profiles normalized to 100 % of contact time. Data are representative means (SD) for Subject 8 (forward) and Subject 3 (Sideways) Magnitude of ground reaction force presented in Newtons per kilogram of body mass (N/kg).

Table 3.2 Maximal $GRFz^{max}$ magnitude and location:
$GRFz^{max}$ corresponds to the peak vertical GRF and Location to percentage of contact time (see Figure 3.1). Magnitude of vertical ground reaction force ($GRFz^{max}$) presented relative to body mass.

$GRFz^{max}$	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	34.11 N/kg (4.47)	2.236	0.050	0.417
	Control	32.48 N/kg (3.22)			
Sideways	ASO	37.76 N/kg (3.50)	3.113	0.017	0.513
	Control	36.12 N/kg (2.89)			
$GRFz^{max}$ Location ($TGRFz^{max}$)					
Forward	ASO	12.71% (5.25)	-3.024	0.019	0.337
	Control	14.71% (6.56)			
Sideways	ASO	10.27% (4.16)	-2.746	0.029	0.417
	Control	12.42% (5.97)			

3.2.2 Braking GRF

During the braking phase of movement, defined as the time period when the anterior/posterior force was directed in the posterior direction, there were trends in the magnitude of the braking force between brace conditions. The anterior/posterior GRF profiles for the forward and sideways maneuvers are given in Figure 3.2. The early braking GRF is indicated by a large positive (posterior) spike within the first 5% of contact time. During the forward jump, the braced condition demonstrated a trend for a larger maximal braking force (F_GRFy^{max}) along with a trend for an earlier time to reach maximal braking force (F_TGRFy^{max}). During the sideways jump the braking force was larger between conditions (S_GRFy^{max}), however no difference was apparent for the time to reach maximal braking force (S_TGRFy^{max}). (Table 3.3)

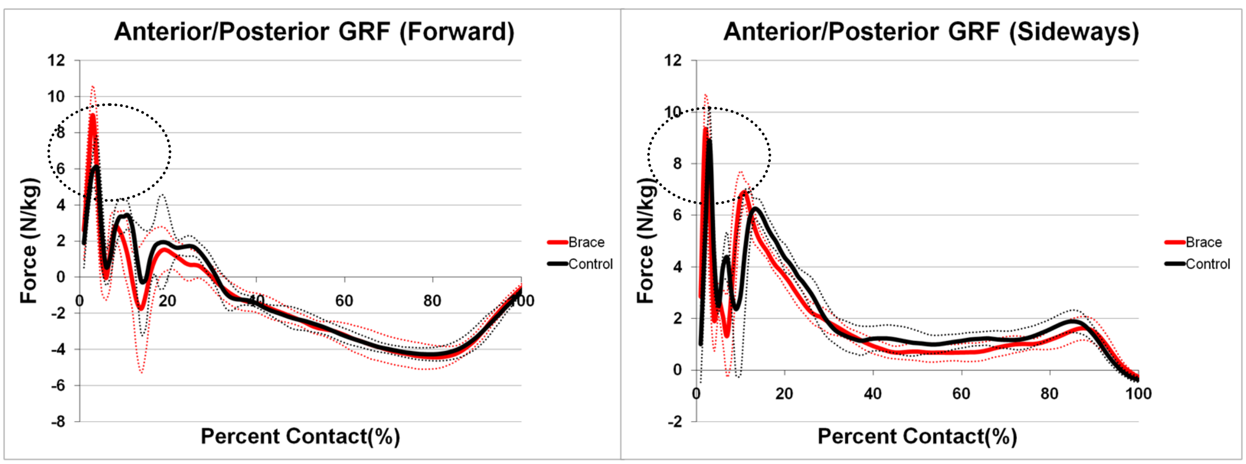

Figure 3.2 Maximal GRFy^{max} magnitude and location:
Anterior/Posterior GRF normalized to 100 % of contact. Data are representative means (SD) for Subject 5 (Forward) and Subject 7 (Sideways). Magnitude of ground reaction force presented in Newtons per kilogram of body mass (N/kg). The posterior direction is positive.

Table 3.3 Maximal Posterior GRFy^{max} magnitude and location:
The maximal posterior ground reaction force corresponds to peak posterior GRF and Location to percentage of contact time (see Figure 3.2). Magnitude of vertical ground reaction force (GRFy^{max}) presented relative to body mass.

GRFy^{max}	Brace	Mean (±S.D.)	*t*	*p*	ES
Forward	ASO	8.59 N/kg (1.51)	2.987	0.020	0.746
	Control	7.36 N/kg (1.76)			
Sideways	ASO	9.20 N/kg (1.86)	2.629	0.034	0.451
	Control	8.39 N/kg (1.70)			
GRFy^{max} Location					
Forward (F_TGRFy^{max})	ASO	2.00% (0.56)	-2.646	0.033	0.481
	Control	2.25% (0.48)			
Sideways (S_TGRFy^{max})	ASO	1.92 % (0.65)	-1.994	0.086	0.921
	Control	3.15 % (1.77)			

3.2.3 Propulsive GRF

No bracing effects were observed for either the time or magnitude of the maximal propulsive ground reaction force for either forward of the sideways maneuver. Propulsive ground reaction forces were measured as either the maximal anterior ground reaction force observed during the forward (GRFy^{min}) or as the maximal medial ground reaction force (GRFx^{min}) during the sideways maneuver. The GRF profiles for the forward and sideways maneuvers are given in Figure 3.2. Negative values represent an anterior GRFy force and a medial GRFx with respect to how the participants were oriented during the jumping maneuvers. As such, the values are reported as 'min' values to differentiate propulsive forces from braking forces. (Table 3.4a and Table 3.4b).

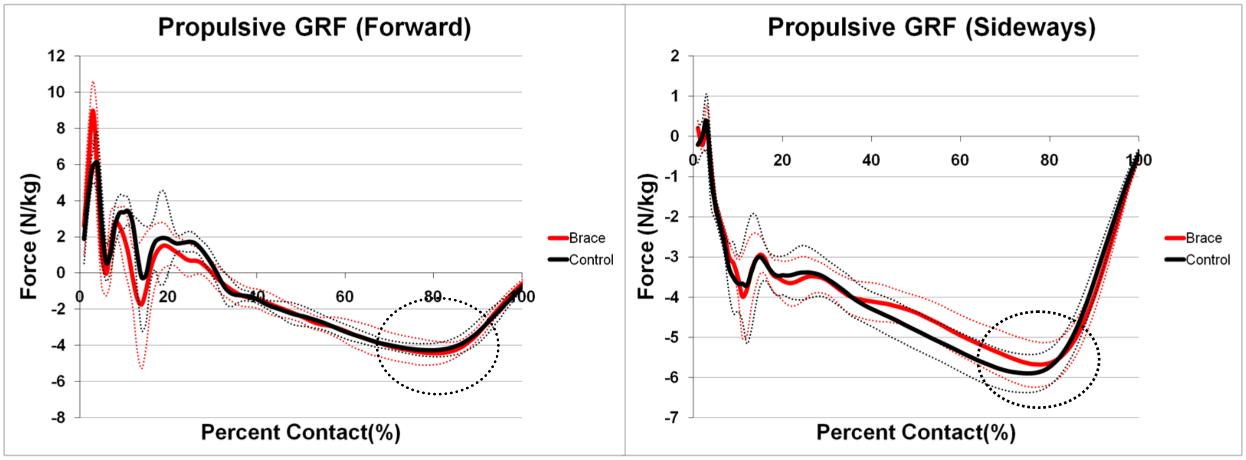

Figure 3.3 Maximal Propulsive GRFy^{min} GRFx^{min} magnitude and location: Propulsive GRF profiles normalized to 100 % of contact time. Data are representative means (SD) for Subject 5 (Forward) and Subject 3 (Sideways). Magnitude of ground reaction force presented in Newtons per kilogram of body mass (N/kg). For the forward jump, negative is the anterior direction and for the sideways jump negative is the medial direction. Highlighted is the maximal propulsive phase.

Table 3.4a Maximal Propulsive GRFy^{min} Forward magnitude and location: The propulsive ground reaction force corresponds to push off phase of movement with location indicated by percentage of contact time (see Figure 3.3). Magnitude of vertical ground reaction force (GRFy^{min}) presented relative to body mass and negative GRF indicates anterior direction.

GRFy^{min}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	-3.75 N/kg (0.54)	.234	0.822	0.035
	Control	-3.77 N/kg (0.51)			
GRFy^{max} Location (TGRFy^{min})					
Forward	ASO	76.46 % (5.79)	1.063	0.323	0.275
	Control	74.46 % (8.48)			

Table 3.4b Maximal Propulsive GRFx^{min} Sideways magnitude and location: The propulsive ground reaction force corresponds to push off phase of movement with location indicated by percentage of contact time (see Figure 3,3). Magnitude of vertical ground reaction force (GRFx^{min}) presented relative to body mass and negative GRF indicates medial direction.

GRFx^{min}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Sideways	ASO	-5.64 N/kg (1.10)	.482	0.645	0.209
	Control	-5.70 N/kg (1.10)			
GRFx^{min} Location (TGRFx^{min})					
Sideways	ASO	73.42 % (8.50)	.530	0.612	0.077
	Control	72.87 % (6.95)			

3.3 Joint Angles

3.3.1 Overall Range of Motion

The overall sagittal plane joint ROM was measured for the ankle (ROM^{ank}), knee (ROM^{knee}) and hip (ROM^{hip}) joints for both the forward and sideways maneuvers (Figure 3.4). Overall joint ROM was calculated as the difference between the minimum and the maximum joint angle observed across contact time. All ROM values are presented as an absolute value for change in joint ROM. The only bracing effects observed for the change in overall sagittal joint range of motion were observed for ROM^{ank}. The brace condition displayed a significant decrease in the overall ROM^{ank} for sideways maneuvers only, with a strong trend for a difference in the forward maneuver. No differences were observed for the sagittal plane ROM^{knee} or ROM^{hip} between bracing conditions for either maneuver. (Table 3.5)

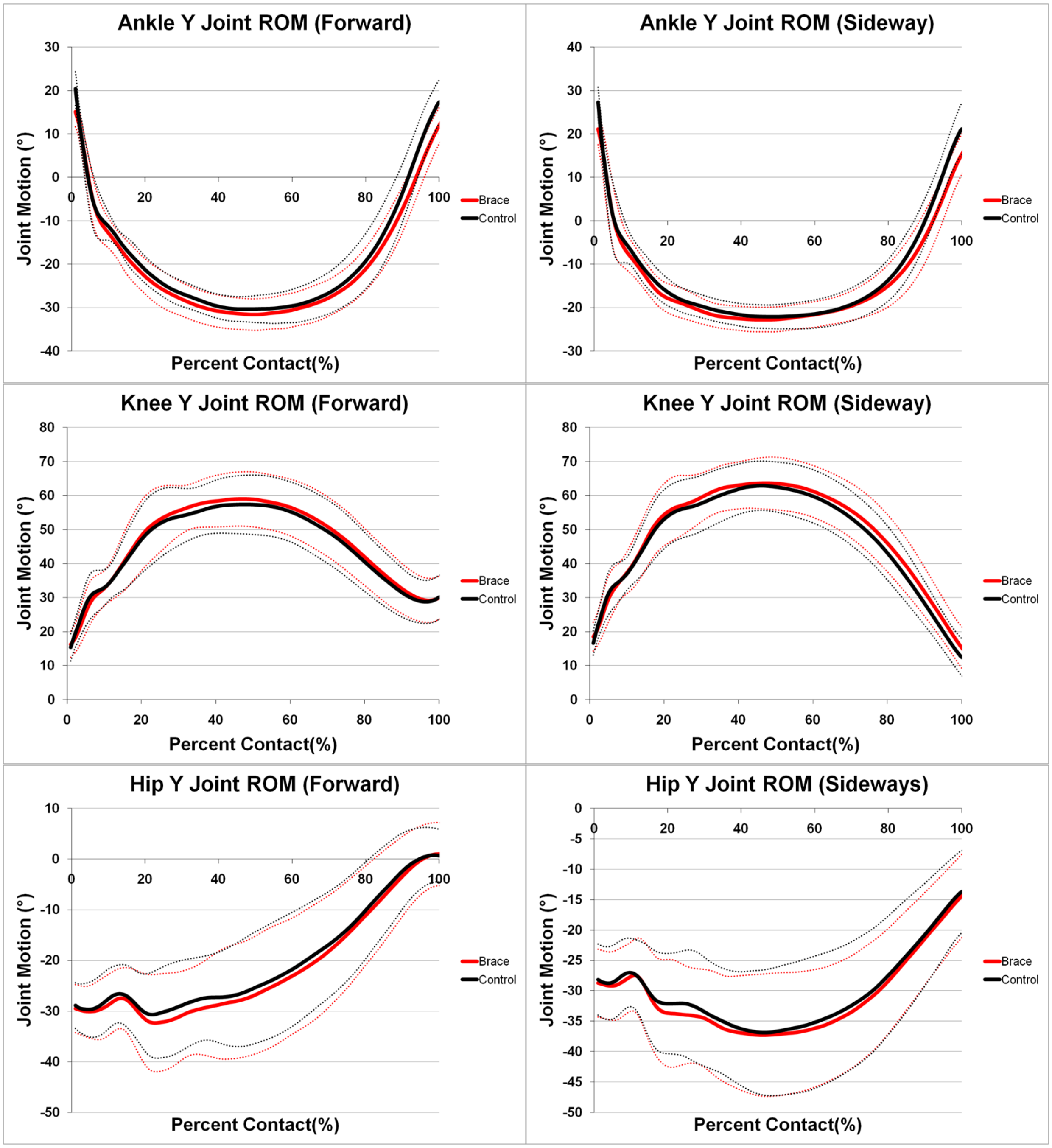

Figure 3.4 Overall Sagittal Joint Range of Motion:
Joint angles are presented relative to angles during quiet standing and normalized to 100% of contact time. For the ankle, knee and hip positive values represent dorsi flexion, flexion and extension respectively. Note: These graphs are overall means of all subjects.

Table 3.5 Overall Sagittal Joint Range of Motion:

Overall joint ROM corresponds to the difference in joint angle between initial contact and maximal displacement observed during contact. Values are presented as the absolute change in joint ROM with larger values corresponding to a larger change in joint ROM.

ROM^{ank}	Brace	Mean (±S.D.)	*t*	*p*
Forward	ASO	44.23° (2.30)	-4.070	0.005
	Control	48.61° (4.38)		
Sideways	ASO	40.77° (2.71)	-5.775	0.001
	Control	46.78° (3.74)		
ROM^{knee}				
Forward	ASO	43.57° (6.00)	0.681	0.517
	Control	42.88° (6.87)		
Sideways	ASO	45.82° (5.16)	-0.771	0.466
	Control	46.61° (5.67)		
ROM^{hip}				
Forward	ASO	5.89° (4.45)	1.119	0.300
	Control	4.78° (2.82)		
Sideways	ASO	10.55° (5.73)	0.588	0.575
	Control	10.12° (5.78)		

3.3.2 Angles at Impact

In examining the effects of an ankle brace on ankle, knee and hip kinematics, significant differences were observed at the time of initial foot contact (i.e. 0% - 5% of contact). The largest differences observed in the joint kinematics were seen in the ankle and knee with no differences observed for the hip angles.

3.3.2.1 Ankle Angles Impact

The brace significantly decreased plantar flexion angle (AAy^{imp}) and resulted in a significantly increased external rotation angle (AAz^{imp}) at impact while displaying a trend for increased ankle inversion angle (AAx^{imp}). (Refer to Appendix D; Figure D.1 for complete ankle joint angle figures). For the forward and sideways maneuvers, ankle inversion increased by approximately 3.32° and 2.94° respectively with the ankle brace. The ankle also displayed a significant decrease in ankle plantar flexion of 5.34° and 6.19° for the forward and sideways movement respectively. The use of ankle ASO brace changed the ankle's transverse axis position from an internally rotated position at impact to an externally rotated position, with this result being significantly different for both the forward and sideways maneuvers. (Table 3.6)

Table 3.6 Ankle Angles Impact:
Ankle angles at initial ground contact (i.e. 0% - 5% of contact). Angles are expressed relative to angles during quiet standing. AAx is inversion/eversion with positive values corresponding to eversion. AAy is dorsi flexion/plantar flexion with positive values corresponding to dorsi flexion. AAz is internal/external rotation with positive values corresponding to external rotation.

AAx^{imp}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	-7.62° (5.78)	-3.885	0.006	0.632
	Control	-4.30° (4.68)			
Sideways	ASO	-8.09° (6.66)	-4.702	0.002	0.474
	Control	-5.15° (5.71)			
AAy^{imp}					
Forward	ASO	15.08° (3.41)	-12.76	0.000	1.455
	Control	20.42° (3.90)			
Sideways	ASO	21.13° (3.58)	-8.805	0.000	1.755
	Control	27.32° (3.47)			
AAz^{imp}					
Forward	ASO	3.71° (1.95)	9.331	0.000	2.143
	Control	-0.93° (2.36)			
Sideways	ASO	4.12° (2.79)	6.204	0.000	1.801
	Control	-1.36° (3.28)			

3.3.2.2 Knee Angles Impact

There were less kinematic differences with the brace at the knee joint at impact than were observed at the ankle joint (Refer to Appendix D; Figure D.2 for complete knee joint angle figures). During the forward maneuver the braced condition participants displayed a trend for increased internal rotation (KAz^{imp}) at impact. During the sideways jumping maneuver participants displayed a trend for increased knee joint flexion (KAy^{imp}) and abduction (KAx^{imp}).

Table 3.7 Knee Angles Impact:
Knee angles at initial ground contact (i.e. 0% - 5% of contact). Angles are expressed relative to angles during quiet standing. KAx is abduction/adduction with positive values corresponding to abduction. KAy is flexion/extension with positive values corresponding to flexion. KAz is internal/external rotation with positive values corresponding to external rotation.

KAx^{imp}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	2.02° (2.05)	-2.082	0.076	0.357
	Control	2.71° (1.79)			
Sideways	ASO	1.48° (1.69)	-3.168	0.016	0.633
	Control	2.48° (1.48)			
KAy^{imp}					
Forward	ASO	16.31° (3.84)	1.159	0.285	0.237
	Control	15.37° (4.02)			
Sideways	ASO	18.54° (4.26)	2.949	0.021	0.478
	Control	16.66° (3.58)			
KAz^{imp}					
Forward	ASO	-1.24° (4.41)	-2.923	0.022	0.413
	Control	0.51° (4.04)			
Sideways	ASO	-4.59° (4.80)	-1.843	0.108	0.423
	Control	-2.34° (5.81)			

3.3.2.3 Hip Angles Impact

Use of ankle bracing had no kinematic effect on the hip abduction/adduction (HAx^{imp}) flexion/extension (HAy^{imp}) or internal/external rotation (HAz^{imp}) joint angles during impact (Table 3.8). Refer to Appendix D; Figure D.3 for complete hip joint angle figures.

Table 3.8 Hip Angles Impact:
Hip angles at initial ground contact (i.e. 0% - 5% of contact). Angles are expressed relative to angles during quiet standing. HAx is abduction/adduction with positive values corresponding to abduction. HAy is extension/flexion with positive values corresponding to extension. HAz is internal/external rotation with positive values corresponding to external rotation.

HAx^{imp}	Brace	Mean (±S.D.)	*t*	*p*	ES
Forward	ASO	-2.32° (4.55)	0.800	0.450	0.212
	Control	-3.15° (3.15)			
Sideways	ASO	4.74° (3.43)	-0.680	0.518	0.229
	Control	5.48° (3.04)			
HAy^{imp}					
Forward	ASO	-29.43° (4.79)	-0.748	0.479	0.136
	Control	-28.78° (4.70)			
Sideways	ASO	-28.49° (5.56)	-0.480	0.646	0.079
	Control	-28.03° (5.87)			
HAz^{imp}					
Forward	ASO	5.87° (5.81)	-1.789	0.117	0.215
	Control	7.17° (6.31)			
Sideways	ASO	5.35° (6.29)	-1.487	0.181	0.225
	Control	6.97° (6.89)			

3.3.3 Angles at Max Propulsive Force

In addition to the observations occurring during impact, a second time variable was chosen as an indication of the participant generating the force that would propel them in the desired direction through the second half of contact. This phase was defined as the maximal propulsive phase and corresponded to the percentage of contact time which maximal anterior force (forward maneuver) and maximal medial force (sideways maneuver) were achieved. This phase occurs at approximately 75% of stance (see Tables 3.4 and 3.5). In examining the effects of an ankle brace on joint angles there were only small trends observed at the ankle.

3.3.3.1 Ankle Angles at Max Propulsive Force

Two ankle angle variables, inversion and external rotation, show the largest trend for a difference to occur between bracing conditions during the maximal propulsive phase of movement. Participants displayed a trend for increased ankle inversion (AAx^{prop}) during the braced condition for both the forward and sideways manoeuvre. For the sideways manoeuvre the braced condition displayed a trend for increased external rotation (AAz^{prop}). No difference was observed for ankle plantar/dorsi flexion (AAy^{prop}) for either maneuver. (Tables 3.9)

Table 3.9 Ankle Angles Max Propulsive:
Ankle angles at the timing of maximal propulsion (i.e. during $GRFx^{min}$ or $GRFy^{min}$). Angles are expressed relative to angles during quiet standing. AAx is inversion/eversion with positive values corresponding to eversion. AAy is dorsi flexion/plantar flexion with positive values corresponding to dorsi flexion. AAz is internal/external rotation with positive values corresponding to external rotation.

AAx^{prop}	Brace	Mean (±S.D.)	*t*	*p*	ES
Forward	ASO	-1.00° (5.01)	-4.431	0.004	1.167
	Control	4.58° (4.53)			
Sideways	ASO	-10.07° (4.24)	-2.688	0.031	0.945
	Control	-6.76° (2.57)			
AAy^{prop}					
Forward	ASO	-22.61° (5.58)	0.026	0.980	0.006
	Control	-22.64° (5.32)			
Sideways	ASO	-17.82° (3.24)	0.048	0.963	0.014
	Control	-17.87° (3.39)			
AAz^{prop}					
Forward	ASO	8.30° (2.04)	-0.356	0.732	0.136
	Control	8.63° (2.85)			
Sideways	ASO	4.73° (3.43)	3.954	0.006	0.352
	Control	3.54° (3.28)			

3.3.3.2 Knee Angles at Max Propulsive Force

Use of ankle bracing had no effect on the knee abduction/adduction (KAx^{prop}) flexion/extension (KAy^{prop}) or internal/external rotation (KAz^{prop}) joint angles during timing of maximal propulsive phase. (Table 3.10)

Table 3.10 Knee Angles Max Propulsive:
Knee angles at the timing of maximal propulsion (i.e. during $GRFx^{min}$ or $GRFy^{min}$). Angles are expressed relative to angles during quiet standing. KAx is abduction/adduction with positive values corresponding to abduction. KAy is flexion/extension with positive values corresponding to flexion. KAz is internal/external rotation with positive values corresponding to external rotation.

KAx^{prop}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	-4.70° (2.65)	1.624	0.148	0.448
	Control	-5.77° (2.08)			
Sideways	ASO	-7.40° (3.03)	0.875	0.410	0.118
	Control	-7.79° (3.51)			
KAy^{prop}					
Forward	ASO	43.24° (5.52)	-0.411	0.693	0.094
	Control	43.74° (4.97)			
Sideways	ASO	50.81° (4.84)	0.504	0.630	0.136
	Control	50.17° (4.54)			
KAz^{prop}					
Forward	ASO	-11.98° (4.01)	-0.507	0.628	0.036
	Control	-11.84° (3.73)			
Sideways	ASO	-16.34° (3.97)	0.468	0.654	0.041
	Control	-16.50° (3.79)			

3.3.3.3 Hip Angles at Max Propulsive Force

Use of ankle bracing had no effect on the hip abduction/adduction (HAx^{prop}) flexion/extension (HAy^{prop}) or internal/external rotation (HAz^{prop}) joint angles during the maximal propulsive phase. (Table 3.11)

Table 3.11 Hip Angles Max Propulsive:
Hip angles at the timing of maximal propulsion (i.e. during $GRFx^{min}$ or $GRFy^{min}$). Angles are expressed relative to angles during quiet standing. HAx is abduction/adduction with positive values corresponding to abduction. HAy is extension/flexion with positive values corresponding to extension. HAz is internal/external rotation with positive values corresponding to external rotation.

HAx^{prop}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	-3.39° (5.04)	-0.116	0.911	0.017
	Control	-3.29° (6.64)			
Sideways	ASO	8.09° (3.50)	-2.255	0.059	0.623
	Control	10.52° (4.25)			
HAy^{prop}					
Forward	ASO	-12.03° (8.34)	1.306	0.233	0.124
	Control	-12.95° (6.56)			
Sideways	ASO	-30.63° (7.42)	0.157	0.880	0.026
	Control	-30.83° (8.64)			
HAz^{prop}					
Forward	ASO	3.19° (5.23)	-0.704	0.504	0.132
	Control	3.86° (4.86)			
Sideways	ASO	10.40° (4.66)	0.058	0.955	0.015
	Control	10.32° (5.87)			

3.4 Joint Moments

3.4.1. Moments at Impact

In examining the effects of an ankle brace on joint moments there were no significant differences observed at the ankle, knee and hip. There were trends observed between the brace condition with the largest trends were observed to occur around the ankle joint with fewer trends observed with the knee and hip moments. Figures are presented in the axis in which the largest trends were observed between bracing conditions.

3.4.1.1. Ankle Moments at Impact

Increased ankle joint moments were observed across all three planes of motion during the ankle brace condition with participants displaying a trend for increased ankle eversion moment (AMx^{imp}) and an external rotation moment (AMz^{imp}) for the sideways maneuver. During the forward maneuver the participants also displayed trend for increased external rotation moments (AMz^{imp}) in addition to a trend for increased plantar flexor moment (AMy^{imp}) under the braced condition. (Table 3.12) Refer to Appendix D; Figure D.4 for complete ankle joint moment figures.

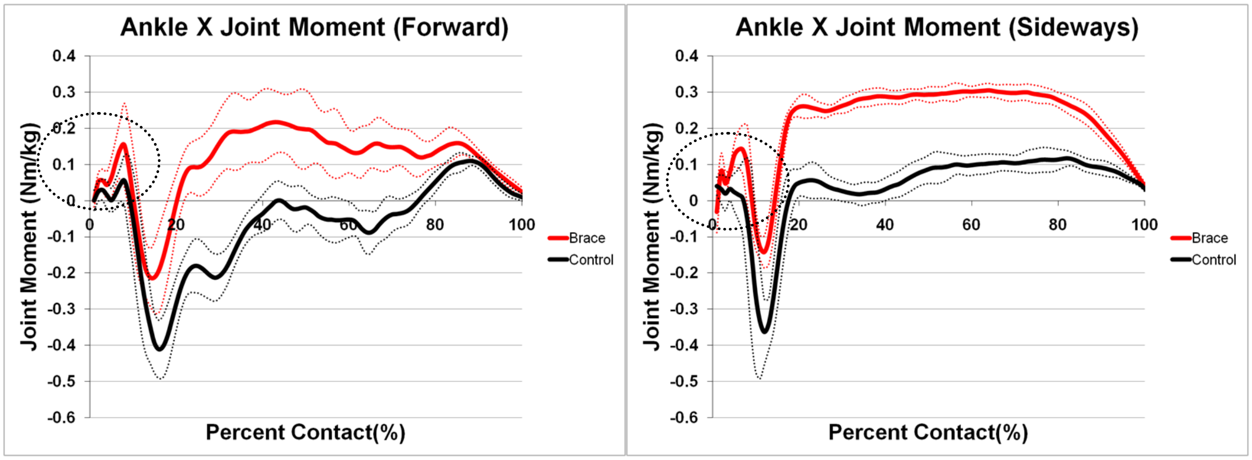

Figure 3.5 Frontal Plane Ankle Moments (AMx^{imp})
Ankle moments for the frontal plane of motion normalized to 100% of contact time Data are representative means (SD) for Subject 1 (Forward) and Subject 2 (Sideways
Joint moments are presented as Newton meters per kilogram of body mass (N·m/kg). Ankle moment peaks identified at impact are highlighted (AMx^{imp}). Positive values correspond to eversion moments.

Table 3.12 Ankle Moments Impact:
Ankle moment at initial ground contact (i.e. 0% - 5% of contact). Moments are presented relative to body mass. AMx is inversion/eversion with positive values corresponding to eversion. AMy is plantar/dorsi flexion with positive values corresponding to plantar flexion. AMz is internal/external rotation with positive values corresponding to external rotation.

AMx^{imp}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	0.07 Nm/kg (0.04)	2.270	0.058	0.674
	Control	0.05 Nm/kg (0.02)			
Sideways	ASO	0.06 Nm/kg (0.04)	4.506	0.003	0.752
	Control	0.04 Nm/kg (0.03)			
AMy^{imp}					
Forward	ASO	0.42 Nm/kg (0.18)	3.916	0.006	0.380
	Control	0.36 Nm/kg (0.16)			
Sideways	ASO	0.55 Nm/kg (0.26)	0.488	0.641	0.103
	Control	0.52 Nm/kg (0.31)			
AMz^{imp}					
Forward	ASO	0.25 N/kg (0.19)	3.013	0.020	0.590
	Control	0.15 N/kg (0.12)			
Sideways	ASO	0.43 N/kg (0.27)	3.845	0.006	0.554
	Control	0.30 N/kg (0.19)			

3.4.1.2. Knee Moments at Impact

The main difference observed between bracing conditions for knee joint moments occurred in the sagittal plane (KMy^{imp}). Figure 3.6 outlines the joint moments observed at the knee in the sagittal plane highlighting the difference in flexion moment at impact. For both maneuvers, there were trends for increased knee extensor moment at impact (KMy^{imp})with a brace. During the forward maneuver bracing displayed a trend for increasing the participant's knee abduction moment (KMx^{imp}) (Table 3.13). No consistent peaks were identified at impact for ankle internal/external rotation moments. Refer to Appendix D; Figure D.5 for complete knee joint moment figures.

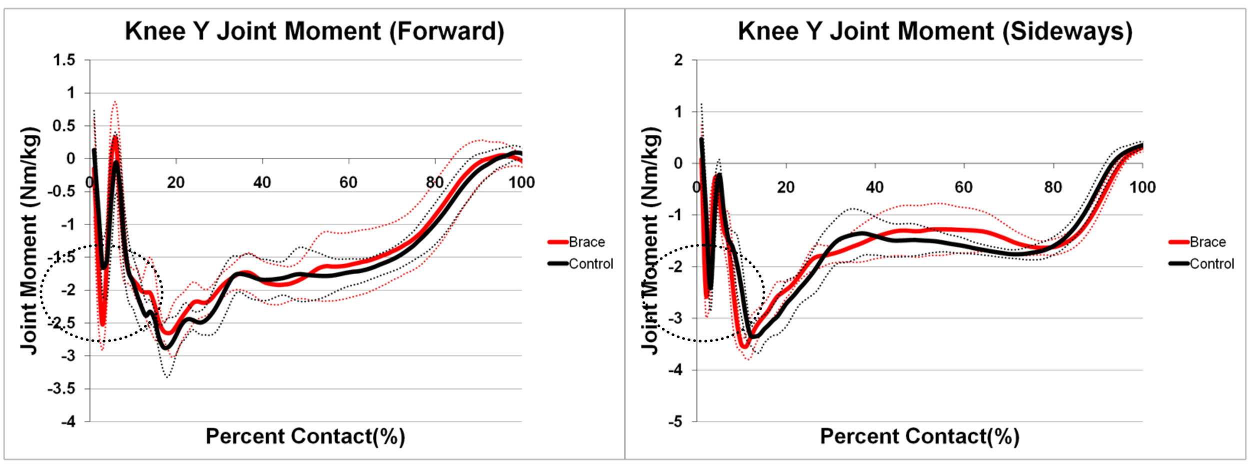

Figure 3.6 Sagittal Plane Knee Moments (KMy^{imp})
Flexion/extension knee moments normalized to100 % of contact. Data are representative means (SD) for Subject 5 (Forward) and Subject 7 (Sideways). Joint moments are presented as Newton meters per kilogram of body mass (N·m/kg). Knee moments at impact are highlighted (KMy^{imp}) with a negative values corresponding to an extensor moments.

Table 3.13Knee Moments Impact:
Knee moments at initial ground contact (i.e. 0% - 5% of contact). Moments are expressed relative body mass. KMx is abduction/adduction with positive values corresponding to abduction. KMy is flexion/extension with positive values corresponding to flexion. KMz is internal/external rotation with positive values corresponding to external rotation.

KMx^{imp}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	0.56 Nm/kg (0.18)	2.444	0.045	0.722
	Control	0.42 Nm/kg (0.21)			
Sideways	ASO	-0.28 Nm/kg (0.18)	-0.046	0.964	0.0185
	Control	-0.27 Nm/kg (0.07)			
KMy^{imp}					
Forward	ASO	-2.54 Nm/kg (0.50)	-4.423	0.003	0.895
	Control	-2.06 Nm/kg (0.57)			
Sideways	ASO	-2.65 Nm/kg (0.63)	-2.926	0.022	0.522
	Control	-2.33 Nm/kg (0.58)			

3.4.1.3. Hip Moments at Impact

The hip joint only displayed a trend for the joint moments occurring in the sagittal plane (HMy^{imp}) during the braced condition at impact. Figure 3.7 outlines the joint moments observed at the hip in the sagittal plane, highlighting the difference in flexion moment at impact. A trend was observed for increased flexion (HMy^{imp}) during the forward maneuver. The difference observed was not significant during the sideways manuver. There were no significant differences observed for internal/external rotation hip joint moments (HMz^{imp}) occurring for either maneuver (Table 3.14). There were no consistent frontal plane hip moment peaks at impact. Refer to Appendix D; Figure D.6 for complete hip joint moment figures.

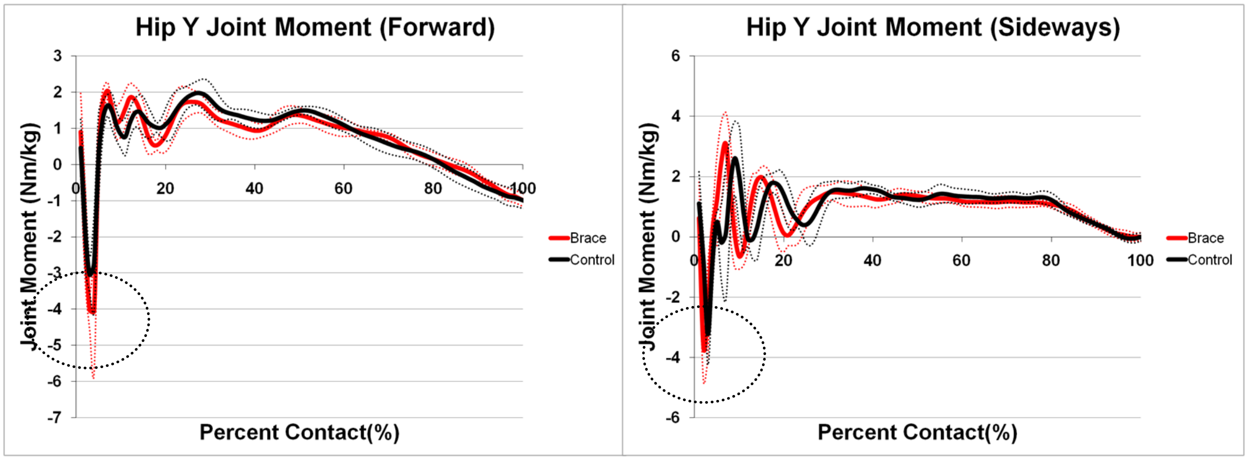

Figure 3.7 Sagittal Plane Hip Moments (HMy^{imp})
Flexion/extension hip moments normalized to 100 % of contact. Data are representative means (SD) for Subject 6 (Forward) and Subject 3 (Sideways). Joint moments are presented as Newton meters per kilogram of body mass (N·m/kg). Hip moments at impact are highlighted (HMy^{imp}) with a negative values corresponding to flexor moments.

Table 3.14 Hip Moments Impact:
Hip Moments at initial ground contact (i.e. 0% - 5% of contact). Moments are expressed relative to body mass. HMx is abduction/adduction with positive values corresponding to abduction. HMy is extension/flexion with positive values corresponding to extension. HMz is internal/external rotation with positive values corresponding to external rotation.

HMy^{imp}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	-4.08 Nm/kg (0.77)	2.444	0.003	0.892
	Control	-3.26 Nm/kg (1.06)			
Sideways	ASO	-4.04 Nm/kg (0.92)	-0.046	0.079	0.470
	Control	-3.56 Nm/kg (1.11)			
HMz^{imp}					
Forward	ASO	-0.002 Nm/kg (0.09)	1.507	0.175	0.411
	Control	-0.003 Nm/kg (0.08)			
Sideways	ASO	0.09 Nm/kg (0.10)	1.554	0.164	0.340
	Control	0.06 Nm/kg (0.07)			

3.4.2Moments at Max Propulsive Force

In examining the effects of a brace on ankle, knee and hip joint moments it was observed that ankle bracing had no significant effect during the time of maximal propulsive force. The largest trends observed between conditions were seen at the ankle joint with only the knee displaying a single trend in the sagittal plane moment during the forward maneuver.

3.4.2.1 Ankle Moments at Max Propulsive Force

The strongest trend observed during the propulsive phase was the participants displaying a strong trend for increased ankle eversion joint moment (AMx^{prop}) for both the forward and sideway maneuvers for the braced condition. Figure 3.8 outlines the joint moments observed at the ankle in the frontal plane highlighting the difference in eversion moment during the propulsive phase. The ankle brace also displayed a trend for increased external rotation (AMz^{prop}) moment during the forward jump maneuver exclusively. There were no observed bracing effects on the ankle's flexion/extension joint moments (AMy^{prop}) (Table 3.15). Refer to Appendix D; Figure D.4 for complete ankle joint moment figures.

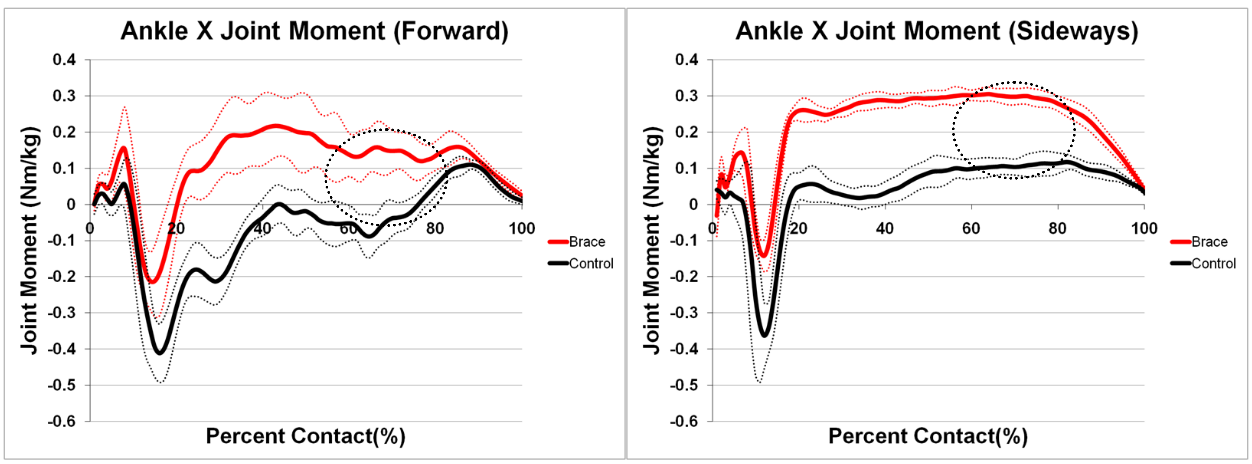

Figure 3.8 Frontal Plane Ankle Moments (AMx^{prop})
Inversion/eversion ankle moments normalized to 100 % of contact. Data are representative means (SD) for Subject 1 (Forward) and Subject 2 (Sideways) Ankle moments at propulsive phase are highlighted (AMx^{prop}) with a positive values corresponding to eversion moments.

Table 3.15 Ankle Moments Max Propulsive:
Ankle moments at the timing of maximal propulsion (i.e. during $GRFx^{min}$ or $GRFy^{min}$). Moments are presented relative to body mass. AMx is inversion/eversion with positive values corresponding to eversion. AMy is plantar/dorsi flexion with positive values corresponding to plantar flexion. AMz is internal/external rotation with positive values corresponding to external rotation.

Ankle X	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	0.05 Nm/kg (0.09)	2.980	0.021	0.983
	Control	-0.08 Nm/kg (0.17)			
Sideways	ASO	0.25 Nm/kg (0.04)	4.332	0.003	1.944
	Control	0.13 Nm/kg (0.07)			
Ankle Y					
Forward	ASO	2.27 Nm/kg (0.44)	-1.455	0.189	0.117
	Control	2.32 Nm/kg (0.43)			
Sideways	ASO	1.93 Nm/kg (0.40)	-1.740	0.125	0.138
	Control	1.99 Nm/kg (0.42)			
Ankle Z					
Forward	ASO	0.41 Nm/kg (0.23)	2.477	0.042	0.812
	Control	0.23 Nm/kg (0.20)			
Sideways	ASO	0.63 Nm/kg (0.27)	1.802	0.115	0.392
	Control	0.54 Nm/kg (0.16)			

3.4.2.2 Knee Moments at Max Propulsive Force

The application of an ankle brace only had a small effect on knee joint moments during the propulsive phase. Figure 3.9 outlines the joint moments observed at the knee in the frontal plane highlighting a trend for the braced condition to displayed a decrease in knee abduction moment (KMx^{prop}) for the forward maneuver exclusively. No differences were observed in knee flexion/extension (KMy^{prop}) (Table 3.16). Refer to Appendix D; Figure D.5 for complete knee joint moment figures.

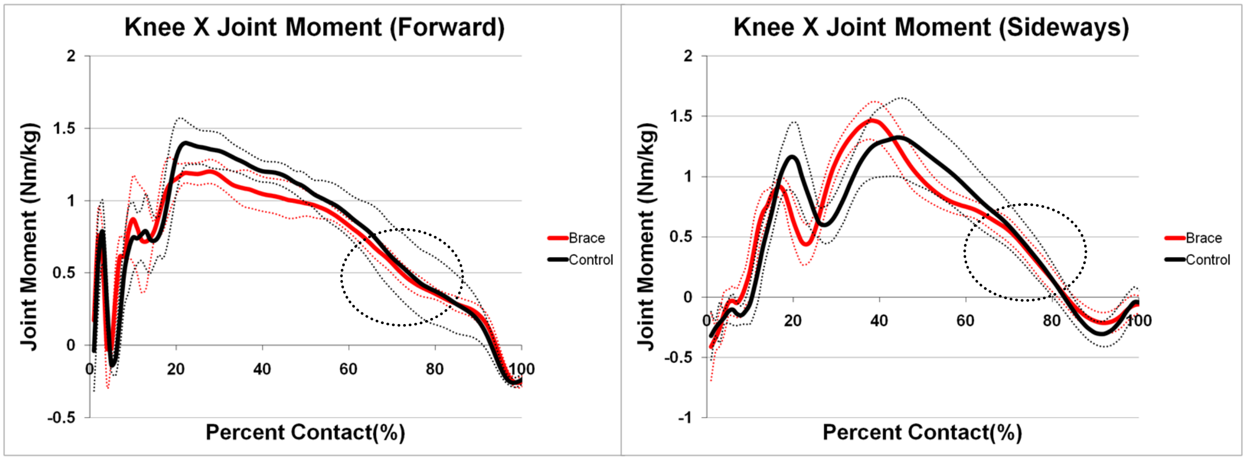

Figure 3.9 Frontal Plane Knee Moments (KMx^{prop})
Knee moments are displayed for the frontal plane of motion over 100 % of contact. Data are representative means (SD) for Subject 6 (Forward) and Subject 6 (Sideways). Joint moments are presented as Newton meters per kilogram (N·m/kg). Knee moments at impact are highlighted (KMx^{prop}) with a positive moment corresponding to an adduction moment.

Table 3.16 Knee Moments Max Propulsive:
Knee moments at the timing of maximal propulsion (i.e. during GRFx^{min} or GRFy^{min}). Moments are expressed relative body mass. KMx is abduction/adduction with positive values corresponding to abduction. KMy is flexion/extension with positive values corresponding to flexion.

KMx^{prop}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	0.56 Nm/kg (0.22)	-2.918	0.022	0.430
	Control	0.68 Nm/kg (0.31)			
Sideways	ASO	0.55 Nm/kg (0.24)	-0.574	0.584	0.087
	Control	0.58 Nm/kg (0.35)			
KMy^{prop}					
Forward	ASO	-0.94 Nm/kg (0.38)	0.904	0.396	0.236
	Control	-1.04 Nm/kg (0.47)			
Sideways	ASO	-1.68 Nm/kg (0.41)	-0.094	0.928	0.013
	Control	-1.68 Nm/kg (0.44)			

3.4.2.3 Hip Moments at Max Propulsive Force

Use of ankle bracing had no effect on the hip abduction/adduction (HMx^{prop}) flexion/extension (HMy^{prop}) or internal/external rotation (HMz^{prop}) joint moment during timing of maximal propulsive phase (Table 3.17). Refer to Appendix D; Figure D.6 for complete hip joint moment figures.

Table 3.17 Hip Moments Max Propulsive:
Hip moments at the timing of maximal propulsion (i.e. during $GRFx^{min}$ or $GRFy^{min}$).Moments are expressed relative to body mass. HMx is abduction/adduction with positive values corresponding to abduction. HMy is extension/flexion with positive values corresponding to extension. HMz is internal/external rotation with positive values corresponding to external rotation.

Hip X^{prop}	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	1.27 Nm/kg (0.46)	-0.777	0.462	0.140
	Control	1.35 Nm/kg (0.57)			
Sideways	ASO	0.62 Nm/kg (0.24)	-0.175	0.866	0.055
	Control	0.64 Nm/kg (0.27)			
Hip Y^{prop}					
Forward	ASO	0.29 Nm/kg (0.44)	-0.255	0.806	0.054
	Control	0.31 Nm/kg (0.42)			
Sideways	ASO	0.57 Nm/kg (0.32)	-0.126	0.903	0.026
	Control	0.58 Nm/kg (0.45)			
Hip Z^{prop}					
Forward	ASO	-0.23 Nm/kg (0.08)	1.884	0.102	0.494
	Control	-0.29 Nm/kg (0.14)			
Sideways	ASO	-0.13 Nm/kg (0.09)	0.779	0.462	0.163
	Control	-0.15 Nm/kg (0.13)			

3.4.3 Peak Moments not occurring at Impact or Max Propulsive Force

In examining the effects of an ankle brace, there were trends observed between the time points of impact and maximal propulsive force. These trends were observed as peak overall moments and acted on the ankle and knee during the forward manoeuvre exclusively. Figure 3.10 outlines the peak joint moments observed at the ankle in the frontal plane highlighting the ankle brace condition displaying a trend for decreased maximal ankle inversion, and a trend for larger ankle eversion moment throughout contact. Only a small difference was observed post impact during the forward jump condition with the brace condition displaying a decrease in knee abduction moment. The knee abduction moment difference can be observed in Figure 3.9 occurring at approximately 40% of contact for the forward movement and at 20% of contact for the sideways movement. (Table 3.18)

Table 3.18 Peak Moments Post Impact/ Pre Propulsive:

Peak moments occurring for the knee and hip joints. Moments are expressed relative to body mass. AMx^{inv} is inversion/eversion with negative values corresponding to an inversion moment. KMx^{abd} is abduction/ adduction with positive values corresponding to an abductor moment.

AMx^{inv}	Brace	Mean (±S.D.)	*t*	*p*	ES
Forward	ASO	-0.47 Nm/kg (0.21)	2.643	0.033	0.439
	Control	-0.56 Nm/kg (0.17)			
Sideways	ASO	-0.27 Nm/kg (0.20)	1.915	0.097	0.406
	Control	-0.35 Nm/kg (0.17)			
KMx^{abd}					
Forward	ASO	1.37 Nm/kg (0.36)	-2.278	0.057	0.418
	Control	1.50 Nm/kg (0.24)			
Sideways	ASO	1.31 Nm/kg (0.34)	0.415	0.619	0.048
	Control	1.29 Nm/kg (0.30)			

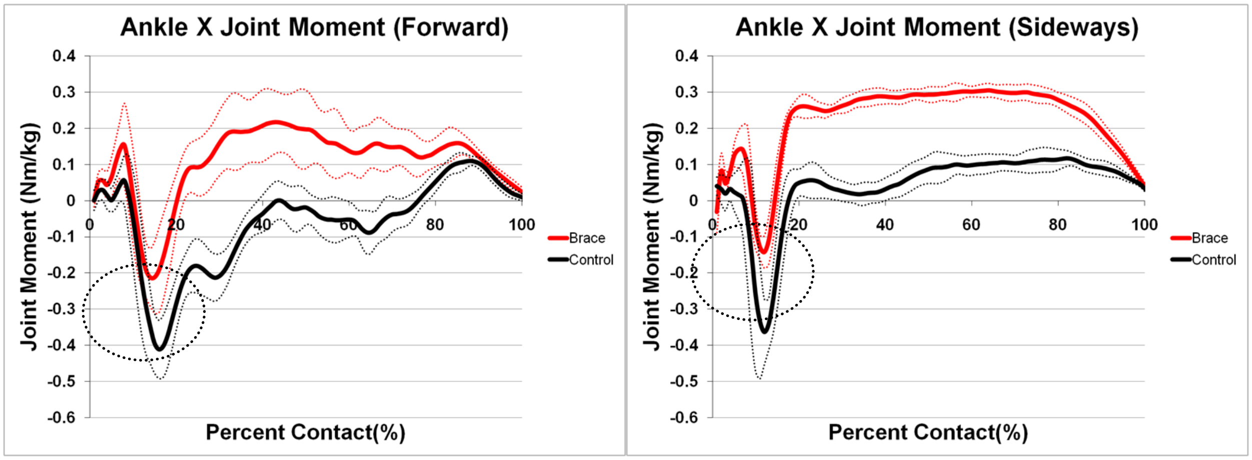

Figure 3.10 Frontal Plane Ankle Moments Post Impact (AMx^{inv})
Inversion/eversion ankle moments normalized to 100 % of contact. Data are representative means (SD) for Subject 1 (forward) and Subject 2 (Sideways). Joint moments are presented as Newton meters per kilogram of body weight (N·m/kg). Ankle inversion moments post impact are highlighted (AMx^{inv}) with a negative values corresponding to inversion moments.

3.5 Stiffness

3.5.1 Ankle and Knee Stiffness

The ASO ankle brace displayed a strong trend for increased sagittal plane stiffness at the ankle joint. This increase in stiffness was apparent for both movements at impact as well as averaged across the early contact phase from impact until the timing of maximal vertical ground reaction force. The brace application did not result in any difference observed in sagittal knee joint stiffness. (Tables 3.19 and 3.20)

Table 3.19 Ankle Joint Stiffness in the Sagittal Plane:
Ankle joint stiffness values are given for impact (Ankle Imp.) and averaged across stance until timing of maximal vertical force (Ankle Average). Negative values indicates resistance to ankle dorsi flexion

Ankle Imp.	Brace	Mean (±S.D.)	*t*	*p*	ES
Forward	ASO	-4.85 N·m/° (1.44)	-3.228	0.014	0.390
	Control	-4.28 N·m/° (1.32)			
Sideways	ASO	-4.91 N·m/° (1.05)	-3.270	0.014	0.549
	Control	-4.39 N·m/° (0.83)			
Ankle Average.					
Forward	ASO	-3.55 N·m/° (1.48)	-2.888	0.023	0.244
	Control	-3.22 N·m/° (1.23)			
Sideways	ASO	-4.91 N·m/° (1.06)	-2.654	0.033	0.406
	Control	-3.07 N·m/° (0.79)			

Table 3.20 Knee Joint Stiffness in the Sagittal Plane:
Knee joint stiffness values are given for impact (Knee IMP.) and averaged across stance until timing of maximal vertical force (Knee Average). Negative values indicates resistance to knee flexion

Knee Imp.	**Brace**	**Mean (±S.D.)**	***t***	***p***	**ES**
Forward	ASO	-6.36 N·m/° (0.82)	-0.046	0.965	0.020
	Control	-6.33 N·m/° (1.87)			
Sideways	ASO	-7.50 N·m/° (1.44)	-1.210	0.265	0.475
	Control	-6.84 N·m/° (1.32)			
Knee Average.					
Forward	ASO	-6.03 N·m/° (0.84)	1.105	0.306	0.325
	Control	-6.38 N·m/° (1.27)			
Sideways	ASO	-7.12 N·m/° (1.04)	-0.046	0.965	0.014
	Control	-7.10 N·m/° (1.12)			

Chapter 4

Discussion

The main hypotheses were that ankle bracing would alter ankle joint kinematics, modify the ground reaction forces incurred by changing the normal loading patterns of the ankle, knee and hip as well as change the movement strategy and the net joint moments of leg muscles when compared with a non-stabilized condition. As well it was hypothesized that the ankle brace would increase joint stiffness and subsequently decreases knee joint stiffness to compensate. The multidimensional hypothesis was based on the premise that alterations in ground reaction force profiles, proximal joint loading patterns, joint range of motion and joint moments are specifically caused by the added brace effect.

Overall, with the Bonferroni correction, the only statistically significant differences were observed for the change in ankle joint position at impact. Data collected during the current investigation indicates ankle bracing has a significant ability to alter the sagittal and transverse ankle joint kinematics along with a non-significant trend for altering the frontal plane kinematics. These findings confirm previous ankle brace literature that ankle bracing does restrict normal ankle joint motion. This confirmation of a change in ankle dynamics may provide new insights into how ankle braces affect highly trained athletic female populations.

Within this study the participants displayed trends for increasing both the GRF magnitudes in the vertical and anterior-posterior direction as well as displaying a trend for a decreased rate of time to reach the peak force.

The ankle braced condition displayed trends to modify the knee and hip joint kinematics and joint kinetics. And although not significant these trends were observed primarily at the impact phase.

There was a trend for the ankle brace condition to increase the ankle joint stiffness, both at impact and across contact time, however against our hypothesis there were no differences observed for the knee joint.

Although the primary results occurred at impact the small difference observed prior to take-off during the propulsive phase may provide new insight into how ankle stabilizers may affect athletic performance.

In an attempt to generalize the results observed in the current investigation, a clinical relevance section has been developed to generalize the findings into implications towards enhancing the knowledge base for practitioners and athletes who wish to prescribe or wear ankle braces. The clinical relevance section includes the following: 1) Common injury mechanisms found in female athletes, specifically anterior cruciate ligament (ACL) injuries. This section integrates the observed effect of ankle stabilizing and how ankle stabilizing my interact with ACL mechanisms 2) How the application of an ankle stabilizer may affect performance of athletic maneuvers and 3) How training could be incorporated to offset the changes in joint dynamics caused by the application of an ankle stabilizer.

4.1 Ground Reaction Force

Our results displayed strong trends for changes in ground reaction force profiles during the braced condition for both the forward and sideways maneuvers. Broken down into the three phases of interest, maximal posterior braking force ($GRFy^{max}$), maximal vertical force ($GRFz^{max}$) and maximal propulsive force ($GRFy^{min}$ and $GRFx^{min}$- for the forward and sideways jump maneuver respectively), the application of the ASO ankle brace had the largest effects during the breaking and maximal vertical force phases. The $GRFy^{max}$ and $GRFz^{max}$ correspond to the impact and deceleration phase of landing. For both the forward and sideways maneuvers the braced condition displayed a trend indicating larger anterior-posterior breaking force ($GRFy^{max}$) and larger maximal vertical force ($GRFz^{max}$) than the control condition. These increased breaking and vertical ground reaction forces may contribute to the risk of lower extremity injury, as increased joint loads at the ankle, knee and hip, have been shown to increase joint demands, possibly harming the joint ligaments in the process (Yeow et al., 2009). The increase in ground reaction force observed has been confirmed in a recent study examining bracing effects. With protocols similar our current investigation, Cordova et al. (2010) found that during a single leg drop landing from 0.305 meters, participants wearing a semi-rigid ankle brace developed vertical ground reaction forces of greater magnitude than a control condition. Hodgson et al. (2005) indicated a significant increase in maximal vertical ground reaction force when the participant's ankles were braced during a bilateral leg impact. However their protocol examined impact magnitude from a 0.61 meter height, which was higher than the height currently examined. DiStefano et al. (2008) did not find a brace effect on the magnitude of the vertical ground reaction force generated during double legged landing from 0.30 meter height. This result is also in accordance with Riemann et al. (2002) who did not find a brace effect on vertical ground reaction force magnitude before or after an exercise protocol. While both these two studies did not find a brace effect, both protocol stipulated a two foot landing strategy. This was different from the single foot landing protocol used in the current investigation. The mixed results on ground reaction force magnitude may therefore be a factor of testing protocol, i.e. two legged landing or single legged landing. Additionally, impact height must also be considered with landing protocols. When examining single leg landings versus dual leg landings, dual leg landings may require increased jump height to observe a significant brace effect compared with a single leg impact.

When examining changes in ground reaction forces, the peak magnitude does not represent the entire landing sequence of events. A second variable, time (percentage of contact time) at which the peak force magnitude is produced also provides information about the landing sequence. The time to peak vertical ground reaction force may indicate how rigid the ankle structure is during landing (Cordova et al., 2010). This indication of ankle rigidity is supported by the results of Riemann et al. (2000) who determined that an increase in rigidity will correspond to a decreased time interval to peak force generation. During the current investigation, for both the forward and sideways maneuvers, the maximal vertical ground reaction force displayed a strong trend to occur earlier for the braced condition. Times to peak vertical ground reaction force for the braced condition were found, on average, to occur 2.0 % and 2.14% earlier for the forward and sideways jump maneuvers respectively. Our data is in agreement with previous investigations examining time to peak ground reaction force. In the previous investigations, all report a significant decrease in time interval to peak ground reaction force under stabilized conditions (Cordova et al., 2010; Hodgson et al 2005; Riemann et al., 2002; Sacco et al., 2006). As opposed to peak force magnitude, the decrease in contact time was not dependent to methodology of landing protocol (i.e. dual leg landing or single leg landing). Being methodology independent may indicate that the brace application has larger implications for decreasing time to peak vertical ground reaction force than for increasing the magnitude.

Contrary to our hypothesis, ankle stabilization did not have any effect on ground reaction force generation at the time of maximal propulsive force. Defined as the point in time corresponding to maximal force in the anterior ($GRFy^{min}$) direction for the forward maneuver and generation of maximal force in the medial direction ($GRFx^{min}$) for the sideways maneuver, the maximal propulsive force occurred at approximately 75 % of contact. This phase was important as it corresponded to the instance when the participant generated the force used to propel the body forward or sideways prior to take-off. With an average decrease of 0.02 N/kg and 0.06 N/kg for the forward and sideways maneuvers respectively, the application of ASO ankle brace had no significant effect on the force production prior to take-off. Additionally, no significant differences were observed in the timing of propulsive force generation. This lack of change occurring in the propulsive phase of movement may have implications with respect to performance and as such will be further discussed under the clinical relevance and performance section.

The ground reaction force profiles are important as they relate to the absorption and transmission of energy onto the different tissues comprising the musculoskeletal system of the lower limb (McCaw et al., 1999). Although the dampening characteristics of bone, cartilage and tissue are associated with impact energy absorption, the joint kinematic patterns of the lower limb have a larger influence on the ground reaction force characteristics (Riemann et al., 2002). Generally impact absorption within the limb is considered to occur distally to proximally in sequence with joint motion playing an essential role in reducing impact magnitudes (McCaw et al., 1999). The ankle joint, in addition to being one of the first major joints loaded in the distal to proximal sequence, has been identified as playing a significant role in controlling impact forces at landing (Riemann et al., 2002). There are two theories as to what specifically a brace does to ankle mechanics that can cause an increased magnitude and a decreased interval to peak ground reaction force. First, the ankle joint complex needs to be compliant for the ground reaction forces to be attenuated through the joint at a normal rate. Increasing the rigidity of the ankle joint in the sagittal plane through the use of a brace can cause an increase in the ground reaction force (Cordova et al., 2010). Secondly, a decrease in the time interval required to reach the maximal GRF may be caused by the reduced plantar flexion at impact, with a brace applied, mimicking a floor contact similar to a flat-foot landing strategy (Riemann et al., 2002). Decreasing the range of ankle motion will cause an increase in GRF to occur earlier in contact. Therefore observing the ankle joint kinematic parameters may provide insight into why changes in ground reaction force were different between bracing conditions at impact without an associated change at the timing of maximum propulsive force.

4.2 Joint Kinematics

The largest differences observed in joint kinematics occurred at the ankle joint. For both the forward and sideways maneuver the brace effectively positioned the foot and ankle joint in a different position at impact by significantly reducing the allowable plantar flexion, and increasing the external rotation. As well, the ankle brace displayed a strong trend for the participants to impact with increased ankle joint inversion angle in the frontal plane. These changes are in agreement with our hypothesis. In contrast to our hypothesis that changes in ankle kinematics would relate to proximal joint changes; only small differences were observed in the knee and no alterations were observed in the hip joint kinematics at impact.

4.2.1 Ankle Kinematics

The brace condition exhibited a significantly decreased plantar flexion angle at impact when compared to the non-braced condition. The non-braced condition exhibited a 5.33° and 6.19° increase in plantar flexion at impact for the forward and sideways maneuvers respectively. This is in agreement with recent studies which have concluded that, although the primary effect of an ankle brace is to prevent ankle inversion sprains by increasing mechanical support in the frontal plane (Cordova et al., 2002; Santos et al. 2004; Thonnard et al., 1996) there is also a restrictive effect on the dorsi and plantar flexion range of motion as well (Cordova et al., 2010; DiStefano et al., 2008; McCaw et al., 1999; Wright et al., 2000).

Reduced ankle plantar flexion due to the application of an ankle brace is also demonstrated by the decrease in overall joint range of motion available during the entire landing sequence until maximal joint displacement was observed. Sagittal plane ankle motion is one of the primary mechanisms in which individuals absorb and dissipate ground reaction forces when landing (DiStefano et al., 2008; Stoffel et al., 2010), therefore decreasing the initial sagittal joint angle as well as decreasing the overall available ankle range of motion may be the cause of the observed increase in maximal vertical ground reaction forces and larger breaking impact forces generated during the braced condition.

Recent literature has suggested that depending on the ankle stabilization used, ankle kinematics will differ according to the stabilizer's rigidity. Currently it is thought that stiffer tape applications and rigid plastic braces will restrict ankle motion to a larger degree than lace up style braces (Cordova et al., 2010; McCaw et al., 1999). While the stiffer stabilization applications may produce larger restriction in ankle range of motion, our results are in accordance with those of Cordova et al., (2010); DiStefano et al. (2008) and McCaw et al. (1999), which previously reported lace up ankle braces restricting plantar flexion 2° - 8.9° prior to or immediately at impact during drop landing type maneuvers compared to a control condition. The ASO ankle brace used in this study is therefore in the range of what has been examined and can be subsequently compared for kinematic variables. The decrease in ankle plantar flexion at impact has been previously described as a protective mechanism for reducing stretch on anterior talo-fibular ligament (ATFL), one of the primary sprain ligaments, and may increase the required torque for the ankle to supinate, which is one of the primary injury mechanism observed with ankles (Wright et al., 2000).

In addition to the reduced plantar flexed position, the results from the current investigation indicate that the ASO brace also decreased ankle motion in the transverse plane at impact. Increase in internal rotation is also a known mechanism for lateral ankle sprains (Eils et al., 2002), and the ability of the ASO ankle brace to limit internal rotation may offer extended protection to the ankle during impact. Increased external ankle rotation further protects the ATFL by decreasing the ability of the tibia to externally rotate on an inverted foot and reduce the stress placed on the ligament (Wilkerson, 2002). Reduced plantar flexion and increased external rotation with the ASO ankle brace offers two mechanisms for enhancing protection against lateral ankle ligament sprains.

One kinematic variable that is not easily explained but was consistently observed was the increased ankle inversion angle observed throughout both jumping maneuvers. This observation was in disagreement with previous research examining ankle brace effects on ankle kinematics. Within the current body of ankle brace literature it has almost unanimously held that ankle braces restrict ankle inversion (Meana, et al., 2008; Gudibanda et al., 2005; Verhagen et al., 2001), as a decrease in ankle inversion ROM has been suspected to be the primary method in which ankle braces protect the lateral ankle joint ligaments (Eils et al., 2002; Eils, Imberge, Völker, & Rosenbaum, 2007; Simpson et al., 1999). For the current study both of the jumping maneuvers displayed increases in ankle inversion during the braced condition with the sideways maneuver displaying larger inversion ankle motion at both time points of impact and maximum propulsive force generation. Only two previous studies have observed similar changes in ankle inversion, with Zhang et al. (2009) reporting that an ASO ankle brace did not effectively reduce peak contact inversion angle and Simpson et al. (2009) indicating one brace (not an ASO) displaying increased peak inversion angle, both during lateral cutting maneuvers. The increase in ankle inversion may indicate a preference for a specific range of kinematic movement at the ankle during landing. Increasing inversion at the ankle, specifically in the sideways jump, will allow for the body center of mass to be shifted toward the direction of movement, perhaps allowing for a benefit in movement quality. In other words the brace may have allowed the participant to increase ankle inversion during the movement over and above what they would do in an unsupported situation. This increased inversion may therefore have been a preferred joint control strategy used by the participants to complete the movement protocols with the greatest success. The application of the ankle brace may have supported the ankle where the participant determined the brace rigidity would allow for a safe increase in ankle inversion to complete the movement protocol.

The increase in inversion angle may also indicate that the ASO ankle braces are less effective in preventing inversion injury. By allowing for a larger joint ROM in the frontal plane the ankle brace would not reduce strain on the lateral ankle joint ligaments (Eils et al., 2002; Zhang, et al., 2009). However, based on the kinematic analysis of the participant's foot position it is unclear if the observed inversion increase will relate directly to increase sprain potential. As the foot was observed to be flat on the floor and did not display medial rollover in relation to the ankle joint during the two maneuvers, the increase in ankle inversion cannot be directly related to an injury stimulus (i.e. slant board landing) observed in previous examinations. As this outcome was highly unexpected, further examination is necessary to determine the true mechanism and implication of the increase in inversion angle.

4.2.2 Knee and Hip Kinematics

It was hypothesized that ankle bracing would change the kinematics at the knee and hip to compensate for the kinematic changes at the ankle joint. Results indicate that during impact, for the brace condition the participants may have modified (not significantly) the knee and hip kinematics but not to the same degree of change exhibited in the ankle. During impact the only trend for a difference in sagittal plane knee kinematics occurred during the sideways jump maneuver, with the ankle brace condition demonstrating 1.88 ° greater knee flexion than the control. The forward maneuver showed a trend of increased knee flexion during impact but again it was not significant. Utilizing the distal to proximal theory for absorbing ground reaction forces, it was hypothesized the decrease in ankle sagittal plane kinematics would lead to an increased sagittal knee range of motion both at impact and overall. However this theory was not found to be true as the overall knee range of motion was not altered during the time of impact until the maximal displacement had been observed.

Landing from a jump with a more extended knee angle has been hypothesized to increase the risk of knee injury at impact (Cordova, et al., 2010; Fagenbaum et al., 2003). Specifically, by decreasing knee flexion, landing in a more upright contact decreases the ability of the hamstring musculature to prevent the quadriceps from increasing the anterior pull on the tibia (Yu et al., 2007). Increased quadriceps force, with insufficient hamstring co-contraction will increase strain on the ACL and increasing the risk of knee injury (Fagenbaum et al., 2003). An increased knee flexion during the sideways jump at impact with the application of an ankle brace may enhance the landing position for the applicant, possibly decreasing one risk factor associated with knee ligament injury. This result is in accordance with DiStefano et al. (2008) and Stoffell et al. (2010) who reported increased knee flexion angles at impact while participants were subjected to ankle bracing. Our observations in the sagittal plane may indicate that, although our results were not significantly different, the trends of increased knee flexion with brace application may promote a safer landing strategy at impact.

During impact the hip must also be considered due to its large range of motion in the sagittal plane. Our results indicate that in the sagittal plane the hip joint demonstrated the largest degree of flexion ROM for any of the leg joints, but did not differ significantly between braced conditions for either the forward or sideways maneuver at impact. Additionally the hip also did not differ in the overall range of joint motion. These findings are in agreement to those reported by Cordova (2010) who found no significant difference in hip ROM during a single legged jump under any braced condition. This lack of change at the hip may be attributed to the impact load having already been attenuated by the ankle and knee joints. In addition it is also conceivable that the hip joint ROM remains relatively unchanged between conditions based on a lower contribution of the hip joint to the total lower extremity force absorption during a drop landing (Cordova, 2010). Because there was an observed increase in ground reaction force with the braced condition with no change in sagittal hip ROM at impact, it could be argued that the trend for increased in knee flexion did not fully compensate for the alterations in ankle motion disruption. Consequently, increased knee flexion at impact may not have been able to fully compensate for the brace effect at impact without subsequent increases in hip flexion at impact or increases in overall hip and knee flexion over the period of contact.

For frontal plane kinematics, executing jumping maneuvers while braced resulted in a strong trend for decreased knee abduction during the sideways maneuver. Excessive frontal plane motion, specifically abduction, can potentially aid in the dissipation of ground reaction forces (McLean, et al., 2005). However, at the knee, excessive abduction can be detrimental and is thought to be one of the potential mechanisms for ACL injuries (Quatman, et al., 2010; Yeow, et al., 2009). With an ankle brace applied, the knee joint may be in a safer position as observed by the decrease in knee abduction. This decrease in abduction angle may be a response in an attempt to prevent the ankle joint from further inverting during the impact phase, potentially preventing ankle roll over. This decrease in knee abduction is more pronounced during the sideways maneuver, possibly in an attempt to accommodate the larger ankle inversion angle observed for the sideways maneuver. There seems to be no association between the knee and hip kinematics in the frontal plane that would further explain the knee trend to decrease abduction at impact as the hip displayed no significant changes or consistent trends in the frontal plane.

The application of an ASO ankle brace resulted in a trend for a change in the participant's transverse axis position of the knee at impact. During the forward maneuver there was a trend for the knee to shift from an externally rotated position during the control condition to an internally rotated position under the brace condition at impact. For the sideways maneuver the knee also did not demonstrate any significant difference around the transverse axis, however there was still a trend for the braced condition to display increased internal rotation. Subjects may increase internal rotation at the lower extremity in an attempt to reduce the joint loading demands in the sagittal plane (Sigward, et al., 2007). A larger degree of internal rotation may position the body in a more effective manner to accomplish a forward or sideways direction of movement and potentially increase chance of success (Sigward, et al., 2007).

As opposed to impact phase, there were no significant differences between brace conditions for knee and hip kinematics during the maximal propulsive phase. At this time point it appears that the ankle brace has no substantial effect on altering normal knee and hip joint kinematics. For this reason it is believed that the ankle brace has a smaller effect during the propulsive phase than it does on impact. The effect of bracing on the propulsive phase of contact will be presented in detail as part of the clinical relevance section on performance outcomes.

4.3 Joint Moments

The main trends in joint moments at impact were observed to occur at the ankle and knee, with smaller changes in the hip. For the forward maneuver the application of an ASO ankle brace resulted in a trend for increased ankle plantar flexor moment, increased external rotation moment, increased knee abductor and extensor moment and increased hip flexor moment. For the sideways maneuver under the braced condition, the participants displayed a trend for increased ankle eversion moment and external rotation moment, increased the knee extensor moment and increased the hip abductor moment. These results demonstrate that application of ankle brace may potentially alter the muscle action, possibly in an attempt to stabilize the joints of the lower leg during contact phase of simulated athletic maneuvers.

4.3.1 Ankle Moments

Our findings partially support our hypothesis that an ankle stabilizer can display trends for altering the moments around the ankle joint. The largest trends observed for joint moments at the ankle occurred in the plane of motion (i.e. sagittal plane during forward jump and frontal plane during the sideways jump). Increases in ankle joint moments are due to a combined effect of the material properties of the ankle brace restricting motion and with alterations in muscle contractions. In contrast to the increase in ankle joint inversion observed in the ankle kinematics, strong trends were observed that the ankle eversion moment was increased during the braced condition for the sideways maneuver. This is a possible indication that the lateral ankle structures and/or the brace were attempting to limit ankle inversion, possibly in compensation for the observed increase in ankle inversion ROM at impact. This eversion moment is relevant to ankle joint injury prevention, and as opposed to the observed increase in inversion ROM corresponds to previous literature (Ubell et al., 2003). Increased ankle eversion moments at impact indicate that the application of the brace was supporting the lateral ankle ligaments and, according to Venesky et al. (2006), indicates a safer landing strategy to prevent ankle sprain mechanisms. The trend for increased eversion moments was stronger during the sideways maneuver than the forward. This discrepancy between maneuvers may relate to the increased demands of frontal plane muscle contribution during a lateral movement. Support for the ankle joint (preventing further impact inversion) while also preparing the ankle for a lateral takeoff may require larger joint moments than supporting the ankle during a primarily flexion extension movement as observed in the forward jump.

For the forward maneuver the ankle brace condition displayed a strong trend for increased plantar flexor moment at impact, which was not observed during the sideways maneuver. This increase in plantar flexion may relate to the challenge of the forward jump associated with the application of the brace reducing the range of motion available at the ankle. Under the braced condition the ankle joint does not have the same sagittal range of motion (decreased plantar flexion at impact) as the control and subsequently the ankle plantar flexor muscle group has to provide sufficient eccentric muscle force to decelerate the body mass at impact (Yeow, et al., 2009; Cordova, et al., 2010). The plantar flexor muscles must increase in the moment to counteract the decreased ROM.

While this may not have provided a significant protective effect, the increase in plantar flexion moment may be a precursor to the change in knee or hip dynamics. For example, this increase in ankle plantar flexion may have provided a stimulus for the increase in knee flexion observed during impact. Because the ankle plantar flexor muscle group also crosses the knee joint, with the gastrocnemious providing knee flexion in addition to ankle plantar flexion, the increase in ankle plantar flexion moment may have translated proximally to the knee. If the knee became more compliant to flexion (i.e. during the sideways maneuver observing increased knee flexion) the effectiveness of the plantar flexor muscle group may have not have displayed changes at the ankle joint. With the forward maneuver not displaying an increase in knee joint flexion, the moment created by the ankle plantar flexors may have had a larger effect on the ankle joint, causing the large the strong trend in ankle plantar flexor joint moment at impact. For both movement directions the braced ankle condition demonstrated a strong trend for an increase in ankle external rotation moment. This increase in external rotation moment agrees with our kinematic observation for the ankle brace positioning the ankle joint in an externally rotated joint angle at impact, which due to the concentric nature of the contraction, may further enhance the protective mechanism of the ankle ligaments. The concentric contraction is caused by the joint moment and joint angle occurring in the same direction, and as an externally rotated position has been previously examined and determined to decrease stress applied to the lateral ankle ligaments (Eils et al., 2002; Wilkerson, 2002), provides a strong indication that the brace offers joint ligament protection.

The frontal plane ankle joint moments from this study also display differences across the contact time. Although not significant, there was a trend observed across contact time indicating ankle joint inversion/eversion may be shifted with the application of the brace. Specifically, post impact the brace condition displays a decreased inversion moment. This decrease in inversion moment occurs at the same time point as eversion ROM. This combination would indicate a decreased eccentric moment for the brace condition. Possibly due to the limited ROM, under the brace condition the ankle moments did not have to compensate to the same extent as the control condition and possibly decreased the muscle activation required. Having the brace condition display a decrease in inversion moment may also indicate that the brace application aided in facilitating movement by decreasing the required torque to transition into the propulsive phase, especially during the sideways maneuver where an inversion moment would be a precursor for lateral movement. Subsequently the ankle brace displays an obvious change in ankle joint inversion-eversion moments across the entire contact period. This bracing shift across the entire contact period is difficult to explain as no previous research has reported an ankle inversion position change for that extent of contact. The results may indicate a participant preference for an overall eversion moment, which the brace seems to allow, for both movement directions. If the brace limits the ankle inversion moment compared to the control condition for overall contact, then the results may further support previous literature regarding ankle braces having the potential to restrict inversion range of motion (Alves et al., 2002; Eils et al., 2002), by providing a shift towards an overall eversion joint support moment.

4.3.2 Knee and Hip Moments

In the sagittal plane, for both the forward and sideways maneuvers our participants displayed a trend for increased knee extensor moment generated during impact during the braced condition. Knee extensor moments, primarily caused by a greater quadriceps to hamstring muscle activation, can increase the strain on the ACL by increasing the tibial translation and therefore increasing the anterior shear force in the knee (Sigward et al., 2006). The increase in braced knee extensor moment may be related to the change in foot position and ankle angle at impact. Due to the brace possibly preventing the preferred ankle range of motion at impact, the knee extensor joint moment may have increased in response. Decreased ankle motion has been previously observed by Stoffel et al. (2010) to influence the position of the ground reaction force vector and subsequently alter the knee joint moments which potentially influence ACL strain. This is in accordance to previous authors who have proposed that increased knee extensor moments can influence ACL strain (Yu et al., 2007). Across our population, a trend was observed indicating the female participants increased their knee extension joint moment while subjected to the brace condition during the forward jumping maneuver only. The increase in knee extension moment may also be a result of the larger posterior ground reaction force at impact also observed for the brace condition. A posterior ground reaction force creates a flexion moment around the knee joint which needs to be balanced by increasing the knee extensor moment. This trend for a change in sagittal knee joint moment is in partial support of our hypothesis but not at our adjusted level of significance. While an increase in knee extensor moments has been thought to be a primary cause of knee ligament injury, recent investigations have speculated that the contribution of knee extensor moments alone may not be of significant magnitude to result in an ACL injury (Sigward et al., 2006). Therefore, despite similarities between our data indicating an increase in extensor moment, it is still unclear if these patterns would automatically place female at a greater risk for ACL injuries.

In combination with increased extensor moments for the forward jumping maneuver the knee joint also displayed a strong trend for increased abduction moment at impact during the braced condition. According to Venesky et al. (2006), the application of ankle braces has been suspected to increase knee abduction moments by limiting knee adduction motion. However, our results do not indicate an increase in knee joint frontal plane kinematics, as both the forward and sideways maneuvers decreased abduction ROM at impact. Although previous literature has indicated excessive knee abduction to be potentially harmful, and that women compared with men appear to land from a jump with increased knee abduction (Griffin. et al, 2006), there is not one clear explanation as to why the frontal moments increase during a sagittal plane forward movement (McLean, et al., 2005). Abduction at the knee is commonly observed and associated with cutting maneuvers (Quatman, et al., 2010) and although females have been observed to demonstrate significantly larger peak abduction knee moments than males this trend was found to be dependent on initial contact abduction angle (Hewett et al., 2005). Therefore our results may indicate a potential for increased knee injury risk due to the increase abduction observed at impact when combined with an increase in abduction angle. However without the associated change in knee joint angle our result of increased abduction moment with brace application may not be significantly meaningful.

Although not statistically significant, the brace condition almost demonstrates a trend for decrease in knee abduction moment post impact. This result, more clearly defined in the forward jump, is an indication that the lateral knee joint structures do not require the same support as in the control condition. This may subsequently reduce the stress on the medial joint ligaments, and is in accordance with Venesky et al., (2006) who also reported no difference between the brace and no brace condition during their movement trials. They concluded that during drop landings an ankle brace poses no more risk to the lateral knee structures than a control condition. Our results would agree with their findings.

The hip joint only displayed only a small trend in joint moment changes between the brace condition. The only trend that participants displayed was observed to be an increase in sagittal plane hip moments during the forward maneuver. For the forward movement the hip joint only displayed a trend for an increase in the flexion moment. The hip controls a large amount of body mass as it can facilitate movement of both the thigh segment, pelvis and to some degree trunk. Therefore the brief increase in hip flexion may be directly related to the large posterior GRF at impact requiring a rapid change in body mass. The posterior directed GRF will cause a rotation about the body which must be limited by muscle activation. Because the hip flexor has the ability to flex the trunk, the hip flexion moment may be favored by the participants due to previous training adapting the hip to offset the braking GRF and stabilize the body.

There were no differences that occurred in the knee or hip joint moments during the maximal propulsive phase of contact. This may be an indication that the knee and hip joint muscles are independent of brace effects during the propulsive phase. This effect may be related to performance. Performance could potentially be mediated by alterations in knee and hip joint moments. During this phase the knee and hip musculature are the largest muscle groups and produce the majority of the force that propels the body through take-off. The performance aspects of the propulsive phase are discussed separately below.

4.4 Stiffness

The stiffness of the ankle and knee joint were measured to examine the contribution of the brace to the joint stability both at impact and averaged across contact up to maximal vertical ground reaction force. The results of our study indicate that there is a trend for the increase in stiffness at the ankle joint during both maneuvers for the braced condition compared to the non-braced condition. Theses stiffness trends were observed both at impact and averaged across the contact phase for the ankle joint. However, there was no difference in alteration of knee stiffness for any maneuver between braced conditions.

Researchers have shown that active muscle stiffness properties are essential to controlling the dynamic stability of the joints, and recent evidence has revealed that there can be adjustments in the coordinating pattern of lower limb stiffness by modulating ankle stiffness (Farley & Morgenroth., 1999; Zinder, Granata, Shultz and Gansneder, 2009). The effect of bracing increasing the stiffness of the ankle joint was hypothesized to predispose the knee to display decreased stiffness. This inverse relationship (increased ankle stiffness with decreased knee stiffness) was thought to be a mechanism by which the participant could mediate overall leg stiffness. However, the lack of change in knee stiffness indicates to the brace is not the primary cause of increased ankle stiffness. In an examination of the passive mechanical stiffness provided by an ASO ankle brace, Smith, Lanovaz and Barss (*in progress*) determined that the passive torque provided by the ankle brace changes over the range of ankle motion allowable. The results of this examination, presented in Table 4.1 outline the passive characteristics of an ASO ankle brace, and indicate that the ASO ankle brace provides the largest passive resistance during peak dorsi flexion and peak plantar flexion with a peak stiffness ranging approximately between 0.08 and 0.11 N·m/degree. Within the current study, the peak brace stiffness (calculated by subtracting the control condition) was 0.57 N·m/degree and0.52 N·m/degree for the forward and sideways maneuvers respectively at impact. This result was much larger than what the passive brace stiffness should have been based on the work completed by Smith, et al. (*in progress*), indicating the muscle around the ankle joint was activated in a different manner with the brace applied than under the control condition. This increase in stiffness would have been caused by an increase in the net ankle joint plantar flexor moment at the time of impact. Average ankle brace stiffness was closer to that of the measured passive brace characteristics. Specifically, in the forward direction the average ankle brace stiffness alone accounting for a 0.02 N·m/degree increase. This result indicates that post impact, the change in ankle joint stiffness in accordance to the passive brace stiffness characteristics.

Table 4.1 Passive Stiffness Measures for ASO Ankle Brace:
The values are reported as N·m/deg (standard deviation). The values become larger with a larger stabilizing effect.

Brace Type	-10° to -20° Dorsi	-5° to +5° Neutral	+10° to +20° Plantar
ASO	0.110 (0.031)	0.050 (0.002)	0.083 (0.010)

Observing no relationship between ankle and knee stiffness outcome, the possibility exists that the knee stiffness may not be altered as much initially hypothesized as a secondary effect to changes in ankle stiffness. This lack of result may indicate that the lower limb joints, while working together as an interconnected system to absorb loads and optimize movement performance, do not necessarily work in a distal to proximal manner. According to Williams and Riemann (2009) increase in knee and hip stiffness may only be observed for activities involving large impact forces, and therefore, for activities that involve smaller impact forces the stiffness changes in the proximal joint may be minimal. Therefore the lack of stiffness change at the knee joint may be related to the lack of impact from our drop height. Subsequently, the increase in GRF caused by the ankle brace may have not been of significant magnitude to cause knee stiffness changes, and a larger magnitude of force at impact may be needed to affect knee stiffness. As well, our measures did not take into account the potential differences in hip stiffness between brace conditions. If the hip compensates for the increased ankle stiffness the knee stiffness may maintain a neutral pattern over both brace conditions. The hip joint was not measured due to difficulties in obtaining consistent and measurable data from the hip joint.

4.5 Clinical Relevance

4.5.1 ACL Injury Mechanisms

Numerous studies have found female athletes participating in pivoting and jumping sports possess a higher rate of non-contact anterior cruciate ligament (ACL) injury compared to males (Decker et al., 2003; Hewett et al., 2005). Research aiming to determine the risk factors for sustaining non-contact ACL injuries is increasing as concerns grow about the large number of female ACL injuries. The currently accepted mechanism for ACL injury is a knee extension moment combined with increased joint moment peaks in internal rotation and frontal plane axes. The increase in joint moments is typically observed during impact and deceleration, with both sagittal and frontal plane biomechanical factors having the potential to increase ACL loading mechanisms (Stoffel et al., 2010). Our data supports previous literature that the main increase in joint moments are found during the early percentage of contact, as these were the locations of the largest trends observed between braced conditions during both the forward and sideways maneuvers. In observing the brace effects during impact it was observed that participants had a strong trend for increased knee extension joint moment. This increased extensor moment has been described by Yu et al. (2007), as an indication of an athlete at high risk for ACL injury. They indicate that the increase in quadriceps pull will add significant anterior shear force on the proximal end of the tibia through the patellar tendon and by increasing peak posterior ground reaction force caused by decreased knee flexion (knee flexion between 15° and 30 °) the ACL is at larger risk for injury.

In addition to sagittal dynamic differences, our results also trended to indicate ankle bracing caused the participants to change their frontal plane knee kinematics at impact, specifically during the forward jump. In the current investigation the participants tended to impact the ground with a smaller knee abduction angle (smaller knee valgus) and therefore landed with their leg in a more vertical alignment in the frontal plane while under the braced condition compared to the control condition. In a study identifying ACL knee injury mechanisms, Hewett et al. (2005) investigated the neuromuscular control parameters in the lower limb during drop landings and reported increased knee abduction joint angles at impact would increase ACL strain. Their report compared non-injured females to females that went on to an ACL injury and determined that on average injured females exhibited 8.34° greater knee abduction angle, along with greater peak abduction moment averaging -45.3±28.5 N·m during drop jump landings. Increased knee abduction kinematics and moments have also been linked to increased knee ACL injury rates when observing sideways cutting as opposed to drop jumping (McLean et al., 2005). Based on the data reported by Hewett et al. (2005), and McLean et al., (2005), females would be less likely to incur knee injury while impacting the ground with their knee alignment in a neutral position in the frontal plane. Ankle bracing may be effective in reducing ACL strain by limiting knee motion in the frontal plane during impact. Without excessive joint motion at the knee joint, participants landing from a jump can utilize healthy and optimal neuromuscular coordination patterns.

There must be caution in interpreting the observations of the current investigation and their association with knee injury risk. Knee injuries, specifically ACL injuries, are complex mechanisms with multiple factors playing a role in ligament damage. Factors other than knee joint muscle stability and joint kinematics can play a role in contributing to ACL injuries during athletic activity. Such factors may include muscle fatigue, mechanical interface between the playing surface and the shoe, previous injury or overuse chronic joint strain (Renstrom et al., 2008). These factors may play a significantly larger role than an ankle stabilizer and injury mechanisms need to be examined on a case by case basis to determine the impetus to the true cause of injury. For female athletes, physiological differences, such as hormonal changes or anthropometric joint differences have been linked as predisposing stimuli to joint injury (Fagenbaum et al., 2003). Therefore addition of an ankle brace may have both risks and benefits to knee ACL structures depending on all possible combinations of movement constraints, physiological factors and environmental stimuli prior to an injury onset.

4.5.2 The Propulsive Phase and Brace Effects on Performance

Clinicians prescribing ankle stabilizers do so with the intent of decreasing ankle joint injuries. However when dealing with elite athletes competing at provincial, national or international levels, the prevention of joint injuries must be balanced against how the athlete can perform with the ankle stabilizer applied. Performance measures must therefore be considered when prescribing ankle stabilizers to an athletic population (Mackean, et al, 1995). Although the current investigation did not utilize a specified performance measure, the maximal propulsive phase of movement for the two jumping maneuvers were examined to determine the effect of ankle stabilization on the participant’s ability to prepare for take-off. The results of the present study indicate that the effect of ankle bracing has limited effect on the joint kinematics, ground reaction force and proximal joint moment during the propulsive phase in either the forward or sideways directions for a single jumping procedure. The results showed that, except for the small changes in ankle frontal and transverse plane range of motion and associated moments, no significant differences were observed in any of the kinematics or kinetics at the knee or the hip joint. As well, there was no significance found in the timing or magnitude of the maximal propulsive ground reaction forces generated for either maneuver between conditions.

Our result showed no difference in time to peak propulsive force generation, suggesting that, at the time participants were preparing for their take-off, the lower limb was generally unaffected by the brace application. Further supporting this theory, were the observations during the maximal propulsive force phase the knee and hip kinematics and kinetics displayed no significant differences and only limited trends for change. Also no significant difference was found between ankle brace conditions on propulsive force magnitude. Therefore, for the jumping protocol, the participants may have been able to control and match the amount of force they were going to generate prior to take-off for the given task. Potentially over the course of 100% of contact, participants may be able to adapt to the ankle brace's rigidity to achieve the same propulsive strategy and subsequently achieve similar performance under braced and non-braced conditions. With similar lower limb mechanics there is no reason to speculate that the brace will affect the overall joint coordination pattern prior to take-off, and therefore have little effect on take-off performance during a real life game situation. More research is required to fully determine the effect ankle braces have on performance effects, specifically how the single isolated jumping protocol results can be expanded into more reactive game situations. Investigation of a stronger performance measure along with the inclusion of multiple braces may further clarify the performance discrepancy. Based on literature and results from the current investigation, use of an ankle brace by competitive athletes does show the ability to radically detriment performance.

4.5.3 Training Adapted Specifically for Ankle Brace Wear

The population examined was that of a trained, competitive (CIS) athletic population. The use of high performance training techniques is common within this population, and typically the training programs aim to enhance performance while having a secondary focus on injury prevention. An ASO ankle brace can affect proximal joints and muscles. Incorporation of training programs adapted to the kinematic and joint moment effects of ankle stabilizers would be beneficial to the athlete's performance and could have the added potential effect, if designed correctly, of reducing the risk of joint injuries in general. In a review on the concepts of ACL injury Renstrom et al. (2008) specify that successful programs have common elements including education, traditional stretching, strengthening, awareness of high risk positions, technique modifications, aerobic conditioning, sport specific agilities, proprioceptive and balance training and plyometrics. If the aim of these components can be modified to the specific mechanisms of ACL injury in females, and if the participants are going to be prescribed an ankle brace, then the training should be modified in accordance to the bracing effects as well. As an example, the incorporation of these components to the specific risk factors can be viewed below.

Table 4.2: Simple Training Adaptations:
Training should incorporate the effect of the ankle brace. The following is a brief suggestion of how training would be adapted and applied for two results outlined within this study.

ASO Bracing Modification	Intervention Strategy	How to Incorporate
Increase vertical GRF	Synergistic joint motion modification	Full joint ROM exercises, strengthen muscles at flexed positions
Increased ankle inversion	Proprioceptive enhancement	Stability training with proprioceptive alignment

The current Huskie Women's Basketball training program (Appendix F) designed by the Human Performance Center (HPC) at the University of Saskatchewan is a series of 5 programs broken down into first and second training regimens. The program is very comprehensive and incorporates whole body strength with a different group of muscles as the target area per program. The program does not incorporate any specified training aspects using ankle bracing and does not take into account alterations caused by the brace. Combining the information gathered from the current study with previous research aimed at developing injury prevention programs (specifically ACL prevention programs) a specified and modified training regimen could be theoretically incorporated into the current training programs with the aim of enhancing performance while preventing injury.

4.6 Limitations

There were several limitations to the current investigation. The first limitation is the sample size used within this study only examining eight participants. Given the specific sport and gender population criteria for acceptance into this study it was difficult to recruit a large number of participants. Due to the variability of the movement parameters and the large degree of movement freedom each participant had in completing the forward and sideways jumping maneuvers, a larger number of participants may have allowed for a stronger interpretation of the results. Secondly, this study only used two movements to examine the effect of an ankle brace. Although the movements were meant to replicate basketball game situations, there are differences between laboratory and in game settings. These differences may include participants having the ability to think or prepare for a movement in a laboratory, whereas in the game the movement would be a reaction to the defender or opposition or not moving as the same velocity in the lab as they would be in the actual game. These differences between games and laboratory studies limits the ability to directly relate the results to game situations and may have a direct effect on the magnitude of joint dynamic changes observed.

A third limitation of this study would be regarding the specific population chosen for analysis. Choosing a specific female athletic population, who are familiar with ankle brace wear, limits the generalization of the study results. As there are many individuals who wear ankle braces for recreation sports or daily activities, many of which are not elite trained athletes, caution should be exercised when attempting to generalize the results to all populations.

Lastly the ankle brace itself has inherent limitations within the validity of application. Ankle braces are difficult to tighten consistently and are applied based on the individual preference for brace tightness. The first brace limitation within this study was allowing the participants to lace up their own ankle braces with researchers supervision. Although each participant was observed to tighten and apply the ankle braces in a similar manner, there may have been slight variations between participants, possibly leading to increased variability within the joint kinematics or joint moment data. The only possible way to fully monitor the brace tightness level would have been to build a tightening device with a calibrated instrument measuring the force applied to the laces. However such a device was not feasible to design or build within the course of this study. The second brace limitation is in regards to applying brand new braces to each participant for the jumping maneuvers. As a brace is worn over time the material comprising the boot of the brace will ultimately break down and like a shoe or other pieces of sports equipment, become "worn-in". Typically a brace is worn over the entire course of a single season or multiple seasons, therefore the results of using a new brace may not fully represent the true brace effects observed with a participant using a used brace.

Lastly there were limitations with the statistical analysis techniques chosen for this study. Due to the exploratory nature of the study design and the vast number of variables examined Type I error was likely with the number of paired *t-test* used. To protect against the Type I error a conservative Bonferroni correction was implemented in adjusting the level of significance to the $p<0.001$ level. With such a conservative approach it is likely that Type II errors may have occurred, and real significant differences were not observed. For a study with implication for possible injury occurrence due to the application of a brace, the potential of Type II error may have an effect on an athlete's choice to wear an ankle brace based on the perceived outcome. An example within this study would be the trend for increased knee extension observed for the brace condition during the sideways manoeuvre. With a trend of $p=0.021$ the increased knee extension may be considered significant under a less conservative approach, and due to the implications of increased knee extension at impact and potential knee injury, a significant finding may determine if the potential for knee injury is greater than the benefit the brace provides to the ankle. To remove the potential for Type II error trends in the data were presented. However these results were not defined as significant, therefore the limitation still applies.

4.7 Future Directions

The main area for future research regarding the effects of ankle bracing on the ankle knee and hip dynamics involves researching a variety of populations. One of the main limitations of this study was with the specific population used to assess bracing effect. It will be important to confirm that the results observed within this study are transferable to the general population of non-elite athletes. Further it is important to determine if the results have inherent gender effects. Because of the unique injury rates and physiological differences, women athletes have been a focus for lower limb biomechanical studies; however, it is unlikely that a female is any more likely to wear an ankle brace than her male counterpart. For these two reasons it is important to determine the effect of ankle bracing on non-elite as well as male athletes to generalize the result to the large amount of recreational and competitive males and females that use ankle braces to prevent and recover from ankle injuries.

A second area for future research may involve altering methodology with the aim of the increasing the specificity of the results observed within our current population of the elite female athlete. One method for strengthening the results may be through modifying the movement parameters to increase real world application. With our study, the movements were chosen to mimic common athletic movements within the confines of our laboratory and laboratory runway. If it were possible it may be advantageous to institute running and or cutting maneuvers that were more difficult and could constrain the possible movement options available to the participants. These more advanced maneuvers may be better at mimicking the participant's reaction to defenders and or the constraints of moving around other teammates on the floor during the game. These movements may give clarity as to where the largest effect of the ankle brace is in a game situation, which would be beneficial to athletic therapists and coaches in understanding the choice in prescribing an ankle brace.

Chapter 5

Conclusions

5.1 Conclusion

The largest differences observed between brace conditions were at impact, during the early phase of contact, with minimal differences observed after the impact phase. As hypothesized and consistent with previous literature the ankle brace displayed a significant ability to alter the ankle joints kinematics and displayed strong trends in altering the ankle joint moments for both movement procedures, with the largest changes occurring during impact. The observations from this study indicate that an ASO brace can alter the position of the ankle at initial contact, in both the frontal, sagittal and transverse axes, as well as reduce the overall ROM available from contact until maximal joint dorsi flexion. The brace caused a shift in ankle position by rotating the joint into a more dorsi flexed and externally rotated position during the forward and sideways maneuver. These positional changes have been previously examined and are often associated and concluded to be a 'safer' position for landing. The ankle brace also displayed trends for the ability to modify the ankle's joint moments. The largest differences were observed in the sagittal and transverse axes during the forward jump and in the frontal and transverse axes during the sideways jump. The ankle displayed larger joint moments in the plane of movement, again corresponding to previous literature. As well for both the forward and sideways maneuvers, the ankle brace trended to increase ankle joint stiffness. This increase in stiffness was observed both at the instant of contact and averaged across the contact percentage.

The change in ankle dynamics is thought to be the precursor for the why the brace application displayed strong trends for the increases observed for both the maximal vertical and breaking GRFs magnitude as well as decreased time rate at which those forces are generated for both the forward and sideways maneuvers. These findings were consistent with previous literature and there were no differences in the propulsive GRF magnitude of timing for either movement.. The knee and hip joints were not altered to the same degree as the ankle. Trends in the knee and hip, like the ankle joint, primarily occurred during the impact phase and in the plane associated with the direction of movement. Specifically, during the forward jump, the knee displayed a trend for increased flexion and an increase in joint extension moment, while the hip displayed increased flexion during the forward jump and increase abduction during the sideways jump. However there were no changes in the knee and hip kinematics or joint moments during the propulsive phase of movement. There were also no differences in the knee joint stiffness averaged across contact time, or instantaneously at impact.

The unique results of this study indicate bracing to be significantly effective in altering ankle kinematics, and may have a smaller effect on the proximal joints than previously expected or previously reported. Overall the largest effects observed with the ankle brace occurred at the beginning stages of contact, possibly enhancing the safety of the ankle joint structure during impact. These results support previous literature suggesting that protection is a primary purpose of applying an ankle brace. Our study strengthens the research supporting the use of ankle stabilizers to prevent injuries without strong evidence of proximal joint injury.

References

Alves J. W., Alday, R. V., Ketcham D. L., & Lentell, G. L. (1992) A comparison of the passive support provided by various ankle braces. *Journal of Orthopeadic & Sports Physical Therapy 1*, 10–18.

Borotikar, B.S., Newcomer, R., Koppes, R., & McLean, S. G. (2008). Combined effects of fatigue and decision making on female lower limb landing postures: Central and peripheral contributions to ACL injury risk. *Clinical Biomechanics, 23(1),* 81-92.

Bot, S. M., & van Mechelen, W. (1999). The effect of ankle bracing on athletic performance. *Journal of Sports Medicine, 27(3),* 171-178.

Cloak, R., Galloway, S., & Wyon, M. (2010). The effect of ankle bracing on peak mediolateral ground reaction force during cutting maneuvers in collegiate male basketball players. *Journal of Strength & Conditioning Research, 24 (9)*, 2429 – 2433.

Cordova. M. L., Ingersoll, C. D., & LeBlanc, M. J. (2002). Influence of ankle support on joint range of motion before and after exercise: a meta-analysis. *Journal of Orthopaedic & Sports Physical Therapy, 30(4)*, 170-182.

Cordova, M. L., Scott, B. D., Ingersoll, C. D., & LeBlanc, M. J. (2005). Effects of ankle support on lower extremity functional performance. *Medicine & Science in Sport & Exercise, 37(4)*, 635-641.

Cordova, M.L., Takahashi, Y., Kress, G.M., Brucker, J.B., & Finch, A.E. (2010). Influence of external ankle support on lower extremity joint mechanics during drop landings. *Journal of Sports Rehabilitation*, *19*,136-148.

Decker, M. J., Torry, M. R., Wyland, D. J., Sterett, W. I., & Stedman, J. R. (2003). Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clinical Biomechanics, 18(7)*, 662-669.

DeVita, P., & Skelly, W. A., (1992). Effect of landing stiffness on joint kinetics and energetic in the lower extremity. *Medicine and Science in Sport and Exercise, 24(1),* 108-115.

DiStefano, L.J., Padua, D. A., Brown, C. N., & Guskiewicz, K. M. (2008). Lower extremity kinematics and ground reaction forces after prophylactic lace-up ankle bracing. *Journal of Athletic Training, 4*, 234-241.

Dizon, J. R., & Reyes, J. B. (2010) A systematic review on the effectiveness of external ankle supports in the prevention of inversion ankle sprains among elite and recreational players. *Journal of Science & Medicine in Sport, 13(3),* 309-317.

Duysens, J., & Levin, O. (2010). Ankle sprains: getting off on the wrong foot. *Exercise & Sport Sciences Reviews, 38(3),* 143-149

Eils, E., Demming, C., Kollmeier, G, Thorwesten, L., Völker, K. & Rosenbaum, D. (2002). Comprehensive testing of 10 ankle braces evaluation of passive and rapidly induced stability in subjects with chronic ankle instability. *Clinical Biomechanics, 17,* 526-535.

Eils, E., Imberge, S., Völker, K. & Rosenbaum, D. (2007). Passive stability characteristics of ankle braces and tape in barefoot and shod conditions. *The American Journal of Sports Medicine, 35,* 282-287.

Fagenbaum, R., & Darling, W.G. (2003). Jump landing strategies in male and female college athletes and the implications of such strategies for anterior cruciate ligament injury. *American Journal of Sports Medicine, 31(2),* 233-240.

Farley, C.T., & Morgenroth, D.C. (1999). Leg stiffness primarily depends on ankle stiffness during human hopping. *Journal of Biomechanics, 32,* 267-273.

Gilchrist, J., Mandelbaum, B., Melancon, H., Ryan, G. W., Silvers, H. J., Griffin, L. Y., Watanabe, D. S., Dick, R. W., & Dvorak, J. (2008). A randomized control trial to prevent noncontact anterior cruciate ligament injury in female collegiate soccer players. *The American Journal of Sports Medicine, 36(8),* 14761483.

Gribble, P. A., & Robinson, R. H. (2009). Alterations in knee kinematics and dynamic stability associated wth chronic ankle instability. *Journal of Athletic Training, 44(4),* 350- 355.

Gudibanda, A., & Wang, Y. T. (2005). Effect of the ankle stabilizing orthosis on foot and ankle kinematics during cutting maneuvers. *Research in Sports Medicine, 13,* 111-126.

Griffin. et al, (2006). Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting. *American Journal of Sports Medicine, 34,* 1512–1532

Grood, E. S., & Suntay, W. J. (1983). A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *Journal of Biomechanical Engineering, 105(2),* 136-145.

Hewett, T.E., Myer, G.D., Ford, K.R., Heidt, R.S., Colosimo, A.J., McLean, S.G., van den Bogert, A.J., Paterno, M.V., & Succop, P. (2005). Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes. *The American Journal of Sports Medicine, 33(4),* 492-501.

Hughes, G., Watkins, J. & Owen, N. J. (2008). Gender differences in the lower limb frontal plane kinematics during landing. *Sports Biomechanics, 7(3),* 333-341.

Hodgson, B., Tis, L., Cobb, S., & Higbie, E. (2005). The effect of external ankle support on vertical ground-reaction force and lower body kinematics. *Journal of Sports Rehabilitation*, *14*, 301-312.

Houck, J., Duncan, A., & De Haven, K. (2006). Comparison of frontal plane trunk kinematics and hip and knee moments during anticipated and unanticipated walking and side step cutting tasks. *Gait and Posture 24*,314–322.

Kellis, E., & Kouvelioti, V. (2009) Agonist versus antagonist muscle fatigue effects on thigh muscle activity and vertical ground reaction during drop landing. *Journal of Electromyography and Kinesiology, 19(1),* 55-64.

Kernozek, T. W., Torry, M. R., Van Hoof, H., Cowley, H., & Tanner, S. (2005). Gender differences in frontal and sagittal plane biomechanics during drop landings. *Medicine & Science in Sport and Exercise, 37,* 1003-1012.

MacKean,L., Bell, G., & Burnham, R. (1995). Prophylactic ankle bracing versus taping: effects on functional performance in female basketball players. *Journal of Orthopaedic & Sports Physical Therapy, 22(2),* 77-81.

McCaw, S. T., & Cerullo, J. F. (1999). Prophylactic ankle stabilizers affect ankle joint kinematics during drop landings. *Medicine & Science in Sport & Exercise 31(5),* 702-707.

McLean, S.G., Huang, X., & van den Bogert, A.J. (2005). Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: Implications for ACL injury. *Clinical Biomechanics, 20*, 863-870.

Medina, J.M., et al.(2008). Timing of neuromuscular activation of the quadriceps and hamstrings prior to landing in high school male athletes, female athletes, and female non-athletes. *Journal or Electromyography Kinesiology*, *18(4)* 591-597.

Meana, M., Alegre, L. M., Elvira, J. L. L., & Aguado, X. (2008). Kinematics of ankle taping after a training session. *International Journal of Sports Medicine, 29(1),* 70-76.

Mickel, T. J., Bottoni, C. R., Tsui, G., Chang, K., Baum, L., & Tokushige, K. A. S. (2006). Prophylactic bracing versus taping for the prevention of ankle sprains in high school athletes: A prospective, randomized trial. *The Journal of Foot & Ankle Surgery, 45(6),* 360-365.

Midgley, W., Hopkins J. T., Feland, B., Kasier, D., Merrill, G., & Hunter, I. (2007). The effects of external ankle support on dynamic restraint characteristics of the ankle in volleyball players. *Clinical Journal of Sports Medicine,17(5),* 343-348.

Myer, G.D., Ford, K.R., Palumbo, J.P., & Hewett, T.E. (2005). Neuromuscular training improves performance and lower extremity biomechanics in female athletes. *Journal of strength and Conditioning 19(1)*, 51-60.

Myklebust, G., Engebretsen, L., Brækken, I.H., Skjolberg, A., Olsen, O., & Bahr, R. (2003). Prevention of anterior cruciate ligament injuries in female team handball playersP: A prospective intervention study over three seasons. *Clinical Journal of Sports Medicine, 13,* 71-78.

Paris, D, L., Kokkaliaris, J., Vardaxis, V. (1995). Ankle ranges of motion during extended activity periods while taped and braced. *Journal of Athletic Training, 30(3),* 223-228.

Popadopoulos, E. S., Nicolopoulos, C., Anderson, E. G., Curran, M., & Athanasopoulos, S. (2005). The role of ankle bracing in injury prevention, athletic performance and neuromuscular control: a review of the literature. *The Foot 15*, 1-6.

Pollard, C. D., Sigward, S. M., & Powers, C. M. (2010). Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. *Clinical Biomechanics, 25(2).*142-146.

Quatman, C. E., Quatman-Yates, C. C., & Hewett, T., E. (2010). A plane explanation of anterior cruciate ligament injury mechanisms: a systematic review. *Sports Medicine, 40(9),*729-746.

Renstrom, et al. (2008). Non-contact ACL injuries in female athletes: An international Olympic committee current concepts statement. *British Journal of Sports Medicine, 4*, 394-412.

Riemann, B. L., Schmitz, R. J., Gale, M., & McCaw S. T. (2002). Effect of ankle taping and bracing on vertical ground reaction forces during drop landings before and after treadmill jogging. *Journal of Orthopeadic & Sports Physical Therapy 32*, 628-635.

Rosenbaum, D., Kamps, N., Bosch, K., Thorwesten, L., Völker, K., & Eils, E. (2005). The influence of external ankle braces on subjective and objective parameters of performance in a sports-related agility course. *Journal of Sports Medicine, 13,* 419-425.

Sanna, G., & O'Connor, K.M.(2008). Fatigue-related changes in stance leg mechanics during sidestep cutting maneuvers. *Clinical Biomechanics, 23,* 946–954.

Santos, M. J., McIntire, K. Foecking, J., & Liu, W. (2004). The effects of ankle bracing on motion of the knee and the hip joint during trunk rotation. *Journal of Clinical Biomechanics, 19,* 964-971.

Sacco I., Takahasi, H. Y., Suda, E. Y., Battistella, L, R., Vavamoto, C. A., Lopes, J. A. F., & de Vasconcelos, J. C. P. (2006). Ground reaction force in basketball cutting maneuvers with and without ankle bracing and tape. *Sao Paulo Medical Journal, 124(5),* 245-252.

Sefton, J.M., Hicks-Little, C. A., Koceja, D. M., & Cordova, M. L. (2007). Modulation of soleus H-reflex by presynaptic spinal mechanisms during varying surface and ankle brace conditions. *Clinical Neurophysiology 37*, 15-21.

Siegler, S., Liu, W., Sennett, B., Nobilini, R. J., & Dunbar, D. (1997). The three-dimensional passive support characteristics of ankle braces. *Journal of Orthopaedic & Sports Physical Therapy, 26(6),* 299-309.

Sigward, S.M., & Powers, C.M. (2007). The influence of gender on knee kinematics, kinetics, and muscle activation patterns during side-step cutting. *Clinical Biomechanics, 21*, 41-48.

Sigward, S.M., & Powers, C.M. (2007). Loading characteristics of females exhibiting excessive valgus moments during cutting. *Clinical Biomechanics, 22*, 827-833.

Simpson, K.J., Craven, S., Theodorou, C., & DelRay, P. (1999). A comparison of the sport stirrup, malleoloc, and swede-o ankle orthoses for the foot-ankle kinematics of a rapid lateral movement. *International Journal of Sports Medicine, 20*, 396-402.

Shapiro, M. S., Kabo, J. M., Mitchell, P. W., Loren, G., & Tsenter, M. (1994). Ankle sprain prophylaxis: an analysis of the stabilizing effects of braces and tape. *The American Journal of Sports Medicine, 22*, 78-82.

Smith, M. J., Barss, T. S., & Lanovaz, J. L. (*in progress*). Bracing effects on resistance torque and stiffness.

Stoffel, K.K., Nicholls, R.L., Winata, A.R., Dempsey, A.R., Boyle, J.J.W., & Lloyd, D.G. (2010). Effect of ankle taping on knee and ankle joint biomechanics in sporting tasks. *Medicine and Science in Sport and Exercise, 42(11),* 2089-2097.

Thonnard, J.L., Bragard, D., Willems, P.A., & Plaghki, L. (1996). Stability of the braced ankle. *American Journal of Sports Medicine, 24*, 356-361.

Ubell, M. L., Boylan, J. P., Ashton-Miller, J. A., & Wojtys, E. M. (2003). The effect of ankle braces on the prevention of forced ankle inversion. *The American Journal of Sports Medicine, 31*, 935-940.

van den Bogert, A. J., & de Koning, J. J. (1996). On optimal filtering for inverse dynamics analysis. *Proceedings of the IXth Biennial Conference of the Canadian Society of Biomechanics,* 214-215.

Venesky, K., Docherty, C. L., Dapena, J. & Schrader, J. (2006). Prophylactic ankle braces and knee varus-valgus and internal-external rotation torque. *Journal of Athletic Training, 41(3),* 239-244.

Verhagen, E. A., van der Beek, A. J., & van Mechelen W. (2001). The effect of tape, braces and shoes on ankle range of motion. *Journal of Sports Medicine, 31(9),* 667-677.

Verbrugge, J. D. (1996). The effect of semirigid air-stirup bracing vs. adhesive ankle taping on motor performance. *Journal of Orthopaedic & Sports Physical Therapy, 23(5)*, 320-325.

Wilkerson, G. (2002). Biomechanical and neuromuscular effects of ankle taping and bracing. *Journal of Athletic Training, 37(4)*, 436-445.

Williams, S., & Riemann, B.L. (2009). Vertical leg stiffness following ankle taping and bracing. *International Journal of Sports Medicine, 30,* 383-386.

Winter, D. A. (2005). Biomechanics and motor control of human movement (3rd edition). New Jersey (NY): John Wiley & Sons, Inc.

Wright, I. C., Neptune, A. J., van den Bogert, A. J., & Nigg, B. M. (2000). The influence of foot positioning on ankle sprains. *Journal of Biomechanics, 33,* 513- 519.

Yeow, C.H., Lee, P.V.S., & Goh, J.C.H. (2009). Effect of landing height on frontal plane kinematics, kinetics and energy dissipation at lower extremity joints. *Journal of Biomechanics*,*42*,1967-1973.

Yu, B., & Garrett, W.E. (2007). Mechanisms of non-contact ACL injuries. *British Journal of Sports Medicine, 41(1)*, i47-i51.

Zhang, S., Bates, B. T., & Dufek, J. S. (2000). Contributions of lower extremity joints to energy dissipation during landings. *Medicine and Science in Sport and Exercise*, *32*, 812-819.

Zhang, S., Wortley, M., Chen, Q., & Freedman, J. (2009). Efficacy of an ankle brace with a subtalar locking system in inversion control dynamics movements. *Journal of Orthopeadic & Sports Physical Therapy, 39(12),* 875-883.

Zinder, S.M., Granata, K.P., Shultz, S.J., & Gansneder, B.M. (2009). Ankle bracing and the neuromuscular factors influencing joint stiffness. *Journal of Athletic Training, 44(4),* 363-369.

Appendix A: Ethics and Copy of Certificate of Approval

Research Participant Information and Consent Form

Title: **The effect of ankle stabilization on knee and hip mechanics and muscle activation in female athletes**

Name of Researcher:

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Introduction

You are being invited to participate in a research study because we want to measure how ankle braces may alter lower leg muscle activation and knee and hip joint loading mechanics.

Your participation is entirely voluntary, so it is up to you to decide whether or not you wish to take part. If you decide not to take part, you do not have to provide a reason and it will not affect your relationship with any of the researchers. Your academic standing will not be affected in any way. If you decide to take part in this study, you are still free to withdraw at any time without any consequences or giving any reasons for your decision.

This consent form may contain words that you do not understand. Please ask any of the researchers listed above to explain any words or information that you do not clearly understand. You may ask as many questions as you need to understand what the study involves. Please feel free to discuss this with your family, friends or family physician.

Note that neither the institution nor any of the investigators or staff will receive any direct financial benefit from conducting this study.

The study will be conducted at the Physical Activity Complex on the University of Saskatchewan campus and we expect to enroll 20 participants.

Study Purpose

The purpose of this experimental study is to collect and analyze the forces, movement and muscle activation patterns in your legs during basic athletic movements while your ankle is supported by a range of ankle braces. This study is investigating whether ankle braces have a more global effect on leg mechanics than just decreased range of motion localized at the ankle.

Who Can Participate?
You are eligible to participate in this study if you are female, healthy and between the ages of 18 and 28. You must be a current player in post secondary, Canadian Interuniversity Sport level athletics (Basketball, Volleyball, or Soccer). You need to be free of any significant physical or neurological impairment as well as being free of any significant lower body injury (such as broken bones or sprained or torn ligaments) for the previous 6 months leading up to the study. You also need to be capable of performing mild physical activity such as walking, jogging and jumping for short periods of time.

Study Procedures
If you agree to participate in this study, you will come to Room 355 at the Physical Activity Complex, University of Saskatchewan for one visit and the following will take place:

1. The study procedures will be explained and you will have an opportunity to ask questions before signing the consent form.

2. Your height and weight will be recorded.

3. You will be asked to complete a brief questionnaire to determine your lateral foot preference (dominant foot). The questions are taken from the Waterloo Footedness Questionnaire- Revised (Elias et al., 1998), and will asses foot preferences for performing a series of tasks.

4. A series of small reflective spheres, used by the motion capture system, will be attached to your lower limbs and arms using two-sided hypoallergenic tape.

5. Electromyography (EMG) electrodes, used to passively record muscle activity, will be attached on the major muscle groups of your legs using hypoallergenic adhesive.

6. For each ankle brace being tested, the following will happen:

    a. You will be asked to remove your socks and shoes.
    b. A thin plastic force transducer will be taped to one of your feet.
    c. You will put on the ankle brace and adjust it until you are comfortable.
    d. You will put your shoes back on and be allowed 5- 10 min of walking to get comfortable with the brace.
    e. The force transducer and EMG leads will be connected to a small pack secured around your waist.
    f. You will be asked to perform some basic physical activity movements including walking, jogging, jumping, jumping sideways, landing and squatting (in a randomized order).
    g. You will then be asked to remove your shoes and the brace.

7. You will be asked to repeat step number 5 for up to six different braces.

Data will be gathered using a motion capture system that tracks the movements of your limbs. At the same time, we will record forces that you apply to the ground using an instrumented platform embedded in the floor. Also a high speed video camera will be used to record your movements as reference data during analysis. The EMG system uses the small sensors taped to your skin to passively record the natural electrical activity produced by your muscles. The areas where the electrodes will be placed may need to be shaved.

The total visit will take approximately 2 hours. You will be allowed to take as many rest periods as you require during the testing.

What are the Benefits of Participating in this Study?
There are no anticipated benefits from this study to you directly. It is hoped that the information gathered in this study can be used in the future to better understand how ankle bracing alters the muscle activity of the leg during movement compared to a non-braced condition.

What are the possible Risks and Discomforts?
The risks from this study are minimal and are no more than what you would have in normal everyday activity. The movements that you will be performing do not require much physical exertion. However, if you feel tired or uncomfortable, you may ask to rest at any time and for as long as you need.

There may be some discomfort on your skin from the adhesive tape that temporarily sticks the spheres, or from the brace rubbing against the skin, but this is rare.

There may also be unforeseen and unknown risks during the study, or possibly after the study has been completed.

Are there any alternative treatments?

You do not have to participate in this study to use an ankle brace. They are commonly available at many sports supply stores.

What happens if I decide to withdraw?
Your participation is entirely voluntary, so it is up to you to decide whether or not to take part in this study. If you do decide to take part in this study, you are free to withdraw at any time without giving any reasons for your decision and your refusal to participate will not affect your relationship with any of the researchers or the University of Saskatchewan, and will not affect your academic standing if you are a student at the university. If you choose to enter the study and then decide to withdraw at a later time, all data collected about you during your enrolment will be retained for analysis.

What happens if something goes wrong?
In the case of a medical emergency related to the study, you should seek immediate care and, as soon as possible, notify the study doctor. Inform the medical staff you are participating in a clinical study. Necessary medical treatment will be made available at no cost to you. By signing this document, you do not waive any of your legal rights against the sponsor, investigators or anyone else.

What happens after completion of this study?
The data from this will be presented by the researchers at academic conferences and published in peer-reviewed academic journals. If you wish to receive a lay person's summary of the results of this study after it is complete, please contact Dr. Joel Lanovaz by phone (306-966-1073) or e-mail (joel.lanovaz@usask.ca). This summary will be an aggregate of all results and not your individual results.

What will the study cost me?
You will not be charged for any research-related procedures. You will not be paid for participating in this study. Reimbursement for study-related expenses (e.g. travel, parking, meals) is not available.

Will my information be kept Confidential?
While complete subject anonymity cannot be guaranteed, every effort will be made to ensure that the information you provide for this study is kept entirely confidential. Your name will not be attached to any information, nor mentioned in any study report, nor be made available to anyone except the research team. It is the intention of the research team to publish results of this research in scientific journals and to present the findings at related conferences and workshops.

Most research findings will be reported in aggregate form without reference to specific participants. In the event individual data are used, only participant codes will be referenced and your identity will not be revealed. Some digital still images and video are taken during data collection for reference. These images are kept confidential. If an image is used for publication purposes, it will be altered to remove all information that could be used to identify a specific individual.

Data are stored on a password protected digital media (i.e., DVD) in a locked lab/office in the College of Kinesiology to which only the researchers will have access. The data will be used for dissertation and publication purposes only, and will be retained for a minimum of five years. Normally data is retained for a period of five years post-publication, after which time it may be destroyed.

Who do I contact if I have any Questions about the study?
If you have any questions concerning the study, please feel free to ask them at any point; you are also free to contact the researchers at Dr. Joel Lanovaz at 306-966-1073 (collect calls accepted) or by e-mail provided if you have any questions at a later time. This research project was reviewed and approved on ethical grounds by the University of Saskatchewan Biomedical Research Ethics Board.

If you have questions about your rights as a research subject, you should contact the Chair of the Biomedical Research Ethics Board, University of Saskatchewan at (306) 966-4053. Again, this number can be called collect if you are phoning long distance.

Subject Consent to Participate

I have read (or someone else has read to me) the information in this consent form. I understand the purpose and procedures, the possible risks and benefits of the study. I was given sufficient time to think about it, and to ask questions, receiving satisfactory answers to all of my questions.

I am free to withdraw from this study at any time for any reason and the decision will not affect your relationship with the researchers.

I voluntarily consent to take part in this research study and give permission to the use and disclosure of my de-identified personal health information collected for the research purposes described above.

By signing this document I do not waive any of my legal rights. I will be given a signed copy of this consent form.

Printed Name of Participant:______________________________

Participant's Signature:_________________________ Date: ______________________

Individual conducting the consent process:_________________________

Date: _______________________

Appendix B: Kinematic Collection Details

This appendix gives the details regarding the motion capture data collection. The locations of the landmarks used are given along with details regarding calibration, calculation of functional joint centres and definitions for segment coordinate systems.

B.1 Landmark and Marker Descriptions

A total of 39 markers were used to collect the kinematic data using the motion capture system. Markers are classified into two types; *required* and *calibration* only. Required markers were used in the full collection of 3D joint and limb movement throughout all movement trials (Table B.1). The calibration only markers were used primarily for the purpose of defining joint axis in reference to the required markers. The calibration only markers were placed on the anatomical landmarks for the calibration pose only (Table B.2). The calibration pose, which is used to allow the computer and 3D motion capture system the ability to accurately define the current participants marker placement, includes all required and calibration markers.

Table B.1: Required Markers – Used for the motion analysis of participants body limbs and joint positions for both movement trials as well as calibration. Markers were not removed after calibration file

Marker Name	Anatomical Position
Back	Cervical Vertebra 7 (C7)
Right Shoulder	Right Acromion Process
Right Elbow	Lateral Epicondyle of Right Humerus
Right Wrist	Right Ulnar Styloid Process
Left Shoulder	Left Acromion Process
Left Elbow	Lateral Epicondyle of Left Humerus
Left Wrist	Left Ulnar Styloid Process
Pelvis Marker 1	Pelvis Cluster- Attached to belt around Pelvis. Resting on Posterior Ilium and Posterior Superior Iliac Spine(Figure. 2.3)
Pelvis Marker 2	
Pelvis Marker 3	
Pelvis Marker 4	
Right Femur Marker 1	Right Femur Cluster- Attached to lower 1/3 of lateral thigh, below hand resting length (Figure. 2.3)
Right Femur Marker 2	
Right Femur Marker 3	
Right Femur Marker 4	
Right Tibia Marker 1	Right Tibia Cluster- Attached above maximal brace height on lateral side of shank (Figure. 2.3)
Right Tibia Marker 2	
Right Tibia Marker 3	
Right Tibia Marker 4	
Right Foot Marker 1	Markers arranged in triangle formation on lateral aspect of foot posterior to Phalanges and attached to shoe (Figure 2.3)
Right Foot Marker 2	
Right Foot Marker 3	
Right Heel	Proximal aspect of Right Calcaneous
Left Femur Marker 1	Left Femur Cluster- Attached to lower 1/3 of lateral thigh, below hand resting length (Figure. 2.3)
Left Femur Marker 2	
Left Femur Marker 3	
Left Femur Marker 4	
Left Tibia Marker 1	Left Tibia Cluster- Attached above maximal brace height on lateral side of shank (Figure. 2.3)
Left Tibia Marker 2	
Left Tibia Marker 3	
Left Tibia Marker 4	
Left Foot Marker 1	Markers arranged in triangle formation on lateral aspect of foot posterior to Phalanges and attached to shoe (Figure 2.3)
Left Foot Marker 2	
Left Foot Marker 3	
Left Heel	Proximal aspect of Left Calcaneous

Table B.2: Calibration Only Markers –Used for calibration of joint centers into human body motion capture model. Removed prior to movement trials (after calibration file)

Marker Name	Anatomical Position
Right Medial Femoral Condyle	Right Medial Femoral Condyle
Right Lateral Femoral Condyle	Right Lateral Femoral Condyle
Right Medial Malleolus	Right Medial Malleolus
Right Lateral Malleolus	Right Lateral Malleolus
Right Toe	Second Metatarsophalangeal Joint of Right Foot
Left Medial Femoral Condyle	Left Medial Femoral Condyle
Left Lateral Femoral Condyle	Left Lateral Femoral Condyle
Left Medial Malleolus	Left Medial Malleolus
Left Lateral Malleolus	Left Lateral Malleolus
Left Toe	Second Metatarsophalangeal Joint of Left Foot
Right Jig	In line with heel, located on Calibration Board
Left Jig	In line with heel, located on Calibration Board

Calibration:

There were three calibration files required prior to collection of movement data. The calibration files allowed the computer and camera system to calibrate and determine where each marker was located relative to adjacent markers located on a given participant's body. A common pose was used for both static calibration files. The pose used for calibration was to have the participants stand with their feet flat on the wooden calibration standing 'jig'. Between their feet was a 50.8cm wooden block, which maintained a consistent stance width between participants as well as a wooden heel ridge that maintained consistent foot position front to back. Calibration markers were installed along either side of the right and left heel along the heel ridge line. The participants were instructed to position their arms to be raised comfortably away from midline of the body and face directly forward. The calibration files were also used for the determination of joint centres. The three calibration files are as follows:

Ankle Joint Calibration File

The first calibration procedure was to locate the ankle joint centres relative to the tibia tracking clusters. Because the application of the ankle brace completely covered the ankle malleoli; an ankle calibration was required to locate the medial and lateral malleoli using the tibia marker clusters located on the lateral aspect of the left and right shank. This calibration was taken with the participant standing on the calibration jig.

Full Body Calibration File

The second calibration was used to obtain a static pose of each participant with the full marker setup. The calibration file consisted of the participant standing motionless on the calibration jig, in the calibration pose for a few seconds in the centre of the data collection area. This pose was use to calibrate the marker tracking algorithm in the motion capture system and to obtain reference data from the subject in a neutral static pose.

Functional Joint Calibration File

Following the capture of the static pose, each participant performed a series of movements to allow for the identification of hip and knee joint centres. The functional joint calibrations required the participants to move dynamically through both the knee and hip joints with enough range to accurately locate the joint centres. Each subject performed two successful trials for the hip and the knee calibration of each limb. The hip calibration was defined as a combination of hip flexion and extension in the sagittal plane adduction and abduction in the frontal plane. The participants completed the hip calibration by moving through both sagittal and frontal plane motions in one complete trial, attempting to achieve comfortable end ranges in both planes. The knee calibration was defined as pure flexion and extension in the sagittal plane. Both movements were demonstrated to each participant prior to collection of the calibration trials. The dynamic movements were designed to mimic and achieve motion through all degrees of freedom available at the hip and the knee. (Figure B.1 and B.2)

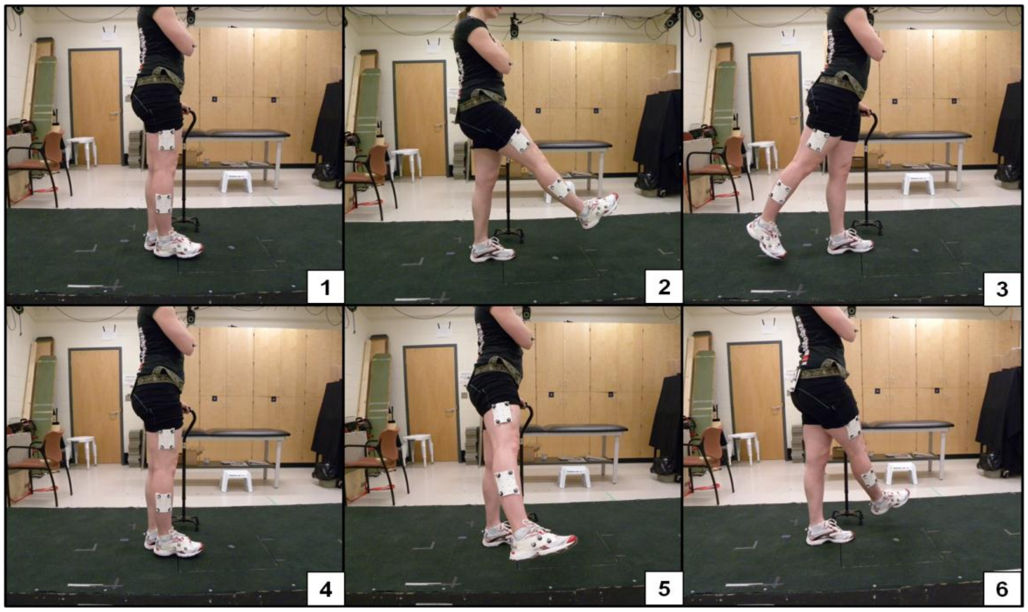

Figure B.1: Representative trial of a functional hip calibration (*Numbers indicate progression movement sequence) outlining the key components of the movement including 1) ready position; 2) hip flexion; 3) hip extension ; 4) brief pause and transition to frontal plane movement; 5) hip abduction; 6) hip adduction.

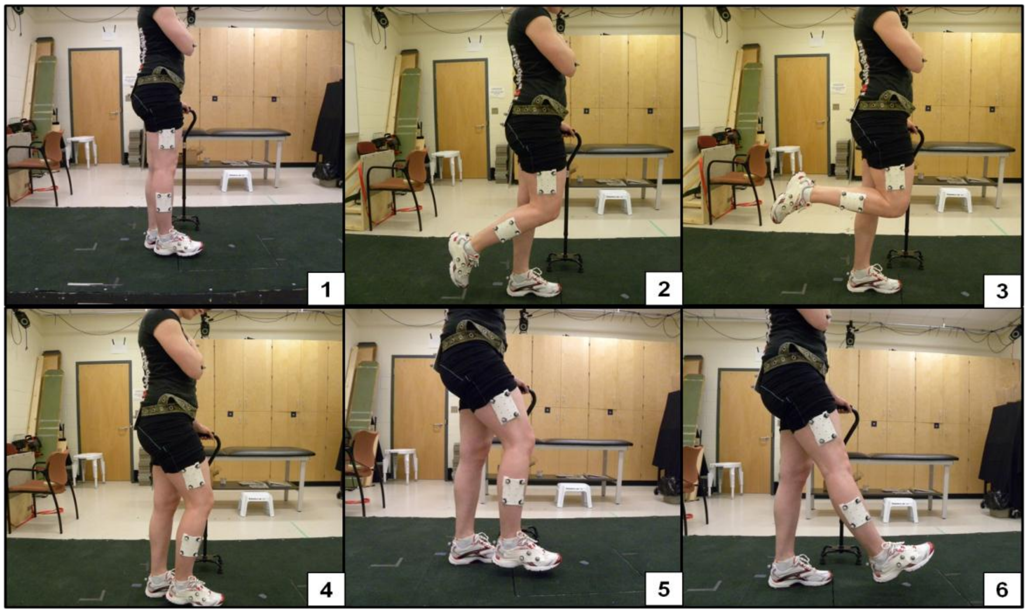

Figure B.2: Representative trial of a functional knee calibration (*Numbers indicate progression movement sequence) outlining the key components of the movement including 1) ready position; 2) initial knee flexion; 3) full knee flexion ; 4) brief pause and transition to extension; 5) initial knee extension; 6) full knee extension.

Joint Segment Definitions:

The marker position data obtained during the three calibration types were used to define the location for the ankle, knee and hip joints axis. With the kinetic data obtained, defining the functional joint centres was possible along with defining the anatomical coordinate systems and subsequently the joint rotations could be described.

Functional Joint Centres

The functional joint centres were estimated based on the calibration files using the methods described by O'Brien (2000) and Ehrig, Taylor, Duda, & Heller (2007). Using the prominent aspects of the medial and lateral malleoli as well as the medial and lateral condyles of the femur, anatomical locations of the joint were defined. Motion capture markers were placed onto the palpated landmarks to define the anatomical frames of the ankle and knee axis respectively for the definition of the anatomical functional calibration (Cappozzo, Catani, Croce,

& Leardini, 1995). The midpoint between the medial and lateral malleoli was identified as the ankle joint centre in the tibial coordinate reference frame.

For the knee joint centre, the midpoint between the condyles was identified as a temporary joint centre. From the flexion extension movement performed during the functional knee calibration session, a flexion extension (F/E) axis was determined based on the method described by O'Brien (2000). The temporary joint centre was then projected perpendicular onto the functional F/E axis to define the functional knee joint centre (Ehrig, Taylor, Duda, & Heller, 2007). The hip joint centre was identified following the method described by O'Brien (2005) using the functional hip calibration, and imputing the data into a rotary joint model.

Anatomical Coordinates

The pelvis coordinate system was created with the full body standing calibration file using the standing jig and the global coordinate system. The origin of the pelvis is located halfway between the hip joint centres. The vertical axis (z) is created from the global vertical axis. The lateral axis (y) of the pelvis goes from the right side to the left side of the body based on the markers placed along the heel ridge of the standing calibration jig. The anterior posterior (AP) axis (x) is the y-z cross product result.

For the femur coordinate systems the vertical axis (z) goes from the knee centre to the hip centre. For the left femur the lateral axis (y) goes from lateral to medial away from midline. The AP (x) is the result of the cross product of the y and z-axes. The tibia's coordinate systems are defined by the vertical (z) axis going from the ankle joint centre to the knee joint centre. For the left tibia the lateral axis (y) is going from medial to lateral. The AP (x) is the result of the cross product of the y and z-axes.

For the foot coordinate system the vertical axis (z) is a translation of the global vertical axis. The AP axis (x) is directed from the heel to the toe, defined by the heel and toe markers used in calibration. The lateral axis (y) is the result of the cross product of z and x-axes.

Appendix C: Visual Cues

This appendix outlines the 3 visual cues which the participants observed prior to movement. The visual cue indicated the direction in which to proceed.

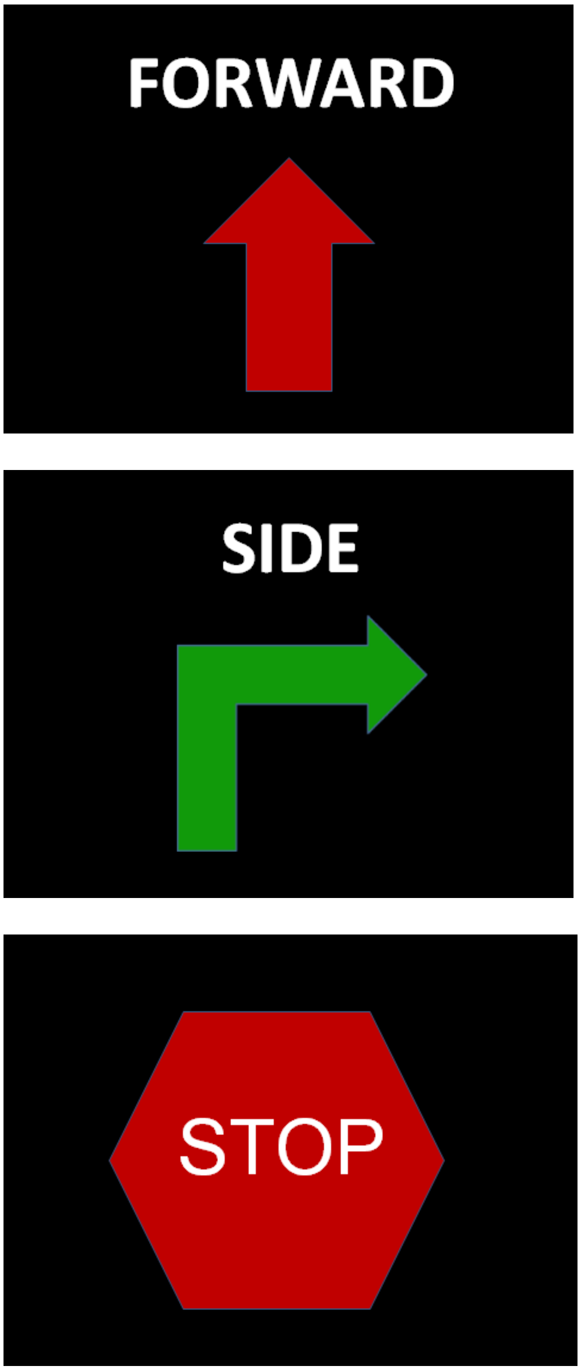

Figure C.1: Visual Cues used to direct participant movement

Appendix D: Joint ROM and Moment Figures:

This appendix contains all joint kinematic and joint motion information for reference use.

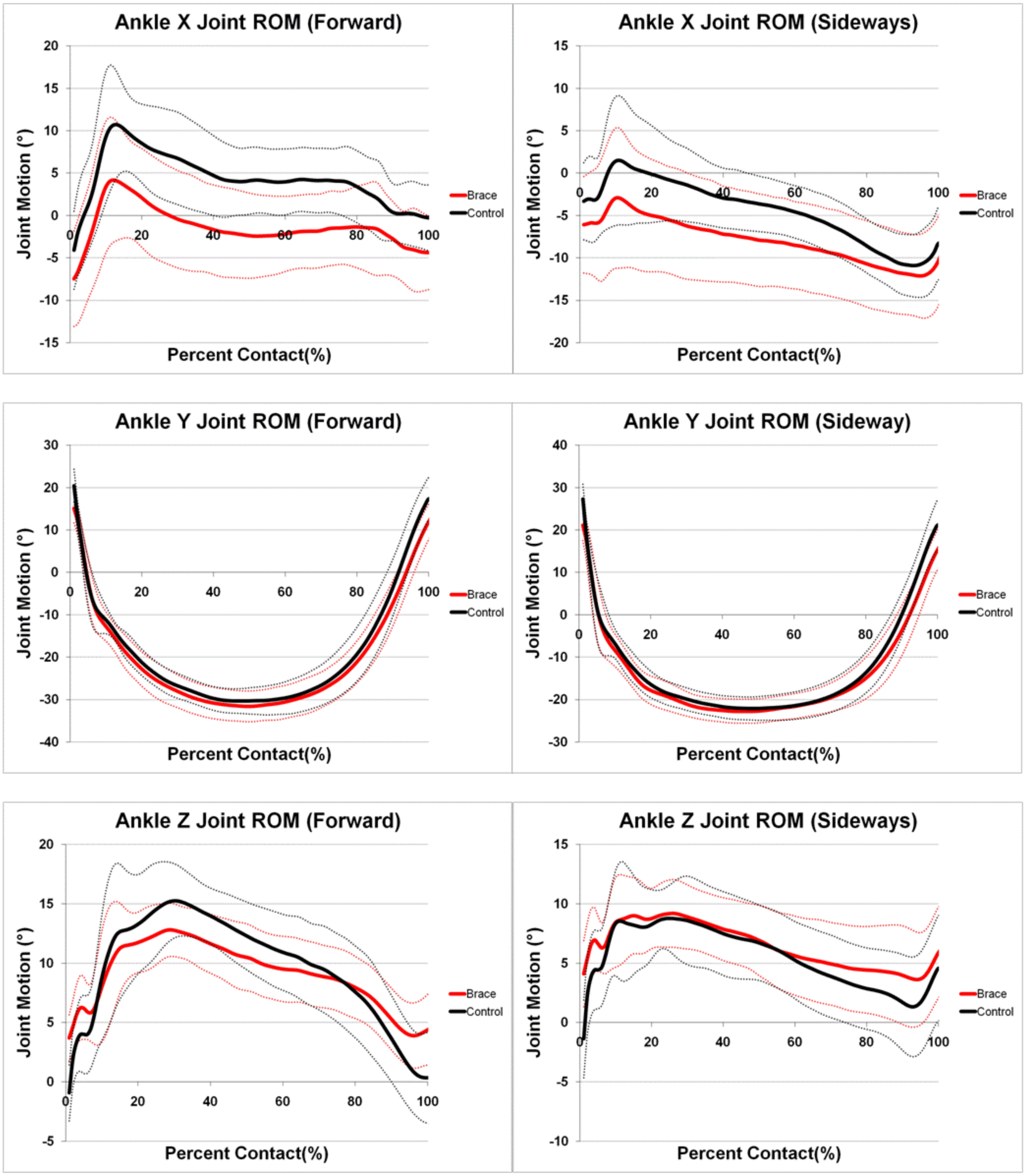

Figure D.1 Overall Ankle Joint Kinematics (Forward & Sideways) Note: These graphs are overall means of all subjects. Angles are presented as the difference from quiet standing angle:

Ankle X- (+) represents eversion
Ankle Y- (+) represents plantar flexion
Ankle Z - (+) represents external rotation

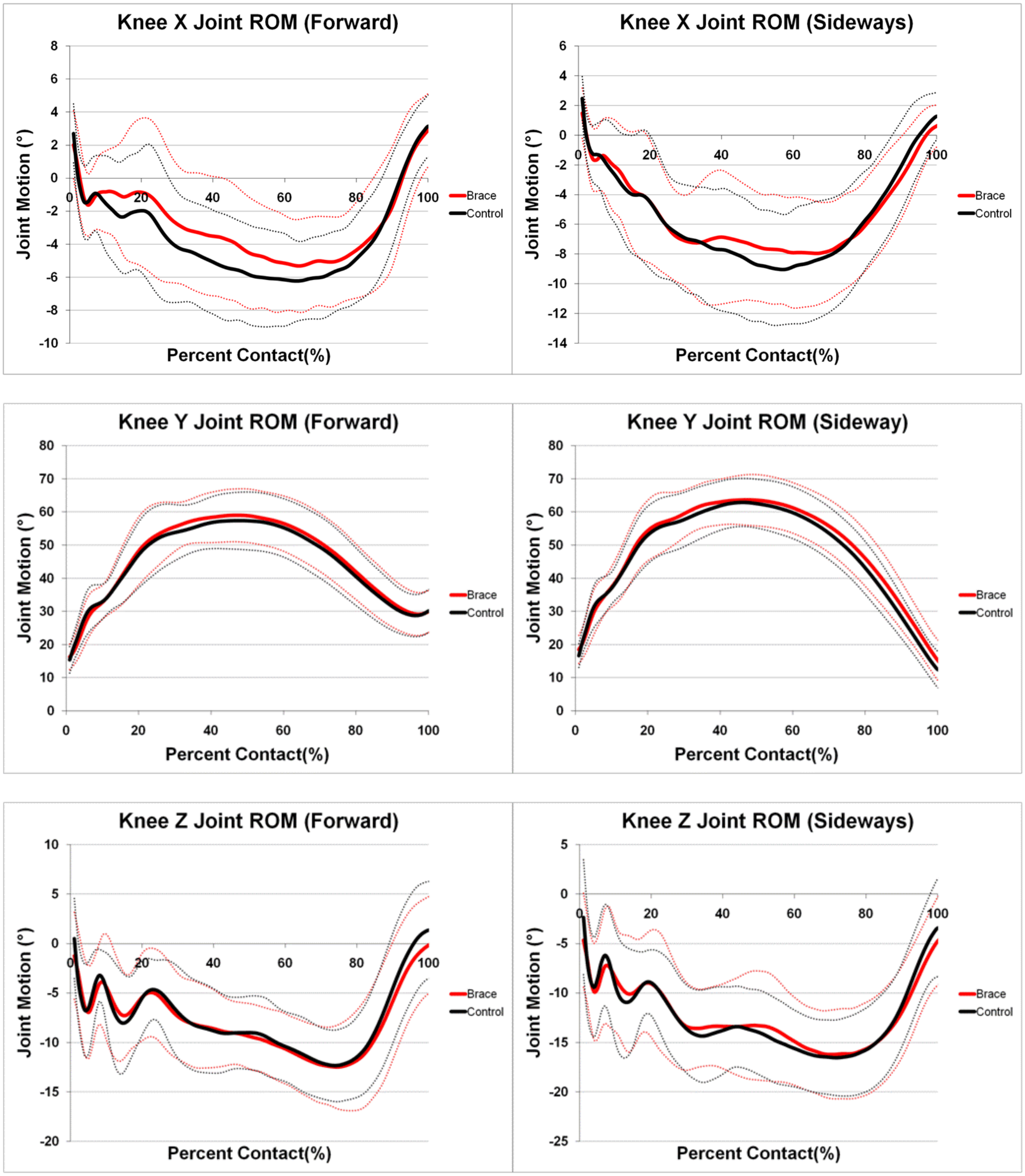

Figure D.2: Overall Knee Joint Kinematics (Forward & Sideways) Note: These graphs are overall means of all subjects. Angles are presented as the difference from quiet standing angle:

Knee X- (+) represents abduction
Knee Y- (+) represents flexion
Knee Z - (+) represents external rotation

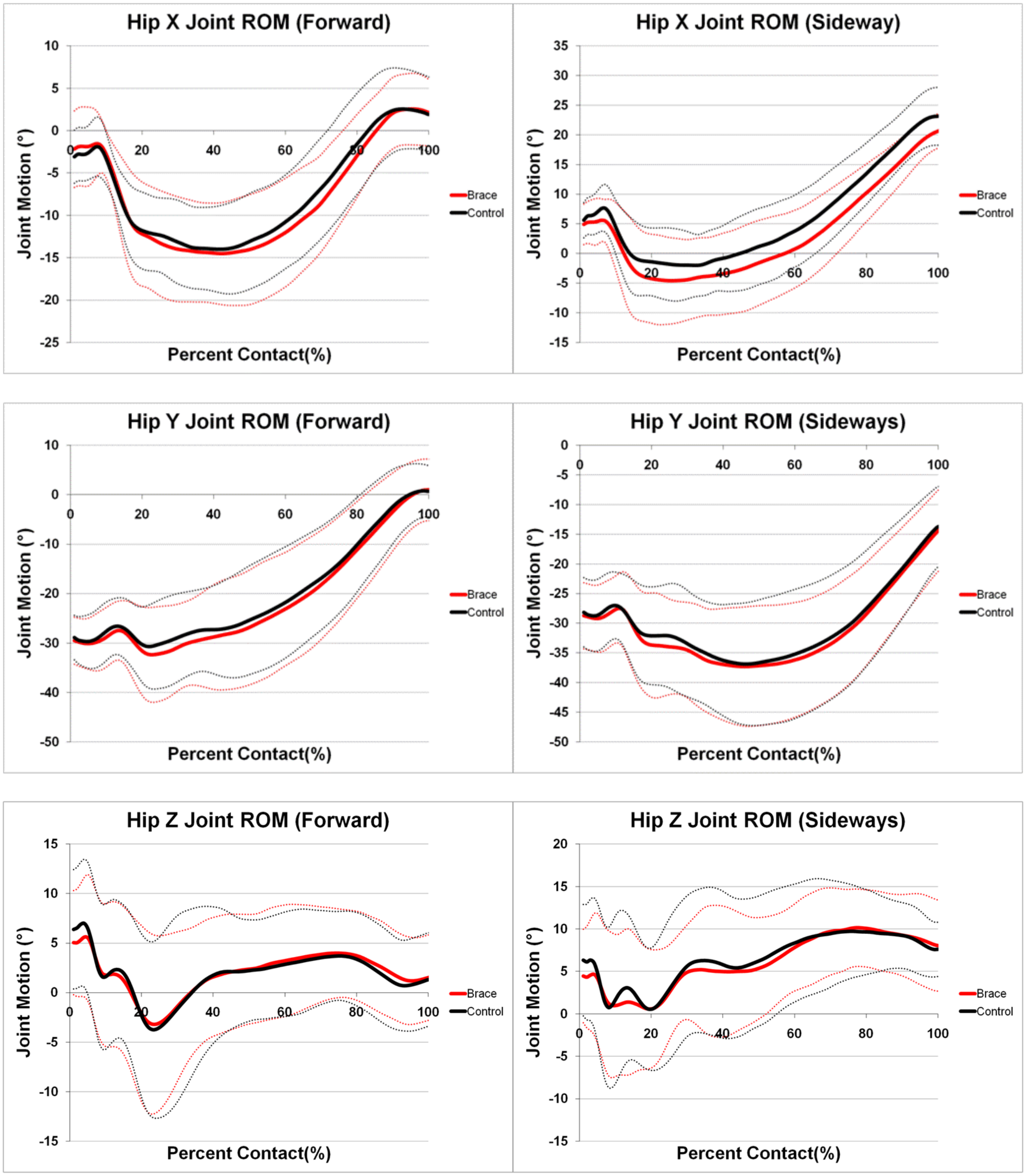

Figure D.3: Overall Hip Joint Kinematics (Forward & Sideways) Note: These graphs are overall means of all subjects. Angles are presented as the difference from quiet standing angle:

Hip X- (+) represents abduction
Hip Y- (+) represents extension
Hip Z - (+) represents external rotation

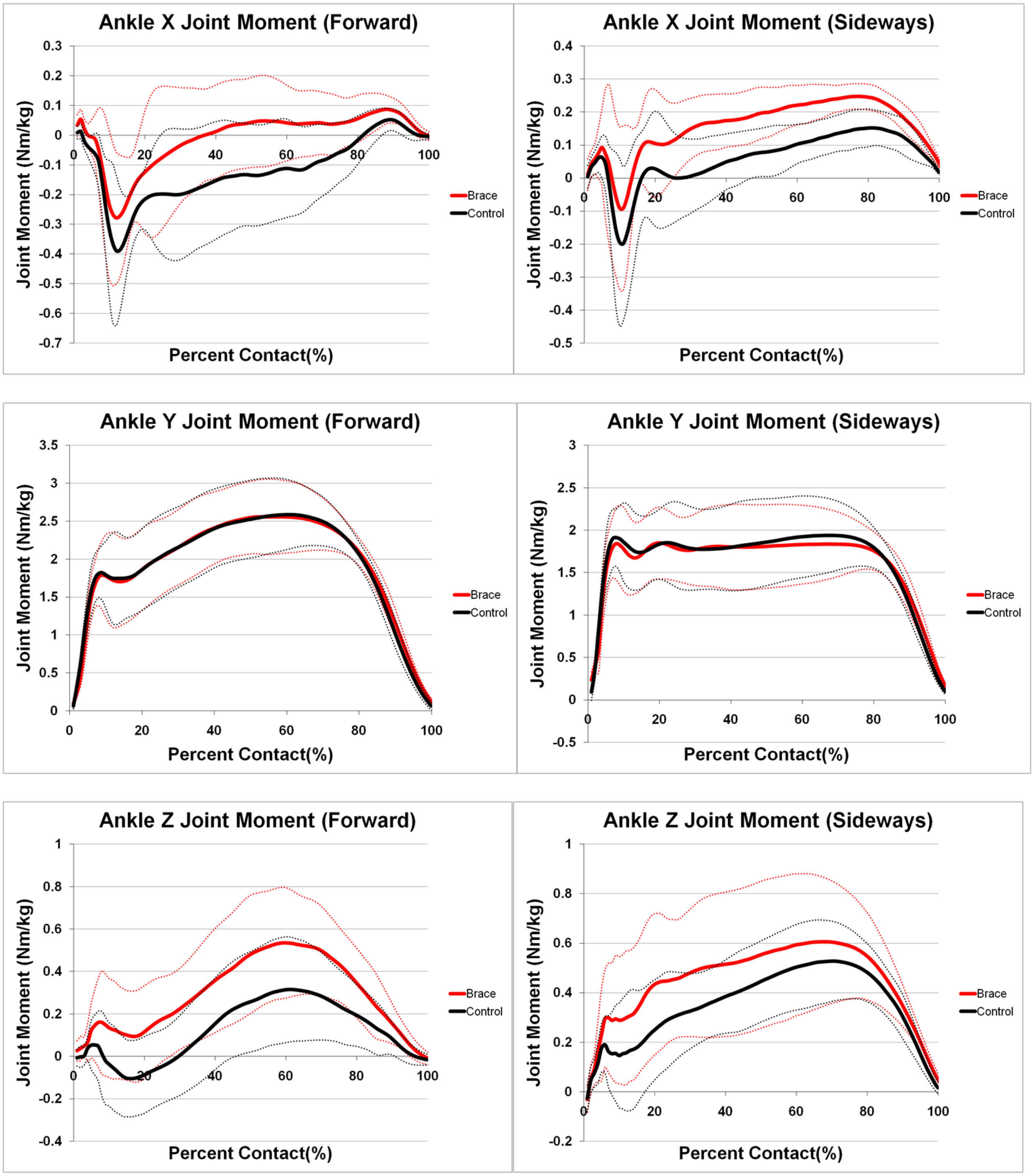

Figure D.4: Overall Ankle Joint Moments (Forward & Sideways) Note: These graphs are overall means of all subjects. Moments values are presented as Newton meters per kilogram (N·m/kg):

Ankle X- (+) represents eversion moment
Ankle Y- (+) represents plantar flexion moment
Ankle Z - (+) represents external rotation moment

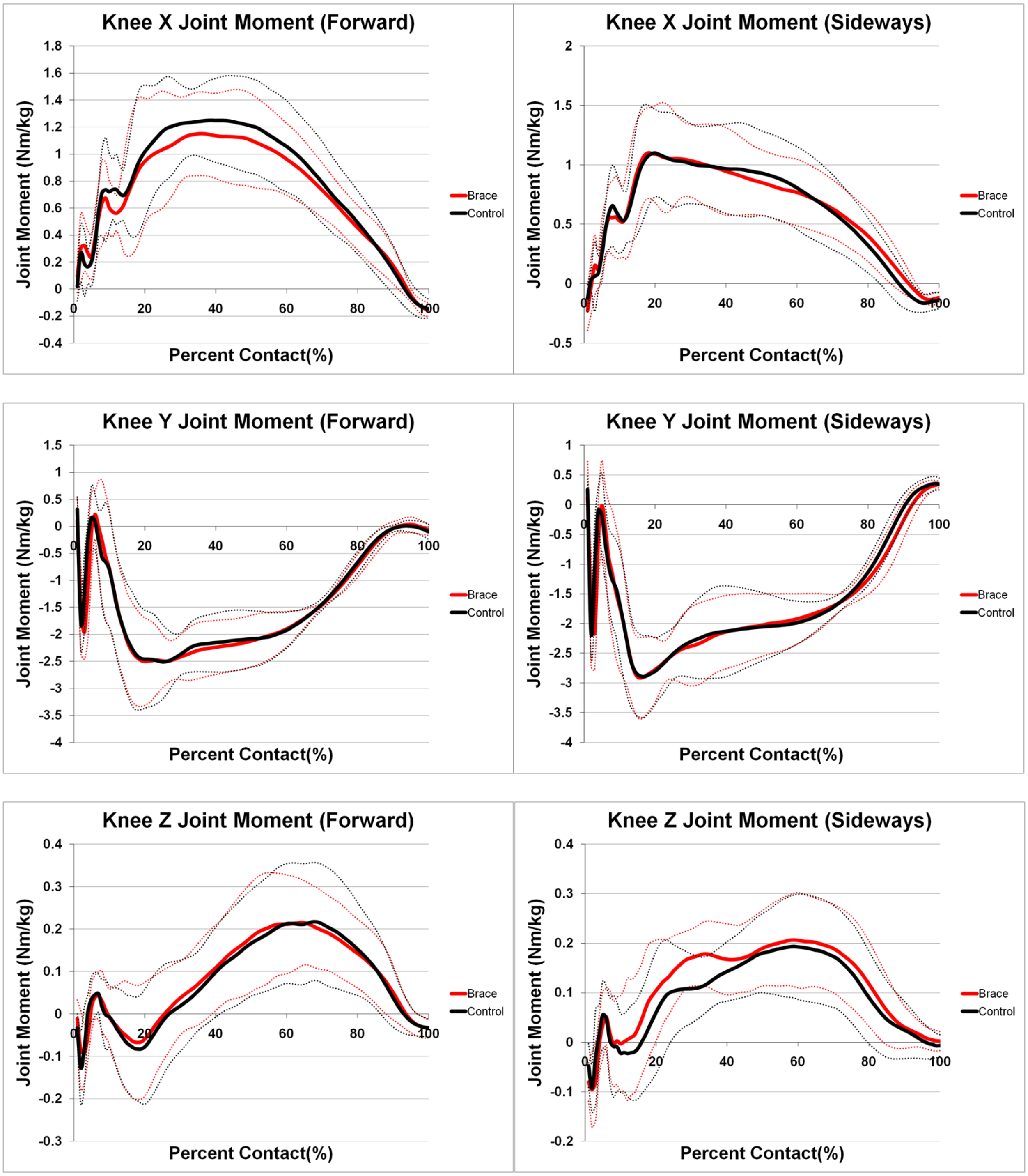

Figure D.5: Overall Knee Joint Moments (Forward & Sideways) Note: These graphs are overall means of all subjects. Moments values are presented as Newton meters per kilogram (N·m/kg):

Knee X- (+) represents abduction moment
Knee Y- (+) represents flexion moment
Knee Z - (+) represents external rotation moment

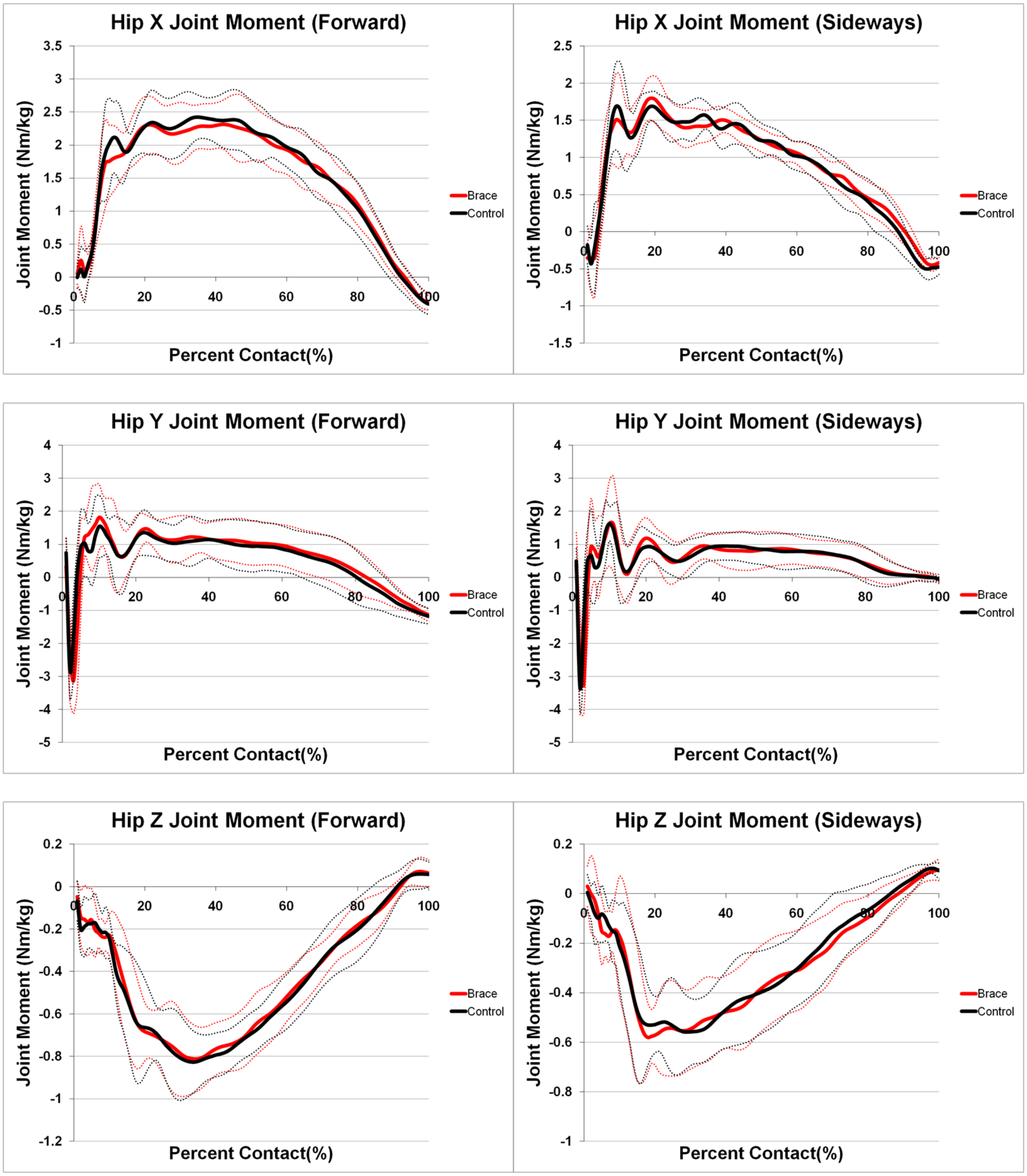

Figure D.6: Overall Hip Joint Moments (Forward & Sideways) Note: These graphs are overall means of all subjects. Moments values are presented as Newton meters per kilogram (N·m/kg):

Hip X- (+) represents abduction moment
Hip Y- (+) represents extension moment
Hip Z - (+) represents external rotation moment

Appendix E: Current Huskie Women's Basketball Training Program

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Huskie Women's Basketball
1

Warm-Up:

Your choice – make sure you do it!!!

Strength:

EXERCISE	SETS & REPS
DB Bench Press	3 x 10
Towel Landmine Row	3 x 10
Arm Only 2 Arm Jammer	3 x 10
2 Arm Pulldown – elbows tight	3 x 10
REST	1:00 min
3 Point Lunge (fwd/side/bkwd)	3 x 6 each leg
DipShits	3 x 6 each leg
Sumo Squat X-Over Step Up	3 x 3 each leg
Single Leg Hockey Stride Slides	3 x 6 each leg
REST	1:00 min
Push Ups	3 x 15
Turkish Get Up	3 x 6 each side
Pull Ups	3 x 5
REST	1:00 min

Core

Huskie Women's Basketball
2

Warm-Up:

Your choice – make sure you do it!!!

Strength:

EXERCISE	SETS & REPS
DB Bench Press (1 arm at a time)	3 x 8 (each arm – keep 1 arm straight)
Pull Ups	3 x 4
Push Ups	3 x 10
Med Ball to the Wall	3 x 20
Seated Row	3 x 10
REST	1:00 min
Front Squats	3 x 10
DipShits	3 x 6 each leg
Step Up with (25 lb plate press)	3 x 6 each leg
Lunge Walk (holding 25 lb DB's)	3 x 6 each leg
REST	1:00 min
Turkish Get Up	3 x 6 each side
Dips	3 x 6
REST	1:00 min

Core

Huskie Women's Basketball
3

Warm-Up:

Your choice – make sure you do it!!!

Strength:

BB Bench 4 x 10 + 10 push ups

BB Squat 4 x 10 + 5 squat jumps

Circuits:

EXERCISE	SETS & REPS
Cross Body Jammer	3 x 8 each way
TBall Hamstring Curls	3 x 10
Cleans or High Pulls	3 x 5
Landmine Press with Lunge	3 x 8 each way
1 Arm DB Swings (explode hips)	3 x 8 each side
Bulgarian Squats (holding 25 lb)	3 x 8 each side
Curl and Press	3 x 10
Turkish Get Up	3 x 6 each side

Core

Huskie Women's Basketball
4

Warm-Up:

Your choice – make sure you do it!!!

Strength:

EXERCISE	SETS & REPS
DB Bench Press (1 arm at a time)	3 x 8 (each arm – keep 1 arm straight)
Dipshit with 1 Arm Row	3 x 8 (8 rows in dipshit pos'n)
Push Ups	3 x 10 (Clap every 2^{nd})
Pull Ups	3 x 5
REST	1:00 min
Lunge Walk with Overhead Press	3 x 5 each leg
BB Front Squats	3 x 10
Sumo Squat X-Over Step Up	3 x 5 each leg
REST	1:00 min
1 Arm DB Snatch	3 x 3 each side
Turkish Get Up	3 x 3 each side
REST	1:00 min

Core

HPC

HUMAN PERFORMANCE CENTER

Huskie Women's Basketball
5

Warm-Up:

Your choice – make sure you do it!!!

Strength:

EXERCISE	SETS & REPS
Push Ups	3 x 20
Pull Ups	3 x 5
Med Ball to the Wall	3 x 20
Close/Reverse Grip Pulldown	3 x 10
REST	1:00 min
Bulgarian Squats	3 x 5 + 5 jumps
T-Ball Hamstring Curls	3 x 10
3 Point Lunge (fwd/side/bkwd)	3 x 3 each leg
REST	1:00 min
2 Arm DB Swings	3 x 10
Turkish Get Up	3 x 3 each side
REST	1:00 min

Core