JENNIFER WILDER
A comparison of kinematic restraint of two prophylactic ankle braces provided during flat and inverted drop surface landings
(Under the direction of KATHY J. SIMPSON)

The passive restraint provided by two prophylactic ankle braces during drop landings was compared. The angular kinematics of 27 participants were generated for three brace (Malleoloc™ = modified stirrup design, Active Ankle™ = hinge design and no brace = control) and two platform (flat and 30° inverted) conditions. From the 3 x 2 repeated measure ANOVAs ($\underline{p} < 0.05$), no significant differences were detected between the braces for in/eversion motion. However, the braces demonstrated less maximum inversion and inversion displacement than the control. The Malleoloc™ brace exhibited less maximum dorsiflexion and dorsiflexion angular displacement than either the Active Ankle™ or the control condition. Therefore, while there were no in/eversion differences between the hinge and the modified stirrup design, the hinge design allowed more natural dorsiflexion motion.

INDEX WORDS: Ankle brace kinematics, Passive restraint, Drop landings

A COMPARISON OF KINEMATIC RESTRAINT OF TWO PROPHYLACTIC ANKLE BRACES PROVIDED DURING FLAT AND INVERTED DROP SURFACE LANDINGS

by

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CHAPTER I

INTRODUCTION

For individuals who have had a previous ankle injury, ankle braces can reduce the number of people who will re-sprain their ankles by 40% (Sitler et al., 1994; Surve, Schwellnus, Noakes & Lombard, 1994). This preventative measure is beneficial, as 20% to 45% of individuals who have experienced an ankle sprain will continue to experience reoccurring ankle sprains (Hollis, Blasier & Flahiff, 1995; Löfvenberg, Kärrholm, Sundelin & Ahlgren, 1995; Renstrom & Konradsen., 1997). Thus, it is desirable to further reduce the incidence of chronic ankle sprains, which can be accomplished by understanding the factors related to effective brace design.

Although the exact cause of ankle sprains is not known, one situation that has been reputed to create an ankle sprain is landing with the foot-ankle complex in a plantarflexed and slightly inverted position onto an uneven surface e.g., another person's foot (Garrick, 1977; Shapiro, Kabo, Mitchell, Loren & Tsenter, 1994). As the landing continues, the foot-ankle complex rapidly inverts and dorsiflexes. The resulting stress placed on the foot-ankle complex is particularly detrimental to two ligaments. The ligament surmised to be torn first is the anterior talofibular ligament (ATFL), followed by the calcaneofibular ligament (CFL) (Renstrom & Konradsen, 1997; Rubin & Sallis, 1996). The location of these ligaments over the lateral malleolus makes them susceptible to injury due to tensile loading during inversion. While inversion displacement stresses both the ATFL and CFL, the tensile loading increases in the ATFL when the foot is in the plantarflexed position, whereas dorsiflexion of the foot stresses the CFL (Colville Marder, Boyle & Zarnis, 1990; Siegler, Chen & Schneck, 1988; Stormont, Steger, Stüssi & Reinschmidt, 1985). Thus reducing strain to these ligaments via a brace involves changes in foot position and reducing displacement in the plantar/dorsiflexion direction as well as to the inversion/eversion direction during high impact landings. Changes in the mechanical properties of the ATFL and CFL and other tissues damaged during an inversion sprain (e.g., joint capsule) are a primary reason that prophylactic ankle bracing is effective only for preventing sprains to previously sprained ankles (Alves, Alday, Ketcham & Lentell, 1992). An acute inversion sprain or repeated strains loads applied at or above the yield point can damage the ligament, thereby changing its inherent stiffness and permanent length (Nordin & Franklin, 1989). After such damage, the ligament exhibits changed mechanical properties: less stiffness and increased laxity due to permanent deformation (Karlsson, Peterson, Andreasson, & Högfors, 1992). For individuals with chronic ankle instability, the passive restraint provided by an ankle brace during situations of high tensile loading to the ATFL and CFL (e.g., landing on an uneven surface) is thought to effectively reduce the strain to these tissues (Alves et al., 1992; Siegler, Liu, Sennett, Nobilini & Dunbar, 1997). Consequently, in the current investigation only individuals who had previously sprained their right ankles participated.

The inherent properties of viscoelastic tissues, such as ligaments, can be modified not only by injury but temporarily by strain rate (Nordin & Franklin, 1989). Therefore, to determine the passive restraint of a prophylactic ankle brace, it is important to simulate actual landing conditions, i.e., typical, non-injurious landings onto a flat surface as well as landings on an uneven surface similar to atypical, injurious landing. To investigate the effects of typical and uneven surface landings on the ATFL and CFL, Self (1996) performed mechanical drop tests using the lower leg and foot of cadavers while measuring the strain and strain rates for the ATFL and CFL. For landing conditions requiring more inversion, i.e. landing onto a 30º inverted V platform compared to landing onto a flat platform, the strain rates and strain displacements for both ligaments increased, particularly for the ATFL. Similar to Self, the landing conditions of this study are comprised of a 0º and 30º inverted landing surface, which simulates, respectively, a typically non-injurious landing and a landing similar to landing on another person's foot.

Therefore, it is assumed for this study that the maximum angular inversion/eversion displacements and time to maximum inversion of the rearfoot segment relative to the lower leg are indirect measures related to the strain and strain rate for the ATFL and CFL (Self, 1996).

During its use, a prophylactic ankle brace provides passive restraint to the foot-ankle complex and reduces ROM (in/eversion and in/external directions) and angular velocity for inversion motion (Alves et al., 1992). To date, it has been established for laboratory situations that semi-rigid brace designs provide greater passive restriction than other brace designs, e.g., tape and non-rigid designs (Alves et al., 1992). Also, the semi-rigid brace has been shown to significantly decrease angular velocity during landing activities compared to a non-rigid brace (Podzielny & Hennig, 1997) and to not wearing a brace (Podzielny & Hennig, 1997). The decreased angular velocity is surmised to delay the time to maximum inversion angle, thereby giving more time for the peroneal muscles to create an opposing evertor torque (Podzielny &Hennig, 1997). Therefore, for this study increased time to maximum inversion was anticipated to occur while wearing either the Malleoloc™ or the Active Ankle™ compared to not wearing a prophylactic ankle brace.

While motion restraint for the inversion/eversion direction of motion while wearing a prophylactic ankle brace is surmised to help protect the foot-ankle complex (Alves 1992; Siegler 1997), motion restraint provided by a prophylactic ankle brace for the plantar/dorsiflexion direction of motion has potential positive and negative consequences. Limiting plantarflexion with a semi-rigid brace was observed to reduce the plantarflexion angular velocity by 110 °/second compared to a no brace condition, which was hypothesized to reduce the plantar/dorsiflexion external torque acting on the foot-ankle complex (McCaw & Cerullo, 1998). A second, potentially positive consequence of limiting plantarflexion is to decrease the moment arm of the applied external ground reaction force to the foot-ankle complex during landing (Shapiro et al., 1994). Therefore, for this study, it was assumed that the foot-ankle complex should have a lower plantarflexion angle at touchdown to help reduce the strain of the ATFL at contact with the platform.

In contrast, one hypothetical negative effect of limiting plantarflexion is that the body's natural ability to absorb the external torque through the musculature of the ankle, knee and hip may be hindered. When the ROM of the ankle plantarflexion is restricted, the energy to be absorbed by the knee and hip extensors increases (McCaw & Curello, 1998). However, no data exist to date that support the premise that limited plantarflexion causes knee injury (Feuerbach, Ludin & Grabiner, 1993).

Therefore, while it is likely that, in general, semi-rigid braces can provide greater passive motion resistance to external inversion torques than non-rigid or tape, it is not known if design enhancements to the semi-rigid design can improve or decrease inversion motion restraint or change the magnitude of plantar/dorsiflexion motion. Within the category of semi-rigid braces, to date no comparisons have been made between a modified stirrup (Malleoloc ™) brace and a hinged stirrup (Active Ankle™) brace during a dynamic situation, e.g., drop landing. The introduction of a hinge on a stirrup brace may allow increase plantar/dorsiflexion movement compared to the modified semi-rigid brace design. However, it is not known how a hinge influences passive motion restraint during dynamic situations in all directions of motion.

The Malleoloc™ is of a modified stirrup design (Bauerfeind Corporation) for use as a preventive orthosis. The design is specifically constructed to fit such that the lateral stirrup is anterior to the lateral malleolus and the medial stirrup is posterior to the medial malleolus. The lateral stirrup is positioned over the ATFL, theoretically for greater passive restraint in the combined inversion and dorsiflexion directions of motion. The Active Ankle™ is a stirrup design hypothesized to allow free range of motion in the plantar/dorsiflexion direction, due to the placement of a mediolateral axis hinge at the lateral malleolus height, while being able to restrict inversion/eversion motion of the foot-ankle complex.

Purpose of the Study and Hypotheses

There are two purposes for this research regarding semi-rigid braces. The first purpose was to determine whether there are differences in rearfoot kinematics while wearing different prophylactic ankle brace designs. The second purpose is to determine the effect of landing on a sideward slanted slope (similar to landing on an uneven surface) verse landing on a flat surface on the passive restraint provided by the two semi-rigid braces. The following hypotheses will be used to test the first purpose:

1. Both braces (Malleoloc™ and Active Ankle™) will allow less maximum inversion and inversion angular displacement than the control condition (no brace) during both landing conditions.
2. The Active Ankle™ will exhibit a longer time to maximum inversion than the Malleoloc™ brace and control condition.
3. The Malleoloc™ will exhibit a greater time to maximum inversion than control condition.
4. The Malleoloc™ will demonstrate less maximum plantarflexion, maximum dorsiflexion and dorsiflexion angular displacement than the Active Ankle™ brace and control condition.
5. Due to the hinge design, the Active Ankle™ brace will not exhibit more than 2° difference from the control condition for maximum plantarflexion and maximum dorsiflexion and 4° for dorsiflexion angular displacement.

The following hypothesis will be used to test the second purpose:

1. Compared to landing on a flat surface, landing on a 30° inverted slope will cause an increase in maximum inversion, time to maximum inversion, inversion angular displacement, but will not change the maximum plantarflexion or maximum

dorsiflexion by more than 2°, dorsiflexion angular displacement by more than 4° and time to maximum dorsiflexion by more than 5% of the total time for all brace conditions.

Significance of the Study

Dynamic tests simulating actual landing conditions while wearing semi-rigid prophylactic braces provide similar kinematic movement patterns seen during physical activity, in contrast to the movement using limited, closed chain movements, e.g., passive joint ROM evaluations. Furthermore, passive ROM measurements are of limited use, as a direct link between the results of passive measurements and dynamic measurements does not exist (Siegler et al., 1997). As further evidence, decreased range of motion in the plantar flexion/dorsiflexion and inversion/eversion directions during passive range of motion evaluations were reported for the Active Ankle™ (Siegler et al., 1997) and Malleoloc™ brace (Johnson, Veale & McCarthy, 1994; Wiley & Nigg, 1996). However, Simpson, Cravens, Higbie, Theordorou and DelRey (1999) observed that the Malleoloc™ did not restrict maximum inversion or maximum velocity for sideward motions compared to non-rigid brace values or to no-brace condition values. Having participants land onto an uneven surface similar to the uneven surfaces encountered during physical activity provides a better understanding of whether or not these braces constrain movement. Furthermore, it is not known how the design of either a modified stirrup (Malleoloc™ brace) or a hinge design stirrup (Active Ankle™ brace) influences the passive motion restraint.

Until now, it has been difficult to obtain valid measures of the rearfoot motion, due to the triplanar nature of the calcaneal motion. Hence, prophylactic ankle brace studies have only utilized two-dimensional methodology (Nawoczenski, Owen, Ecker, Altman & Epler, 1985) or only measured the motion of the shoe (Nawoczenski et al., 1985; Simpson et al., 1999). However, as investigators believe that to prevent excessive tensile loading to the ATFL and CFL requires a decrease in rearfoot displacement, it is important to measure the displacement of the foot.

Another methodological factor influencing the validity of past studies is the placement of the markers used to track the motion of the rearfoot. Typically, the displacement of the rearfoot has been measured by using markers on the shoe (Nawoczenski et al., 1985; Simpson et al., 1999). However, the brace stabilizes the foot and not the shoe, and therefore skin markers should provide more accurate data (Reinschmidt, Stacoff & Stüssi, 1992). For example, inversion motion exhibited during running and sideward cutting maneuvers have been shown to be overestimated when markers were on the shoe compared to inversion values obtained from foot marker data (Reinschmidt et al., 1992; Stacoff, Steger, Stüssi & Reinschmidt, 1996). Furthermore, the presence of a brace inside the shoe may cause the shoe to move differently than the foot. Therefore, to accurately measure movements of the foot-ankle complex, the markers during this investigation will be placed on the participant’s skin. Thus, this study is the first to obtain a direct measure of the rearfoot motion that occurs during semi-rigid brace wear, and, therefore should provide accurate estimate of the motion restraint provided by two types of prophylactic ankle braces than past studies.

Assumptions

It has been suggested that the reduction of subsequent injury when an ankle brace is worn is due not just to passive motion restraint but also to increased stimulation to cutaneous receptors of the foot-ankle complex (Freeman, 1965). "Enhanced proprioception" is typically described as an increase in peroneal muscular activity or an increase in joint position sense. However, enhanced muscle activity is questioned because of contradictory findings among the studies (Karlsson & Andreasson, 1992; Stüssi, Tiegermann, Gerber, Raemy & Stacoff, 1987), particularly for studies using dynamic movements in which increased peroneal activity due to proprioceptive input is very difficult to detect (Karlsson et al., 1992; Springings, Pelton & Brandell, 1981).

Furthermore, there is conflicting evidence to conclude that passive joint position sense is related to injury prevention during dynamic movements (Feuerbach, Grabiner, Kohn & Weiker, 1994; Simoneau, Degner, Kramper & Kittleson, 1997). For example, in an investigation by Feuerbach et al. (1994), anesthetized and non-anesthetized ligament conditions exhibited no differences between scores for replicated inversion foot-ankle position and original inversion foot-ankle position. Feuerbach et al. (1994), however, observed that increased accuracy occurred for joint position sense for the brace versus the non-braced condition. As the mechanoreceptors were not functioning during the anesthetized ligament condition, Feuerbach et al. concluded that the increased proprioception during the brace wear condition compared to the non-braced condition wear must have been due to increased cutaneous stimulation rather than mechanoreceptors.

In further support of the role of cutaneous stimulation, Simoneau et al. (1997) determined that the ability of the ankle joint to recognize joint position in a non-weight bearing condition was more accurate when athletic tape strips were placed on dorsum of the foot compared to a no tape condition. Yet, when changing to a weight-bearing situation, the presence of tape had no significant influence on joint position (Simoneau et al., 1997). Therefore, due to weight bearing and dynamic nature of the task and due to the lack of evidence supporting the concept of braces causing an increased peroneal muscle activity due to proprioceptive input, the focus of the investigation will be on the motion restraint provided by the brace with minimal regard to any potential proprioception.

CHAPTER II

REVIEW OF LITERATURE

The prescription of a particular prophylactic stabilizer is based on several elements, such as the mechanism of ankle injury and the types and severity of tissues damaged. Hence, this chapter contains the following topics: a) motions of the foot-ankle complex, b) surmised causal mechanisms of ankle sprains, c) tissues damaged during inversion sprains, d) *in vitro* ligament studies, e) opposing muscles to sudden inversion, and f) muscle reflexes during sudden inversion movements. Next, the two rationales underlying the efficacy of prophylactic stabilizer aids are considered. Finally, the methodological considerations unique to this investigation also are addressed.

Motions of the Foot-Ankle Complex

The foot-ankle complex is a combination of the talocrural joint and the subtalar joint. Although the talocrural and subtalar joints are two distinct articulations, they work together to function as a unit (Hamill & Knutzen, 1995). Movements about these two joints and several others within the foot combine to produce the movements of in/eversion, plantar/dorsiflexion and ab/adduction. Specifically, the talocrural (ankle) joint is a collective configuration of the tibia, fibula and talus. This unique complex is usually described as three articulations (the tibiofibular joint, the tibiotalar joint, and the fibulotalar joint) that produces motion, primarily although not solely, in the plantar/dorsiflexion direction (Hall, 1999). Plantarflexion is the extension of the foot away from the lower leg, while dorsiflexion is movement of the foot towards the lower leg.

The movements of the subtalar joint are also complex as this joint utilizes triplanar motion (Rockar, 1995). The two primary motions occurring about the subtalar joint are in/eversion and ab/adduction of the foot-ankle complex. During inversion, the plantar surface of the foot turns inward toward the midline of the body. Eversion is the opposite motion as the plantar surface of the foot turns outward away from the midline of the body. During adduction, the foot moves toward the midline of the body, while abduction moves the foot away from the midline of the body.

The representation of plantar/dorsiflexion and in/eversion axes as cardinal or orthogonal axes do not accurately represent the true directions of these axes. However, determination of the true locations of these non-orthogonal axes is extremely difficult and unique to each individual (Rockar, 1995). Therefore, this is a simplified explanation of each movement and its corresponding axis of rotation. For this study, it was assumed that plantar/dorsiflexion, in/eversion and ab/adduction occur about the mediolateral, longitudinal, and anteroposterior axes of the foot, respectively, from anatomical position.

Injury Mechanisms and Risk Factors

An ankle sprain is defined as damage to soft tissue e.g., ligaments and tendons. However, the exact mechanical cause of an ankle sprain is relatively unknown, consequently making the etiology difficult to quantify. Yet, several situations arise that typically cause the ligaments of the ankle to become injured. One of the most common situations for an ankle to be sprained occurs when an individual lands forcefully and unexpectedly onto an uneven surface or another person's foot, with the foot-ankle complex initially in a plantarflexed and slightly inverted position (Garrick, 1977; Renstrom & Konradsen, 1997; Shapiro et al., 1994). According to Tropp, Askings and Gilgquist, (1986) as the impact force increases the vertical ground reaction force vector (VGRF) also shifts towards the lateral edge of the plantar surface of the foot, increasing the VGRF moment arm. The resulting inversion torque causes excessive loading of the lateral tissues of the foot-ankle complex and possibly to the evertor muscles, i.e., peroneus longus and peroneus brevis (peroneals).

Risk factors for an inversion sprain can be categorized into extrinsic and intrinsic characteristics (Lysens et al., 1984). Extrinsic characteristics include the type of sport, environmental conditions, and playing time. These risk factors explain the prevalence of inversion sprains in basketball, volleyball, and soccer, as the opportunity to land unexpectedly increases due to the parameters of the game. Physical attributes such as age, gender, joint stability and isokinetic strength describe intrinsic characteristics. Of these intrinsic attributes, joint stability and isokinetic strength of the peroneals correlate highly to future ankle injuries, as they are suspected reasons for chronic ankle instability (Baumhauer, Alosa, Renström, Trevino & Beynnon, 1995).

Of those individuals who reported having had a first time acute ankle sprain, 20 to 45% report reoccurrence or chronic instability (Hollis et al., 1995; Löfvenberg et al., 1995; Lysens et al., 1984; Renstrom & Konradsen, 1997). However, the symptoms or signs of instability within each classification overlap or are not easily quantifiable, making it difficult to determine if someone indeed does have chronic ankle instability or to determine the underlying structural problems causing recurrent sprains (Karlsson, Eriksson & Renström, 1997). Various classifications for chronic ankle instability are used by clinicians to describe an ankle that has excessive range of motion and or talar malalignment (Karlsson et al., 1997). For example, anterior drawer tilt and talar tests are often used to help determine if the participant has instability due to mechanical and subtalar instabilities (Karlsson et al., 1997; Renstrom & Konradsen, 1997). To date, no physical test can be used to accurately identify functional instability; therefore, functional instability is defined as a reoccurrence of ankle sprains (Renstrom &Konradsen, 1997).

Furthermore, it is puzzling why functional instability is not correlated highly with mechanical instability (Renstrom & Konradsen, 1997). Perhaps this suggests that there is a variety of structural problems that can produce ankle instability. Thus, different types of ankle instability can also produce a myriad of foot-ankle mechanics. For example for gait, individuals who exhibit functional instability significantly demonstrate increased lateral plantar pressures, while individuals exhibiting mechanical instability demonstrate tendencies of increased medial pressures compared to individuals without instability (Becker, Rosenbaum, Claes & Gerngross, 1997).

Ligament Strain

The engineering definition of strain is the "change in length of material in reference to its original length" (Nordin &Frankel, 1989). The amount of strain a material can handle before failure can be described using a stress/strain curve, or a load-deformation curve. This curve has four main areas: toe, elastic region, plastic region, and maximum load (Hamill & Knutzen, 1995; Nordin &Frankel, 1989). For ligaments, the toe response is the region of the stress/strain curve where the slope is low. This is because the initial change in length is due to straightening out the crimps in the collagen fibers, with little tensile force produced by the ligament (Hurschler, Loitz-Ramange & Vanderby Jr., 1997). Increasing the load to the ligament causes the ligament to undergo tensile loading and elongation (Nordin & Frankel, 1989). During loading in the elastic region, the ligament continues to elongate in a relatively linear fashion when the force applied to it is increased linearly. Damage to fibrils (microfibers) can occur at the end of the elastic region (Nordin & Frankel, 1989). After the elastic limit is reached, any further deformation of the ligament would increase the permanent length of that ligament; hence, this region is termed the plastic region. Any additional load to the ligament during the plastic region will cause major failure of the ligament until finally complete failure or rupture of the ligament occurs (Hamill & Knutzen, 1995; Nordin & Frankel, 1989).

Another valuable measure of ligament integrity is the strain rate, as ligaments exhibit time-dependent behaviors (Nordin & Frankel, 1989). Based on mechanical stress-strain tests, as the strain rate for a particular ligament increases, the force to failure also increases (Attarian, McCrackin, DeVito, McElhaney & Garret, 1985). Thus, in the elastic region, a ligament stretched at a higher strain rate can resist a greater force but allow less deformation (Nordin & Frankel, 1989).

Ligament Properties

The properties of elastic region of the stress-strain curve of a ligament are based upon a ligament fiber's inherent stiffness. The magnitude of fiber stiffness is deduced from the fiber constitutive law, wherein fibril volume, fibril stiffness and fibril orientation concentration can affect the elastic region of the stress-strain curve (Hurschler et al., 1997). An acute inversion sprain or repeated strain loads at or just below the elastic limit can damage the ligament, whereby the ligament decreases in stiffness and/or the ligament demonstrates a permanently increased length. (Hintermann, 1998; Nordin & Frankel, 1989). After an injury, the scar tissue formed can reduce the overall stiffness of the fibrils, as the fibril orientation becomes disorganized compared to its non-injured state. The decreased fiber stiffness and greater elongation state is hypothesized to delay the time when the ligament mechanoreceptors can detect potentially damaging ligament displacement as well as the maximum magnitude of strain applied to the ligament (Karlsson et al., 1992). Therefore, the degree of ligament displacement and the magnitude of strain are thought to endanger the ankle joint to future traumatic and/or chronic injury (Karlsson et al., 1992).

Tissues Typically Injured During a Sprain

An ankle sprain typically involves visual signs and sensory signs of swelling (edema), broken blood vessels (hematoma), pain and tenderness (Rubin & Sallis, 1996). The severity of the injury, despite the presence of these symptoms, does not necessarily reflect the impairment to the underlying tissues e.g., ligaments and muscles (Rubin & Sallis, 1996).

Ligaments

Each articulation of the foot-ankle complex has numerous ligaments to maintain joint stability. However, two ligaments in particular often become injured during sudden inversion motions. They include the anterior talofibular (ATFL) and the calcaneofibular ligaments (CFL) (Figure 1) (Karlsson, Eriksson & Renström, 1997). Among 110 patients who exhibited chronic ankle instability (as defined by having experienced one prior ankle sprain and other instability symptoms for a minimum of six months), 64% and 41% of patients exhibited a complete rupture of the ATFL and/or CFL, respectively (Schäfer & Hintermann, 1996).

After these two ligaments become damaged during a sprain, the order in which other ligaments are injured is of some controversy, but include the posterior talofibular and lateral talocalcaneal and cervical ligaments (Karlsson et al., 1997; Renstrom & Konradsen, 1997; Rubin & Sallis, 1996). Therefore, only the ATFL and CFL are of primary interest in this study.

Anterior Talofibular Ligament (ATFL)

The ligament surmised to be torn first is the ATFL (Attarian et al., 1985; Hollis et al., 1995; Renstrom & Konradsen, 1997). This small ligament originates at the lateral malleolus and inserts on the neck of the talus (Hamill & Knutzen, 1995). The ligament's dimensions (6-10 mm wide, 20 mm long and 2 mm thick) infer a low tensile strength (140 +/- 24 N) compared to other lateral ankle ligaments (Attarian et al., 1985; Hollis et al., 1995; Renstrom & Konradsen, 1997). When the foot is in the neutral position, the ATFL runs parallel to the long axis of the talus. As the foot moves into a plantarflexed position, the ATFL begins to run parallel to the tibia and fibia. The strain of the ATFL increases as the foot moves from dorsiflexion to plantarflexion (Colville et al., 1990; Self, 1996). The ATFL also experiences increases in strain as the foot-ankle complex is inverted (Self, 1996) and internally rotated. Conversely, the strain decreases as the foot is everted and externally rotated (Colville et al., 1990). The importance of the ATFL during

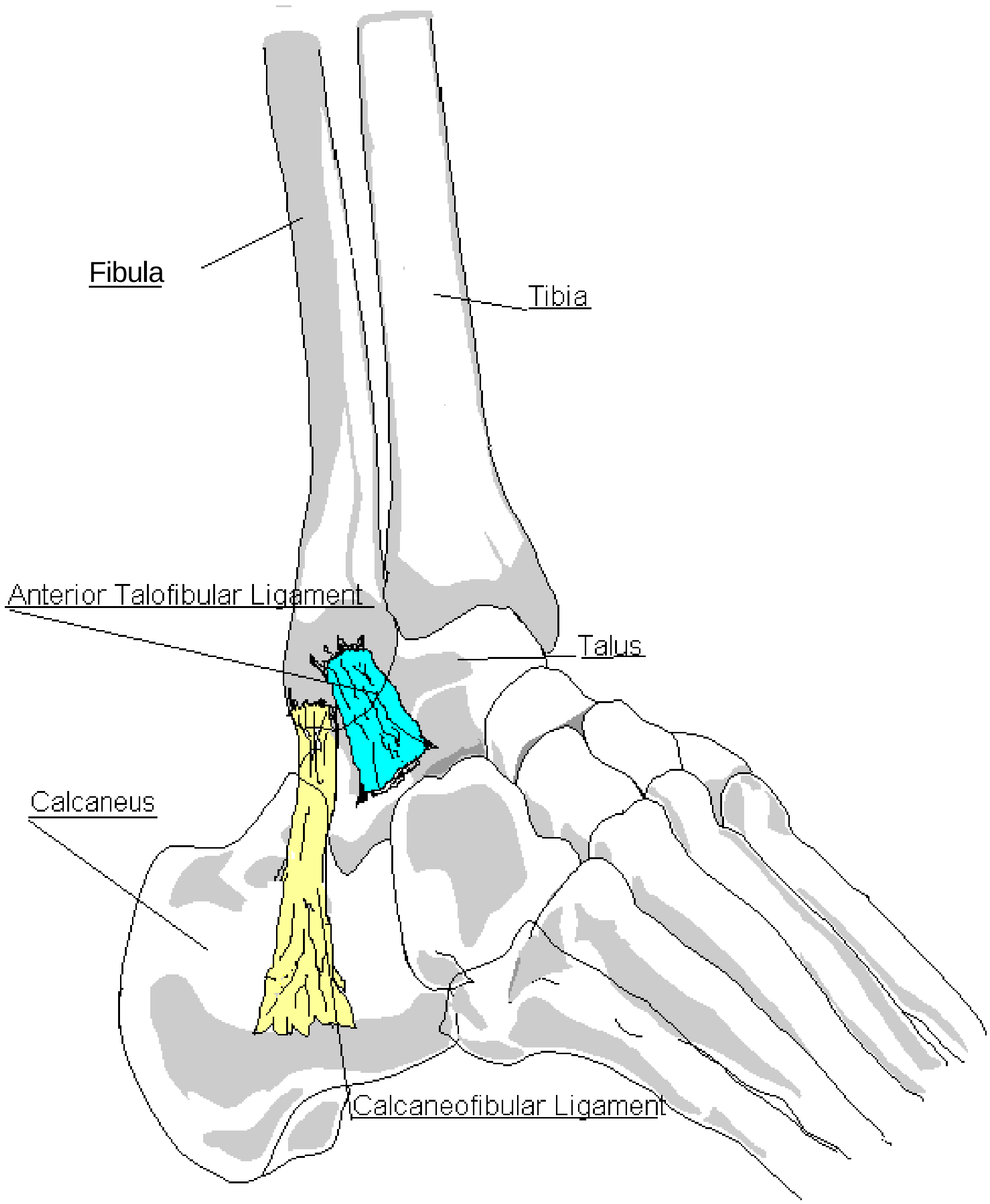

Figure 1. The location of the ATFL and CFL are represented relative to the calcaneus, fibula, tibia and talus bones of the foot-ankle complex.

dynamic movements becomes apparent, as it is an important factor in limiting inversion in conjunction with the CFL (Chen, Siegler & Schneck, 1988), particularly when the foot-ankle complex is in a plantarflexed position.

Calcaneofibular Ligament (CFL)

The calcaneofibular ligament is the other most commonly injured ligament in the lateral ankle complex during movements of rapid inversion. The CFL is a long, round ligament about 20-25 mm in length and 6-8 mm in diameter (Renstrom & Konradsen, 1997). As the CFL is associated with the peroneal tendon sheath, damage to the ligament often causes damage to the peroneal tendon and tendon sheath (Renstrom & Konradsen, 1997).

As shown in Figure 1, the CFL runs obliquely distally and posteriorly from the lateral malleolus to the lateral surface of the calcaneus (Rockar, 1995). The CFL exhibits an increase in strain as the foot-ankle complex inverts or externally rotates (Colville et al., 1990). Movement from dorsiflexion to plantarflexion has been observed to decrease the strain present in the CFL (Colville et al., 1990; Self, 1996), while movement from 20 degrees of plantarflexion to 30 degrees plantarflexion has been observed to increase the strain (Colville et al., 1990). Whether the strain continues to increase while moving from 30° plantarflexion to the full range of motion of plantarflexion (50 degrees) is not known. However, the extent to which the CFL limits inversion movement is questioned due to the low strain measurements of the ligament compared to the ATFL (Colville et al., 1990).

Simulated Ligament Injury

An indirect approach to determine the contributions of ligaments to ankle stability is to measure the maximum rearfoot displacement of a cadaver, section a given ligament in the cadaver, and then observe the increases in rearfoot displacement. After the ATFL is sectioned compared to pre-sectioned values, greater inversion (Chen et al., 1988), anterior drawer flexibility (Lapointe, Siegler, Hillstrom, Nobilini & Mlodzienski, 1997) and talar tilt (Johnson & Markolf, 1983) is observed. Simulating an actual injury by sectioning both the ATFL and the CFL also creates increases in inversion range of motion and increases the coupling between internal rotation motion and inversion motion compared to the measures of the intact cadavers (Lapointe et al., 1997; Rosenbaum, Becker, Wilke & Claes, 1996).

However, the probability of injury during landing when a given ligament is lax cannot be determined from ligament sectioning studies. Thus, the measurement of strain on a ligament during a flat drop landing and inverted drop landing may give more insight into the actual loading that may occur during landings typically exhibited during physical activity. Hence, the potential for injury to the ligaments can be ascertained for high impact landings. As reported in an unpublished dissertation, Self (1996) dropped lower extremity cadavers from two heights of six and twelve inches onto a 30° inverted-V platform and onto a flat platform from a six inch height. At the six inch height, significant differences ($p = 0.10$) were found between the flat and inverted landings, whereby the inverted landing condition exhibited greater strain for the ATFL and CFL than the flat landing condition. However, no differences for maximum strain were found between the two ligaments. Due to the small number of cadavers used for the inverted 12 inch drop ($n = 4$), the ATFL and CFL maximum strain values were not statistically compared. However, the strain during the 12 inch inverted landing condition did exhibit increases of approximately 50% for both ligaments when compared to the 6 inch inverted landing condition.

The time dependent behavior of a ligament is another measurement that provides insight into ligament integrity. In addition to measuring strain values, Self (1996) also measured strain rates of both the flat and inverted platform landings. Significant differences ($p=0.10$) were found at the flat and 6 inch inverted landing, whereby the six inch inverted landings exhibited greater values for strain rates of the ATFL and CFL compared to the values observed for the flat landing. In addition, the ATFL had higher strain rates than the CFL. At the 12 inch inverted drop height, the strain rates also increased compared to the 6 inch inverted drop height, although the difference in strain rates were not significant due to the low number of cadavers (n = 5) used. For both inverted landings, higher strain rate values observed for the ATFL in comparison to the CFL exhibits compensation in the ATFL for its lower pure tensile strength observed by another investigator (Attarian et al., 1985). The higher strain rate of the ATFL may allow the ligament to attain a higher magnitude of force before the elastic limit is reached in comparison to the CFL.

Muscles

Another structural unit that can be injured during a sprain is the evertor muscle group, which counteracts the inversion motion of the foot-ankle complex. The peroneal muscles are of particular interest as they are strong evertors of the foot that can be used to counteract the external inversion torques applied to the foot. Therefore, injury to these muscles may decrease the amount of protection against sudden inversion of the ankle (Baumhauer, Alosa, Renström & Beynnon, 1995).

Simultaneous stretching of the tendon and contraction of the muscles during eccentric action can cause the muscle to become susceptible to high tensile loading (Hamill & Knutzen, 1995). Injury to muscles in adults usually occurs at the myotendinous junction of the muscle (where the myofibrils of the muscle join collagen fibers of the tendon) or the belly of the muscle (Bassett & Speer, 1993; Hamill & Knutzen, 1995). The unexpected sudden inversion of the foot-ankle complex while in a plantarflexed position has been surmised to strain the peroneal muscles and damage the peroneal tendon at the myotendinous junction (Bassett & Speer, 1993).

Role of Muscle Reflexes

For a high impact event, the peroneal muscles pre-contract approximately 60-90 ms before landing occurs to prepare to attenuate the ground reaction forces (Karlsson et al., 1992; Konradsen & Højsgarrd, 1999; Springings et al., 1981). During a normal drop landing situation, two peak ground reaction forces occur within the first 50 ms of the landing event (Dufek & Bates, 1990; Dufek & Bates, 1991; Reinschmidt et al., 1992). Therefore, before the two peak ground reactions forces occur, the peroneal muscles are producing a contractile force to counteract the impact forces.

Thus, it has been proposed that a correct prediction of the timing of the landing is needed to generate the necessary evertor muscle contractile force (Santello & McDonagh, 1998). However, during an unexpected landing, contact with a surface may happen earlier than the participant expected. Another type of unexpected landing could arise from the change in the slope of the landing surface, where a flat landing was expected but the landing actually occurred on another person’s foot. The change in the slope of the landing surface introduces unanticipated, complex inversion torques to act on the foot-ankle complex in addition to the high impact vertical ground reaction forces.

Although the exact biomechanics that occurs during an unexpected landing is not well understood, the sources of forces that can create high inversion torques are the vertical, medio-lateral and possibly anterio-posterior ground reaction forces, joint reaction forces, and invertor muscle forces. Therefore, landing on a sloped surface, the evertor muscles need to compensate for the increasing external inversion torques with opposing eversion torques. For stable ankles, Konradsen, Voigt and Højsgarrd (1997) observed that an active eversion (goniometric evidence of eversion movement) of the ankle occurs 176 ms after a sudden inversion movement is induced via a trapdoor test. Thus, the latency period of the evertor muscles is such that increased muscle activity cannot occur in response to a sudden external inversion moment (Isakov, Mizarahi, Solzi, Susak & Lotem, 1986; Ottaviani, Asthon-Miller, Kothari & Wojtys, 1995). Furthermore, the issue also is controversial as there is no agreement on the peroneal muscle movement time that occurs in response to an inversion motion stimulus if the muscle has been previously damaged (Beckman & Buchanan, 1995; Ebig, Lephart, Burdett, Miller & Pincivero, 1996; Hollis et al., 1995; Isakov et al., 1986; Isakov & Mizrahi, 1998; Nawoczenski et al., 1985). However, the breadth of this controversy is beyond the scope of this review.

Prevention of Sprains

Damage to the ATFL, CFL and peroneal muscles of the lower extremity from an ankle sprain is approximated to occur once for every 10,000 persons each day (Baumhauer, Alosa, Renström, Trevino et al., 1995). In a four-year study of 138 participants, 162 ankle sprains occurred (Lysens et al., 1984). Of the individuals who experienced a sprain, 44% experienced a reoccurrence of another ankle sprain (Lysens et al., 1984). Fortunately, under certain conditions, the reduction of ankle injury incidents can be accomplished with the use of prophylactic ankle stabilizers (Karlsson & Andreasson, 1992; Rovere, Clarke & Yates, 1988; Tropp et al., 1985). Several clinical studies have shown that the frequency of injury among previously injured participant decreases with the application of stabilizing aids (Sitler et al., 1994; Surve et al., 1994).

For one prospective study, Surve et al. (1992) categorized 516 soccer players into two groups: a) a previously injured group (at least one previous ankle sprain within two seasons) and b) a no injury group (no previous ankle sprain history). The two injury groups were then randomly assigned to one of two brace conditions: a brace (Aircast Sport-Stirrup™) or a no-brace condition. The amount of exposure time was reported as the number of injuries/1,000 hours for each participant. The previously injured participants who wore the Sport-Stirrup™ exhibited a significantly lower rate of injury incidence in comparison to those participants who were previously injured but did not wear the Sport-Stirrup™. In addition, the severity of the ankle sprain was significantly reduced for those with previous ankle injury when compared to the previous injured control group. However, a reduction in injury incidence was not observed for those individuals who had no previous injury and who wore the brace in comparison to the no-brace uninjured participants.

For another prospective study, Sitler et al. (1994) conducted a two-year study on military cadets who played intramural basketball at the United States Military Academy, West Point, New York. Of the 1,601 participants, 177 cadets were assigned to the previously injured ankle group, while the remaining cadets served as the non-injured control group. The two injury groups were randomly assigned to a brace group (the Aircast Sport Stirrup™) or to a no-brace group (the control). During the two-year investigation, 46 ankle sprains were reported, 11 of which occurred in the ankle stabilizer group. Thus, using the semi-rigid Aircast Sport Stirrup™ significantly reduced the frequency of injury in comparison to not wearing a brace. However, due to the small sample size of the previously injured group, a reduction in injury severity was not observed for those who wore the brace compared to a control. A previously injured participant who did not wear a brace was reported to have a 1.4 times greater risk of injury than a non-injured participant.

Two other prospective studies also investigated the efficacy of ankle stabilizers. Tropp et al. (1985), randomly assigned 425 soccer players to one of three groups, 1) semi-rigid brace 2) proprioceptive disc training and 3) control (no brace or disc training). After six months, the semi-rigid and disc training conditions were equally effective in reducing the frequency of ankle injury when compared to the control group individuals who had previous injury.

Another investigation by Rovere et al. (1988), incorporated a 7 year retrospective study where shoe design and brace type were individually chosen by the participants. Rovere et al. determined that those individuals who chose the brace/low-top shoe combination had significantly fewer ankle injuries than those players who used no tape or who wore high-top shoes.

Prophylactic Ankle Stabilizers

For rehabilitation, therapists/trainers commonly prescribe prophylactic ankle stabilizers to controlling swelling and range of motion after inversion injury to the foot-ankle complex (Callaghan, 1997). In addition to the evidence of injury frequency reduction with the aide of an ankle stabilizer (Sitler et al., 1994; Surve et al., 1994), the commercial ankle stabilizer is commonly prescribed to those with chronic ankle instability to provide reinforcement to the foot-ankle complex (Hume & Gerrard, 1998). Scientific verification through passive range of motion and dynamic situations determines the stabilizing device's ability to provide and maintain restrictive properties for the foot-ankle complex.

Effectiveness of Tape

In 1946, the usage of an ankle stabilizer was first prescribed (Quigley, Cox, & Murphy, 1946). The first material used is referred to today as "athletic tape," which can be applied in various wrapping techniques. This costly method of stabilization has received mixed reviews for its efficacy, as tape does not maintain its tensile strength throughout a regimen of exercise (Callaghan, 1997; Garrick, 1977; Greene & Wight, 1990). An extensive review on the comparisons between taping methods can be found in Callaghan’s review of taping versus bracing (Callaghan, 1997).

Non-Rigid brace designs and Semi-Rigid brace designs Versus No Brace

Besides taping methods, several stabilizing devices are available to the consumer. Various choices of material and attachment devices help characterize the non-rigid and semi-rigid brace styles (Callaghan, 1997). The non-rigid devices typical involve a sleeve design of cloth or pliable plastics that are tightened with laces. The semi-rigid devices involve thermoplastic or plastic polymers to encase the ankle with a stirrup design that is tightened by Velcro® straps.

One method to test the effect of an ankle stabilizer on the foot-ankle complex's range of motion is to measure the passive range of motion. A machine is used to move the foot while the participant's muscles are in a relaxed state. When using a passive evaluation method for a particular movement, e.g., inversion, the angular displacement values are compared among the different styles of prophylactic devices as well as to a no brace condition. Typically, both types of braces provide significantly greater restraint for passive motion in comparison to a no brace condition in the inversion/eversion, plantar/dorsiflexion, and internal/external rotation directions (Alves et al., 1992; Bruns, Scherlitz &Luessenhop, 1996; Greene & Wight, 1990; Gross, Ballard, Mears, & Watkins, 1992; Gross, 1998; Hartsell & Spaulding, 1996; Johnson, Veale & McCarthy, 1994; Shapiro et al., 1994; Siegler et al., 1997).

Non-Rigid Brace Designs Versus Semi-Rigid Brace Designs

During a passive evaluation, differences in displacement values between the non-rigid and semi-rigid braces also are noticed for ankle range of motion (ROM) measurements (Alves et al., 1992; Greene & Wight, 1990; Gross et al., 1992). The semi-rigid braces, such as the Air-Stirrup™, passively restrict total inversion/eversion range of motion by at least 42% (Greene & Wight, 1990). In comparison to the semi-rigid brace, the non-rigid brace e.g., Swede-O™, provides passive restriction of 30% for inversion/eversion (Alves et al., 1992). For internal and external rotation, passive restriction also is significantly greater for the semi-rigid design compared to the non-rigid design (Siegler et al., 1997).

Although both designs limit the amount of range of motion in in/eversion and in/external rotation during a passive test, with a period of exercise, e.g. 20 minutes (Greene & Wight, 1990), the non-rigid brace loosens, which allows more in/eversion motion (Alves et al., 1992; Greene & Wight, 1990). For the Greene and Wight (1990) study, the non-rigid design (Swede-O™) allowed 15 more degrees of motion after 90 minutes of exercise. However, the semi-rigid design maintained restrictive properties to the foot-ankle complex. Only a 6 % increase in in/eversion range of motion was observed for the semi-rigid design (Air-Stirrup™) after exercise compared to the pre-exercise value (Alves et al., 1992; Greene & Wight, 1990).

For this particular study, two semi-rigid braces are of interest to prevent ankle sprains are the Active Ankle™ brace and Malleoloc™ brace. The Active Ankle™ (Figure 2) of Active Ankle System, Inc (Louisville, KY) is designed with a medio-lateral axis hinge located at the approximate height of the lateral malleolus. The hinge design allows unrestrained range of motion in the plantar/dorsiflexion direction while inhibiting in/eversion and ab/adduction of the foot-ankle complex. The Malleoloc™ brace (Figure 3) of Bauerfeind USA, Inc., (Kennesaw, GA) is a semi-rigid brace that incorporates a modified-stirrup design. The design is specifically constructed to fit such that the lateral stirrup is anterior to the lateral malleolus and the medial stirrup is posterior to the medial malleolus. According to the manufacturer, the location of lateral stirrup is positioned superficially over the ATFL to prevent excessive tensile loading to the ATFL.

The Malleoloc™ and Active Ankle™ braces have shown variations in restricting motion during passive range of motion evaluations in the plantar/dorsiflexion and in/eversion directions of motion compared to a no brace condition. For movement in the sagittal plane, the Malleoloc™ brace has been shown to limit passive range of motion for the plantar/dorsiflexion direction compared to a no brace condition (Wiley & Nigg, 1996). In contrast, the Active Ankle was reported not to significantly restrict passive plantar/dorsiflexion when compared to a no brace condition (Lindley & Kernozek, 1995; Siegler et al., 1997). For passive motion, both the Active Ankle™ and Malleoloc™ braces have shown reduction in the range of motion for inversion/eversion directions when compared to a no-brace condition (Johnson et al., 1994; Siegler et al., 1997; Wiley & Nigg, 1996). For the Wiley and Nigg investigation, when the foot was placed in positions of 20° dorsiflexion, neutral, 20° plantarflexion and 40° plantarflexion, the inversion passive range of motion decreased 45% or more. After a period of exercise, the

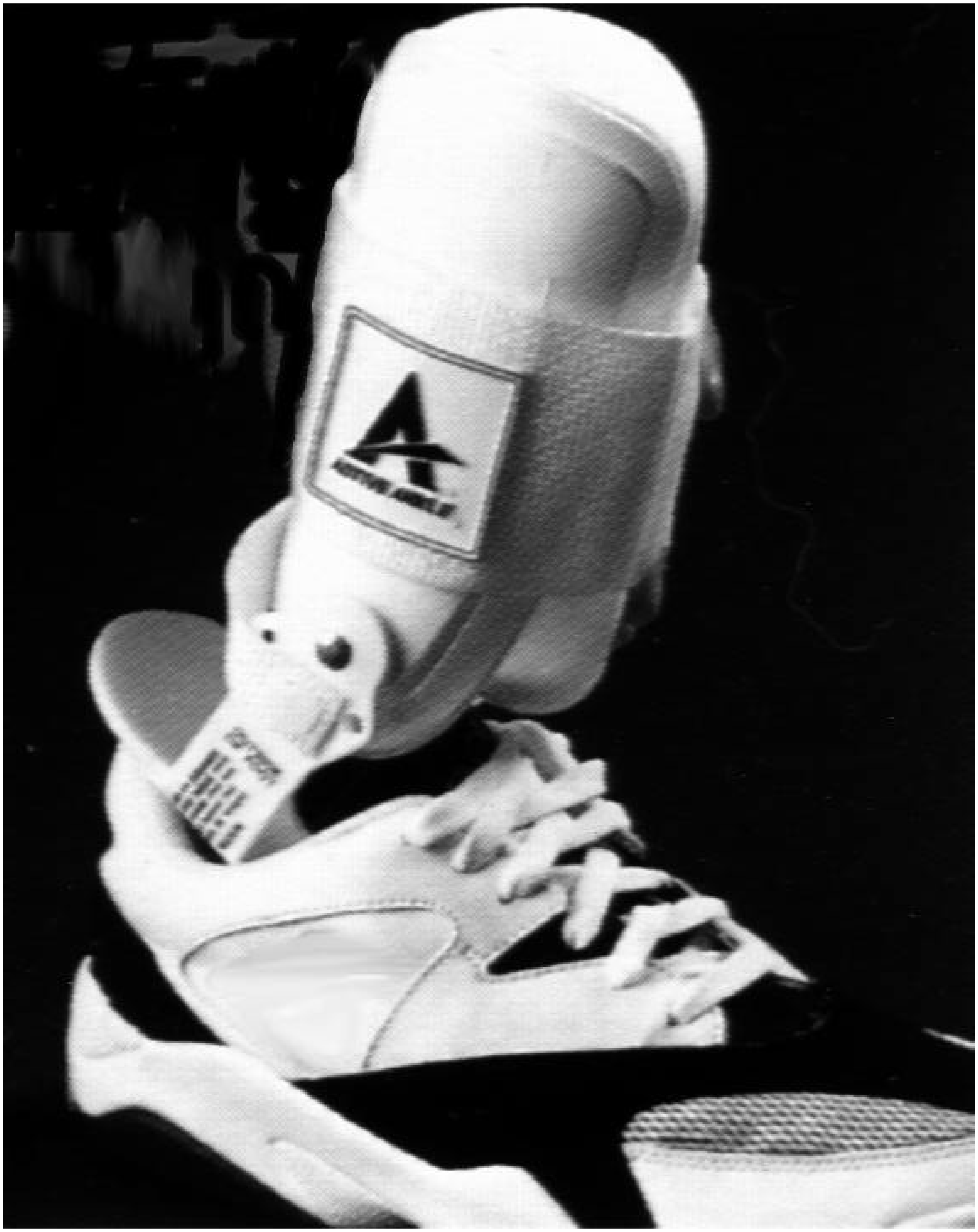

Figure 2. A representation of the Active Ankle™ brace before application to the foot-ankle complex.

Figure 4. A representation of the Malleoloc™ brace as applied to the foot-ankle complex.

passive range of motion values were not significantly different for any direction compared to the pre-exercise values. Thus, it was concluded that the Malleoloc™ brace maintained its restrictive properties to the foot-ankle complex even with exercise. In a direct comparison of the Malleoloc™ and Active Ankle™ braces, Johnson et al. (1994) reported the Active Ankle™ brace restricts inversion motion more significantly than the Malleoloc™ brace before and after an hour long exercise bout.

Another method to assess the range of motion restriction is an active movement in which the participant is instructed to move the foot-ankle complex at a maximum rate in the desired direction. Wiley and Nigg (1996) reported that the Malleoloc™, in comparison to a no-brace condition, restricted the ankle joint range of motion 11° more for inversion, 3° more for eversion, 6° more for plantarflexion, and 3° more for dorsiflexion. Furthermore, the range of motion measurements for the Malleoloc™ were not significantly different for any direction after an exercise period.

However, the stress of a dynamic situation is thought to be greater than those incurred during passive and active range of motion evaluations (Simpson et al., 1999). Simpson et al. compared the angular kinematics of the Malleoloc™ (semi-rigid) Aircast Sport Stirrup™ (semi-rigid), Swede-O™ (non-rigid) and a no-brace condition exhibited during a sideward cutting maneuver. Simpson et al. did not observe the same results as Johnson et al. (1994) and Wiley and Nigg (1996), as the Malleoloc™ exhibited a significantly higher maximum inversion value than the non-rigid (3°) and no brace (3°) conditions. In addition, although not significantly different from the control condition, the other semi-rigid brace (Aircast™) also exhibited a slightly higher maximum inversion value (2°) than the Swede-O™ and no brace conditions. The unusual results of higher maximum inversion were explained potentially by the individual's perception of more stability provided by the semi-rigid brace design compared to the lack of motion restraint when wearing the Swede-O™ or no brace. Therefore, the participant subconsciously landed more tentatively during the non-rigid and no brace conditions than when wearing the either semi-rigid brace (Malleoloc™ or Aircast™).

This injury avoidance phenomenon has been observed previously. Xia and Robinson (1997) compared inversion values during running of a typical running shoe condition to values for a shoe designed especially to increase inversion. However, the participants exhibited significantly lower inversion values for the prototype shoe compared to the control shoe (Xia & Robinson, 1997).

It has been surmised that a reduction in angular velocity should occur when a brace is worn compared to not wearing a brace (Podzielny & Hennig, 1997). Semi-rigid braces have been observed to reduce angular velocity by at least 200 º/second for inversion (Podzielny & Hennig, 1997) and 120 º/second for plantarflexion (McCaw & Cerullo, 1998) compared to a no brace condition. The reduction of angular velocity represents a delay in the time at which maximum inversion is reached (Podzielny & Hennig, 1997). Secondly, a reduced angular velocity is hypothesized to indicate a decrease in torque to be absorbed by the ankle, knee and hip joints (McCaw & Cerullo, 1998). However the reduction in the inversion velocity was not observed during an investigation by Simpson et al., as the semi-rigid (Malleoloc™ and Aircast™), non-rigid (Swede-O™) and no brace conditions were not significantly different. Yet, there was a high degree of inter-participant variability for angular velocity, leading to insufficient statistical power.

In designing a brace that can effectively reduce inversion motion, the motion most often compromised is plantarflexion/dorsiflexion motion (Sitler & Horodyski, 1995). During running and jumping activities, the full range of motion of flexion and extension in the foot-ankle complex is surmised to be a necessity to maintain performance effectiveness e.g., jump as high as possible (McCaw & Cerullo, 1998; Sitler & Horodyski, 1995). Therefore, the ultimate purpose for a brace is to provide stability while not sacrificing performance. Testing the functional performance of a participant while wearing a semi-rigid brace incorporates jumping (Bocchinfuso, Sitler & Kimura, 1994; Johnson & Veale, 1994; MacKean, Bell & Burnham, 1995; Wiley & Nigg, 1996), running (Bocchinfuso et al., 1994; Gross et al., 1997; MacKean et al., 1995) and agility tests (Johnson & Veale, 1994; Wiley & Nigg, 1996).

The functional performance of the Malleoloc™ brace and the Active Ankle™ brace have been tested with vertical jump test (Bocchinfuso et al., 1994; Wiley & Nigg, 1996) and various running courses (Bocchinfuso et al., 1994; MacKean et al., 1995; Wiley & Nigg., 1996). Although the angular displacements for plantar/dorsiflexion were not measured, during a performance test of the Malleoloc™, the brace did not inhibit the performance of 12 participants during a figure-eight run course or vertical jump test (Wiley & Nigg, 1996). In a similar fashion to the Malleoloc brace, the Active Ankle™ brace did not inhibit the participants performance for vertical jump (Bocchinfuso et al., 1994), running (shuttle, sprint, and four-point run) (Bocchinfuso et al., 1994; MacKean et al., 1995), and basketball jump shot (MacKean et al., 1995).

Neuromuscular Considerations While Wearing a Prophylactic Ankle Stabilizer

Another interpretation behind the reduction the frequency of ankle sprain by the use of a stabilization aid is due to a controversial idea of enhanced somesthesia (Freeman, 1965). The idea of somesthesia, includes body sensations of touch, pain, temperature and limb position (Rose, 1997). Cutaneous receptors and proprioceptors are the two subdivisions of somesthesia. The cutaneous receptors detect touch and pressure via physical deformation of a particular receptor within the different layers of skin. The proprioceptors detect motion and joint position through specialized mechanoreceptors located in ligaments, tendons, joints, and in the vestibular apparatus. These specialized receptors provide continuous input about general position of the body in space prior to and during movements. These sensory organs gain input for the central nervous system in order to generate motor responses.

Due to injury, the ankle is thought to "lose proprioceptive properties around the ankle joint" (Freeman, 1965; Perrin P.P., Béné, Perrin C. A. & Durupt, 1997). The addition of a prophylactic device is thought to play a role in adding cutaneous stimulation and mechanical pressure to the subcutaneous tissue of the foot-ankle region (Simoneau et al., 1997). To prove this premise, researchers have measured postural control of participants who were wearing a prophylactic device and compared the results obtained when the participant did not wear a brace during a balance test. Feuerbach & Grabiner (1993) discovered a lower mean sway while wearing a brace condition in reference to a no-brace condition during a static test.

However, there is also evidence to suggest ankle braces do not improve motor response. Within the same investigation by Feuerbach and Grabiner (1993), when utilizing a dynamic test (the apparatus moved in a circular motion), no differences between the brace and no-brace conditions were observed. Furthermore, Bennell and Goldie (1994) measured touchdown frequency in which wearing a brace caused the participant to increase the number of corrective posturing touchdowns by the opposite foot in comparison to the number of touchdowns of a no brace condition.

Testing ankle joint position sense is another method of investigating proprioception around the ankle joint by quantitatively having the individual match a reference ankle joint angle or to sense initial joint movement. It has been claimed that the ability to sense joint position is inhibited by previous injury to the ankle complex (Lentell, Baas, Lopez, McGuire, Sarrels & Snyder, 1995). In an investigation by Feuerbach et al. (1994), anesthetized and non-anesthetized ligaments conditions revealed no differences for accuracy of matching joint positions to the referenced positions. Thus, the mechanoreceptors in the ligaments were surmised not to be the receptors that provide proprioceptive feedback to match joint position (Feuerbach et al., 1994). Yet, a significant difference was detected between the brace and no brace conditions of both the anesthetized and non-anesthetized ligament conditions. Therefore, Feuerbach et al.

(1994) concluded that an increase in cutaneous stimulation may have enhanced the awareness of joint position (in all directions) during matching of reference positions for both anesthetized and non-anesthetized ligaments.

In further support of this concept, Simoneau et al. (1997) determined that the ability of the ankle joint to recognize joint position in a non-weight bearing condition was more accurate when athletic tape strips were placed on dorsum of the foot compared to a no tape condition. Yet, when changing to a weight-bearing situation, the presence of tape had no significant influence on joint position (Simoneau et al., 1997).

Peroneal Muscle Activity

Although it is not clear whether braces improve proprioception, wearing a brace also is postulated to provide decreased onset times for the peroneal muscles. The ability of the brace to provide added cutaneous stimulation to the ankle complex ideologically may enhance the muscle activity of the peroneal muscles, although this is not proven (Feuerbach et al., 1994). Compared to not wearing a brace, when wearing semi-rigid braces faster latency periods of the peroneal muscles have been observed (Karlsson & Andreasson, 1992; Nishikawa & Grabiner, 1995; Nishikawa & Grabiner, 1996; Springings et al., 1981). However, another inquiry found no increase in the latency of the peroneal muscles during a brace condition compared to a no brace condition (Stüssi et al., 1987).

The conflicting findings of these studies may be due to the different methods and movements used during the experiments. The majority of investigators (Karlsson & Andreasson, 1992) have used a passive closed chain movement, while Stüssi et al. (1987) used running, an open-chain skill. Karlsson and Andreasson, (1992) noted that the degree of mechanical instability of participants during a sudden inversion via a trapdoor test influenced the length of the latency periods. Thus, the differences between the findings of Stüssi et al.'s open-chain skill and the other closed-chained skills (Karlsson &

Andreasson, 1992; Nishikawa & Grabiner, 1995; Nishikawa & Grabiner, 1996, Springings et al., 1981) may be due partly to differing ankle stability among the participants of these investigations. Thus, the hypothesis that ankle braces provide proprioception and/or enhance peroneal muscle onset time is difficult to assess due to the complexity underlying neuromuscular response during actual landings.

Although the idea of proprioception may not accurately explain the decrease in ankle injury due to prophylactic bracing, one idea does hold true about ankle bracing. Evidence from nearly all studies reviewed showed that inversion ankle range of motion is limited when a brace is worn. Whether injury prevention when wearing a prophylactic ankle brace is due to the reduction in motion in one or more directions is not known. However, to improve the efficacy of braces to prevent injury, we need to understand the mechanisms that underlie protection against sudden inversion of the ankle.

Methodological Consideration of Brace Studies

The ground reaction force absorption during landing phases of a physical activity begins with the foot-ankle complex and then passes to the connecting joints of the knee and hip. Each individual incorporates his/her unique style to attenuate the ground reaction forces (Caster& Bates, 1995; Dufek & Bates, 1990; Schot, Bates & Dufek, 1994). The amount of knee flexion can increase or decrease the amount of ground reaction force that needs to be absorbed (Devita & Skelly, 1992). Thus, methodology used for studies investigating mechanisms underlying brace efficacy is of importance.

Knee Flexion

Typical angles of knee flexion are those distinguished as low knee flexion (<100°) and high knee flexion (>170°) (Devita & Skelly, 1992; Gross & Nelson, 1988). The vertical ground reaction forces are influenced by the magnitude of knee flexion during landings, as decreased vertical ground reaction forces occur with greater knee flexion angle (Devita & Skelly, 1992; Dufek & Bates, 1990). The amount of force that each joint contributes to attenuate the force is highly dependent on the magnitude of knee flexion. During a high knee flexion landing, the angular negative work absorbed across the ankle joint is greatest (50%) when compared to the hip and knee joints (20% and 31%, respectively). In comparison, a low knee flexion landing increases the negative work provided by the hip and knee joints; thus the energy absorbed is more evenly distributed across all three joints (25%, 37%, and 37%, respectively) (Dufek & Bates, 1990). Therefore, for the current study, to ensure consistency of the ankle mechanics across all landing conditions for each participant, the maximum knee flexion angle for any landing trial was ±3° of the maximum knee angle exhibited during a natural landing.

Marker Placement

During investigations using static or dynamic movements, the locations for the markers on the participant must be made relative to the methodology selected for generating segment coordinate systems and joint coordinate systems (Areblad, Nigg, Ekstrand, Olsson & Ekstrom, 1990; Grood & Suntay, 1983). Furthermore, valid estimates of marker displacements occurring during the experiment must be considered. Not placing the markers directly on the skin or bones enlarges the magnitude of error to the kinematic measurements (Wilkerson, Pinerola, Caturano, 1997). Various other kinematic analyses of inversion have been done with markers placed on the shoe (Nawoczenski et al., 1985; Simpson et al., 1999). However, marker placement on the shoe does not accurately describe the motion of the foot during a sudden inversion (Stacoff et al., 1996). Reinschmidt et al., (1992) demonstrated that greater inversion angles were generated from shoe markers compared to the angle calculated using skin markers. By cutting holes into the shoe, markers can be placed on the skin to give greater accuracy for measuring the location of the markers; hence, greater validity of other measures e.g., the inversion angle (Stacoff et al., 1996). Although skin movement may introduce error commonly referred to as "skin movement artifact," tibiocalcaneal rotation in all three planes (in/eversion, add/abduction, and plantar/dorsiflexion) was adequately represented during a running activity when skin markers were used (Reinschmidt, van den Bogert, Nigg, Lundberg & Murphy, 1997).

Compared to skin markers, a better estimation of the bone movement can be gained through the use of pins that are inserted into the bone (Lafortune, Cavanagh, Sommer & Kalenak, 1994; Reinschmidt, van den Bogert, Murphy, Lundberg & Nigg, 1997). This method, however more accurate, requires surgical intervention and limits the types of experiments that can be performed.

Summary

The exact etiologies of ankle sprains are not known, in regard to the loading that occurs to the anterior talofibular and calcaneofibular ligaments. Nevertheless, the way in which the foot lands during movement creates several situations that increase the probability of spraining the ankle. After one ankle injury, the natural stabilizing agents, such as ligaments, are thought to be altered due to inherent changes experienced by the tissue (Nordin & Frankel, 1989). These changes in the mechanical properties create an environment in which the chance for the reoccurrence of ankle sprains is about 20 to 45% (Hollis et al., 1995; Löfvenberg et al., 1995; Lysens et al., 1984; Renstrom & Konradsen, 1997).

Two main theories have been proposed to account for the observed reduction of ankle sprains when the prophylactic device is worn. For one theory, ankle braces are thought to increase cutaneous stimulation that subsequently improves proprioception or enhances muscle response. The evidence for this theory is based on results of studies demonstrating decreased postural sway, improved joint position sense, and decreased onset of muscle activity when participants perform static or actual movements while wearing braces compared to not wearing a brace. Another theory explaining brace efficacy is the device’s ability to passively restrain the ankle from excessive movement. The motion restraint theory has been examined on numerous occasions through passive tests of foot-ankle motion in which results of brace and no brace conditions range of motions are compared. However, the presence of the brace does not explain the ankle sprain frequency reduction. Thus, as it is desirable to further reduce the incidence of chronic ankle sprains, understanding the factors related to effective brace design is relevant.

CHAPTER III

A COMPARISON OF KINEMATIC RESTRAINT EXHIBITED BY TWO PROPHYLACTIC ANKLE BRACES DURING FLAT AND INVERTED DROP SURFACE LANDINGS[1]

[1] Wilder, J. A. and Simpson, K. J. To be submitted to Journal of Applied Biomechanics.

For individuals who have had a previous ankle injury, ankle braces can reduce the number of subsequent ankle sprains by approximately 40% (Sitler et al., 1994; Surve, Schwellnus, Noakes & Lombard, 1994). While the exact cause of ankle sprains is not known, one situation that has been reputed to create an ankle sprain is landing with the foot-ankle complex in a plantarflexed and slightly inverted position onto an uneven surface e.g., another person's foot (Garrick, 1977; Shapiro, Kabo, Mitchell, Loren & Tsenter, 1994). The ligament surmised to be torn first is the anterior talofibular ligament (ATFL), followed by the calcaneofibular ligament (CFL) (Renstrom & Konradsen, 1997; Rubin & Sallis, 1996). While inversion displacement stresses both the ATFL and CFL, in the plantar/dorsiflexion direction, the tensile loading is greatest in the ATFL when the foot is a plantarflexed position, whereas dorsiflexion motion stresses the CFL (Colville, Marder, Boyle & Zarins, 1990; Siegler, Chen & Schneck, 1988; Stormont, Steger, Stüssi & Reinschmidt, 1985). Thus, reducing strain to these ligaments during high impact landings via a brace must consider foot position and displacement primarily in the inversion direction but also in the plantar/dorsiflexion direction of motion.

The effectiveness of semi-rigid designs are based on several criteria. The brace should: a) provide inversion restraint during a potentially injurious landing, b) not restrict dorsiflexion motion, and c) be perceived by the user as comfortable and efficacious. For this study, the perceived comfort by the user of a particular brace is not being investigated, therefore; the investigation will focus on the first two criteria stated above. Although it has been shown for constrained and/or slow laboratory movements that wearing a semi-rigid orthosis can significantly reduce maximum inversion displacement (Alves, Alday, Ketcham & Lentell, 1992; Bruns, Scherlitz & Luessenhop, 1996; Greene & Wight, 1990; Gross, Ballard, Mears & Watkins, 1992; Hartsell & Spaulding, 1996; Johnson, Veale & McCarthy, 1994; Shapiro et al., 1994; Siegler et al., 1997) and inversion angular velocity (Podzielny & Hennig, 1997) compared to wearing a non-rigid brace or no brace, it is not known if the efficacy varies among semi-rigid designs.

Within the category of semi-rigid braces, to date, no comparisons have been made between a modified stirrup (Malleoloc™) brace and a hinged stirrup (Active Ankle™) brace during a dynamic situation, e.g., a drop landing. For the Malleoloc™ (Bauerfeind Inc.), the lateral stirrup is anterior to the lateral malleolus and superficial to the ATFL and the medial stirrup runs posterior to the medial malleolus. The lateral stirrup position has been surmised to provide greater passive restraint of the foot in all directions. The Active Ankle™ (Active Ankle Systems Inc.) is a stirrup design hypothesized to allow free range of motion in the plantar/dorsiflexion direction, due to the placement of a mediolateral axis hinge at the height of the midpoint of the lateral malleolus, while being able to restrict inversion/eversion motion of the foot-ankle complex.

Several investigations have evaluated the effectiveness of the Malleoloc™ and Active Ankle™ relative to the criteria listed above. For constrained movements, both braces have been shown to restrict inversion but not dorsiflexion (Siegler et al., 1997; Wiley & Nigg, 1996). Both the Malleoloc™ and Active Ankle™ brace designs have been deemed comfortable by the user; however, preferences among designs vary (Siegler et al., 1997; Simpson, Cravens, Higbie, Theodorou & DelRey, 1999). However, for either design it is not known whether excessive ankle inversion is restricted during a dynamic situation (Simpson et al., 1999), particularly during landings similar to those that could produce an ankle sprain.

Therefore, using a drop landing movement onto an uneven surface i.e., sideward sloped surface may give better insight into the strain placed on the ATFL and CFL that occurs during landings typically exhibited during physical activity than passive, closed chain ROM tests. Landing on an 30° inverted V surface has been shown to significantly increase strain in the ATFL and CFL of cadavers when compared to landing on a flat surface (Self, 1996), demonstrating the increased ligament loading that occurs while landing on an inverted surface compared to a flat surface. Thus for this study, drop landing onto a 30° inverted V platform as well as a flat surface were used to simulate landings similar to atypical, potentially injurious and typical, non-injurious landings.

Thus, the objective of this study is to determine whether there were differences in motion restraint, i.e., rearfoot kinematics between the modified stirrup (Malleoloc™) and the hinge brace (Active Ankle™) for a flat and an inverted landing. It was hypothesized that less maximum inversion, less inversion angular displacement and more time to maximum inversion would be exhibited by both braces compared to the control condition (no brace) with the Active Ankle™ brace exhibiting less inversion motion than the Malleoloc™ brace. In addition, it was hypothesized that the Malleoloc™ condition would demonstrate less maximum plantarflexion, maximum dorsiflexion and dorsiflexion angular displacement than the Active Ankle™ brace and control conditions. It was also hypothesized that compared to the flat landing surface, the inverted landing condition would exhibit greater values for maximum inversion angle, time to maximum inversion, and greater inversion angular displacement; but the values for the two surfaces would not vary by more than 2° for maximum plantarflexion and maximum dorsiflexion, by more than 4° for dorsiflexion angular displacement, and by more than 5% of time to maximum dorsiflexion relative to total landing time.

In addition to evaluating if different stirrup designs affect passive motion restraint to the foot-ankle complex, it also was of interest to better understand the foot motion that occurs during brace wear, as this is not known. Typically, the displacement of the rearfoot during brace studies has been measured by using markers on the shoe (Nawoczenski, Owen, Ecker, Altman & Epler, 1985; Simpson et al., 1999). However, the brace stabilizes the foot, not the shoe, consequently, markers placed on the skin rather than the shoe should provide more accurate data (Reinschmidt, Stacoff & Stüssi, 1992). For example, inversion motions exhibited during running, walking and sideward cutting maneuvers have been shown to be overestimated when markers were on the shoe compared to inversion values obtained from foot marker data (Reinschmidt et al., 1992;

Stacoff, Steger, Stüssi & Reinschmidt, 1996). In addition, the presence of a brace in a shoe could create shoe motions that are different than rearfoot movements. Therefore, to accurately measure movements of the foot-ankle complex during a non-injurious landing and potentially injurious landing, the markers during this investigation were placed on the participant's skin.

Methods

Participants

Potential participants were recruited from the general population of the University of Georgia. A questionnaire (Appendix A) was used to determine previous recreational experience and injury history. Thus, only those participants who had experience in physical activities involving impact landings (Appendix B) and who had a previous ankle sprain to the right ankle were considered for potential participation. However, for a Grade I, II, or III sprain, the participant could not have had a sprain within 3 months, 6 months and 1 year, respectively, of the physical exam date. After signing the consent form, the potential participant was evaluated for lower extremity dysfunction and recent injuries to other body segments by a physical therapist. Therefore, in addition to the previously stated criteria, only those participants whose lower extremity range of motion (ROM) values were within the American Academy of Orthopaedic Surgeons (AAOS) (Greene & Heckman, 1994) expected range of motion of the ankle, knee and hip joints and who were free from injury were eligible to participate in the investigation.

Of the potential participants evaluated, 27 participants (mean +/- SD: age = 22.5 +/- 6.4 yr., ht. = 174.2 +/- 9.4 cm, mass = 73.8+/-14.9 kg), (see individual participant data in Appendix C) were accepted to participate in the study. Paired sample t-tests of the right and left limbs ROM means were found non-significant (p-value range = .204 -.823) for all ROM measurements (Table 1). None of the participants examined were found to have a positive anterior drawer or talar tilt test result, excessive tibial torsion, femoral

Table 1
Means (M), Standard Deviations (SD) and Range of Values of the Participants for Selected Range of Motion (ROM) Variables from the Participant Screening

Ankle ROM				Range	
Variable	Leg	M	SD	Min.	Max.
Plantarflexion	Right	52.6	7.0	30	60
	Left	53.0	5.9	40	60
Dorsiflexion	Right	12.4	4.9	0	20
	Left	12.7	4.5	5	20
Inversion	Right	31.5	5.7	20	45
	Left	31.9	4.8	20	45
Eversion	Right	21.0	4.6	15	30
	Left	21.9	4.2	15	30
Subtalar Inversion	Right	5.2	0.8	5	8
	Left	5.6	1.3	3	7
Subtalar Eversion	Right	4.9	1.1	3	8
	Left	4.5	0.9	2	5

torsion or forefoot valgus/varus. The participants were fitted for the Malleoloc™ and Active Ankle™ braces and laboratory shoes for both feet.

During two practice sessions, the participants were accommodated to the task of dropping onto the flat and two inverted landing platforms and to the brace conditions. Once the participant felt comfortable dropping onto a 15º inverted practice platform, the participant was then introduced to the 30º inverted test platform. Five drop landings onto the 30º platform were practiced.

Experimental Setup

Cameras

Three genlocked high-speed video cameras (Pulinex TM 640) operating at a sampling rate of 120 fields/s and a shutter speed of 1/1000 s were used to capture the positions of the markers of the right extremity. The experimental setup, including the locations of the cameras, is shown in Figure 4. The field of view of the cameras was a truncated pyramid (base = 1.59 m x 1.91 m; top surface = 1.21 m x 1.02 m; perpendicular distance between the two bases = 0.60 m).

Task

The participant climbed three steps, grasped an adjustable drop bar which was mounted from a cement beam in the ceiling. Then, the participant hung from the bar, the steps were removed and the performer was stabilized by the researcher. The height of the drop was 0.40 m as measured from the distal end of the lateral malleolus to the landing point on the platform. After the landing was complete, the participant remained in a static position in order for the researcher to obtain the estimated maximum knee flexion angle via a goniometer (left knee) and foot landing angle relative to the mid-sagittal axis of the platform. To obtain the segment coordinate systems and relative displacement of a given

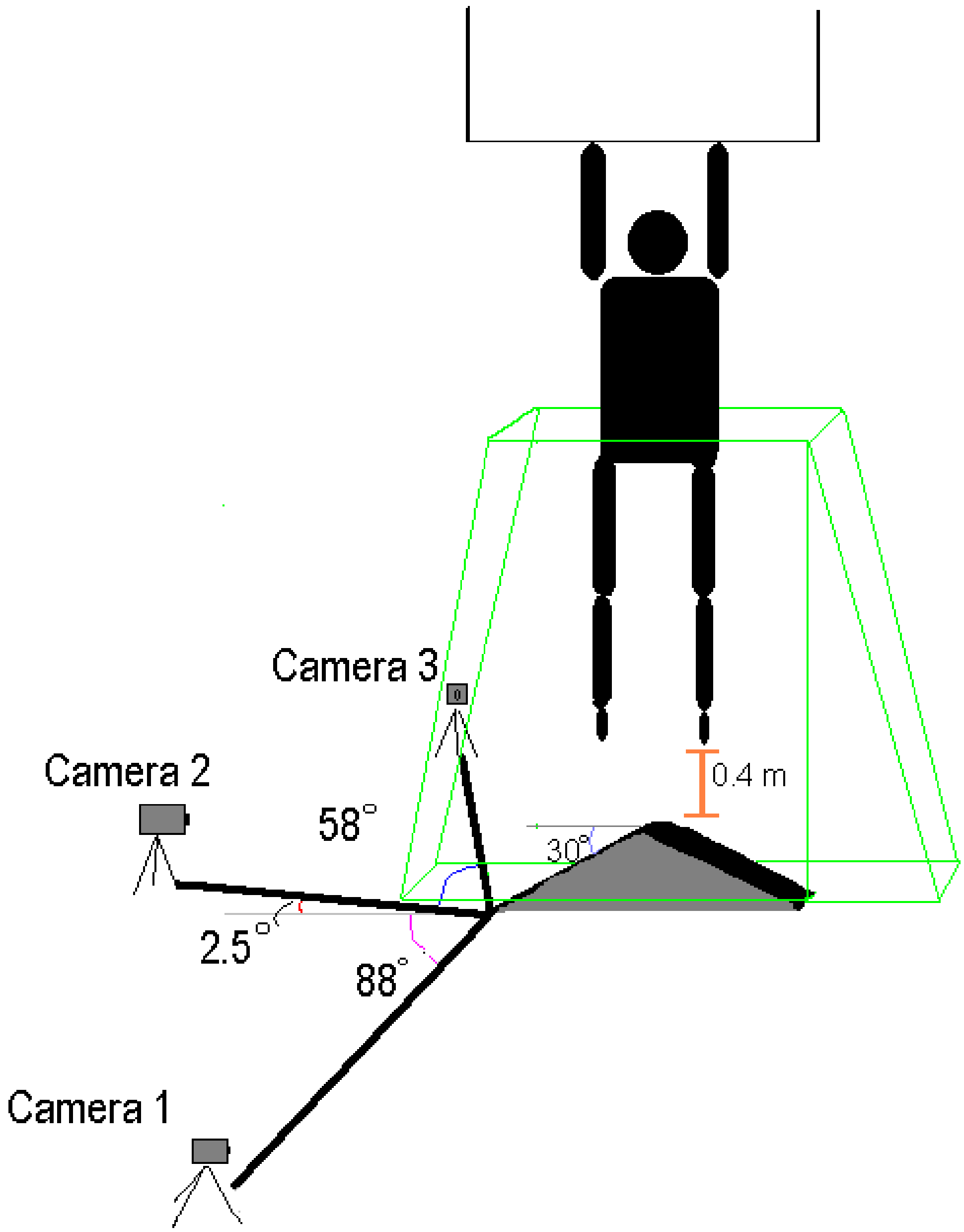

Figure 4. Experimental Setup. The locations of cameras 1, 2 and 3 are represented in relation to the right front corner of the landing platform. The green lines represent the three cameras field of view of a truncated pyramid. Camera distances = 5.4 m, 5.5 m and 5.7 m, respectively. The participant was stabilized over the landing platform and initiated a drop landing of 0.4 m (from platform to lateral distal malleolus).

Markers

For the right leg, three non-collinear markers were placed on each segment (thigh, lower leg, rearfoot) and a marker was placed on the head of the fifth metatarsal (Figure 5). The markers placed on the foot segment were made from a T-nut, machine screw and reflective ball (Figures 6, 7, 8). The right shoe had elliptical holes cut into the heel counter and side of shoe that were no larger than 3.0 cm x 3.5 cm to insure visibility of the markers (Reinschmidt et al., 1992; Stacoff et al., 1996). For a given foot segment marker: 1) a T-nut was applied to the skin, 2) the brace and shoe were applied, 3) the machine screw was attached and 4) the reflective marker was attached.

Protocol

A warm-up session similar to the two practice sessions was performed by the participant. Before testing began of a particular brace condition, the participant stood in a natural position on the flat platform while the lower extremity was videotaped. Then, six acceptable trials were performed for a given brace-surface condition. For an acceptable trial: 1) the participant must have landed in a balanced position, based on researcher's visual assessment and the participant's self report, 2) the maximum knee angle must have been within $\pm 3^{o}$ of the warm-up trial average and 3) the foot landing angle must have been similar to the angle of the test day warm-up trials. Both platform conditions were performed for a given brace condition before another brace condition was performed in order to minimize the number of times the shoe and machine screw was removed during testing. The test order was counterbalanced across participants, first for brace condition, then for platform condition.

Data Reduction

The coordinate data for each reflective marker were smoothed using an optimal smoothing factor quintic spline (Peak Motus 4.3.3 ™). To obtain the segment coordinate

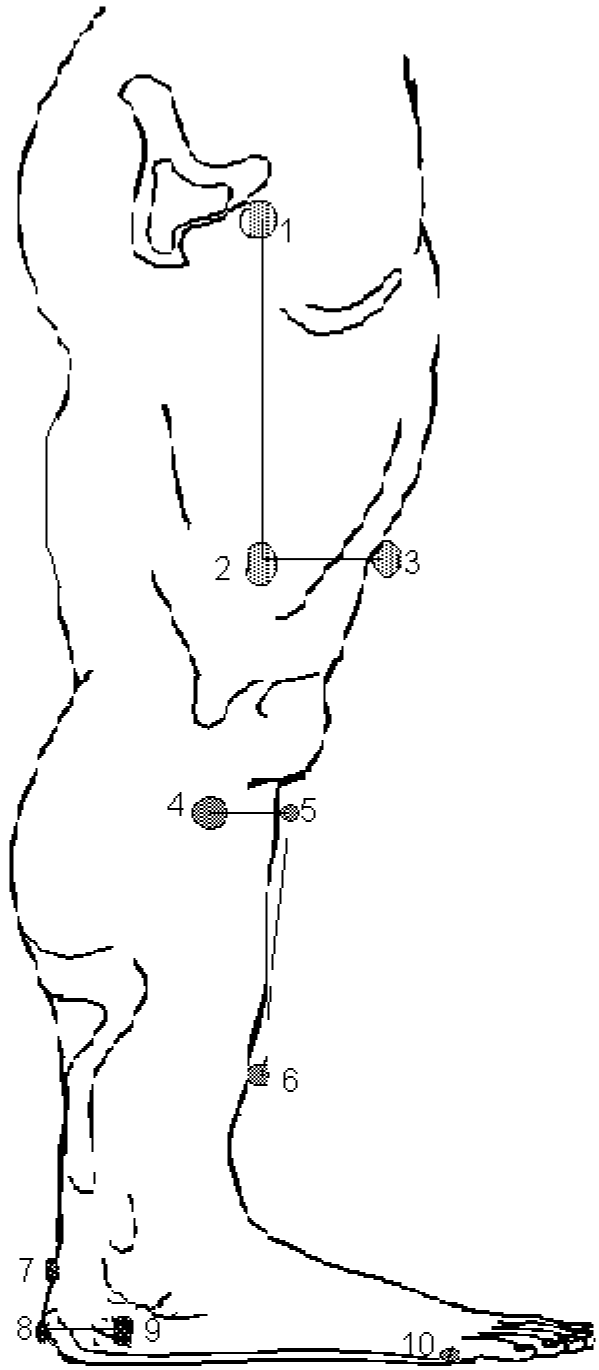

Figure 5. The marker numbers identify location of the reflective marker placed on the skin of the participant: 1) greater trochanter, 2) lateral thigh, 3) anterior thigh, 4) lateral lower leg, 5) anterior lower leg, 6) distal lower leg, 7) proximal calcaneus, 8) distal calcaneus, 9) lateral calcaneus and 10) head of fifth metatarsal.

Figure 6. Angled lateral view of T-nut application to the skin of a participant for the foot segment markers.

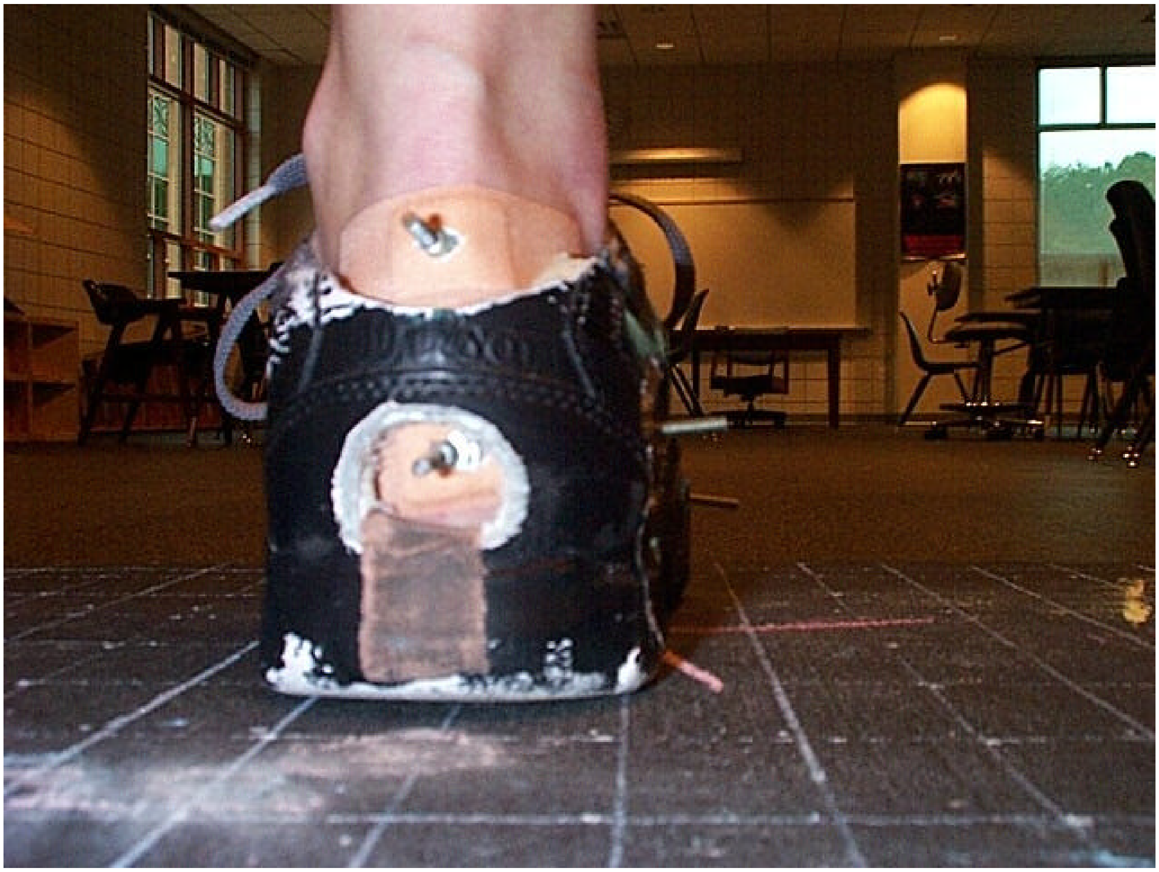

Figure 7. Posterior view of the foot segment markers comprised of the T-nut, machine screw and shoe as applied to a participant

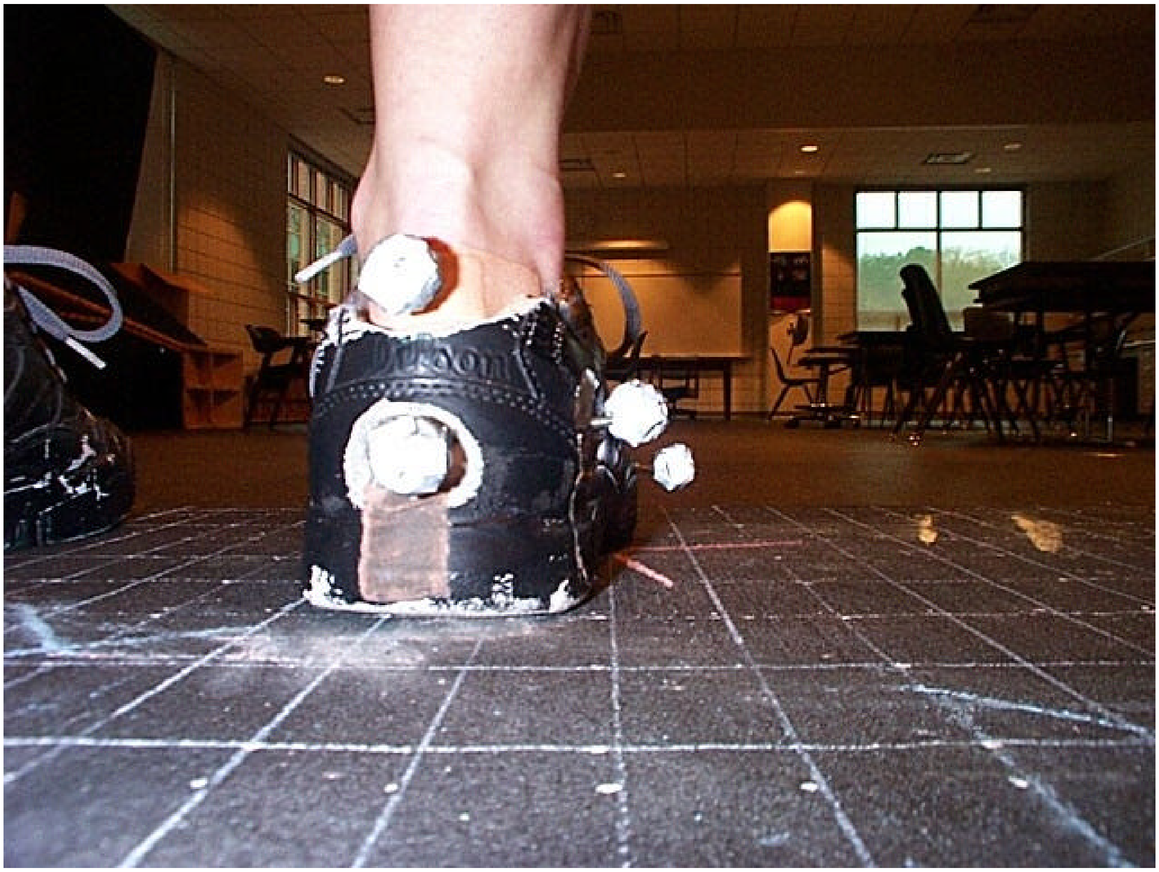

Figure 8. Posterior view of the foot segment markers comprised of the T-nut, machine screw, reflective ball and shoe as applied to the participant.

and relative displacement of a given segment, the methods of Soutas-Little, Beavis, Verstraete & Markus (1986) and Grood and Suntay (1983), respectively, were used.

A right-handed room coordinate system was created with X perpendicular to the landing surface, Y parallel to the landing direction and Z orthogonal to X and Y. Joint coordinate system configurations were created for each segment, where $<x_i, y_i, z_i>$ were segmental coordinate systems (segment = i), similar to the room coordinate system $< X, Y, Z >$ configuration. Euler angles were created for the right foot-ankle complex and for the right knee (Appendix E). Using the vectors shown in Figure 9, the angles of the right foot-ankle complex and right knee were defined as: plantar/dorsiflexion = 90°- arccos $(\mathbf{k}_{LL} \cdot \mathbf{e}_2)$, in/eversion = arccos $(\mathbf{k}_{LL} \cdot \mathbf{e}_1)$, ab/adduction = arccos $(\mathbf{i}_{FT} \cdot \mathbf{e}_2)$ and knee flexion = arccos $(\mathbf{k}_{Th} \cdot \mathbf{e}_{1'})$, where TH= thigh, LL= lower leg and FT =foot.

Data Analysis

As an indirect measure of the strain applied to the ATFL and CFL, the angular displacements from the time of contact to the maximum angle for dorsiflexion and inversion directions were calculated. The times to these events were also determined, as these variables may be indirectly related to the strain of the tissues of the lateral portion of the foot-ankle complex.

For each of these variables, a (2 x 3) two-way repeated measures ANOVA (Brace x Platform) was performed. The Huynh-Feldt test was used to check sphericity (E = 0.850). Simple comparisons between braces were made using least significant difference (LSD) adjustment methods. All comparisons were evaluated at $p < 0.05$.

Results

No Brace x Platform interactions were detected for in/eversion and plantar/dorsiflexion and knee flexion/extension directions. For the foot-ankle complex significant main effects existed for in/eversion and plantar/dorsiflexion variables. The means, standard error and post hoc analyses results are presented for brace comparisons (Table 2) and platform comparisons (Table 3).

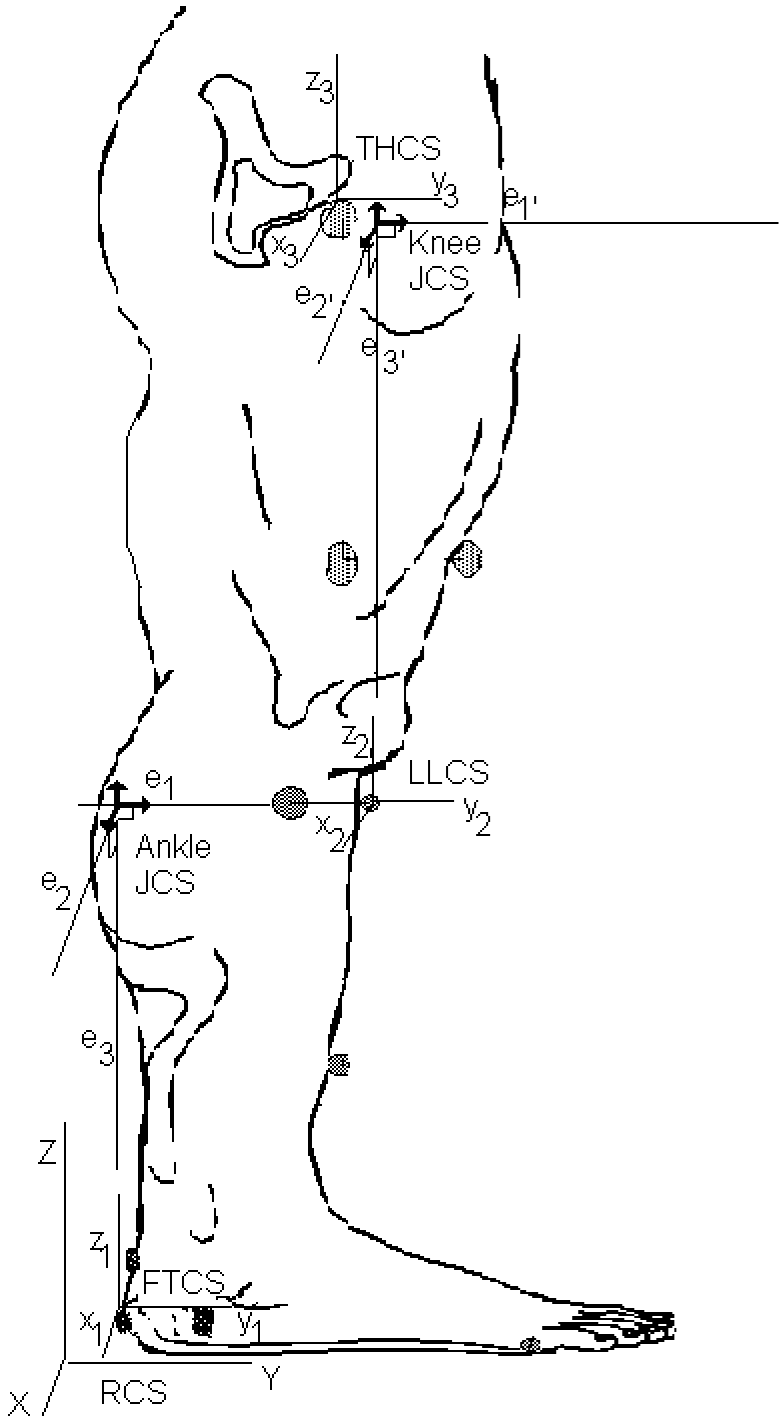

Figure 9. Represented are the coordinate systems for the room, RCS < **X, Y, Z**>, foot, FTCS < $\mathbf{x_1, y_1, z_1}$ >, lower leg, LLCS < $\mathbf{x_2, y_2, z_2}$ > and thigh, THCS < $\mathbf{x_3, y_3, z_3}$>. The corresponding unit vectors for the foot, lower leg and thigh, respectively, are <$\mathbf{i_{FT}}$, $\mathbf{j_{FT}}$, $\mathbf{k_{FT}}$>, <$\mathbf{i_{LL}}$, $\mathbf{j_{LL}}$, $\mathbf{k_{LL}}$>, <$\mathbf{i_{TH}}$, $\mathbf{j_{TH}}$, $\mathbf{k_{TH}}$>. The floating vectors for the ankle and knee joint coordinate systems, respectively, $\mathbf{e_2} = \mathbf{e_3} \times \mathbf{e_1}$; $\mathbf{e_{2'}} = \mathbf{e_{3'}} \times \mathbf{e_{1'}}$.

Table 2

Brace Condition Means (M) and Standard Errors (SE) for Position (deg), Time to Position (% Total Landing Time) and Displacement (deg) Variables and Statistically Significant Comparisons Among Brace Conditions

		Brace		
Kinematic Variable		Malleoloc	Active Ankle	No-Brace
Maximum Pf	M	13^{a}	13^{b}	19^{ab}
	SE	1	1	1
Maximum Df	M	16^{ab}	20^{a}	20^{b}
	SE	2	2	2
Relative Time to Maximum Df	M	74^{ab}	78^{a}	79^{b}
	SE	3	2	2
Pf/Df Displacement	M	28^{ab}	33^{bc}	39^{ac}
	SE	1	1	1
Inversion at Touchdown[1]	M	4	3	5
	SE	1	1	2
Maximum Inversion	M	9^{a}	7^{b}	12^{ab}
	SE	2	2	2
Relative Time to Maximum Inversion[2]	M	23	28	31
	SE	3	4	5
Inversion Displacement	M	12^{a}	11^{b}	14^{ab}
	SE	1	1	1
Knee Flexion at Touchdown	M	13^{a}	14^{b}	10^{ab}
	SE	2	2	1
Maximum Knee Flexion[3]	M	68	67	68
	SE	3	3	3
Knee Flexion Displacement[4]	M	53	53	58
	SE	3	2	3

Note. For a given variable, means sharing a letter (e.g., a, b, c) differ significantly $p < .05$ by the LSD (least significant difference) pairwise comparisons. For a given number (e.g., 1,2,3), variables were not significantly different $p < .05$ for the main effects of a two-way ANOVA. Pf = plantarflexion Df = dorsiflexion.

Table 3

Platform Condition Means (M) and Standard Errors (SE) for Position (deg), Time to Position (% Total Landing Time) and Displacement (deg) Variables and Statistically Significant Main Effects

		Platform	
Kinematic Variable		Flat	Inverted
Maximum Pf	M	13[a]	16[a]
	SE	1	1
Maximum Df	M	23[a]	15[a]
	SE	2	2
Relative Time to Maximum Df	M	76	78
	SE	2	2
Pf/Df Displacement	M	36[a]	30[a]
	SE	1	1
Inversion at Touchdown	M	4	4
	SE	1	1
Maximum Inversion	M	8[a]	11[a]
	SE	2	1
Relative Time to Maximum Inversion	M	18[a]	37[a]
	SE	4	4
Inversion Displacement	M	14[a]	10[a]
	SE	1	1
Knee Flexion at Touchdown	M	13	12
	SE	1	1
Maximum Knee Flexion	M	68	67
	SE	2	3
Knee Flexion Displacement	M	56	55
	SE	2	2

Note. For a given variable, means sharing a letter (e.g., a) differ significantly $p < .05$ by the platform main effect. Pf = plantarflexion Df = dorsiflexion.

Inversion Motion

For the brace conditions, no significant differences were found between the two semi-rigid brace designs, for maximum inversion and inversion angular displacement of the foot-ankle complex (Table 2). However, differences were found between the control condition and one/or both semi-rigid brace conditions for all in/eversion variables. Although there were no significant differences between the brace conditions and the control condition for the touchdown angle, the maximal inversion angle was 3° and 5° less for the Malleoloc™ and Active Ankle™ conditions, respectively, were worn compared to the no brace condition (Figure 10). Therefore, the magnitude of inversion angular displacement also was significantly less for the two braces and the control condition. There was no significant main effect among the brace conditions for the relative time to maximum inversion due to the variability of the times to maximum inversion. However, the Malleoloc™ tended to reach maximum inversion 5% earlier than the Active Ankle™ (p = .172) and 8% earlier than the control condition (p = .032).

For the platform conditions, there was no significant difference for the inversion angle at touchdown between the flat and inverted platform. However, as the maximum inversion angle was less for the flat condition ($M \pm SE = 8 \pm 2°$) compared to the inverted condition ($11 \pm 2°$). Consequently, the inversion angular displacement for the flat platform was 4° significantly greater than the inverted platform (Table 3). The 20% difference in relative time to maximum inversion was significantly greater for the inverted platform compared to the flat platform. Both braces exhibited significantly less maximum inversion, inversion displacement and time to maximum inversion compared to not wearing a brace. In summary, the greatest in maximum inversion angle occurred during landings on the inverted surface when no brace was worn.

Plantar/Dorsiflexion Motion

The significant differences detected among the brace conditions for the plantar/dorsiflexion motion are shown in Table 2. At touchdown, when either brace was

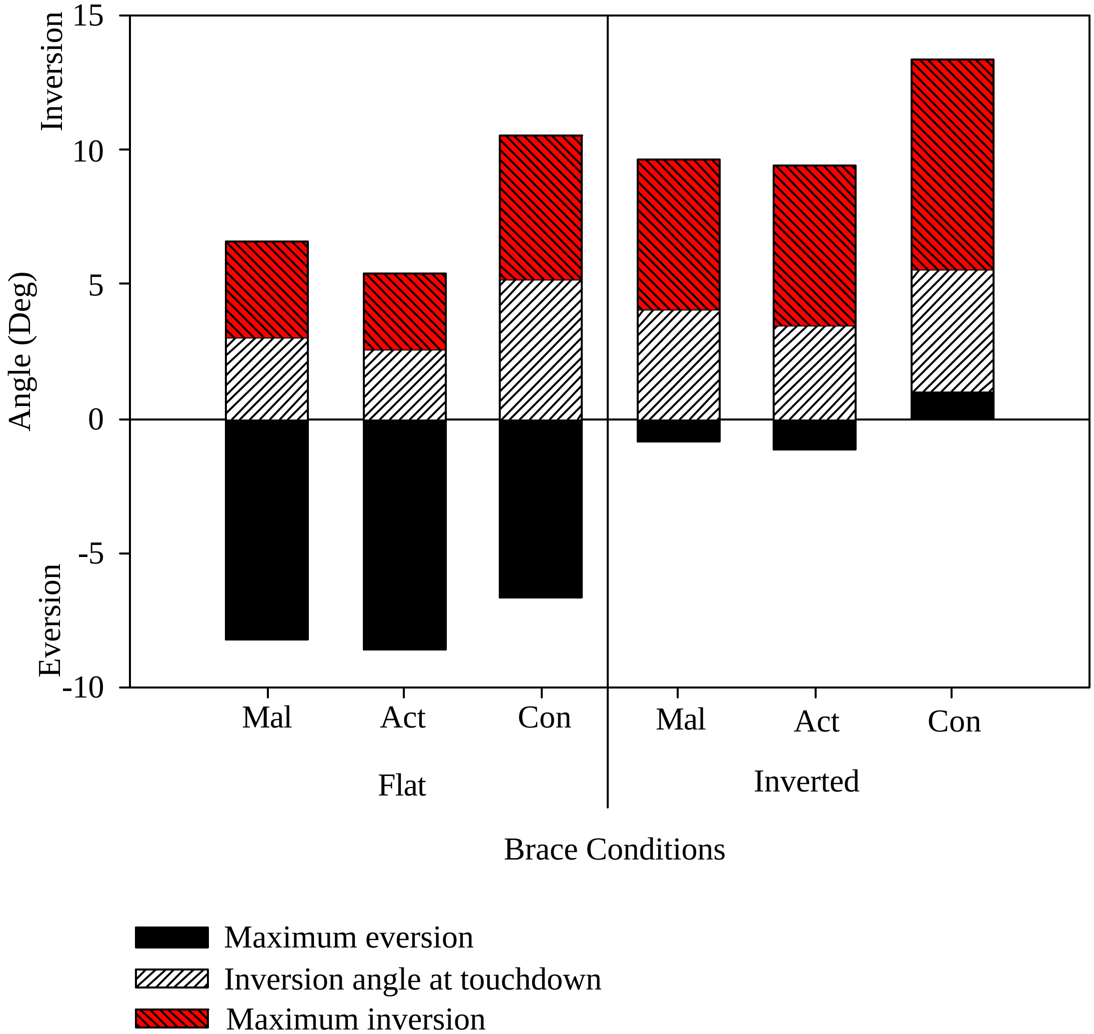

Figure 10. Bar graphs of in/eversion values for angle at touchdown, max inversion and eversion angles and in/eversion displacement (difference between max. eversion and max. inversion) for flat and inverted platform conditions for the Malleoloc (Mal), Active Ankle (Act) and Control (Con) conditions. Significant differences ($p < 0.05$) were observed between platform conditions for max inversion and inversion angular displacement. For brace comparisons, both semi-rigid braces exhibited less max. inversion and inversion displacement compared to the control condition. No significant differences were detected between the semi-rigid braces. Note. All bars begin at 0^{o}.

worn, the foot landed in a 5° less plantarflexed position compared to the control condition. For the maximum dorsiflexion angle, the Malleoloc™ brace value demonstrated 5° less dorsiflexion than either the Active Ankle™ or the control condition angle. Subsequently, significant differences were detected among all the brace conditions for dorsiflexion displacement, with the no brace condition exhibiting the greatest displacement and the Malleoloc™ exhibiting the least displacement (Figure 11). The relative time to maximum dorsiflexion was significantly lower for the Malleoloc™ than the Active Ankle™ or the control brace conditions.

For the platform comparisons (Table 3), at touchdown, the foot-ankle complex landed in a significantly less plantarflexed position (-14± 1°) when landing on the flat compared to the inverted platform (-16 ± 1°). The foot-ankle complex attained a significantly greater mean maximum dorsiflexion angle of 22± 2° for the flat platform compared to 15 ± 2° for the inverted platform (Table 3). In addition, a difference of 6° was observed for dorsiflexion displacement between platform means, with a significantly greater displacement for the flat platform value compared to the inverted platform value.

<u>Consistencies in Landing Kinematics Among Brace and Platform Conditions</u>

Although a visual measurement for foot landing angle and maximum knee flexion angle was obtained during testing, to determine if the participant used a consistent landing technique during brace conditions and within a given platform condition, the kinematic values for foot landing angle and maximum knee flexion were statistically compared. There were no significant brace or platform main effects for the abduction angle at touchdown or abduction displacement (Table 4). However, a Brace x Platform interaction for maximum abduction angle was observed. For each brace condition, greater maximum abduction was exhibited for the sloped surface than the flat surface.

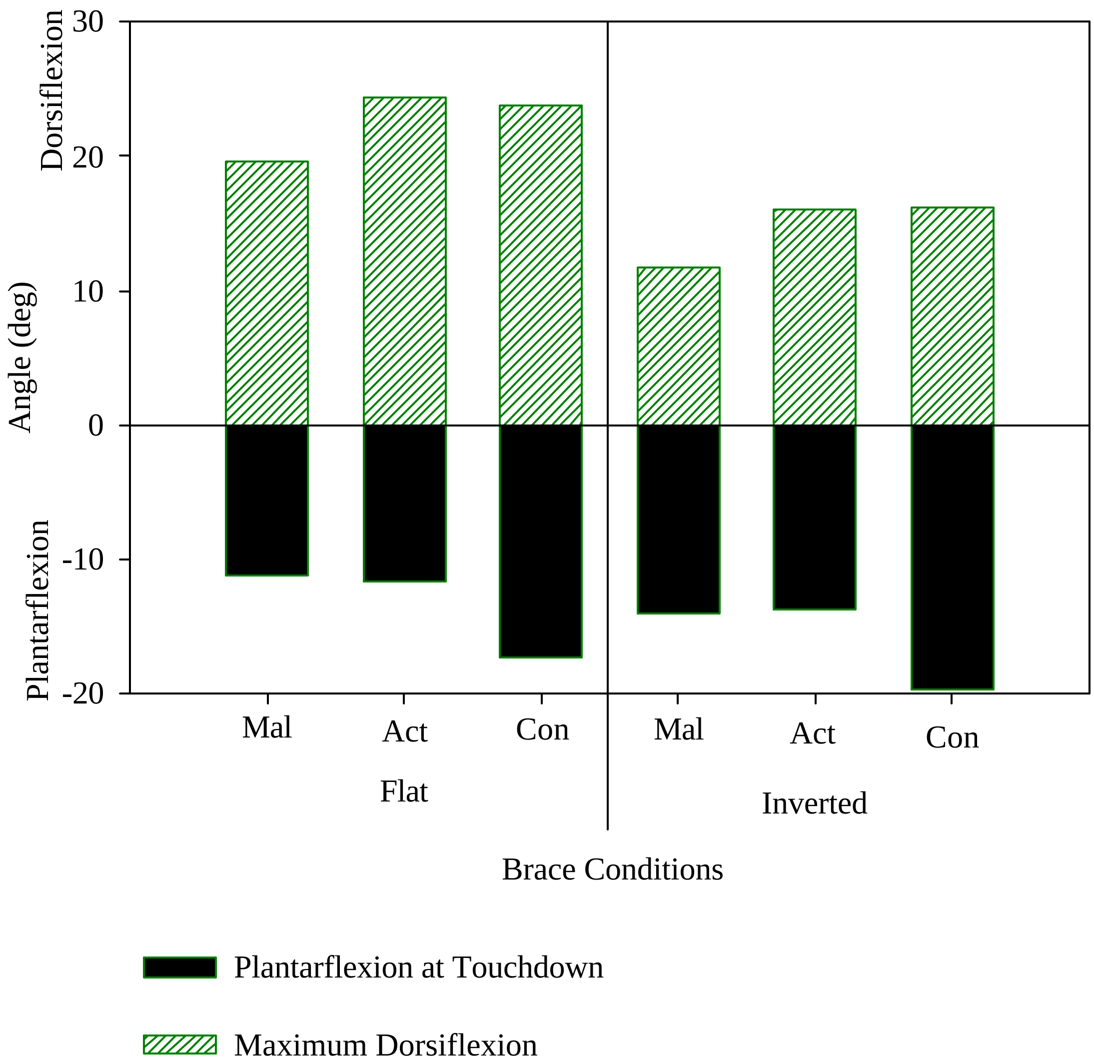

Figure 11. Bar graphs of plantar (pf)/dorsiflexion (df) mean values for angles at touchdown, max df and pf/df displacement (difference between the touchdown angle and max df angle) for flat and inverted platform conditions for the Malleoloc (Mal) Active Ankle (Act) and Control (Con) conditions. Significant differences ($p < 0.05$) were observed between platform conditions for max pf, max df and df displacement. For brace comparisons, both semi-rigid braces exhibited less than control for max pf and df displacement, while the Mal brace exhibited less maximum df and df displacement than both the Act and Con conditions. Note. All bars begin at 0^o.

Table 4

Ab/Adduction Direction of Motion Brace and Platform Condition Means (M) and Standard Errors (SE) for Position (deg) and Angular Displacement (deg) Variables for Brace and Platform Conditions

		Brace			Platform	
Kinematic Variable		Malleoloc	Active Ankle	No Brace	Flat	Inverted
Ab/Adduction at	M	8	4	10	7	8
Touchdown	SE	4	5	4	3	3
Maximum	M	18	13	18	12	20
Abduction	SE	4	4	4	3	3
Abduction	M	12	11	12	11	12
Displacement	SE	1	1	1	1	1

There were no significant main effects for maximum knee flexion angle or knee flexion displacement(Tables 2 & 3). However, the lack of significance for maximum knee flexion and knee flexion displacement may be due to variability in values among the brace and platform conditions. For knee flexion angle at touchdown, significant brace condition differences were observed, with the braces exhibiting 3° to 4° greater knee flexion values than the control condition (Figure 12).

Discussion

An efficacious brace is believed to be one that restricts inversion motion while not inhibiting dorsiflexion motion. However, it is not known if the use of a hinge design accomplishes these goals. Based on cadaver research, the ATFL and the CFL are surmised to be torn first during a sudden inversion of the foot-ankle complex (Renstrom & Konradsen, 1997; Rubin & Sallis, 1996). The strain of the ATFL increases during inversion and plantarflexion direction of motion (Colville et al., 1990; Siegler et al., 1988; Stormont et al., 1985). In contrast, increased deformation of the CFL occurs during inversion and dorsiflexion motions (Colville et al., 1990; Siegler et al., 1988; Stormont et al., 1985). Therefore, to indicate the possible reduction in the strain of the ATFL and CFL during a sudden inversion of the foot-ankle complex, the magnitude of rearfoot displacement in the in/eversion and plantar/dorsiflexion directions were calculated for three brace conditions: two semi-rigid braces (a hinge and no hinge design) and a control condition.

Brace Effects

While the exact mechanical cause of an ankle sprain is relatively unknown, the most common situation for an ankle sprain to occur is landing forcefully in a plantarflexed and slightly inverted position (Garrick, 1977; Renstrom & Konradsen, 1997; Shapiro et al., 1994). The inversion angle at touchdown has been observed to

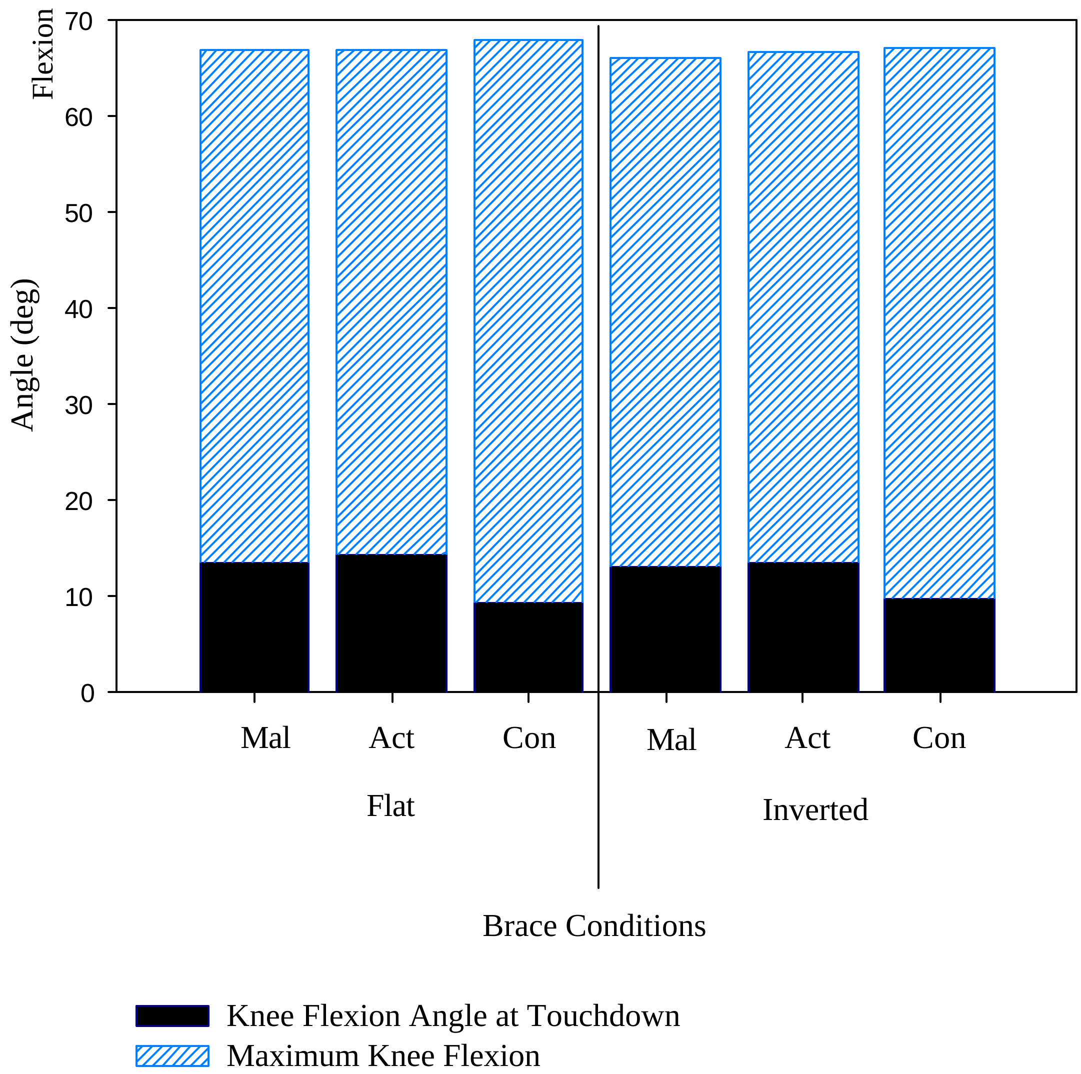

Figure 12. Bar graphs of knee flexion mean values for angles at touchdown, maximum knee flexion and knee flexion displacement (difference between the touchdown angle and maximum angle) for flat and inverted platform conditions for the Malleoloc (Mal), Active Ankle (Act) and Control (Con) conditions. Significant brace condition differences ($p < 0.05$) were observed for knee flexion angle at touchdown for both Mal and Act brace values were greater than the control condition value. Note. All bars begin at 0^o.

affect the values of maximum inversion and inversion angular displacement of the rearfoot during landing by Podzielny and Hennig (1997). Unlike these authors, however, no significant differences were detected among the brace conditions for inversion angle at touchdown. The observed differences in inversion landing angle between the two studies may be attributed to inter-participant variability of this study that reduced the statistical power or to the methodological differences in the landing movement used. Podzielny and Hennig (1997) used a trapdoor method, in which the foot was suddenly released and the trapdoor rotated to 26° of inversion and 13° of plantarflexion. Therefore, many factors, e.g., foot position at touchdown or pre-activation of evertor musculature in response to participant perceptions of the foot-ankle inversion. (Wilkerson & Nitz, 1994), provide alternate explanations for the difference between the two studies.

Predicted differences between the Malleoloc™ brace and Active Ankle™ brace for in/eversion variables were not supported. However, the Malleoloc™ and Active Ankle™ braces were both effective in reducing maximum inversion and consequently, inversion angular displacement of the foot-ankle complex, in comparison to the control condition. The observed decreases in inversion angular displacement that occurred when the braces were worn were potentially due to the passive restraint that the braces provided to the rearfoot during landing and not to the initial inversion landing angle.

An increase in time to maximum inversion has been suggested (Podzielny & Hennig, 1997) although not proven, to allow evertor muscles more time to create counter torque to oppose the sudden external inversion torque applied to the foot when landing on a potentially injurious surface e.g., another person's foot. For this investigation, although no significant difference was detected for relative time to maximum inversion for the brace condition, ($p = 0.063$), the Malleoloc™ and Active Ankle™ braces (absolute time to maximum inversion: $M \pm SE = 0.071 \pm 0.010$s; 0.078 ± 0.010s, respectively) indicated a tendency to reach maximum inversion earlier than the no brace condition ($0.096\pm$

.010s). This suggests that when inversion displacement is reduced there may be an inverse effect on time to maximum inversion.

While motion restraint for the in/eversion direction of motion while wearing a prophylactic ankle brace is commonly believed to help protect the foot-ankle complex (Alves 1992; Siegler 1997), the desired motion restraint to be provided by a prophylactic ankle brace for plantar/dorsiflexion is not known, as restricting motion in this direction has potential positive and negative consequences. As evidence of a positive benefit, limiting plantar/dorsiflexion with a semi-rigid brace was observed to reduce the maximum dorsiflexion angular velocity by 110 °/s compared to a no brace condition during drop landings (McCaw & Cerullo, 1998). The authors hypothesized that this finding was due to reduced dorsiflexion external torques acting on the foot-ankle complex.

In contrast, one hypothetically negative effect of limiting plantar/dorsiflexion is that the body's natural ability to absorb the external torque through the musculature of the ankle, knee and hip may be hindered. When the ROM of ankle dorsiflexion is restricted, the mechanical energy to be absorbed by the knee and hip extensors increases (McCaw & Cerullo, 1998). However, whether this causes lower extremity injury is not known to date (Feuerbach, Ludin & Grabiner, 1993).

Therefore, for this study, it was assumed that during landings, when wearing a brace, the foot-ankle complex could land in a slightly less plantarflexed position compared to non-brace landings to help reduce the strain of the ATFL at contact with the landing surface. Due to a 5° lower plantarflexion angle at touchdown compared to the control condition, the Malleoloc™ and Active Ankle™ brace potentially reduced the strain of the ATFL. Less plantarflexion at touchdown was not expected for the Active Ankle™ brace due to its hinge, which has not been proven to restrict plantar/dorsiflexion motions during a passive ROM test (Siegler et al., 1997).

In addition to restricting plantarflexion at touchdown, the Malleoloc™ brace also exhibited less maximum dorsiflexion compared to both the Active Ankle™ and the control condition values, likely due to the location of the Malleoloc's modified stirrup. Neither the bar of the brace that supports the plantar surface of the foot or the vertical sides of the stirrups align to the plantar/dorsiflexion axis of the talocrural joint, the primary axis for plantar/dorsiflexion motion. Therefore, it is likely that when the foot applied forces to the brace, the brace produced resistive torques on the foot. As the lateral side of the stirrup is located anterior to the plantar/dorsiflexion axis, the lateral portion of the plantar bar and the lateral side of the stirrup may resist plantarflexion of the foot, while the posterior location of the medial side of the stirrup relative to the plantar/dorsiflexion axis would resist foot dorsiflexion.

Compared to not wearing a brace, both semi-rigid braces also were observed to have less dorsiflexion displacement, due partly to less plantarflexion at touchdown for both semi-rigid braces and less maximum dorsiflexion for the Malleoloc™ brace. For the Malleoloc™ brace, by restricting the magnitude of dorsiflexion displacement, it took less time to reach maximum dorsiflexion. Consequently, a possible negative consequence may be increased vertical impact forces when landing while wearing Malleoloc™ brace, although it cannot be proven with these data.

Furthermore, decreased dorsiflexion angular displacement may not produce a deleterious effect on absorbing impact forces. To counteract the decreased dorsiflexion ROM of the ankle while wearing semi-rigid braces compared to the no brace dorsiflexion ROM values, participants have been observed to exhibit greater knee flexion (Feuerbach et al., 1993). For this study, it was observed that participants landed with the knee more flexed while wearing a brace in comparison to the control condition this finding may reflect a subconscious attempt to counteract the decreased range of motion at the ankle joint. As individuals will adjust the magnitude of knee flexion relative to their perceptions of the landing surface (Jameson & Simpson, 1997), we chose to require the participants to flex to a self-selected but consistent degree of knee flexion during all conditions. Therefore, for this investigation, by constraining maximum knee flexion we could allow further confirm Feuerbach et al. (1993) observations.

Platform Differences

One surmised method of an inversion sprain is landing on an uneven surface, e.g., another person's foot (Garrick, 1977; Shapiro et al., 1994) that produces excessive inversion strain and tensile stress to the ATFL and CFL. Self (1996) measured the strain and strain rate of the ATFL and CFL that occurs during a potential ankle sprain situation, i.e., landing onto a 30° inverted V platform as was used in this study. After determining the loading parameters, e.g., maximum inversion velocity from actual participants who performed drop landings onto a flat surface, the lower extremity of cadavers were dropped from 0.13 m onto the flat and inverted platform. Significant increases in strain and strain rate values were observed for the ATFL and CFL of the cadavers for the inverted landing conditions compared to the values of the flat landing condition. Therefore, to indirectly measure the strain of the ATFL and CFL that would simulate landing on an uneven surface, the angular displacement of the rearfoot for in/eversion and plantar/dorsiflexion directions of motion for the brace conditions were compared between the flat and inverted V platform conditions in this study.

As anticipated, there were significant differences between in/eversion variables landing on a typical, flat surface and on a surface similar to landing on an uneven surface that creates a potentially injurious situation. For all brace conditions, the sloped surface landing exhibited greater maximum inversion motion than the flat surface landing. However, there was inversion restraint provided by the braces during the sloped surface landing. To put the maximum inversion brace values into perspective, the means of either brace for the inverted platform condition were slightly less than that of the no brace, flat landing, i.e., a typical landing. This finding suggests that either brace passively restrained inversion when landing on a sloped surface similarly to a typical landing when no brace is worn. Although the actual inversion restraint provided by either brace when landing during a truly injurious situation is not known, this finding suggests that due to passive brace restraint, the amount of inversion exhibited (during a drop landing onto a 30º inverted surface) is similar to a typical landing performed without a brace.

Of further interest regarding landing on an inverted surface versus a flat surface is the surmised differences in foot ankle complex landing mechanics. In addition, the landing mechanics may underlie the observed platform differences for in/eversion variables. During a typical landing, the foot-ankle complex lands on the antero-lateral part of the plantar surface of the foot causing the vertical ground reaction forces (VGRF) to produce an eversion torque. In contrast, during an atypical, sloped surface landing, the foot-ankle complex lands on the medial side of the plantar surface, causing the VGRF to produce an inversion torque. Therefore the maximum magnitude and time to maximum inversion was less but maximum eversion and the total in/eversion angular displacement was greater for the flat versus the inverted platform condition.

Plantar/dorsiflexion angular kinematic differences were also observed among the platform conditions. While the participants landed significantly more plantarflexed on the sloped surface when compared to the flat surface, the difference was not more than 2° as hypothesized. Therefore, the typical and atypical landings were not considered to behaviorally vary from one another. For maximum dorsiflexion and dorsiflexion angular displacement, the lower magnitudes of the sloped surface compared to the flat surface may have been influenced by the tendency for participants to land with greater ankle abduction angle.

As the amount of motion allowed about a given axis is dependent on the positioning of the foot relative to other foot axes, slope differences for dorsiflexion variables may also have been influence by the in/eversion foot position. It is known that while the foot-ankle complex is in a dorsiflexed position inversion is limited due to the talus being wedged into the tibia-fibula mortise (Hamill & Knutzen). Due to anatomical constraints, the wider portion of the talus (anterior) may not fit properly into the mortise while the foot-ankle complex is in an inverted position, is one possible explanation for less maximum dorsiflexion for the inverted compared to the flat platform condition.

In conclusion, although increased inversion did occur for both braces during the 30º inverted surface landing compared to the flat surface landing, the maximum inversion values of the brace conditions were less than the control condition and approximately the same as landing on the flat surface without a brace. Although it was hypothesized that the Malleoloc™ brace would inhibit plantar/dorsiflexion motion due to the locations of the stirrups, it is not known why the Active Ankle™ also inhibited maximum plantarflexion at touchdown. the Active Ankle™, a hinge design, did not exhibit increased motion restraint for the in/eversion direction of motion when compared to the Malleoloc™, a non-hinge design. However, both braces demonstrated inversion restraint compared to not wearing a brace. An efficacious brace is believed to be one that restricts inversion motion while not inhibiting dorsiflexion displacement. Based on the inversion restraint criteria, for both braces, inversion restraint was exhibited, with no significant differences between the braces. Based on the values of the dorsiflexion variables, it is suggested that the second criteria also was fulfilled by the hinge design brace (Active Ankle™).

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CHAPTER IV

SUMMARY AND CONCLUSIONS

The passive range of motion tests used to evaluate inversion restraint of semi-rigid braces in previous investigations (Alves et al., 1992; Greene & Wight, 1990; Lindley & Kernoozek, 1995; Siegler et al. 1997; Wiley & Nigg, 1996) do not necessarily reflect the strain placed on the foot-ankle complex nor the strain placed on the semi-rigid brace during an open-chained movement (Siegler et al., 1997). Therefore, testing the efficacy of semi-rigid braces during a dynamic movement provides a better understanding of whether or not these braces constrain rearfoot movement. The purpose of the study was to compare the rearfoot kinematics that occur when wearing a non-hinged brace (Malleoloc™) versus a hinged brace (Active Ankle™) and to determine the in/eversion and plantar/dorsiflexion motion restriction of these braces relative to not wearing a brace on a potentially non-injurious (flat) landing surface and potentially injurious (inverted) landing surface. Thus the results of the study may help to determine if the deformation of the ATFL and CFL (Self, 1996) can be minimized by the use of a particular brace design.

There were few significant differences for kinematic variables between the two brace types. However, both the Malleoloc™ and Active Ankle™ braces demonstrated significantly less maximum inversion and inversion angular displacement when compared to not wearing a brace. These differences were likely due to the motion restraint provided by the braces during the movement rather than restricting the foot position prior to landing. Both braces were observed to decrease maximum plantarflexion at touchdown compared to not wearing a brace. In addition, the Malleoloc™ brace exhibited less maximum dorsiflexion compared to the Active Ankle™ brace and the control condition. Significantly less dorsiflexion displacement was detected between:

1) the Malleoloc™ and the Active Ankle™, 2) the braces and the control. By wearing either semi-rigid brace, the decrease in plantarflexion motion may also have decreased the strain of the ATFL, as the ATFL lengthens with plantarflexion.

To ensure consistency of landing technique for all brace-platform conditions, the participants were required to flex their knees during landing to the same self-selected angle. Among the brace and landing conditions, the means of the maximum knee flexion angle only ranged from 66° to 70°, confirming that this was accomplished. However, significant differences in knee flexion angle at touchdown while wearing the Malleoloc™ or the Active Ankle™ brace in comparison to the control condition may indicate a subconscious attempt by the participant to counteract the decreased range of dorsiflexion motion at the ankle joint.

Neither semi-rigid brace exhibited greater inversion motion restraint than the other semi-rigid brace. Both brace exhibited greater restraint than the control condition for either landing condition. As landing on an inverted slope surface compared to a flat surface potentially causes greater inversion torques to be applied to the foot-ankle complex, the use of either brace may provide restraint during landings on an inverted surface similar to landing on another person's foot.

An efficacious brace is believed to be one that restricts inversion motion while not inhibiting dorsiflexion motion. Based on these two goals, the Active Ankle™, a hinged brace, accomplished both goals while the Malleoloc™ brace only accomplished the inversion motion restraint goal. However, in choosing a brace to prevent subsequent ankle sprains, the brace must first be proven effective by prospective studies and deemed beneficial and comfortable by the user.

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APPENDIX A

ELIGIBILITY QUESTIONNAIRE

Name ______________________ E-mail _______________________

Age ___________ Gender: M or F Phone: _______________________

Best day & times to reach you at home: _______________________ Shoe Size _______

1. Please check any/all sports that you have participated in and the extent of participation.

	High School Varsity	College Intramural	College Varsity	Recreational (Organized)	Recreational (Unorganized)
Basketball					
Volleyball					
Tennis					
Racquetball					
Baseball					
Softball					
Football					
Soccer					

Please note others not listed above: ____

2. If you checked a box for any sport, how long have you participated in that sport?

	Months	Years
Basketball		
Volleyball		
Racquetball		
Baseball		
Softball		
Football		
Soccer		

Please note others not listed above: ____

3. Which ankle? R or L or BOTH

4. When was the last injury in terms of months or years ago? For sprains to both ankles please indicate separately. _______________________________________

5. If you had to rate the degree of the severity of the ankle sprain, would it be close to a (Circle the number below): 1 (*Grade I*) relates to a mild sprain with minimal swelling; was sore for a few days but did not limit activity; 2 (*Grade II*) relates to a moderate sprain with increased swelling and pain which limited activity and range of motion; 3 (*Grade III*) relates to a severe sprain with total loss of range of motion and no activity.

   Please Circle One Number: R 1 2 3 L 1 2 3

6. Please answer Yes or No to the following for the last ankle sprain.

   Did you seek medical attention? *Y or N*

   Did you ice the injured ankle? *Y or N*

   Did you elevate or use ace bandage to compress the ankle? *Y or N*

   Did you use crutches? *Y or N*

   Did you do any rehabilitation exercises? *Y or N* If Yes, how long __________

7. If you have had a previous ankle injury, do you typically use a brace during physical activity? *Y or N*

8. If yes, do you have a brace preference? Please list type ______________________

9. Have you had any pain or injury in the last year to lower extremity (below hip)?

   Y or N If yes, please briefly describe _______________________________________

   When was the last injury? Month ____ Year ____

10. Have you had any back pain that would hinder from drop landing repetitively?

    Y or N

11. Have you had any shoulder pain that would hinder you from hanging from a bar repetitively? Y or N

12. Have you had any surgery or other medical procedures performed on any region of the lower extremity? *Y or N* If yes, please briefly describe _____________________
    ____

APPENDIX B

PARTCIPANT SELF-REPORTED SPORT PARTICIPATION

Physical Activity									
ID#	BB	VB	TN	RB	BSB	SB	FB	TR	CL
1	HS, I		R	R	HS			CV	
2	HS, I		HS R	R					
3	R	R	R			R	R		
4	R	I	R	R	R	R	R		
5				I					
6		HS,I		R		R			
7	I, R								
8	I, R					I,R	I,R	R	
9	I, R	I, R	I, R	R	I, R	I, R	HS, I	HS	
10	I, R	I, R		I, R		I, R	I, R		
11		R		R			HS, CV		
12	R	R	R		HS	I	HS	HS	
13	R		R	R	HS, R	R	R	HS	
14	R		HS, R						
15			R					R	
16			HS, R			I			

17	HS, I				HS, R	I	I		
18	R	R		R		HS, R		HS	
19		R	R	I, R					
20		CV, R	R	R		R	R	HS	
21									
22			R	R	R		HS		
23	HS, I					HS, R	I		HS
24	HS, I	R	R	R		R			
25					R	I	HS		
26						R			
27			R			R	R		

Note. HS = High School Varsity; I = College Intramural; CV = College Varsity; R = Recreational; BB = Basketball; VB = Volleyball; TN = Tennis; RB = Racquetball; BSB = Baseball; SB = Softball; FB = Football; RT = Running/Track; CL = Cheerleading.

APPENDIX C

PARTICIPANT ANTHROPOMETRIC DATA

Anthropometric Data							
ID #	Gender	Age (yrs)	Height (cm)	Mass (kg)	TH Length (cm)	LL Length (cm)	FT Length (cm)
1	M	24	167	78	42.5	40.5	26.0
2	M	20	190	107	45.0	50.0	29.0
3	M	20	168	61	44.0	39.0	25.0
4	M	30	182	80	47.5	46.0	27.5
5	M	22	177	73	45.0	45.0	30.0
6	F	20	177	66	47.5	42.0	25.5
7	M	20	172	64	45.5	46.0	26.5
8	M	21	177	74	46.8	44.0	25.5
9	M	22	182	80	45.0	43.0	27.5
10	M	29	166	72	42.0	39.0	24.3
11	M	51	172	83	40.0	46.0	25.5
12	M	21	177	93	43.0	43.0	26.0
13	M	19	186	84	45.0	45.5	27.0
14	M	20	188	73	46.5	48.0	29.0
15	F	19	167	68	47.5	41.0	24.0
16	F	22	157	57	42.0	37.0	23.5
17	M	22	179	80	48.0	44.5	26.0

18	F	22	163	61	46.5	41.0	23.0
19	M	19	177	84	48.0	43.0	26.0
20	M	22	175	61	44.0	43.0	26.5
21	F	18	174	68	48.5	43.0	26.0
22	M	23	182	84	41.5	46.0	27.0
23	F	20	177	64	46.5	42.5	26.8
24	F	26	165	59	41.0	41.0	23.0
25	M	19	188	118	45.0	45.0	25.0
26	F	19	157	55	42.0	36.0	22.0
27	F	18	157	56	41.5	38.0	23.0

Note. M = Male; F = Female; Th Length = Thigh Length, LL Length = Lower Leg Length; FT Length = Foot Length, Thigh Length = the distance between head of femur to lateral femoral condyle. Lower Leg Length = the distance between head of lateral tibia to distal tip of lateral malleous. Foot Length = the distance between posterior calcaneus to longest phalanx.

APPENDIX D

RANGE OF MOTION EVALUATION RESULTS

Values of the ankle evaluation for plantar (PF)/dorsiflexion(DF), in(INV)/eversion(EV), and subtalar in(SINV)/eversion(SEV)

ID#	Leg	PF	DF	INV	EV	SINV	SEV
1	Right	40	10	35	20	5	5
	Left	45	10	30	20	5	4
2	Right	50	10	40	20	8	5
	Left	40	10	25	25	5	5
3	Right	55	20	40	30	5	5
	Left	45	20	35	32	5	5
4	Right	50	10	30	20	5	5
	Left	50	5	30	20	8	3
5	Right	50	5	25	20	5	3
	Left	55	10	30	20	5	5
6	Right	50	10	30	25	5	5
	Left	50	10	30	25	6	3
7	Right	45	20	30	20	5	5
	Left	55	20	35	25	8	5
8	Right	50	20	30	15	5	7
	Left	50	15	25	20	7	5
9	Right	55	15	35	20	5	5
	Left	50	20	40	15	3	5
10	Right	55	15	25	20	5	3
	Left	50	15	30	20	5	5
11	Right	50	10	35	20	5	5
	Left	55	15	30	15	5	5

12	Right	55	10	25	15	5	5
	Left	40	10	20	15	5	3
13	Right	60	15	35	20	5	3
	Left	60	20	30	25	5	5
14	Right	30	10	35	20	5	5
	Left	60	15	35	20	5	5
15	Right	50	10	35	20	5	5
	Left	55	10	35	25	5	5
16	Right	55	15	30	15	5	5
	Left	55	10	35	25	5	3
17	Right	55	10	25	20	5	5
	Left	55	10	30	20	5	5
18	Right	50	10	35	20	5	5
	Left	55	10	35	20	5	5
19	Right	60	15	30	15	8	5
	Left	60	15	35	20	8	5
20	Right	60	20	30	20	5	5
	Left	60	20	35	25	5	5
21	Right	45	20	45	20	5	3
	Left	55	15	45	30	7	5
22	Right	60	10	30	15	5	5
	Left	60	15	30	15	5	5
23	Right	60	15	35	20	5	5
	Left	60	10	35	20	5	4
24	Right	55	10	25	25	5	5
	Left	50	10	30	25	5	5
25	Right	55	0	20	30	5	5
	Left	50	5	30	30	5	2
26	Right	60	10	35	30	5	5
	Left	50	10	30	25	5	5
27	Right	60	10	25	30	5	8
	Left	55	10	30	25	8	5

APPENDIX E

COORDINATE SYSTEM AND ANGULAR KINEMATIC METHODOLOGY

Segment Coordinate Systems

See Figure 9.

Thigh Coordinate System (THCS) $< \mathbf{i}_{\mathbf{TH}}, \mathbf{j}_{\mathbf{TH}}, \mathbf{k}_{\mathbf{TH}} >$

Let TH_1, TH_2, TH_3 represent the thigh segment markers of the hip, lateral thigh and anterior thigh, respectively. To generate the TCHS:

$$\mathbf{TH}_{\mathbf{12}} = TH_1 - TH_2$$

$$\mathbf{TH}_{\mathbf{23}} = TH_2 - TH_3$$

$$\mathbf{k}_{\mathbf{TH}} = \mathbf{TH}_{\mathbf{12}} / |\, \mathbf{TH}_{\mathbf{12}} \,|$$

$$\mathbf{i}_{\mathbf{TH}} = \mathbf{TH}_{\mathbf{23}} \times \mathbf{k}_{\mathbf{TH}} / |\, \mathbf{TH}_{\mathbf{23}} \times \mathbf{k}_{\mathbf{TH}} \,|$$

$$\mathbf{j}_{\mathbf{TH}} = \mathbf{k}_{\mathbf{TH}} \times \mathbf{i}_{\mathbf{TH}} / |\, \mathbf{k}_{\mathbf{TH}} \times \mathbf{i}_{\mathbf{TH}} \,|$$

Lower Leg Coordinate System (LLCS) $< \mathbf{i}_{\mathbf{LL}}, \mathbf{j}_{\mathbf{LL}}, \mathbf{k}_{\mathbf{LL}} >$

Let LL_1, LL_2, LL_3 represent the lower leg segment markers of the lateral lower leg, anterior lower leg and distal lower leg, respectively. To generate LLCS:

$$\mathbf{LL}_{\mathbf{23}} = LL_2 - LL_3$$

$$\mathbf{LL}_{\mathbf{12}} = LL_1 - LL_2$$

$$\mathbf{k}_{\mathbf{LL}} = \mathbf{LL}_{\mathbf{23}} / |\, \mathbf{LL}_{\mathbf{23}} |$$

$$\mathbf{i}_{\mathbf{LL}} = \mathbf{LL}_{\mathbf{12}} \times \mathbf{k}_{LL} / |\, \mathbf{LL}_{\mathbf{12}} \times \mathbf{k}_{\mathbf{LL}} \,|$$

$$\mathbf{j}_{\mathbf{LL}} = \mathbf{k}_{\mathbf{LL}} \times \mathbf{i}_{\mathbf{LL}} / |\, \mathbf{k}_{\mathbf{LL}} \times \mathbf{i}_{\mathbf{LL}} \,|$$

<u>Foot Coordinate System (FTCS)</u> $< \mathbf{i_{FT}}, \mathbf{j_{FT}}, \mathbf{k_{FT}} >$

Let FT_1, FT_2, FT_3 FT_4 represent the foot segment markers of the proximal calcaneus, distal calcaneus, lateral calcaneus, and head of fifth metatarsal, respectively. To generate the FTCS:

$\mathbf{FT_{12}} = FT_1 - FT_2$

$\mathbf{FT_{43}} = FT_4 - FT_3$

$\mathbf{k_{FT}} = \mathbf{FT_{12}} / | \mathbf{FT_{12}} |$

$\mathbf{i_{FT}} = \mathbf{FT_{43}} \times \mathbf{k_{FT}} / | \mathbf{FT_{43}} \times \mathbf{k_{FT}} |$

$\mathbf{j_{FT}} = \mathbf{k_{FT}} \times \mathbf{i_{FT}} / | \mathbf{k_{FT}} \times \mathbf{i_{FT}} |$

<u>Joint Coordinate Systems</u>

1. Ankle JCS

   $\mathbf{e_3} = \mathbf{k_{FT}}$

   $\mathbf{e_1} = \mathbf{i_{LL}}$

   $\mathbf{e_2}$ = floating axis = $\mathbf{e_3} \times \mathbf{e_1}$

2. Euler angles of the right foot-ankle complex: (foot displacement relative to the lower leg)

   plantar/dorsiflexion = $90° - \arccos(\mathbf{k}_{LL} \cdot \mathbf{e_2})$

   in/eversion = $90° - \arccos(\mathbf{k}_{FT} \cdot \mathbf{e_1})$

   ab/adduction = $\arccos(\mathbf{i}_{FT} \cdot \mathbf{e_2})$

3. Knee JCS

   $\mathbf{e_{3'}} = \mathbf{k_{LL}}$

   $\mathbf{e_{1'}} = \mathbf{i_{TH}}$

   $\mathbf{e_{2'}}$ = floating axis $\mathbf{e_{3'}} \times \mathbf{e_{1'}}$

4. Euler angle for the right knee (lower leg displacement relative to the thigh):

   knee flexion = $\arccos(\mathbf{k}_{TH'} \cdot \mathbf{e_{2'}}) - 90°$

APPENDIX F

ANOVA RESULTS OF ALL DEPENDENT VARIABLES

Kinematic Variable	Source	df	F	p-value	Eta Squared	Power
Plantarflexion at Touchdown	Brace	2	10.49	<.001	.287	.984
	Error	26				
	Platform	1	11.65	.002	.310	.907
	Error	52				
	Br x Pl	2	0.21	.812	.008	.081
	Error	52				
Maximum Dorsiflexion	Brace	2	5.12	.009	.167	.807
	Error	26				
	Platform	1	234.48	<.001	.900	1.000
	Error	52				
	Br x Pl	2	0.20	.817	.008	.080
	Error	52				
Relative Time to Maximum Dorsiflexion	Brace	2	3.01	.058	.104	.559
	Error	26				
	Platform	1	3.30	.081	.113	.417
	Error	52				
	Br x Pl	2	0.13	.883	.005	.068
	Error	52				
Dorsiflexion Angular Displacement	Brace	2	44.15	<.001	.629	1.000
	Error	26				
	Platform	1	51.55	<.001	.665	1.000
	Error	52				
	Br x Pl	2	0.46	.633	.017	.121
	Error	52				

Inversion Angle at Touchdown	Brace[2]	2	1.02	.356	.038	.201
	Error	26				
	Platform	1	1.22	.609	.010	.079
	Error	52				
	Br x Pl	2	0.14	.870	.005	.070
	Error	52				
Maximum Inversion	Brace	2	5.24	.008	168	.811
	Error	26				
	Platform	1	13.79	.001	.347	.947
	Error	52				
	Br x Pl	2	0.07	.929	.003	.061
	Error	52				
Relative Time to Maximum Inversion	Brace	2	2.91	.063		.545
	Error	26				
	Platform	1	46.70	<.001	.642	1.000
	Error	52				
	Br x Pl	2	1.01	.370	.037	.217
	Error	52				
Inversion Angular Displacement	Brace	2	3.45	.039	.117	.621
	Error	26				
	Platform	1	17.21	<.001	.398	.979
	Error	52				
	Br x Pl	2	0.23	.747	.009	.082
	Error	52				
Ab/Adduction at Touchdown	Brace	2	1.06	.355	.039	.225
	Error	26				
	Platform	1	0.56	.461	.021	.111
	Error	52				
	Br x Pl	2	1.89	.165	.067	.371
	Error	52				
Maximum Abduction	Brace	2	1.24	.298	.046	.258
	Error	52				
	Platform	1	52.27	<.001	.669	1.000
	Error	26				
	Br x Pl	2	3.91	.026	.131	.680
	Error	52				

[2] Sphericity not assumed, Huynh-Feldt = 0.831.

Abduction Angular Displacement	Brace [3]	2	2.41	.115	.085	.402
	Error	26				
	Platform	1	2.08	.162	.074	.284
	Error	52				
	Br x Pl	2	0.79	.459	.030	.148
	Error	52				
Knee Flexion at Touchdown	Brace [4]	2	5.30	.012	.169	.759
	Error	26				
	Platform	1	0.18	.677	.007	.060
	Error	52				
	Br x Pl	2	0.22	.801	.008	.083
	Error	52				
Maximum Knee Flexion	Brace	2	0.14	.870	.005	.070
	Error	26				
	Platform	1	2.29	.142	.081	.308
	Error	52				
	Br x Pl	2	0.70	.463	.026	.146
	Error	52				
Knee Flexion Angular Displacement	Brace	2	2.54	.089	.089	.468
	Error	26				
	Platform	1	0.98	.332	.036	.159
	Error	52				
	Br x Pl	2	0.95	.392	.035	.207
	Error	52				

[3] Sphericity not assumed, Huynh-Feldt = 0.767
[4] Sphericity not assumed, Huynh-Feldt = 0.833