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Biomechanical effects of ankle bracing during gait

Steven R. Casto

To the Graduate Council:

I am submitting herewith a thesis written by Steven R. Casto entitled "Biomechanical effects of ankle bracing during gait." I have examined the final electronic copy of this thesis for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Master of Science, with a major in Human Performance and Sport Studies.

Songning Zhang, Major Professor

We have read this thesis and recommend its acceptance:

Wendell Liemohn, David Bassett

Accepted for the Council:
Carolyn R. Hodges

Vice Provost and Dean of the Graduate School

(Original signatures are on file with official student records.)

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Biomechanical Effects of Ankle Bracing During Gait

A Thesis
Presented for the
Master of Science
Degree
The University of Tennessee, Knoxville

Steven R. Casto, ATC
May 1999

DEDICATION

This thesis is dedicated in its entirety to my mother, Doris Casto, and my late grandmother, Venia Caruthers. These two individuals have taught me more about life than I ever could have hoped to learn, and for this I will be forever grateful. Most importantly, I have learned from them that it is quality, not quantity, that matters most in life.

ACKNOWLEDGEMENTS

This project could not have been completed without the help of many individuals. I would like to first thank my advisor, Dr. Songning Zhang, for his guidance during my stay at UT. Going into a biomechanics program from an athletic training background wasn't easy, but I was able to find some common ground thanks to Dr. Zhang. I would also like to thank my two additional committee members, Dr. Wendell Liemohn and Dr. David Bassett, for their patience and valuable insights into my study. In addition, thanks must be given to Marisa Miller and Traci Haydu, two unbelievable friends who assisted me during my data collection. And finally, to Kerrie. She has made me sit up and enjoy life for what it is, and not to take anything for granted. I'm going to miss you, kid.

ABSTRACT

The purpose of this study was to examine the effects of three different ankle braces on rearfoot motion and ground reaction force (GRF) data. The braces used included the Aircast Air-Stirrup, Aircast Sport-Stirrup, and Active Ankle. Ten healthy and active male subjects, with no history of lower extremity injury, served as subjects for the study. Rearfoot kinematics (Panasonic, 60 Hz) and ground reaction forces (AMTI, 1000 Hz) were sampled simultaneously during data collection. Each subject performed five walking trials (at his own pace) across a walkway without a brace and with each brace in a total of four conditions. Customized software was used to compute variables describing rearfoot motion as well as vertical, anterior-posterior, and medial-lateral GRF. All kinematic variables indicated a trend toward greater rearfoot control with the braces. Of these, time to maximum eversion angle (TMaxEV) and toe-off angle (TOAngle) were found to have significant differences. For the kinetic analysis, the three GRF components indicated a trend toward rearfoot control during the braced conditions. Excursion values from 0-30% (Exc1) and 0-50% (Exc2) of the stance phase were found to have significant differences. Braking impulse (IBrk) was found to be the only significant anterior-posterior GRF variable, while no significant variables were noted for the vertical GRF component. The braces in this study seemed to be able to control and stabilize rearfoot movement to an extent, although this was dependent on the design and intended use of each brace.

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LIST OF ABBREVIATIONS

ROM	Range-of-Motion
GRF	Ground Reaction Force
TDAngle	Touchdown Angle
MaxEV	Maximum Eversion Angle
TMaxEV	Time to Maximum Eversion Angle
EROM	Total Eversion Range-of-Motion
TOAngle	Toe-Off Angle
MaxVel	Maximum Eversion Velocity
TMaxVel	Time to Maximum Eversion Velocity
F1	First Peak Force
T1	Time to First Peak Force
F2	Second Peak Force
T2	Time to Second Peak Force
MaxBrk	Maximum Breaking Force
TMaxBrk	Time to Maximum Breaking Force
MaxProp	Maximum Propulsive Force
TMaxProp	Time to Maximum Propulsive Force
IBrk	Braking Impulse
IProp	Propulsive Impulse
Exc1	Force Excursions (0-30%)
Exc2	Force Excursions (0-50%)
Exc3	Force Excursions (0-100%)

Chapter 1

INTRODUCTION

Few injuries can compare to the lateral (inversion) ankle sprain, in terms of frequency, in athletic populations. A widely cited study [20] first published in 1977 indicated an ankle injury rate of 1 injury for every 17 participants per season (with 85% of these injuries being diagnosed as sprains). In an era where athletes are stronger and faster than their predecessors, these numbers might even be elevated. It is difficult to find an athletic activity that does not involve the foot/ankle complex in some manner, thus the tremendously high potential for injury. Statistics have documented that these injuries can account for up to 14% of all athletic injuries and consequently lead to the greatest number of days lost in athletics [20].

Treatment and preventive methods for the ankle sprain have changed over the years. A player with a first-degree lateral ankle sprain can now return within a matter of days, whereas 10 or 20 years ago he/she would have been out of activity much longer. Advances in treatment methods, however, come along with appropriate choices a sports therapist must make for different athletic situations. The abundance of treatment options presents a unique challenge. A treatment must offer the best healing environment without risk of re-injury and/or developing chronic problems. Taping and, more recently, ankle bracing have become two very popular preventive options when dealing with ankle sprains. The impact of each on controlling extreme ranges of motion (ROM), along with a brief anatomy review of the ankle will be presented in the following sections.

Anatomy

A large percentage of ankle sprains involves the lateral ankle complex, which consists of three primary ligaments: the anterior talofibular (ATF), calcaneofibular (CF), and posterior talofibular (PTF). The structure of the ankle joint tends to make it more susceptible to a sprain of the lateral complex. The talus fits into a mortise formed by the distal ends of the tibia and fibula. The talus itself is narrower posteriorly and naturally becomes weaker and less stable when the foot is placed into plantarflexion. In addition, the medial malleolus is slightly shorter than the lateral malleolus; as a result, an inversion movement can occur easier than eversion. Thus a typical mechanism for an ankle sprain is when the ankle is forced into inversion and plantarflexion; typically the ATF is the first ligament to be injured. The CF and, very rarely, the PTF, can be injured in more severe sprains. Figure 1 presents a review of ankle anatomy with emphasis on those structures typically associated with a lateral sprain of the ankle.

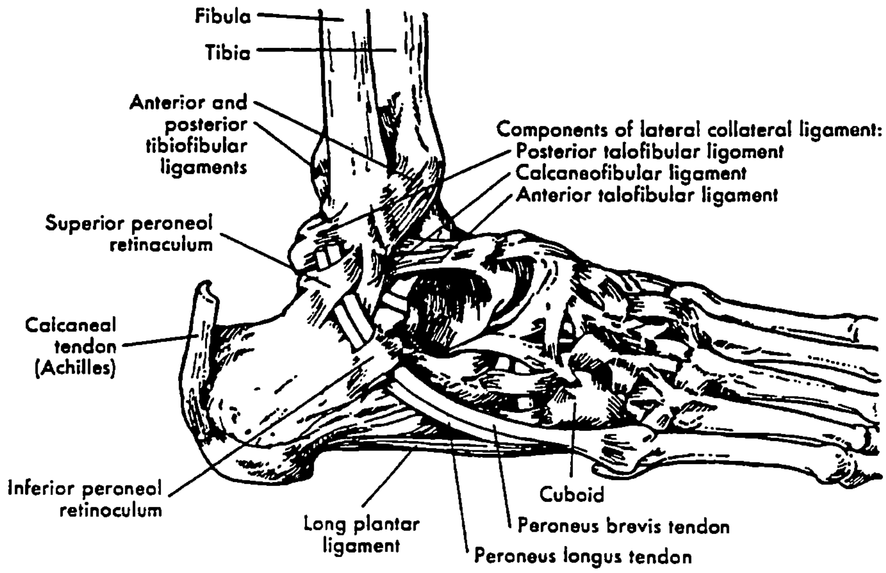


Figure 1. Ligaments of the ankle joint (lateral view). [From: Booher JM, Thibodeau GA. *Athletic Injury Assessment*. St. Louis, MO: Mosby-Year Book, Inc; 1994:408].

Treatment Options

With a variety of treatment and preventive options being employed today, the ankle sprain can present a complex challenge to the sports therapist. Severity of injury, location, previous injury, strength and flexibility of the joint, and available resources are all factors that need to be considered when deciding what is best for the athlete. The therapist must be properly trained and skilled in making accurate assessments so that the correct course of action can be taken. After the athlete has undergone successful treatment for the injury and is ready to begin the final phase of rehabilitation (the return to activity), some form of external assistance is commonly indicated. This not only protects the previously injured structures but also gives the athlete a sense of reassurance and confidence.

The use of athletic tape as a support mechanism during ankle injuries has been a very common practice for decades. However, controversy exists in the literature as to whether tape can effectively serve as an "external ligament" for an extended time during athletic activity in an attempt to prevent ankle injuries. The time that looseness begins to occur is a critical event and depends on a number of factors, including tightness of the taping (as done by the therapist), type of activity, and when the tape is applied in relation to the activity, among others. Studies have suggested that a maximal decrease in tape's ability to restrict inversion/eversion ROM can occur as early as 10 to 20 minutes into the activity [19, 21]. A 69% decrease in support has been shown after a 40-minute exercise bout [30].

In addition, studies comparing the effects of tape versus other external forms of support have shown tape to be an inferior option. One particular study noted that players with taped ankles had twice the risk of injury as did others with alternative methods of support [38]. Fumich [19] has reported that the minimum degrees of restriction that can be expected from tape is 4.39° after an extended activity (2-3 hours) for the motion of plantarflexion/inversion, a primary mechanism for lateral ankle sprains. What is not known is whether this restriction is enough to protect against ankle injuries, either in uninjured patients or those with more chronic problems.

While taping is still a common choice as a form of support, ankle bracing has become more popular in the last two decades. A brace can offer many advantages compared to taping: ease of application, cost-effectiveness, comfort, ability to re-tighten as needed, and a relatively quick application time are a few of the reasons why bracing has become a viable option to ankle taping. Braces can vary in composition and structure, and can be used in a wide variety of situations. Different types of ankle braces have been indicated in the treatment of stable malleolar fractures, distal tibia/fibula stress fractures, and post-operative conditions of the lower extremity [7]. Thus, an ankle brace can be very versatile.

Biomechanical studies involving the use of ankle braces have drawn some interesting conclusions. A landmark study involving the Aircast Air-Stirrup showed a restriction of inversion ROM from an initial (unbraced) range of 5.0-8.0° to a range of 2.0-5.0° with the brace during running [43]. A specially designed shoe was used (medially hard, laterally soft) to prove the effectiveness of the brace. The Active Ankle and Aircast Sport-Stirrup have also been shown to restrict inversion ROM significantly more than other braces before and during an exercise session (allowing an average of only 2.99° of inversion compared with the control values [27].

A study evaluating different biomechanical parameters during the stance phase of walking demonstrated even further the restricting effects of an ankle brace [23]. A total of three variables were significantly affected as a result of the brace: maximum calcaneal eversion angle, time to maximum calcaneal eversion velocity, and total rearfoot motion. In this situation the brace was able to prevent an extreme range of rearfoot motion, but still allow functional movements to occur (i.e., dorsiflexion, plantarflexion). Studies have also shown a less inverted position during touchdown (during the stance phase of gait) with the use of a brace [10, 12]. Translating this finding to athletics could mean a reduced risk of inversion ankle injuries.

As with taping, however, the question that arises with ankle bracing is that of the amount of restriction and the ability of the brace to keep this restriction for an extended period of time. Braces appear to be superior to taping in this regard, although the restriction is individual in nature and depends on a number of factors [43]. Studies presenting pre- and post-exercise data (objectively measuring the effects of the brace) are very prevalent in the literature, but often are specific to a particular group of athletes (i.e., volleyball or basketball players). It would be ideal to be able to test every athletic team in this manner; however, time, cost, and available resources make this virtually impossible. If an expected amount of restriction could be determined, the process of choosing a brace for a particular situation could be made easier. Depending on the nature, severity, and stage of an injury, a brace might be needed to accommodate increased movement (final stages of rehabilitation) or maximum restriction (acute injury).

This study addressed the need for baseline data in foot/ankle biomechanics associated with the use of ankle braces. The purpose was to evaluate the effects of different ankle braces on biomechanical responses of the ankle complex. Specifically, the following purposes were to be accomplished:

1) Examination of rearfoot inversion and eversion during the stance phase of a gait cycle.
2) Evaluation of ground reaction force (GRF) data during the stance phase of gait.

Delimitations

This study was conducted within the following delimitations:

1) Ten male subjects were selected from the student population at the University of Tennessee, Knoxville with no history of significant lower extremity injuries.
2) Testing was performed under laboratory conditions (and not in a true athletic environment).
3) One testing session was required of all subjects.
4) Data were collected and analyzed through use of a force platform (AMTI, 1000 Hz) and video system (Panasonic, 60 Hz) during the stance phase of gait.

Limitations

This study was limited by the following factors:

1) Subjects were selected from the University of Tennessee, Knoxville student population.

2) Only three different ankle braces were used in the study.

3) Difference in sampling frequency of the force platform (1000 Hz) and video system (60 Hz).

4) Possible errors arising from manual digitization of trials.

5) Accuracy of the involved instrumentation: potential problems were acknowledged and accepted based on previous work and manufacturer's recommendations.

Hypotheses

The results were analyzed according to the following hypotheses:

1) Use of an ankle brace will result in reduction of total rearfoot eversion ROM during a gait cycle.

2) The Aircast Air-Stirrup will show greater rearfoot control according to the kinematic analysis than the other braces due to its design and intended usage (acute injuries).

3) Application of an ankle brace will affect the medial-lateral ground reaction force.

Assumptions

The following assumptions were made concerning this study:

1) There was no difference in the walking style of subjects.

2) Use of a manual goniometer was an accurate device for measuring passive ankle inversion/eversion ROM.

Chapter 2

REVIEW OF LITERATURE

Introduction

Studies involving the use of ankle braces are frequently seen in sports medicine literature. Many have documented effects of ankle braces on ankle ROM using instruments specifically designed to measure such motion (usually during exercise). However, fewer have attempted to examine effects through biomechanical analysis. This method provides an accurate, objective measure of the effect of the brace during a dynamic activity. This review of literature will mainly focus on two aspects: ROM restriction in conjunction with exercise and athletic performance, and restriction during gait. The effect on functional activities as the result of ankle braces will also be discussed, as will the effects of the brace on ground reaction force (GRF) parameters.

Biomechanical Studies

Hamill et al. [23] used the Aircast Air-Stirrup to investigate rearfoot kinematic variables before and after an exercise bout. Testing consisted of 10 walking trials in each of four conditions: pre-exercise, pre-exercise with the brace, post-exercise, and post-exercise with the brace. The exercise regimen consisted of 70 maximal eccentric actions of the ankle evertor muscles with 15 seconds between each movement. Force platform data were sampled at 500 Hz while a video system simultaneously filmed each trial at 100 Hz. As part of the kinematic analysis, it was found that the exercise put the foot into a more inverted position upon contact, yielding significant differences in touchdown angle. The exercise also reduced total rearfoot motion ($p<0.05$). During the braced conditions, three kinematic variables were found to be affected: maximum calcaneal eversion angle and total rearfoot motion were reduced, whereas time to maximum calcaneal eversion velocity was delayed ($p<0.05$). The authors concluded that the orthosis was effective in preventing the ankle from moving into extreme ranges of motion while still allowing normal functional movements (dorsiflexion and plantarflexion).

Using a total of four different braces, Scheuffelen et al. [39] studied the Achilles tendon angle (indicative of rearfoot motion) using treadmill conditions. Four different braces were used: the Aircast Air-Stirrup, Mikros Ankle Brace type 'FT,' MHH-Splint Caligamed, and a Push® brace type 'Heavy.' A total of 15 gait cycles were evaluated for each brace, with the velocity for the conditions being set at 8 and 12 km/hr. In contrast to kinematic analysis, a twin axis goniometer (positioned from the Achilles tendon to the dorsal calcaneus) placed directly over the subject's skin was used to record the Achilles tendon angle. At the beginning of ground contact (touchdown), it was found that the Aircast and Caligamed braces slightly inverted the foot. None of the braces were able to prevent inversion and eversion ROM totally; however, most of them were able to reduce it during the course of the testing. The braces did not affect the total range of angular displacement (for each gait cycle). This value, however, did not exceed 10.0°.

Using only the Aircast Air-Stirrup, Stuessi et al. [43] evaluated running strides of 11 subjects while wearing a specially designed shoe (medially very hard, laterally very soft) in an effort to supinate the foot immediately after touchdown. It was hypothesized that an effective brace would counter this movement and restrict the forced supinatory movement. Kinematic analysis (frontal view camera operating at 50 frames per second) of the Achilles tendon angle showed a reduction in motion from an original range of 5.0-8.0° to a range of 2.0-5.0° with the brace. It was concluded that the brace does indeed have a stabilizing effect on the ankle and a restricting effect on rearfoot movement, although this restriction is very individual. Muscle tone and control may also play a part in the amount of restriction offered by the brace.

De Clercq [12] tested seven trained long-distance runners to examine rearfoot movement during the stance phase of locomotion. Subjects performed three trials at 4.5 m/s along a 25-meter track using a Push® brace type 'Medium' (with an unbraced condition for comparison). Frontal plane video recording was done at 250 Hz. Rearfoot kinematic variables that were examined included: Achilles tendon angle (representing subtalar movement), rearfoot angle (representing calcaneal movement), lower leg angle (representing orientation of the lower extremity) and subtalar eversion velocity. The analysis showed a less inverted position at touchdown while wearing the brace. During touchdown and midstance, all joint angle variables listed above demonstrated significantly less subtalar eversion while wearing the brace. In addition, maximum subtalar eversion ($t\beta_{max}$) occurred faster and with a greater magnitude in the unbraced condition. In the unbraced condition, $t\beta_{max}$ reached a peak at approximately 28% into the stance phase. When wearing the brace, however, $t\beta_{max}$ occurred later during the stance phase. The statistical comparisons of the Achilles tendon angle found a significantly greater value for the maximum angle as compared to the 28% stance phase value.

A study showing the versatility of ankle braces was performed by Burdett et al. [10], who examined the gait patterns of hemiplegic patients while wearing the Aircast

Air-Stirrup. These individuals were diagnosed with hemiplegia as a result of a cerebrovascular accident (CVA), and often had a modified gait pattern due to muscle weakness or general limitations of the lower extremity. Frontal plane video recording was used to determine rearfoot joint angles. The calcaneal angle was defined as the angle formed by the intersection of 1) a straight line connecting the posterior knee with the Achilles tendon, and 2) a straight line connecting the superior and inferior midline of the calcaneus. The kinematic analysis showed that the brace significantly reduced touchdown angle compared to an unbraced condition (from 5.8° to 2.9°, $p<0.05$). The brace also showed this reduction in total angular ROM change (from 10.0° to 6.3°, $p<0.05$). In addition, the brace did not effect gait speed, distance, or time for any of the subjects. Thus, the authors believed that the Air-Stirrup was effective in controlling mediolateral instability of subjects with hemiplegia, and can be used as a temporary or long-term device to help control inversion or eversion instabilities.

McIntyre et al. [31] used a Castiglia ankle device to study rearfoot motion of subjects during treadmill walking. The brace was compared with other conditions including closed basketweave taping, Louisiana wrap, and barefoot in a kinematic analysis (100 Hz). Speed of the treadmill was set at 4.5 mi/hr; an incline of 15% was used to ensure extreme dorsiflexion and plantarflexion (sagittal view kinematics were also examined). The posterior reference system included a line drawn from the popliteal fossa to the lower border of the calcaneus. Four points on this line (two above a line linking the right malleoli, and two on the posterior calcaneus) were used for joint angle calculation of the foot, ankle, leg, and thigh. Each subject walked for 10 minutes under each of the conditions. Results showed that all of the experimental conditions were effective in maintaining “immobilization” of the ankle during the activity, as instantaneous angular displacements of the ankle joint were not significantly different among the test conditions. The smallest amount of inversion was found in the barefoot condition. It was suggested that if these devices keep the foot in excessive inversion prior to/at touchdown, the intrinsic eversion mechanism usually seen after touchdown might be limited to a certain extent. This action may encourage ongoing inversion and perhaps predispose users to lateral ankle sprains.

Morin [32] performed a study similar in design to that of Hamill [23] to study rearfoot movement. Subjects were examined during both pre- and post-exercise conditions using the Aircast Air-Stirrup. The exercise regimen consisted of 70 eccentric contractions designed to fatigue the peroneal muscles. Two-dimensional filming from the posterior view was used at 100 frames per second, and a total of ten trials were performed for each condition using a pre-set velocity of 1.4 m/s. As with previous studies, two points on the posterior leg and two on the calcaneus were used to determine rearfoot inversion and eversion angles. The brace significantly reduced maximum eversion angle, total rearfoot motion, maximum eversion velocity, and increased the time to maximum eversion velocity. It was concluded that the brace was effective in controlling rearfoot movement, and should serve as a useful device to aid in the rehabilitation process.

Effects on Athletic Performance

The question behind every ankle brace is relatively simple: can it provide the restriction necessary to prevent injury (or re-injury) and, if so, can it provide it over an extended period of time? This question has been the topic of numerous research studies.

Sports such as football, basketball, and tennis often require athletes to be on the field for over two hours, justifying the need for a brace that can maintain its effectiveness throughout the duration of activity. This topic is the focus of the literature presented in this section.

The ROM restriction of an ankle brace during a three-hour volleyball session was studied by Greene and Hillman [21]. The brace of choice was the DonJoy ALP; a semirigid brace commonly used by sports therapists. The authors used an analog ankle stability test instrument to record passive inversion/eversion ROM during five testing sessions: before support, before activity, 20 minutes into activity, 60 minutes into activity, and after completion of the activity. The device was one that permitted positioning of the lower extremity by controlling knee flexion, ankle dorsiflexion and plantarflexion, and internal/external rotational alignment of the foot relative to the tibia. The brace was able to restrict inversion/eversion ROM 42% before the onset of activity. At both the 20-minute and 60-minute testing sessions, the brace showed no significant loss in range restriction capabilities (3% after 20 minutes, 12% after 60 minutes). The same held true for the post-activity measurement; the initial 42% restriction was only reduced to 37% after a highly demanding, three-hour activity. It is interesting to note that subjects were instructed not to re-tighten the braces during the course of the activity; this could have decreased the passive ROM values that were reported.

Johnson et al. [27] used four different ankle-restricting devices to study active inversion ROM. These included the Active Ankle, Aircast Sport-Stirrup, DonJoy ALP, and Malleoloc. Subjects were tested on a measuring apparatus consisting of an L-shaped aluminum foot plate that pivoted at the level of the ankle joint; a Digital Smart Level®

attached to the back of the foot plate allowed the inversion measurement to be taken and recorded. Testing was performed on four separate occasions according to the following protocol: pre-support, pre-exercise, and post-exercise. The exercise session consisted of a one-hour session of full-court, competitive basketball. All four devices were able to keep an inversion ROM restriction post-exercise, with an average increase of only 3.27°. The Active Ankle and Sport-Stirrup restricted more inversion than did the other two devices, although no difference was found between the two braces. The Active Ankle produced an inversion angle of 18.96° pre-exercise and 21.27° post-exercise, whereas the Sport-Stirrup yielded inversion angles of 17.82° and 21.50° for pre- and post-exercise. In addition, it is interesting to note that these two braces received the highest marks in a subjective questionnaire on perceived stability, comfort, and ease of application. The Active Ankle had the highest percentage of subjective preference during competition.

In contrast to a majority of studies dealing with semirigid orthoses, Myburgh et al. [33] used elastic ankle braces (Ace and Futuro braces) to evaluate support before, during, and after a squash match. The subjects in the study were high-caliber, elite athletes. A specially designed goniometer was used to record ankle ROM with electronic digital data recording. Ankle motion was examined before the onset of activity, 10 minutes into the activity, and after one hour (completion of the match). The results showed that neither of the ankle braces were able to significantly support any of the measured ankle motions (dorsiflexion, plantarflexion, plantarflexed inversion, and plantarflexed eversion) before, during, or after the activity. The authors acknowledged that this finding could well be a result of the composition of the guards, not having sufficient elasticity to be pulled over the foot onto the ankle. They suggested, however, that the braces still might offer proprioceptive and psychological benefits.

Limited resources are a problem many sports therapists face today, in particular at the high school level. When an adequate facility budget is not implemented, alternative methods must be used to ensure the best possible care for the athlete. This situation can be applied to the case of ankle bracing, where buying braces for each athlete is often not feasible. An option, however, is a form of support the therapist constructs directly in the clinical setting. Hughes and Stetts [24], who compared their self-constructed ankle brace to a traditional inversion ankle taping, studied one form of this support. The material used for the guard was Surlyn, a thermoplastic substance made by Dupont, and was constructed in two lengths: 24 inches for subjects 60-65 inches in height, and 28 inches for those above 65 inches in height. It remained flat when cut out, was softened by a heat gun prior to application, and was then fitted over a cotton sock and secured by an elastic bandage during activity.

The test placed subjects through 20 minutes of pre-determined exercise: sprints (forward, backward, lateral) and a 1.25-mile run-walk. Inversion ROM was taken at pre-support, pre-exercise, and post-exercise phases for each subject using a Leighton Flexometer. No difference was found between the tape and ankle brace for any of the three testing phases. Both showed a comparable decrease in post-exercise ROM. It was concluded that although one form of support could not be favored over the other according to this study, the constructed ankle brace could possibly be more practical and financially advantageous.

A prospective study following soccer players over the course of one season was performed by Surve et al. [44]. Subjects wore the Aircast Sport-Stirrup during all practices and matches; incidence and severity of injury were tracked throughout the course of the season. Four categories were identified for analysis: those with a previous history of sprains (two groups – brace and control), and those with no previous history of sprains (two groups – brace and control). The incidence of sprains (expressed as injuries per 1000 playing hours) in the previous history brace group was significantly lower compared to sprains in the previous history control group ($p<0.001$). When categorizing by ankle, a significantly lower incidence of sprains was found in the previously injured ankles of the previous history brace group compared with the previously injured ankles of the previous history control group ($p<0.01$). In addition, a significantly lower incidence was found in the ankles of the no previous history control group compared with the previous injured ankles in the previous history control group ($p<0.05$). It is interesting to note that the application of the orthosis did not alter the incidence of sprains in previously uninjured ankles. The authors suggested that the main effect of the orthosis was to assist in proprioceptive function of the previously injured ankle rather than to strictly provide mechanical support.

Effects on Functional Activities

An ankle brace cannot be chosen exclusively on whether it restricts inversion and eversion ROM. Other factors, such as effects on functional performance and subjective responses, must be taken into consideration. The role that the brace plays in permitting functional abilities mainly comes into play during the latter stages of rehabilitation, when the choice of a brace might be altered. The athlete will typically progress to a brace with a different design during this stage, one that is still restrictive yet allows more movement than a brace designed for an acute injury. It is these types of braces that need to be tested for any negative effects on functional performance (i.e., activities that simulate athletic activity). This section will discuss the results that others have documented dealing with this topic.

Bocchinfuso et al. [5] used the Active Ankle Training Brace and Aircast Sport-Stirrup to evaluate performance during four tests: vertical jump, 80-foot sprint, shuttle run, and 4-point run. Subjects completed three testing sessions during which the performance tests were completed a total of three times, for a total of nine cycles. Shoe-type was controlled to an extent, with high-top basketball shoes being used by all subjects (although not the same type). No significant differences were found among the braces for any of the performance tests, although it was noted that during the shuttle-run and 4-point run, improvement was more evident with the Sport-Stirrup than with the Active Ankle or the unbraced control condition.

Similar measures of performance were studied by Pienkowski et al. [35], using high school basketball players as subjects. Testing was performed during the actual season and used four parameters for evaluation: vertical jump, standing long jump, cone running, and shuttle running. Players were divided into groups based on position played so that the data collection took four weeks to complete; one week with each of three braces (Universal [Swede-O], Kallassy, and Aircast Air-Stirrup) and one week with no brace. For performance measures and brace type, no significant differences were found (i.e., the braces had no inhibiting effect on the standardized tests). There was a trend, however, toward enhanced performance in cone running, vertical jump, and long jump for all braces but the Universal. The authors suggested that athletes should have limited concerns about wearing braces because of concerns of decreased performance.

The Swede-O and Kallassy braces were also the subjects of a study by Burks et al. [11], evaluating functional parameters. Varsity athletes were used to test performance in four events: broad jump, vertical jump, 10-yard shuttle run, and 40-yard sprint. Each event was performed twice with each brace, with respective times and distances being recorded for comparisons. The Swede-O brace resulted in a significantly decreased performance in three of four events: the vertical jump, broad jump, and the time of the sprint. Vertical jump was the only event that showed a significantly decreased performance as a result of the Kallassy brace. Of interest is the fact that although few significant differences were found as a result of brace application, all times/distances were decreased during the braced conditions. In addition, 17 of 22 subjective questionnaires indicated the Swede-O brace as being less comfortable than the Kallassy.

Beriau et al. [4] used an agility course to test performance using four ankle braces: the Aircast Sport-Stirrup, Aircast Training Brace, Swede-O, and DonJoy ALP. The course consisted of forward and backward running, lateral shuffling, and directional changes similar to those often used in athletic activity (and rehabilitation as well). Subjects (N=85) were asked to run the course on four separate occasions: 1) Two trials under the control (unbraced condition), 2, 3) Two trials using each brace, and 4) Two trials again in the control condition. Shoe-type was not controlled in the study. A significant difference was found between all four braces ($p<0.001$); post-hoc analysis revealed that subjects were able to complete the agility test quicker with the Aircast

Training Brace than with the DonJoy ALP. In a subjective finding that contrasts with other studies, the Swede-O brace was favored by 42% of the subjects. It was concluded that there should not be a great concern of decreased performance during the process of choosing an appropriate ankle brace.

Ground Reaction Force Data

Hamill et al [23], in a study previously mentioned within the kinematic section (using the Aircast Air-Stirrup), examined force variables during walking trials, both pre- and post-exercise. A total of 13 variables were used for analysis (8 vertical, 2 anterior-posterior, and 3 medial-lateral). Data were collected using a force platform and sampled at 500 Hz. The vertical GRF variables showed no significant differences as a result of the exercise; however, relative time to the second maximum force did show a significant main effect for the brace (a decrease in this time as a result of the brace, at $p<0.05$). No significant differences were found for the anterior-posterior GRF variables. For the medial-lateral GRF data, two variables were found to have a significant main effect for the braced condition: force excursions (0-30% of the stance phase) and force excursions (0-100% of the stance phase). For each variable, mean values were less when wearing the orthosis. This finding suggested that the orthosis was indeed moderating medial-lateral motion.

In another study combining kinetic and kinematic analysis, Stuessi et al. [43] evaluated the Aircast Air-Stirrup during running strides. As mentioned earlier, a special shoe was used (medially very hard, laterally very soft) to help show the effectiveness of the orthosis. The authors used a mechanical model of the foot and lower limb to help explain the effects of the orthosis on the GRF data. This model produced four hypotheses: 1) The brace stiffens the ankle joint, leading to a harder lateral to medial touchdown and, therefore, should result in a higher medial force peak, 2) After touchdown, pronation is stopped because of shoe design, and the foot is put into supination; the brace slows this action and leads to a lower medial to lateral force, 3) Movement of the foot leads to a change in point of application of force under the contact surface/ground; a more stable joint would result in a smaller medial-lateral velocity of displacement of the point of application, and 4) The variability of the medial-lateral curve after 10% of the stance phase should be smaller while using the Aircast. The first hypothesis was found to be true in 7 of 11 subjects, the second also in 7 of 11 subjects, the third in 6 of 11, and the final hypothesis in 9 of 11. These results confirmed that the orthosis did in fact have a stabilizing effect on the ankle.

De Clercq et al. [12] performed another study involving simultaneous kinetic/kinematic data collection. A Push® brace type 'Medium' was used to obtain variables describing the first vertical impact force peak of the overall GRF as subjects performed running trials of 4.5 m/s across the force platform. These variables included maximal amplitude (Fz_i), time to maximal amplitude (t_i), and the average loading rate (Gz_i, defined as Fz_i/t_i). Comparing braced condition to an unbraced control, no significant differences were found for any of the variables. That is, wearing the brace did not affect the characteristics of the impact force at touchdown. This finding is not surprising, being that shoe type was controlled for in this study.

Nigg and Morlock [34] studied the influence of lateral heel flare (in different running shoes) on GRF parameters. Running trials were used in the protocol at 4 m/s;

force platform measurements were collected at 1020 Hz. Three types of flares were used: Shoe A had a conventional flare of 16°, Shoe B had no flare, and Shoe C had a rounded lateral edge (radius 3-4 cm) providing a negative and non-constant flare angle. Analysis of the force data yielded results for 4 vertical, 2 anterior-posterior, and 2 medial-lateral GRF variables, although only the vertical GRF results were reported. The mean values of the vertical impact force peaks (Fz_i) as well as the loading rate (Gz_i, again defined as dFz/dt) showed no significant change due to the change in heel flare. The time of occurrence of the impact force (tz_i) was decreased from 37.6 ms to 29.1 ms for Shoe C, a change of 29%.

Chapter 3

MATERIALS AND METHODS

Subjects

Ten healthy, active male volunteers served as subjects for this study. Each subject signed an Informed Consent Form approved by the Institutional Review Board at the University of Tennessee, Knoxville (Appendix F) and was familiarized with the testing procedures at the beginning of each testing session. Subjects had no history of major ankle injuries (i.e., fractures) and were free of injury at the time of the study. Subject information is provided in Appendix A.

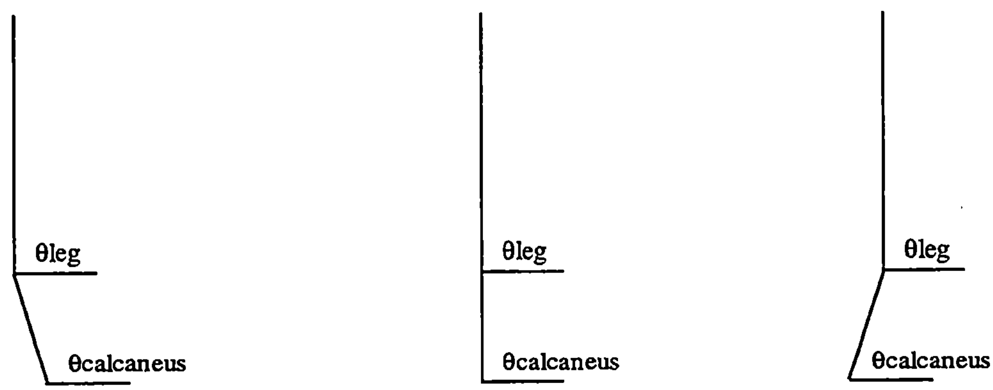
Instrumentation/Equipment

Video System

The posterior view of movement was recorded using two video cameras (Panasonic AG-188U, 60 Hz) and was used to determine subtalar joint motion. Shutter speed for each camera was set at 1/1000 sec. In addition, the cameras were leveled so that they were parallel with the floor. Prior to testing, a reference frame was taped for each view and consisted of four coplanar points that covered the ranges of motion and determined the scale factors for coordinate conversions.

Reflective markers were placed on the right side of each subject for the purpose of determining joint center. For sagittal view, markers were placed on the acromion, greater trochanter of the femur, femoral epicondyle, tibial condyle, lateral malleolus, lateral aspect of the calcaneus (on the outside of the shoe), and head of the fifth metatarsal (on the outside of the shoe). The distance from the floor to the center of the lateral malleolus was used for placement of that specific marker during the braced conditions to increase consistency. For the rear view, markers were placed on the mid-point of the gastrocnemius, 10 cm below this point, and on the top and bottom of the posterior aspect of the calcaneus (on the outside of the shoe). The markers on the video image were digitized using a commercial biomechanical system – APAS (Ariel Performance Analysis System). The digitized coordinates were later used in customized software to determine kinematic variables. Only rearfoot kinematics was processed and analyzed in this study. The definition of rearfoot angle is provided in Figure 2.

Definition of Rearfoot Angle



Eversion
Negative Angle

Neutral
Zero Angle

Inversion
Positive Angle

θrearfoot = θleg - θcalcaneus

Figure 2. Definitions of the rearfoot absolute angles of the leg and calcaneus describing rearfoot movement. Examples shown are for the right foot.

Force Platform

A single force platform (AMTI, American Mechanical Technology Inc.), flush with the surface of a walkway, was used to collect kinetic data for three channels of ground reaction forces (GRF),

Fx – Medial/Lateral Force

Fy – Anterior/Posterior Force

Fz – Vertical Force

three channels of moments,

Mx – Medial/Lateral Moment

My – Anterior/Posterior Moment

Mz – Vertical Moment

and one synchronization channel, for a total of seven channels. The signals were collected for 1.5 seconds and were sampled at 1000 Hz using the same APAS system described previously. Data from the force platform were decoded and analyzed using customized software to obtain peak GRF, time to these peak values, impulse variables, and excursion variables used for later comparisons and analyses. Figure 3 demonstrates the setup of the instrumentation.

Ankle Braces

Three different ankle braces – the Aircast Air-Stirrup, Aircast Sport-Stirrup, and Active Ankle – all widely used in treatment and rehabilitation programs and all of which are commercially available, were used in this study. A description of each is given in this section.

Instrumentation Setup

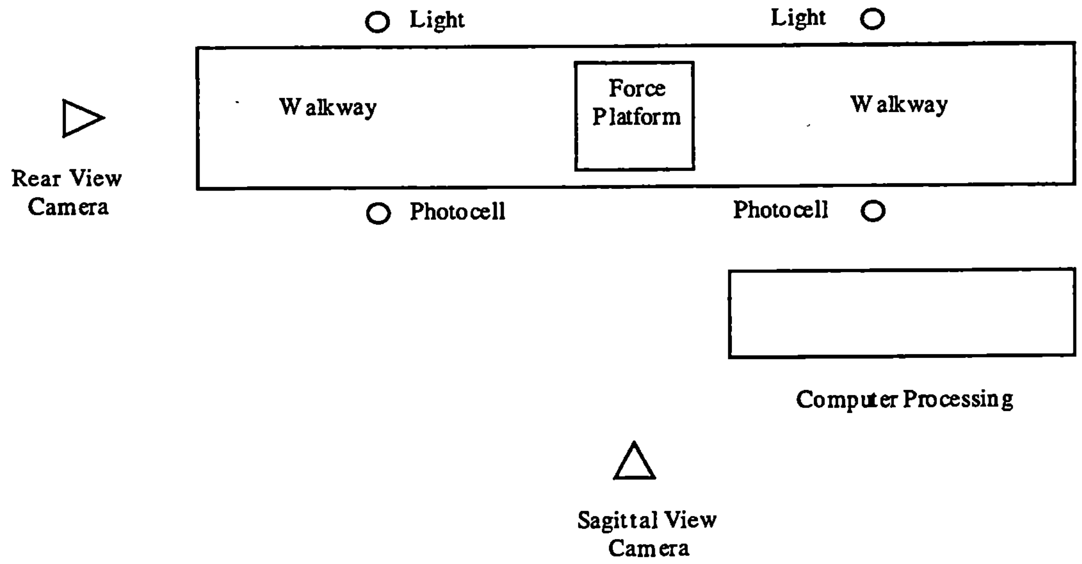


Figure 3. Setup and placement of instrumentation used in the study.

Aircast Air-Stirrup: The Standard Air-Stirrup is constructed from plastic and molded to fit the contours of the ankle and lower leg. It is approximately 11 inches long, and is three inches wide on the proximal two-thirds of both sides. The distal one-third narrows slightly to allow for increased mobility. The outer shell of the brace is contoured in a way that allows adequate space for the medial and lateral malleoli. The interior of the Air-Stirrup is lined with pre-inflated air cells on both sides; one compartment covers the distal one-third, the other covers the full length of the brace. This design has two distinct advantages: 1) It protects the malleoli better by preventing displacement of the air cells, and 2) It provides graduated compression – higher pressure distally, lower pressure proximally. The brace has a heel counter that can be adjusted by velcro to accommodate different foot widths. Securing the brace is done with two velcro straps located on the outside shell. The straps simply wrap around the entire brace and fasten, and can be tightened as needed. The Air-Stirrup comes in three different sizes, depending on height and shoe characteristics of the individual. Left- and right-ankle versions are available. It is intended more for use with acute injuries, including sprains, stable fractures, and certain postoperative conditions [7].

Aircast Sport-Stirrup: The Sport-Stirrup is identical to the Air-Stirrup in composition, but differs slightly in design and intended usage. Its reduction in height (9.5 inches) and reduced malleolar coverage (approximately 2.25 inches) allows for less bulk in the shoe, giving the brace a more functional role in rehabilitation. It follows, then, that the main goal of the Sport-Stirrup is prevention, rather than acute injury treatment.

Active Ankle: The Active Ankle features a slightly different design than the Aircast models. Of the 9.5 inches of its height, the proximal six inches are covered by a plastic molded shell similar to that of the Aircast braces. The insides are lined with a foam-like material that actually covers a greater area than the outer shells. What makes the Active Ankle unique is its hinged design, which constitutes the distal three inches of the brace. This allows for greater freedom of movement within the shoe. The brace fastens with a posterior velcro strap as well as a large velcro strap that winds completely around the circumference. The Active Ankle comes in four different sizes (XS, S, M, and L); heel counter width changes with each size (the heel counter is not adjustable). Each brace can be used bilaterally; no distinctive left- and right-ankle versions are available.

Protocol

At the beginning of each testing session, passive rearfoot inversion/eversion range of motion of the right ankle was measured using a standard flexible goniometer and recorded on each individual's information sheet. Subjects were instructed to lie prone on an examining table as their right ankle was put into subtalar joint neutral position, as described by Elveru et al. [15]. As this position was different for each subject, it represented the starting point (or zero degree) for the goniometric measurement. The ankle was manually inverted to the end range of motion, and the goniometer was read to determine the angle. The same procedure was then used for the eversion measurement. Three trials were performed in each direction, with averages used for further analyses.

Subjects were tested during four conditions in this study: Control (no brace – shoe only), Air-Stirrup, Sport-Stirrup, and Active Ankle. These conditions were randomized in order to eliminate a systematic ordering effect. For each braced condition, the brace was fitted according to the manufacturer's instructions. Subjects were allowed a warm-up period in order to become familiar with the brace and to make any adjustments if necessary. Walking speed was determined in the warm-up trials that preceded each condition by the use of two photocell sensors (Lafayette Instruments). The sensors were placed approximately three meters apart on one side of the platform. The range of the walking speed for each subject was determined by averaging three consistent walking trials with 10% of variation. This range was used to monitor walking speed throughout the actual testing. If the speed for a specific trial was too slow or too fast (as established by the range), feedback was provided and a make-up trial was completed. Within each condition, subjects completed a total of five successful walking trials on a walkway across the force platform. A successful trial was one in which contact of the subject's right foot was made within the surface of the force platform.

Statistical Analysis

Descriptive statistics (means and standard deviations) were calculated for each variable. The statistical design for this study consisted of a repeated measures analysis of variance (ANOVA) with one repeated factor (condition, with four levels). Post-hoc tests were performed using pairwise comparisons with a Bonferroni adjustment (to control the experiment-wise error rate). The significance level was set at $p<0.05$. Due to the small sample size, marginal differences ($p<0.05$-$p<0.10$) will also be discussed in reference to post-hoc tests.

Chapter 4

RESULTS

The primary purpose of this study was to evaluate the effects of ankle braces on rearfoot motion and ground reaction force (GRF) data during the stance phase of gait. In this chapter, the results of the kinematic and kinetic data will be presented for the four conditions in this study: condition 1 (control – no brace), condition 2 (Air-Stirrup), condition 3 (Sport-Stirrup), and condition 4 (Active Ankle).

Passive Range-of-Motion

Means and standard deviations of passive ROM measurements are shown in Table 1. No additional statistical tests were performed for this variable. The inversion motion showed a range of 6.33°-13.33°, while eversion ranged from 5.33°-11.33°. Seven subjects showed an inversion value equal to or greater than that of the eversion value.

Kinematic Variables

The kinematic analysis for this study included a total of seven variables describing rearfoot motion. These variables were touchdown angle (TDAngle), maximum eversion angle (MaxEV), time to maximum eversion angle (TMaxEV), eversion range-of-motion (EROM), toe-off rearfoot angle (TOAngle), maximum eversion velocity (MaxVel), and time to maximum eversion velocity (TMaxVel).

Table 1. Means and standard deviations of passive ROM measurements.

Subject	Inversion (deg)	Eversion (deg)
1	11.67 (0.58)	-10.33 (0.58)
2	11.33 (1.53)	-11.33 (0.58)
3	7.67 (0.58)	-5.67 (0.58)
4	6.67 (1.15)	-5.33 (1.15)
5	6.33 (0.58)	-7.67 (1.53)
6	6.67 (1.15)	-10.00 (0.00)
7	10.67 (1.15)	-9.00 (1.00)
8	6.67 (2.31)	-10.00 (0.00)
9	13.33 (1.15)	-7.33 (1.15)
10	8.00 (2.00)	-6.67 (2.31)
Overall	8.90 (0.59)	-8.33 (0.70)

<u>**Touchdown Angle**</u>: This variable is indicative of the rearfoot position of the foot/ankle at the moment it first makes contact with the ground. In this study the repeated measures ANOVA did not show a significant omnibus ($F=1.15$, $p>0.05$). The mean minimum value was seen in the Active Ankle (-1.47°), while condition 3 (Sport-Stirrup) showed the highest value (2.15°) indicating the greatest amount of inversion. Descriptive statistics for touchdown angle (as well as for all kinematic variables) are listed in Table 2.

<u>**Maximum Eversion Angle**</u>: This value reflects the greatest angle of eversion during the entire stance phase of a gait cycle. No significant differences were found among the conditions ($F=2.85$, $p>0.05$). However, the braced conditions generally demonstrated reduced eversion angles (Figure 4), indicating a trend that each of the braces was able to limit the maximum amount of eversion compared with the control.

Table 2. Means and Standard Deviations of Kinematic Variables by Condition.

Cond	TDAngle (deg)	MaxEV (deg)	TMaxEV (ms)	EROM (deg)	TOAngle (deg)	MaxVel (deg/s)	TMaxVel (ms)
1	0.32 (10.11)	-18.39 (6.73)	472.30 (123.20)	26.28 (9.64)	35.21 (14.10)	-292.57 (79.41)	232.10 (20.92)
2	1.21 (8.00)	-15.04 (6.99)	512.30 (107.90)	22.40 (6.24)	22.51 (17.70)	-258.46 (109.30)	307.70 (82.19)
3	2.15 (6.88)	-15.56 (7.54)	475.80 (114.10)	22.16 (7.09)	27.83 (21.78)	-263.88 (90.22)	297.00 (85.16)
4	-1.47 (8.83)	-15.75 (7.75)	408.60 (101.90)	23.07 (12.00)	26.78 (18.21)	-227.91 (67.17)	295.20 (73.10)

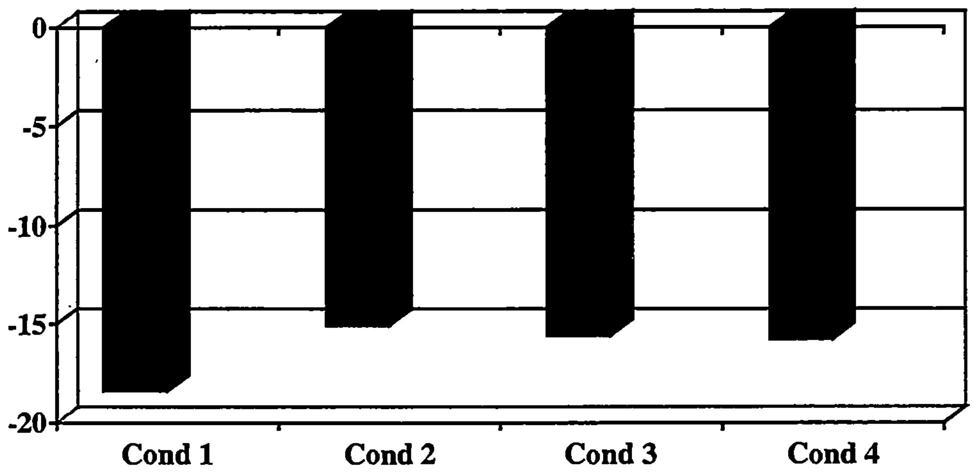


Figure 4. Condition means for Maximum Eversion Angle.

Time to Maximum Eversion Angle: This time represents how long it took for each subject to reach the maximum eversion angle described previously. Across all four conditions, a significant difference was found ($F=4.52$, $p<0.05$). The fastest mean time was found in condition 4 (Active Ankle), and the slowest in condition 2 (Air-Stirrup). Post-hoc comparisons (Table 3) revealed only marginal differences between conditions 1 and 4 ($p=0.055$), and between conditions 2 and 4 ($p=0.063$). No significant differences were found among the other conditions.

Eversion ROM: The value for this variable is defined as the difference of the TDAngle value and the MaxEV value. The resulting figure represents the total range of rearfoot

Table 3. Pairwise comparisons for Time to Maximum Eversion Angle.

Condition	Compared Condition	Mean Difference	Significance Level
1	2	-0.04	1.000
	3	-0.004	1.000
	4	0.06	0.055
2	1	0.04	1.000
	3	0.04	1.000
	4	0.10	0.063
3	1	0.004	1.000
	2	-0.04	1.000
	4	0.07	0.198
4	1	-0.06	0.055
	2	-0.10	0.063
	3	-0.07	0.198

eversion motion during the stance phase. Each braced condition was able to decrease total rearfoot motion (F=0.81, p>0.05) in comparison to the control condition (Table 2).

<u>**Toe-Off Rearfoot Angle**</u>: This value represents the angle of the foot/ankle at the final moment of the stance phase as it prepares to begin the swing phase of a gait cycle. A significant omnibus F was found (F=5.53, p<0.05). Post-hoc comparisons revealed significant differences between conditions 1 and 2 (p=0.017) and marginal differences between conditions 1 and 4 (p=0.056, Table 4). The angle showed a decrease in all of the braced conditions compared to the control. This variable, as with the touchdown angle, showed great variability within conditions.

Table 4. Pairwise comparisons for Toe-Off Angle.

Condition	Compared Condition	Mean Difference	Significance Level
1	2	12.70*	0.017
	3	7.38	0.699
	4	8.43	0.056
2	1	-12.70*	0.017
	3	-5.32	0.974
	4	-4.27	1.000
3	1	-7.38	0.699
	2	5.32	0.974
	4	1.05	1.000
4	1	-8.43	0.056
	2	4.27	1.000
	3	-1.05	1.000

* The mean difference is significant at the p<0.05 level.

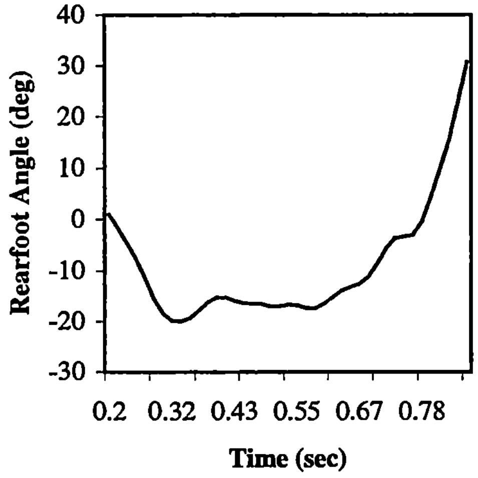
Maximum Eversion Velocity: This value represents the lowest velocity achieved during the stance phase. No significant difference was found for the overall effect ($F=2.91$, $p>0.05$). Large (but expected) standard deviations were seen for this variable. Although not significant, each braced condition was able to decrease the velocity slightly (Table 2).

Time to Maximum Eversion Velocity: This value represents the time associated with the maximum eversion velocity. The ANOVA result demonstrated an insignificant omnibus ($F=2.85$, $p>0.05$). It is interesting to note that each braced condition was able to delay the time it took to reach the maximum velocity.

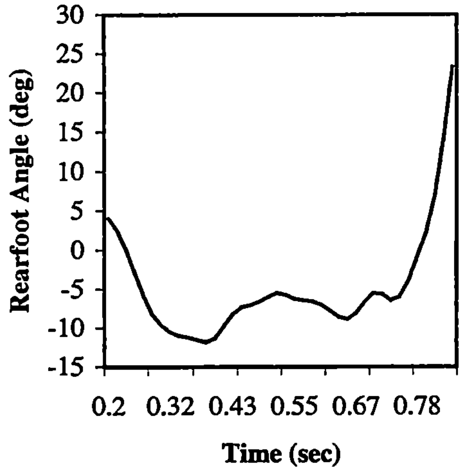
In summary, although only two kinematic variables showed significance (time to maximum eversion angle and toe-off angle), most showed a general trend toward greater rearfoot control for the braced conditions. Representative rearfoot curves for the four test conditions are shown in Figure 5.

Kinetic Variables

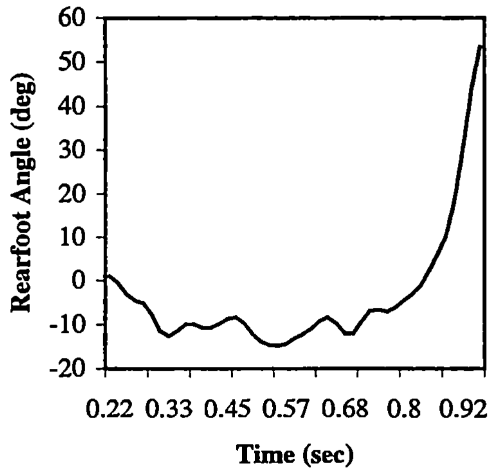
Fx (Medial-Lateral component): This GRF component is perhaps the most relevant in this study, since it can be correlated to the kinematic analysis of rearfoot motion. The variables included in this component are excursions from contact (0%) to 30% of the stance phase (Exc1), excursion from 0% to approximately 50% of the stance phase (Exc2), and excursions from 0% to 100% (Exc3, Table 5). Of these variables, two showed a significant overall effect: Exc1 ($F=5.32$, $p<0.05$), and Exc2 ($F=5.13$, $p<0.05$). For Exc1, post-hoc comparisons showed a significant difference between conditions 1 and 4 ($p=0.037$) and marginal differences between conditions 1 and 2 ($p=0.052$, Table 6).



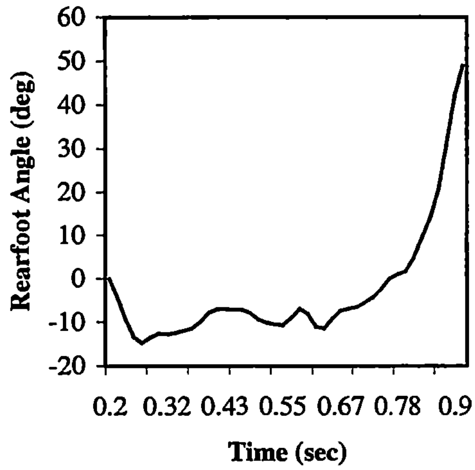
(a). Condition 1 (control).



(b). Condition 2 (Air-Stirrup).



(c). Condition 3 (Sport-Stirrup).



(d). Condition 4 (Active Ankle).

Figure 5. Representative rearfoot curves (by condition).

Table 5. Means and Standard Deviations of Medial-Lateral GRF Results by Condition.

Cond	Exc1 (N/kg)	Exc2 (N/kg)	Exc3 (N/kg)
1	2.87 (0.43)	3.39 (0.45)	4.63 (0.44)
2	2.83 (1.23)	3.30 (1.24)	4.55 (1.22)
3	2.77 (0.97)	3.28 (0.97)	4.53 (1.00)
4	2.76 (0.81)	3.26 (0.82)	4.55 (0.79)

Table 6. Pairwise comparisons for Exc1.

Condition	Compared Condition	Mean Difference	Significance Level
1	2	0.30	0.052
	3	0.24	1.000
	4	0.23*	0.037
2	1	-0.30	0.052
	3	-0.06	1.000
	4	-0.07	1.000
3	1	-0.24	1.000
	2	0.06	1.000
	4	-0.01	1.000
4	1	-0.23*	0.037
	2	0.07	1.000
	3	0.01	1.000

*The mean difference is significant at the p<0.05 level.

For each braced condition, the means were lower than the value seen in the control condition. It should be noted that subject 9 was not included in the analysis of the medial-lateral variables due to unusual values and graphical patterns. The exclusion of this data will be discussed further in Chapter 5.

For Exc2, post-hoc comparisons revealed significant differences between conditions 1 and 2 (p=0.040) and conditions 1 and 4 (p=0.030, Table 7), yielding similar results to Exc1. Means for Exc2 also were lower for each braced condition compared with the control. A representative medial-lateral GRF curve is seen in Figure 6.

Table 7. Pairwise comparisons for Exc2.

Condition	Compared Condition	Mean Difference	Significance Level
1	2	0.36*	0.040
	3	0.25	1.000
	4	0.26*	0.030
2	1	-0.36*	0.040
	3	-0.11	1.000
	4	-0.10	1.000
3	1	-0.25	1.000
	2	0.11	1.000
	4	0.01	1.000
4	1	-0.26*	0.030
	2	0.10	1.000
	3	-0.01	1.000

*The mean difference is significant at the p<0.05 level.

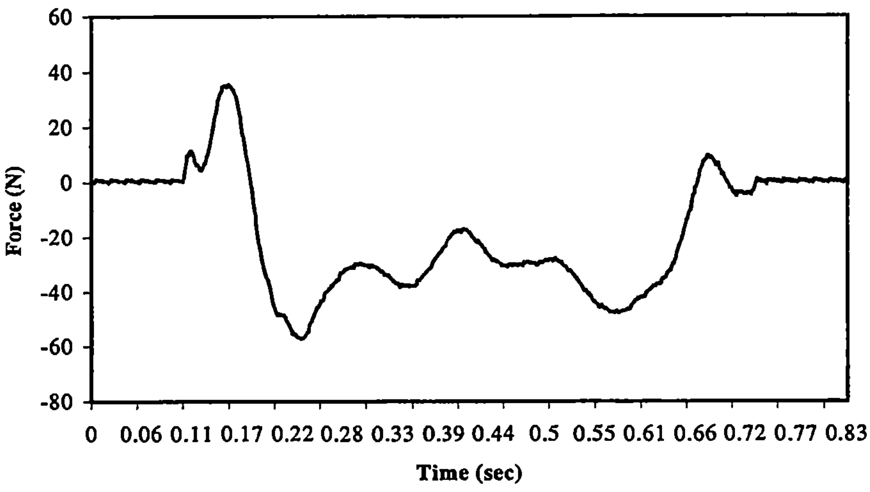


Figure 6. Representative medial-lateral GRF curve.

__Fy (Anterior-Posterior component)__: The variables associated with the anterior/posterior GRF include: maximum breaking force (MaxBrk), time to maximum braking force (TMaxBrk), maximum propulsion force (MaxProp), time to maximum propulsion force (TMaxProp), braking impulse (IBrk), and propulsive impulse (IProp). One variable, braking impulse (IBrk), showed a significant omnibus effect ($F=4.55$, $p<0.05$). Pairwise comparisons revealed only marginal differences between conditions 1 and 3 ($p=0.062$). In general, values for the braced conditions were lower than for the control condition. No other Fy variables were found to be significant across conditions. Means and standard deviations for these variables are shown in Table 8, and a representative curve is shown in Figure 7.

Table 8. Means and Standard Deviations of Anterior-Posterior GRF Results by Condition.

Cond	MaxBrk (N/kg)	TMaxBrk (s)	MaxProp (N/kg)	TmaxProp (s)	IBrk (N-s/kg)	IProp (N-s/kg)
1	-2.19 (0.14)	0.11 (0.01)	4.10 (1.23)	0.60 (0.03)	-0.36 (0.05)	0.31 (0.04)
2	-2.12 (0.19)	0.12 (0.01)	3.95 (1.20)	0.61 (0.04)	-0.35 (0.04)	0.30 (0.04)
3	-2.20 (0.24)	0.11 (0.01)	4.12 (1.24)	0.60 (0.04)	-0.35 (0.05)	0.31 (0.04)
4	-2.17 (0.17)	0.11 (0.01)	4.10 (1.20)	0.61 (0.04)	-0.36 (0.05)	0.31 (0.04)

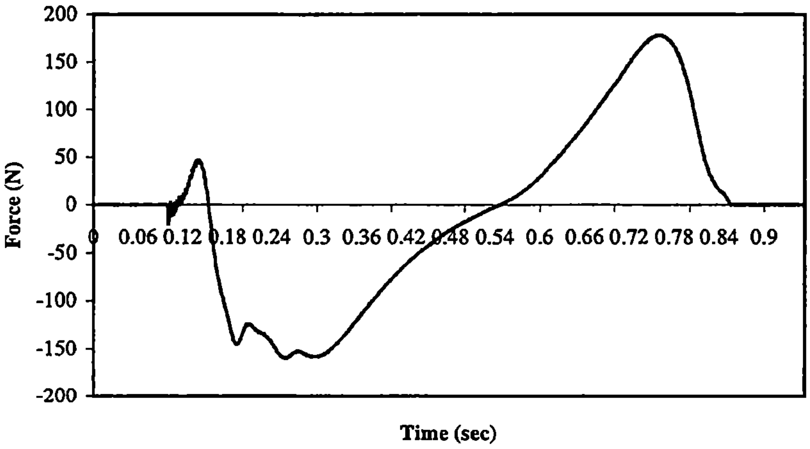


Figure 7. Representative anterior-posterior GRF curve.

Fz (Vertical component): The variables associated with the vertical GRF include: the first peak force (F1), time to first peak force (T1), second peak force (F2), and time to second peak force (T2). The analysis performed on these variables showed no significant differences among the four conditions. A summary of the results for these variables is presented in Table 9; a typical Fz curve is shown in Figure 8.

Table 9. Means and Standard Deviations of Vertical GRF Results by Condition.

Cond	F1 (N/kg)	T1 (s)	F2 (N/kg)	T2 (s)
1	11.20 (0.53)	0.16 (0.02)	10.79 (0.47)	0.54 (0.03)
2	11.08 (0.50)	0.16 (0.02)	10.83 (0.45)	0.54 (0.03)
3	11.18 (0.60)	0.16 (0.02)	10.95 (0.50)	0.54 (0.03)
4	11.11 (0.54)	0.16 (0.02)	10.85 (0.50)	0.54 (0.03)

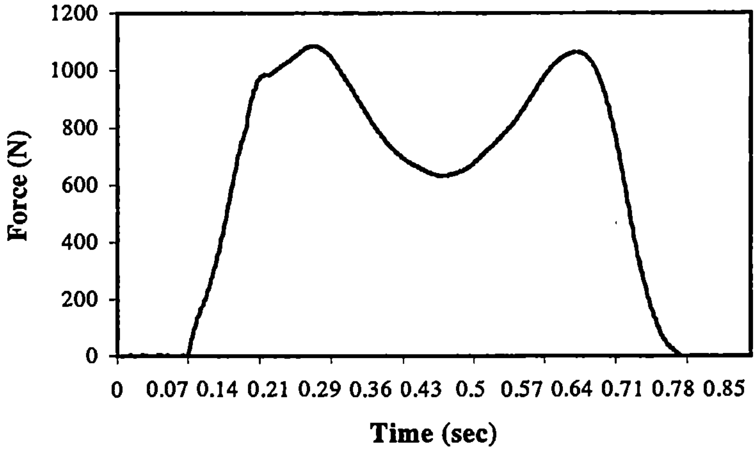


Figure 8. Representative vertical GRF curve.

Chapter 5

DISCUSSION AND CONCLUSIONS

The purpose of this study was to examine the effects of ankle braces on rearfoot movement and GRF data during the stance phase of gait. An effective and well-designed ankle brace must be able to serve many functions. Most importantly, it should be able to restrict the total amount of motion available at the ankle joint in an attempt to prevent serious injury. At the same time, however, the brace must not compromise functional ability or hinder the athlete's performance.

In this study three ankle braces were used to examine rearfoot motion and associated variables: the Aircast Air-Stirrup, Aircast Sport-Stirrup, and Active Ankle. The braces were chosen due to their popularity among athletes today and also because they represent braces used at different stages of an ankle injury. The Air-Stirrup is designed primarily for the acutely injured ankle, while the Sport-Stirrup and Active Ankle are used more by the athlete who is returning to activity. These different braces, along with a comparison to a control (unbraced) condition, should show quantitatively their effects on rearfoot motion through the use of kinematic and kinetic analysis.

Kinematics

The Air-Stirrup and Sport-Stirrup were able to delay the time it took to reach maximum eversion compared to the control by an average of 22 ms. This value is in sharp contrast to other studies [23, 32], who found that the delay with the use of a brace was approximately 220 ms (compared to the unbraced condition). The main difference in this discrepancy can be seen in the control condition. For the two studies previously mentioned, the time to maximum eversion angle occurred quite early during the stance phase. A mean value of 127 ms was found when the maximum value occurred. In the present study, the mean time to maximum eversion angle was 472 ms, or around 43% of the stance phase (Table 2). This finding is in agreement with De Clercq [12], who noted that time to maximum eversion occurred at 44% of the stance phase when using no brace.

When looking at the braced conditions, similarities are seen between the present study and others. In the results of Hamill et al. [23] and Morin [32], the mean brace value for time to maximum eversion was 347 ms. For the Air-Stirrup in this study, the mean value was 512 ms; for the Sport-Stirrup, the mean value was 476 ms (Table 2). These results show general consistency with the previously mentioned data.

A different trend was noted for the Active Ankle during analysis of this variable. The brace showed a faster mean time to reach maximum eversion (409 ms) than did the control condition; this pattern was seen individually for five subjects (Table 10). It is unclear as to why this occurred. The unexpected decrease in time could have come as a result of the touchdown angle, which overall showed an everted foot for this condition (-1.47°). As normal touchdown is in inversion, the subject is getting a "jump start" on reaching maximum eversion. It would seem logical, therefore, that this maximum value would be reached quicker in this type of situation. This was not the case in Hamill et al. [23], where a similar everted touchdown value was found for the braced condition, yet still exhibited a much slower time to maximum eversion angle than did the control. The fact that this pattern existed for some subjects in the Active Ankle condition leads one to

Table 10. Means and Standard Deviations (by subject) for Time to Maximum Eversion Angle.

Subject	Condition 1	Condition 2	Condition 3	Condition 4
1	357.38 (135.98)	354.04 (165.91)	354.04 (167.99)	254.68 (31.20)
2	434.20 (171.94)	501.00 (174.35)	440.88 (167.25)	297.26 (59.51)
3	347.36 (32.12)	613.73 (189.61)	350.70 (39.17)	354.04 (49.26)
4	521.04 (80.44)	480.96 (94.76)	467.60 (70.85)	484.30 (89.15)
5	344.02 (36.59)	414.16 (115.94)	444.22 (132.87)	330.66 (13.97)
6	584.50 (67.84)	698.06 (41.58)	671.34 (82.99)	457.58 (94.17)
7	354.04 (59.51)	454.24 (143.37)	464.26 (157.28)	377.42 (108.61)
8	504.34 (143.37)	430.86 (75.06)	487.64 (113.51)	480.96 (122.95)
9	688.04 (18.29)	624.58 (83.17)	676.35 (21.56)	584.50 (197.24)
10	587.84 (80.44)	551.10 (186.34)	400.80 (66.80)	464.26 (94.03)

Note: Values are in ms.

believe that it is a result of brace mechanics, with perhaps an unusual amount of sagittal plane motion or the unique hinged design contributing to the faster times.

It is interesting to note that none of the braces was able to significantly reduce the maximum eversion angle. From a control value of -18.39°, maximum values of -15.04°, -15.56°, and -15.75° were noted for conditions 2, 3, and 4. These figures represent an average decrease of 15.6% compared to the control. In contrast, other studies [12, 23, 32] have found a decrease of up to 34% in maximum eversion angle during a braced condition. What remains to be determined is how much of a decrease (if any) is needed to properly protect against lateral sprains. A graphical representation of this angle for the present study is depicted in Figure 9.

As was previously mentioned, the mean maximum eversion values obtained for the braced conditions were decreased over the control (although not significantly). When examining maximum values among subjects, a range of -24.35° to -32.34° is seen. These values could be classified as excessive eversion for some subjects, although for others they could represent normal values (due to individual walking styles). An excessively everted foot at contact can lead to many problems, including plantar fasciitis, Achilles tendonitis, and tibial stress syndrome [6]. It may be necessary to evaluate the braces on an individual basis if possible to detect these abnormalities and avoid any secondary problems [32].

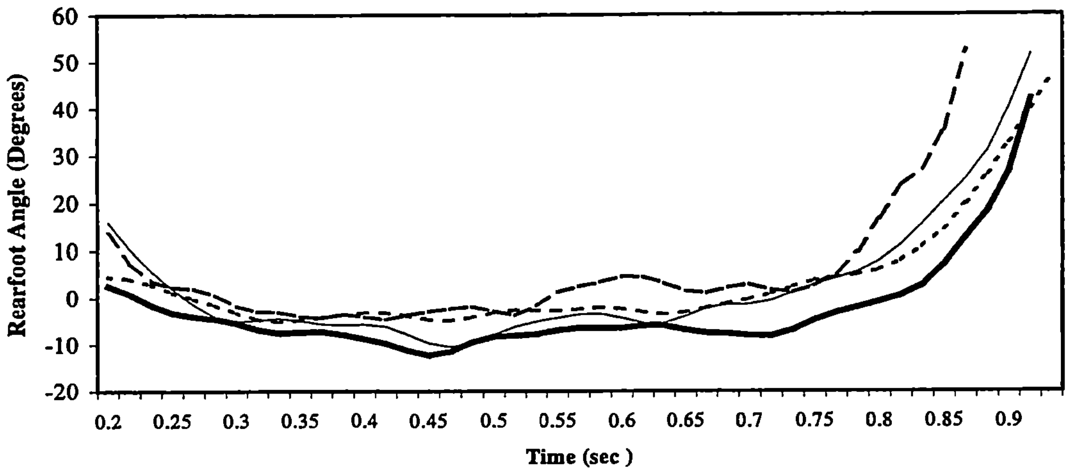


Figure 9. Representative rearfoot curves for each condition (from TDAngle to TOAngle). The darkest line represents condition 4 (Active Ankle).

It was hypothesized that each brace would be able to reduce the total amount of rearfoot eversion motion (EROM) compared to the control, and that the Air-Stirrup would show a greater control of this motion due to its design and intended usage. The Air-Stirrup's greater malleolar coverage and longer design make it preferable for acute injury, whereas the Sport-Stirrup and Active Ankle are primarily designed for protective situations (such as rehabilitation or a return to activity). The EROM variable exhibited values of 26.28°, 22.40°, 22.16°, and 23.07° for conditions 1, 2, 3, and 4, respectively. Although not significant, each of the braces was able to limit this motion to a certain extent (compared to the control). This finding is in partial agreement with other studies [12, 23, 32], which also found a decrease for this variable with the use of a brace (significantly different from the control). These studies have shown a decrease from 15.00° to 10.90° [23] and from 18.10° to 13.30° [12]. The maximum eversion angles for these studies were fairly close; the slight difference in the total motion can be seen in the touchdown angle. Hamill [23] and Morin [32] found an everted touchdown angle (-2.70°) for the braced condition, while an inverted angle (2.30) was shown by De Clercq [12]. In the present study the difference appears to be in the maximum eversion angle, which was decreased by an average of nearly 16% from other studies [12, 23, 32]. Touchdown angles, although showing both inversion and eversion characteristics, were consistent with the previously mentioned studies. A comparison of braces showed the Air-Stirrup and Sport-Stirrup reduced EROM the most compared with the control (Table 2). The result for the Sport-Stirrup is relatively surprising, being that this brace is designed more for functional activity (shorter, less malleolar coverage, etc.).

The velocity parameters in this study showed results that were expected. Maximum eversion velocity was decreased, and time to maximum eversion velocity was increased with the use of a brace (Table 2), although these results were not significant for either variable. From a control value of -292.57°/s, the maximum velocity was decreased by an average of 42.49°/s for the braced conditions. The Active Ankle showed the lowest maximum velocity of any condition (-227.91°/s). Although a similar pattern is seen in other studies (decreased values with the brace), it has not always been a significant difference. Hamill [23] found that values decreased from –184.70°/s to –79.20°/s with the brace (a larger difference than what was seen in the present study), but the change was not significant. However, maximum velocity values were significant for De Clercq [19], and ranged from -533.0°/s to –309.0°/s with a braced condition. A relationship between maximum eversion velocity and maximum eversion angle has been suggested by some authors [14]. The braces might help limit maximum eversion by reducing maximum eversion velocity. This was certainly the case in this study, as braced conditions revealed reduced maximum eversion angles, and decreased maximum velocity values. The braces were able to decrease the time it took to reach the maximum eversion velocity by an average of 68 ms, which is in good agreement with other studies. Hamill [23] found a decrease of 72 ms with the use of a brace. These findings support the idea of rearfoot control during braced testing conditions.

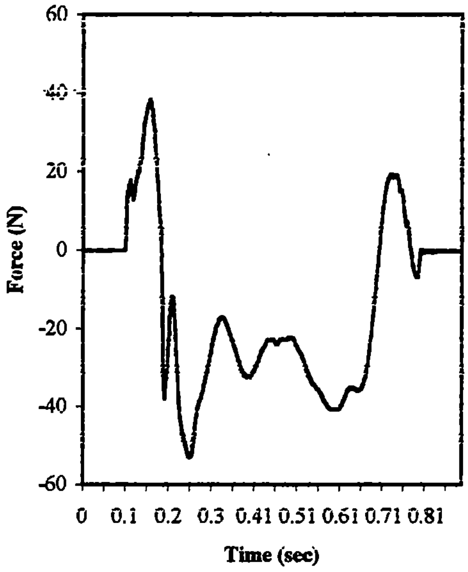
Kinetics

The kinetic analysis was performed in an attempt to correlate the results with the kinematic data. The three components of the GRF were used to help explain rearfoot motion during the stance phase. These components were: Fz (vertical GRF), Fy (anterior-posterior GRF), and Fx (medial-lateral GRF).

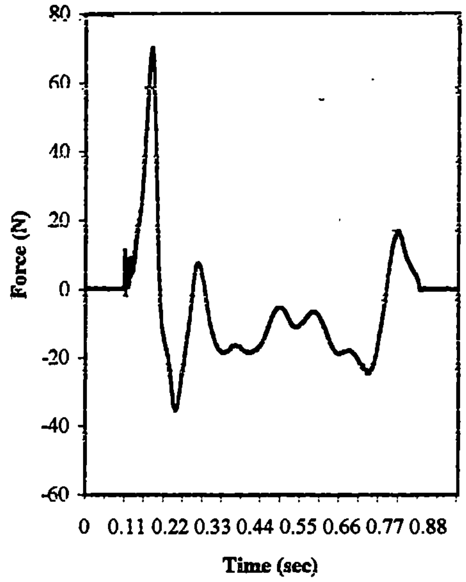
The fact that no vertical GRF variables were significantly different as a result of the braces is not a surprising finding. This component describes the forces applied by and absorbed through the leg during the support phase, and generally is bimodal in shape [16, 23]. The first peak typically occurs during the first half of support, and represents the pattern as the body is lowered after foot contact [16]. The second peak, occurring later during support, represents the active push against the ground in preparation to move into the swing phase [16]. Although the non-significant finding is in agreement with other studies [12, 23], shoe type was controlled in those studies and was not in the present study. Hamill [23] has suggested that no differences should be expected for vertical GRF variables when the same shoe type is used. It was thought that because shoe type was not controlled in this study, some variables in this component could show significant differences. However, this was not the case. Although not exactly the same, the shoes used by the participating subjects were mainly common running shoes (no sport-specific shoes, such as in volleyball or basketball).

Only one variable, braking impulse, showed significant differences for the anterior-posterior component of the GRF. This variable describes the area under the first half of the anterior-posterior curve. This represents the braking (or negative) portion of the curve, and is used to show that the subject has slowed down; that is, that the velocity has decreased [22]. In this study each braced condition was able to decrease this value, meaning that the braces were contributing to the lower velocity values seen previously. This also indicates that the braces were allowing a greater force to be applied over a longer amount of time. This value is not frequently reported in the literature; therefore, comparisons to other studies are difficult. It has been suggested that non-significance for anterior-posterior GRF variables indicates that subjects are walking at a constant velocity, and are under no/minimal influences as a result of the testing devices [23].

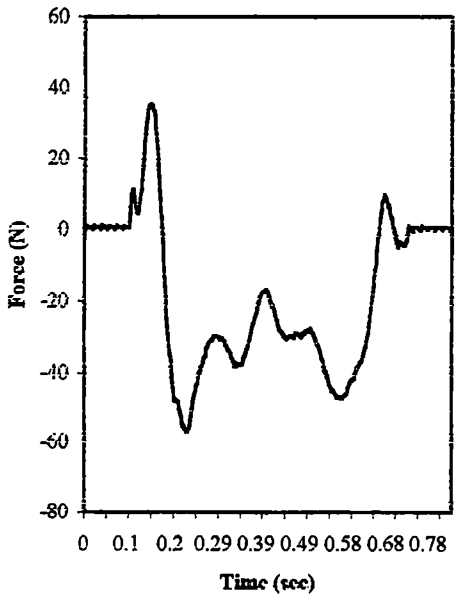
Perhaps the most interesting GRF component in this study is the medial-lateral (Fx). This component is extremely variable among subjects, and has been suggested to show no consistent pattern from person to person [16]. In this study, although the range of values differed for each subject, a distinctive pattern was seen (Figure 10). It was hypothesized that the braces would have an effect on this force, decreasing excursion values. If this were true, it would provide further evidence of rearfoot control from ankle braces. The initial analysis of the medial-lateral variables indicated an unusual graphical pattern for one subject (S9). This pattern was consistent among all three variables, and showed relatively high values compared to other subjects. An example of this can be seen in Figure 11, for Exc1.



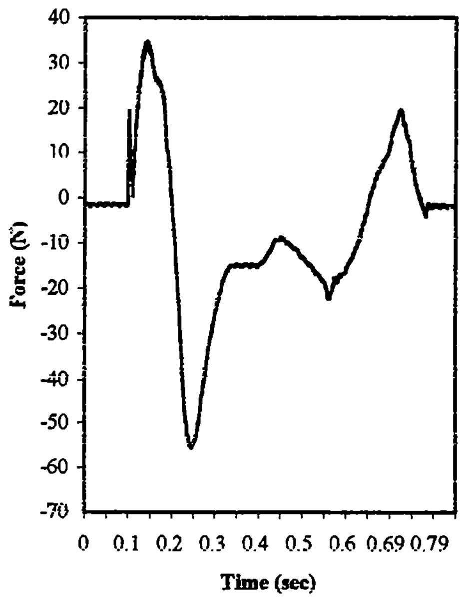
(a). Subject 1.



(b). Subject 2.

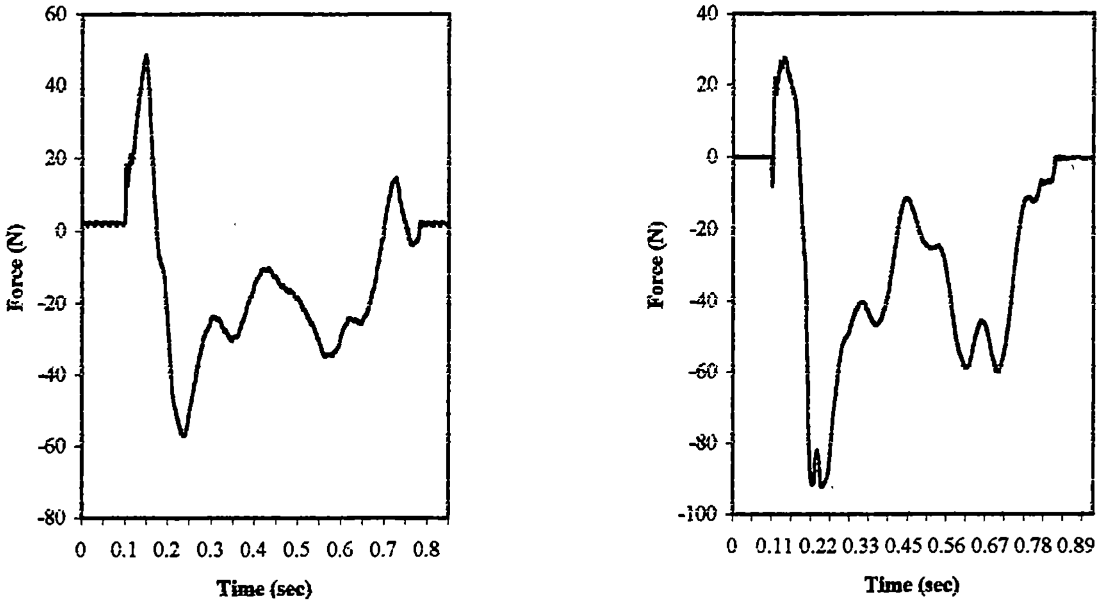


(c). Subject 3.



(d). Subject 4.

Figure 10. Medial-lateral GRF curves for six subjects.



(e). Subject 8.

(f). Subject 9.

Figure 10. (continued).

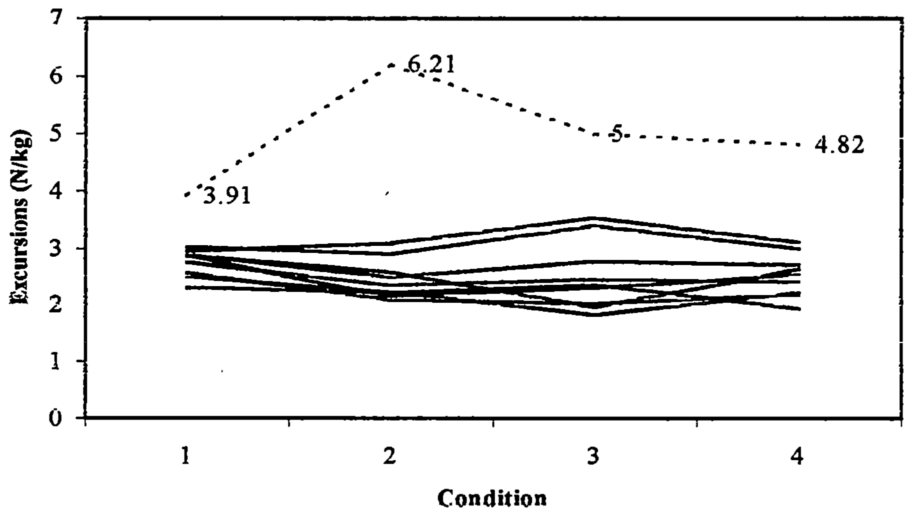


Figure 11. Condition means of all subjects for Exc1. The dashed line represents subject 9.

Due to the "abnormal" pattern exhibited by S9, the data were excluded and the repeated-measures ANOVA was ran again for the excursion variables. The results produced mean values that were much closer to what was anticipated, and these were used for the subsequent analysis. It is difficult to speculate why the subject exhibited this unusual pattern for the medial-lateral variables. A realistic explanation could be provided by shoe type. Subjects were encouraged to wear low-top athletic shoes (i.e., those designed for walking or running) during the data collection. This was done in an attempt to prevent large variations occurring as a result of different shoes. Even so, the fact that the shoes were not of the same brand/model could quite possibly explain some of the differences found in this study, and could be a reason subject 9 showed such high values for these variables. During measurement of passive ROM, this subject showed the highest inversion ROM, at 13.33°, while the average eversion ROM was consistent with the other subjects at 7.33°. The high inversion result could indicate that the subject had a history of minor injury to the right ankle that might not have healed properly (but denied any in the past year). Thus, the joint could have been chronically weak and slightly unstable as a result. This could, therefore, have led to the extreme excursion values for the medial-lateral GRF.

For Exc1, a significant difference was found among the conditions. Force excursions describe the sum of the absolute deviations of the force components in the medial-lateral direction [23]. In the present study, all excursion values were decreased for the braced conditions. Significant differences were found between the control and the Active Ankle for this variable, although each mean brace value was quite similar.

The other significant excursion variable, Exc2, represents the movement from 0% to approximately 50% during the stance phase. As was seen in the other Fx variables, values were decreased for every braced condition. Significant differences were seen again between the control and Active Ankle, and also between the control and Air-Stirrup. All three of the Fx variables show that the medial-lateral function of the foot is definitely moderated with the use of a brace in a manner that helps to control rearfoot movement. Although the third excursion variable (Exc3) was not significant, values were still always lower with the use of a brace, indicating at least some control during rearfoot movement.

Conclusions and Recommendations

In this study two kinematic variables were found to have a significant overall effect: time to maximum eversion angle and toe-off angle. The Active Ankle surprisingly showed a faster mean time to reach the maximum eversion angle than did the control. Although not significant, all other kinematic variables indicated a trend toward greater rearfoot control with the braces. The Air-Stirrup showed the greatest control in four of the seven variables (MaxEV, TMaxEV, TOAngle, and TMaxVel) and exhibited mean values very close to the braces showing the greatest rearfoot control for TDAngle and EROM (Sport-Stirrup) and for MaxVel (Active Ankle).

For the kinetic analysis, significant medial-lateral GRF differences were found for force excursions to 30% and 50% of the stance phase. The decreased values during the braced conditions supports the kinematic data in regards to the effect of rearfoot control. Anterior-posterior GRF results indicated one significant variable, braking impulse (IBrk), with no differences seen between individual braces. No significant differences were seen for the vertical GRF variables, which is an expected finding.

It appears, therefore, that each brace is capable of limiting and controlling rearfoot movement during gait. For the variables used in this study, the Air-Stirrup would seem to be a logical choice when attempting to control such movement during a selected activity. Although this particular brace is designed with this control in mind, its bulkiness and length have mandated its use more in the acute stages of an injury than for activity. The Sport-Stirrup and Active Ankle did not show as much rearfoot control as the Air-Stirrup (although close in many categories), but typically are used more on the athletic field where functional ability must not be compromised. It should be pointed out that these results might not apply to the situations in which the athlete is typically involved, although standard gait is a basis for most athletic activities. The speed, duration, and intensity of athletics places a far greater demand on the body than does gait.

It would be ideal to examine rearfoot movement during actual athletic activity. Most studies presented in the literature have attempted to generalize the results of walking and/or running protocols to athletics. While this may be a presently accepted technique, additional methods to accurately measure this movement during activity are needed. In addition, studies looking at the effects of ankle braces on other lower extremity joints (i.e., knee and hip) would help to determine if the braces are altering mechanics in any way. Although the brace might be restricting inversion/eversion, it could be doing so in a manner that is causing secondary problems. This area of research needs to explored further.

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APPENDICES

APPENDIX A

SUBJECT INFORMATION

Table 11. Subject information.

Subject	Age	Body Mass (kg)	Mean Walking Speed (m/sec)
1	24	68.55	1.68
2	24	68.93	1.32
3	25	77.99	1.45
4	24	101.59	1.56
5	26	69.25	1.40
6	24	77.59	1.59
7	29	105.60	1.43
8	21	94.92	1.36
9	20	80.52	1.33
10	20	76.08	1.01
Overall	23.70	82.10	1.41

APPENDIX B

SUBJECT KINEMATIC RESULTS

Table 12. Subject means and standard deviations of kinematic variables.

Subj	Cond	TDAngle (deg)	MaxEV (deg)	TMaxEV (ms)	EROM (deg)	TOAngle (deg)	MaxVel (deg/s)	TMaxVel (ms)
1	1	-19.75	-29.81	357.38	10.06	26.13	-275.53	203.74
		(3.03)	(1.88)	(135.98)	(2.57)	(13.13)	(106.36)	(34.22)
	2	-7.26	-24.35	354.04	17.09	19.09	-514.94	384.10
		(4.20)	(1.72)	(165.91)	(5.32)	(23.35)	(374.87)	(273.65)
	3	-3.54	-24.59	354.04	21.05	43.52	-408.12	313.96
		(8.57)	(5.78)	(167.99)	(7.19)	(5.03)	(126.61)	(255.84)
	4	-19.86	-32.34	254.68	12.48	6.48	-324.01	329.83
		(4.50)	(1.94)	(15.99)	(3.83)	(8.26)	(88.30)	(281.56)
2	1	-2.86	-14.25	434.20	11.39	35.57	-298.37	203.74
		(2.15)	(0.75)	(171.94)	(2.24)	(4.64)	(58.37)	(18.29)
	2	-7.92	-19.22	501.00	11.31	22.87	-205.04	207.08
		(3.08)	(2.94)	(174.35)	(4.70)	(8.25)	(107.33)	(14.94)
	3	-7.63	-21.76	440.88	14.14	15.73	-310.38	270.54
		(3.25)	(2.99)	(167.25)	(5.90)	(10.83)	(47.14)	(203.99)
	4	-10.81	-18.09	297.26	7.28	25.34	-216.31	203.74
		(3.84)	(1.76)	(59.51)	(2.58)	(3.83)	(55.70)	(13.97)
3	1	2.87	-23.03	347.36	25.90	8.55	-379.90	230.46
		(1.70)	(3.01)	(32.12)	(3.54)	(13.60)	(43.42)	(7.47)
	2	4.27	-19.93	613.73	24.20	-16.01	-327.56	237.98
		(4.85)	(1.89)	(189.61)	(3.23)	(3.37)	(64.57)	(31.61)
	3	-5.12	-26.12	350.70	21.00	-23.68	-336.28	227.12
		(3.46)	(2.77)	(39.17)	(5.82)	(16.83)	(34.57)	(9.15)
	4	-1.63	-17.95	354.04	16.32	-11.01	-254.30	230.46
		(2.38)	(1.68)	(49.26)	(2.64)	(6.75)	(72.52)	(7.47)
4	1	-5.03	-21.52	521.04	16.49	35.19	-234.78	250.50
		(1.47)	(4.59)	(80.44)	(4.73)	(5.56)	(48.95)	(74.69)
	2	-8.15	-19.00	480.96	10.84	21.58	-228.21	400.80
		(3.26)	(2.23)	(94.76)	(4.13)	(11.71)	(51.56)	(156.21)
	3	6.48	-11.25	467.60	17.72	32.57	-170.83	397.46
		(1.25)	(1.95)	(70.85)	(0.96)	(10.10)	(34.75)	(195.61)
	4	-5.11	-23.92	484.30	18.81	20.80	-291.21	437.54
		(4.20)	(2.73)	(89.15)	(6.89)	(16.47)	(71.22)	(246.97)
5	1	0.29	-17.33	344.02	17.62	23.44	-234.52	253.84
		(2.07)	(2.46)	(36.59)	(2.31)	(10.70)	(53.81)	(24.77)
	2	7.43	-11.73	414.16	19.16	21.89	-309.89	250.50
		(3.94)	(1.49)	(115.94)	(4.78)	(8.12)	(94.07)	(26.41)
	3	-2.44	-16.91	444.22	14.48	14.36	-216.09	233.80
		(3.56)	(2.42)	(132.87)	(4.33)	(3.49)	(77.16)	(31.24)
	4	6.21	-10.92	330.66	17.13	22.39	-220.94	247.16
		(4.00)	(4.49)	(13.97)	(8.45)	(7.81)	(90.68)	(43.23)

Table 12. (continued).

Subj	Cond	TDAngle (deg)	MaxEV (deg)	TMaxEV (ms)	EROM (deg)	TOAngle (deg)	MaxVel (deg/s)	TMaxVel (ms)
6	1	-6.43	-23.33	584.50	16.89	36.25	-267.15	220.44
		(2.09)	(3.24)	(67.84)	(2.48)	(8.63)	(78.33)	(13.97)
	2	-6.33	-19.46	698.06	13.13	6.44	-103.31	400.80
		(1.88)	(2.39)	(41.58)	(2.75)	(3.20)	(25.54)	(95.21)
	3	0.50	-15.72	671.34	16.22	25.43	-133.95	467.60
		(4.09)	(2.08)	(82.99)	(4.97)	(8.37)	(34.30)	(224.37)
	4	-0.46	-12.34	457.58	11.88	35.61	-137.94	327.32
		(1.96)	(3.51)	(94.17)	(4.60)	(4.02)	(41.80)	(228.25)
7	1	-1.44	-21.56	354.04	20.13	45.07	-270.22	263.98
		(5.62)	(5.04)	(59.51)	(7.18)	(5.83)	(71.69)	(13.83)
	2	1.49	-15.60	454.24	17.09	46.36	-242.50	233.80
		(2.11)	(2.09)	(143.37)	(3.56)	(14.18)	(43.10)	(16.70)
	3	3.09	-13.19	464.26	16.28	48.80	-244.98	257.18
		(2.76)	(2.99)	(157.28)	(3.58)	(3.40)	(39.39)	(48.11)
	4	-2.14	-14.63	377.42	12.49	45.19	-231.28	357.38
		(5.28)	(1.14)	(108.61)	(5.91)	(3.67)	(72.95)	(241.02)
8	1	12.04	-10.51	504.34	22.56	60.96	-312.86	233.80
		(4.00)	(4.48)	(143.37)	(3.61)	(9.54)	(52.28)	(11.81)
	2	6.01	-6.11	430.86	12.13	40.28	-195.06	317.30
		(3.39)	(1.22)	(75.06)	(2.94)	(13.16)	(100.82)	(92.98)
	3	6.93	-7.48	487.64	14.41	48.24	-212.74	243.82
		(4.63)	(2.59)	(113.51)	(5.18)	(8.60)	(88.66)	(57.37)
	4	3.21	-9.40	480.96	12.62	46.97	-130.02	310.62
		(1.16)	(2.24)	(122.95)	(2.04)	(11.57)	(43.20)	(190.63)
9	1	15.27	-15.16	688.04	30.44	45.04	-464.53	217.10
		(3.52)	(2.33)	(18.29)	(5.33)	(8.34)	(128.83)	(16.70)
	2	10.45	-13.56	624.58	24.00	34.81	-244.45	240.48
		(5.44)	(1.32)	(83.17)	(6.26)	(5.39)	(58.06)	(43.55)
	3	11.20	-16.73	676.35	27.93	37.79	-377.67	204.58
		(4.45)	(2.28)	(21.56)	(3.75)	(2.64)	(92.74)	(8.35)
	4	9.28	-12.16	584.40	21.44	40.41	-302.28	213.76
		(11.56)	(2.20)	(197.24)	(9.46)	(5.27)	(52.90)	(21.77)
10	1	8.23	-7.39	587.84	15.62	35.87	-187.84	243.82
		(4.83)	(1.86)	(80.44)	(4.72)	(6.82)	(34.99)	(32.55)
	2	12.16	-1.46	551.10	13.62	27.78	-213.65	404.14
		(2.70)	(2.38)	(186.34)	(3.23)	(9.29)	(17.46)	(241.26)
	3	12.00	-1.88	400.80	13.89	35.53	-227.74	354.04
		(3.49)	(2.19)	(66.80)	(4.25)	(7.23)	(81.87)	(235.11)
	4	6.58	-5.77	464.26	12.35	35.58	-170.80	293.92
		(4.82)	(3.92)	(94.03)	(5.70)	(6.94)	(29.32)	(134.95)

APPENDIX C

SUBJECT GRF RESULTS (MEDIAL-LATERAL)

Table 13. Subject means and standard deviations of medial-lateral GRF variables.

Subj	Cond	Exc1 (N/kg)	Exc2 (N/kg)	Exc3 (N/kg)
1	1	2.88	3.69	5.25
		(0.20)	(0.25)	(0.27)
	2	2.51	3.29	4.79
		(0.25)	(0.36)	(0.45)
	3	2.78	3.56	5.07
		(0.40)	(0.47)	(0.56)
	4	2.72	3.43	4.85
		(0.26)	(0.29)	(0.25)
2	1	2.89	3.37	4.38
		(0.52)	(0.66)	(0.65)
	2	2.14	2.52	3.60
		(0.84)	(0.86)	(0.73)
	3	2.31	2.78	3.79
		(0.83)	(0.77)	(0.97)
	4	2.56	2.92	4.11
		(0.56)	(0.60)	(0.69)
3	1	3.02	3.70	4.94
		(0.20)	(0.29)	(0.28)
	2	2.90	3.53	4.75
		(0.21)	(0.22)	(0.23)
	3	3.40	3.96	5.25
		(0.45)	(0.46)	(0.49)
	4	2.99	3.66	4.92
		(0.53)	(0.61)	(0.58)
4	1	2.87	3.47	4.53
		(0.34)	(0.39)	(0.47)
	2	2.60	3.23	4.51
		(0.22)	(0.29)	(0.32)
	3	1.95	2.58	3.85
		(0.45)	(0.62)	(0.73)
	4	2.65	3.24	4.60
		(0.11)	(0.13)	(0.16)
5	1	2.95	3.30	4.61
		(0.41)	(0.45)	(0.42)
	2	3.09	3.46	4.82
		(0.56)	(0.53)	(0.46)
	3	3.54	4.00	5.42
		(0.79)	(0.77)	(0.77)
	4	3.10	3.40	4.79
		(0.69)	(0.68)	(0.72)

Table 13. (continued).

Subj	Cond	Exc1 (N/kg)	Exc2 (N/kg)	Exc3 (N/kg)
6	1	2.77	3.33	4.41
		(0.23)	(0.22)	(0.21)
	2	2.36	2.81	3.99
		(0.32)	(0.33)	(0.40)
	3	2.47	2.91	4.07
		(0.32)	(0.33)	(0.43)
	4	2.43	2.89	4.05
		(0.33)	(0.26)	(0.30)
7	1	2.52	3.13	4.37
		(0.19)	(0.28)	(0.22)
	2	2.22	2.57	3.68
		(0.15)	(0.16)	(0.16)
	3	1.82	2.21	3.34
		(0.19)	(0.21)	(0.24)
	4	2.21	3.01	4.18
		(0.66)	(0.65)	(0.60)
8	1	2.31	2.71	4.16
		(0.23)	(0.28)	(0.19)
	2	2.21	2.58	3.99
		(0.21)	(0.25)	(0.23)
	3	2.37	3.00	4.44
		(0.14)	(0.16)	(0.21)
	4	1.92	2.28	3.79
		(0.34)	(0.42)	(0.37)
9	1	3.91	4.32	5.47
		(0.43)	(0.52)	(0.59)
	2	6.21	6.64	7.73
		(1.12)	(1.08)	(1.15)
	3	5.00	5.40	6.51
		(0.92)	(0.91)	(0.97)
	4	4.82	5.25	6.43
		(1.46)	(1.44)	(1.46)
10	1	2.58	2.91	4.21
		(0.68)	(0.67)	(0.65)
	2	2.08	2.41	3.68
		(0.35)	(0.37)	(0.34)
	3	2.01	2.39	3.60
		(0.34)	(0.34)	(0.28)
	4	2.18	2.48	3.75
		(0.58)	(0.58)	(0.73)

APPENDIX D

SUBJECT GRF RESULTS (ANTERIOR-POSTERIOR)

Table 14. Subject means and standard deviations of anterior-posterior GRF variables.

Subj	Cond	MaxBrk (N/kg)	TMaxBrk (s)	MaxProp (N/kg)	TMaxProp (s)	IBrk (N-s/kg)	IProp (N-s/kg)
1	1	-2.21	0.11	1.83	0.54	-0.31	0.23
		(0.15)	(0.00)	(0.03)	(0.01)	(0.03)	(0.01)
	2	-2.22	0.11	1.86	0.53	-0.30	0.23
		(0.13)	(0.00)	(0.04)	(0.02)	(0.03)	(0.01)
	3	-2.40	0.11	1.86	0.53	-0.31	0.23
		(0.11)	(0.01)	(0.03)	(0.01)	(0.01)	(0.01)
	4	-2.27	0.11	1.86	0.54	-0.31	0.23
		(0.16)	(0.01)	(0.06)	(0.01)	(0.02)	(0.00)
2	1	-2.29	0.10	1.94	0.61	-0.31	0.29
		(0.17)	(0.00)	(0.07)	(0.01)	(0.03)	(0.02)
	2	-2.13	0.12	1.83	0.63	-0.33	0.28
		(0.12)	(0.01)	(0.04)	(0.01)	(0.02)	(0.01)
	3	-2.03	0.11	2.11	0.59	-0.29	0.30
		(0.31)	(0.00)	(0.23)	(0.02)	(0.04)	(0.04)
	4	-2.23	0.11	2.06	0.60	-0.30	0.30
		(0.22)	(0.01)	(0.08)	(0.01)	(0.03)	(0.02)
3	1	-2.35	0.11	4.93	0.60	-0.35	0.32
		(0.11)	(0.01)	(0.17)	(0.01)	(0.01)	(0.02)
	2	-2.27	0.12	4.70	0.60	-0.32	0.31
		(0.03)	(0.01)	(0.05)	(0.01)	(0.01)	(0.01)
	3	-2.21	0.11	4.60	0.59	-0.31	0.30
		(0.13)	(0.01)	(0.20)	(0.01)	(0.02)	(0.02)
	4	-2.28	0.12	4.87	0.61	-0.37	0.32
		(0.07)	(0.00)	(0.09)	(0.00)	(0.01)	(0.02)
4	1	-2.26	0.13	5.05	0.60	-0.42	0.36
		(0.13)	(0.01)	(0.13)	(0.01)	(0.05)	(0.02)
	2	-2.27	0.13	5.00	0.60	-0.41	0.36
		(0.15)	(0.01)	(0.19)	(0.01)	(0.04)	(0.03)
	3	-2.54	0.12	5.41	0.60	-0.39	0.40
		(0.05)	(0.01)	(0.10)	(0.01)	(0.04)	(0.01)
	4	-2.16	0.13	4.95	0.60	-0.43	0.34
		(0.15)	(0.01)	(0.16)	(0.01)	(0.03)	(0.03)
5	1	-2.30	0.10	5.06	0.58	-0.42	0.32
		(0.11)	(0.01)	(0.16)	(0.01)	(0.03)	(0.02)
	2	-2.30	0.11	5.02	0.58	-0.39	0.32
		(0.09)	(0.00)	(0.17)	(0.01)	(0.02)	(0.02)
	3	-2.52	0.10	5.47	0.56	-0.41	0.32
		(0.08)	(0.00)	(0.16)	(0.01)	(0.03)	(0.01)
	4	-2.28	0.11	4.96	0.58	-0.36	0.32
		(0.10)	(0.01)	(0.10)	(0.01)	(0.01)	(0.02)

Table 14. (continued).

Subj	Cond	MaxBrk (N/kg)	TMaxBrk (s)	MaxProp (N/kg)	TMaxProp (s)	IBrk (N-s/kg)	IProp (N-s/kg)
6	1	-2.12	0.11	4.39	0.57	-0.33	0.29
		(0.09)	(0.01)	(0.15)	(0.00)	(0.01)	(0.02)
	2	-1.87	0.12	4.01	0.58	-0.32	0.26
		(0.05)	(0.00)	(0.07)	(0.00)	(0.02)	(0.01)
	3	-2.05	0.11	4.32	0.56	-0.33	0.29
		(0.05)	(0.00)	(0.08)	(0.01)	(0.02)	(0.01)
	4	-2.07	0.11	4.41	0.57	-0.34	0.29
		(0.13)	(0.01)	(0.21)	(0.01)	(0.01)	(0.02)
7	1	-2.21	0.10	4.81	0.62	-0.40	0.35
		(0.03)	(0.00)	(0.06)	(0.01)	(0.02)	(0.01)
	2	-2.30	0.10	4.90	0.62	-0.37	0.37
		(0.06)	(0.01)	(0.07)	(0.01)	(0.03)	(0.02)
	3	-2.10	0.11	4.58	0.63	-0.40	0.33
		(0.07)	(0.01)	(0.14)	(0.00)	(0.02)	(0.01)
	4	-2.24	0.09	4.93	0.62	-0.38	0.35
		(0.17)	(0.01)	(0.35)	(0.01)	(0.03)	(0.03)
8	1	-2.24	0.12	4.71	0.62	-0.36	0.35
		(0.07)	(0.02)	(0.13)	(0.01)	(0.01)	(0.02)
	2	-2.07	0.12	4.36	0.62	-0.36	0.33
		(0.09)	(0.01)	(0.15)	(0.01)	(0.03)	(0.02)
	3	-2.33	0.11	4.77	0.60	-0.36	0.34
		(0.13)	(0.02)	(0.24)	(0.01)	(0.01)	(0.01)
	4	-2.31	0.11	4.78	0.62	-0.36	0.35
		(0.08)	(0.00)	(0.12)	(0.01)	(0.03)	(0.01)
9	1	-2.04	0.14	4.48	0.65	-0.45	0.30
		(0.10)	(0.02)	(0.14)	(0.01)	(0.02)	(0.01)
	2	-1.98	0.14	4.23	0.64	-0.39	0.29
		(0.12)	(0.01)	(0.22)	(0.01)	(0.04)	(0.02)
	3	-2.04	0.14	4.42	0.65	-0.43	0.29
		(0.11)	(0.01)	(0.19)	(0.01)	(0.03)	(0.02)
	4	-2.08	0.14	4.57	0.65	-0.43	0.28
		(0.05)	(0.01)	(0.10)	(0.01)	(0.02)	(0.01)
10	1	-1.90	0.10	3.78	0.65	-0.30	0.30
		(0.05)	(0.02)	(0.10)	(0.02)	(0.03)	(0.02)
	2	-1.79	0.13	3.62	0.66	-0.28	0.29
		(0.12)	(0.01)	(0.21)	(0.01)	(0.03)	(0.02)
	3	-1.82	0.10	3.71	0.65	-0.28	0.28
		(0.13)	(0.00)	(0.20)	(0.01)	(0.02)	(0.02)
	4	-1.76	0.12	3.58	0.67	-0.30	0.28
		(0.10)	(0.02)	(0.21)	(0.02)	(0.01)	(0.02)

APPENDIX E

SUBJECT GRF RESULTS (VERTICAL)

Table 15. Subject means and standard deviations of vertical GRF variables.

Subj	Cond	F1 (N/kg)	T1 (s)	F2 (N/kg)	T2 (s)
1	1	10.46 (0.22)	0.14 (0.00)	10.07 (0.16)	0.46 (0.01)
	2	10.54 (0.31)	0.13 (0.01)	10.25 (0.08)	0.46 (0.01)
	3	10.96 (0.34)	0.13 (0.01)	10.19 (0.19)	0.46 (0.01)
	4	10.75 (0.32)	0.13 (0.00)	10.02 (0.23)	0.46 (0.01)
2	1	10.70 (0.45)	0.14 (0.03)	10.73 (0.24)	0.54 (0.02)
	2	10.44 (0.18)	0.17 (0.00)	10.90 (0.14)	0.56 (0.01)
	3	10.47 (0.38)	0.17 (0.02)	11.03 (0.29)	0.53 (0.01)
	4	10.49 (0.42)	0.15 (0.04)	10.91 (0.31)	0.53 (0.02)
3	1	11.28 (0.13)	0.15 (0.00)	11.40 (0.10)	0.54 (0.01)
	2	11.23 (0.14)	0.16 (0.00)	11.38 (0.11)	0.53 (0.01)
	3	11.12 (0.35)	0.16 (0.00)	11.36 (0.25)	0.52 (0.00)
	4	11.35 (0.14)	0.16 (0.00)	11.43 (0.06)	0.55 (0.01)
4	1	11.39 (0.11)	0.16 (0.01)	11.67 (0.12)	0.55 (0.01)
	2	11.25 (0.29)	0.16 (0.01)	11.63 (0.14)	0.55 (0.01)
	3	11.41 (0.27)	0.17 (0.01)	11.71 (0.13)	0.55 (0.00)
	4	11.42 (0.20)	0.17 (0.01)	11.48 (0.24)	0.55 (0.00)
5	1	11.60 (0.41)	0.14 (0.01)	10.82 (0.15)	0.54 (0.01)
	2	11.74 (0.36)	0.15 (0.01)	11.04 (0.26)	0.53 (0.01)
	3	11.88 (0.45)	0.13 (0.01)	11.19 (0.12)	0.52 (0.01)
	4	11.35 (0.25)	0.14 (0.01)	10.97 (0.26)	0.53 (0.01)

Table 15. (continued).

Subj	Cond	F1 (N/kg)	T1 (s)	F2 (N/kg)	T2 (s)
6	1	11.62	0.15	10.70	0.52
		(0.17)	(0.00)	(0.14)	(0.00)
	2	11.37	0.16	10.79	0.52
		(0.23)	(0.00)	(0.21)	(0.00)
	3	11.74	0.15	11.06	0.51
		(0.10)	(0.01)	(0.13)	(0.01)
	4	11.52	0.15	11.04	0.52
		(0.13)	(0.00)	(0.08)	(0.01)
7	1	11.12	0.15	10.46	0.54
		(0.51)	(0.02)	(0.32)	(0.01)
	2	10.98	0.15	10.31	0.54
		(0.58)	(0.01)	(0.28)	(0.01)
	3	10.89	0.17	10.28	0.55
		(0.37)	(0.01)	(0.33)	(0.01)
	4	10.88	0.14	10.12	0.54
		(0.26)	(0.02)	(0.12)	(0.01)
8	1	11.31	0.18	10.80	0.57
		(0.23)	(0.01)	(0.29)	(0.01)
	2	10.93	0.18	10.56	0.57
		(0.19)	(0.01)	(0.27)	(0.01)
	3	11.38	0.18	11.12	0.55
		(0.47)	(0.01)	(0.37)	(0.01)
	4	11.08	0.19	11.10	0.57
		(0.15)	(0.01)	(0.23)	(0.01)
9	1	12.06	0.1654	10.87	0.56
		(0.25)	(0.02)	(0.65)	(0.01)
	2	11.84	0.17	10.98	0.57
		(0.40)	(0.01)	(0.45)	(0.01)
	3	11.84	0.17	11.19	0.57
		(0.34)	(0.01)	(0.32)	(0.01)
	4	12.04	0.17	10.96	0.57
		(0.53)	(0.01)	(0.21)	(0.01)
10	1	10.43	0.19	10.34	0.57
		(0.19)	(0.01)	(0.22)	(0.02)
	2	10.50	0.19	10.49	0.58
		(0.30)	(0.02)	(0.28)	(0.02)
	3	10.06	0.19	10.41	0.58
		(0.16)	(0.00)	(0.11)	(0.01)
	4	10.24	0.20	10.48	0.59
		(0.35)	(0.01)	(0.15)	(0.01)

APPENDIX F

INFORMED CONSENT

Biomechanical Effects of Ankle Bracing During Gait
Informed Consent Form

Principal Investigator:
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Faculty Advisor:
Songning Zhang, Ph. D.
Rm. 337, HPER
1914 Andy Holt Avenue
Knoxville, TN 37996
(423) 974-4716
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You are invited to participate in a study entitled "Biomechanical Effects of Ankle Bracing During Gait." This study will look at rearfoot range-of-motion (ROM) using three different, commercially available ankle braces.

You understand that the study will consist of one (1) testing session that should require no more than two (2) hours of data collection. During this session, you will complete a specific series of tasks and participate in four (4) different testing conditions (three braced and one unbraced). At the beginning of the session the passive rearfoot ROM (inversion/eversion) of your right ankle will be measured using standard techniques. Reflective markers will then be placed on specific anatomical landmarks to assist in the data analysis. After explanation of the procedures, you will complete the four (4) testing conditions. Each condition requires 1) the application of the brace (if required) according to the manufacturer's instructions, 2) a brief warm-up, and 3) the completion of five (5) walking trials. You will begin at one end of the walking surface and walk at your normal pace to the end of the surface. In the middle of the walk you will be asked to step with your right foot completely onto a force platform that is used to record kinetic data. During the testing, you will also be filmed from both the side- and rear-views as part of the kinematic analysis.

Risks involved in this study are minimal, and every effort will be made to ensure that the safety of participation is maximized. All testing will be performed by the principal investigator and the qualified personnel of the Biomechanics/Sports Medicine Lab. The principal investigator is a certified athletic trainer, and the lab personnel have knowledge of first aid procedures should any medical problems arise. In the event of physical injury due to your participation in the study, the University of Tennessee does not automatically provide reimbursement for medical care or other compensation.

Your participation in the study is completely voluntary. You may withdraw from the study at any time without penalty or loss of benefits gained as a result of the study. You will be verbally reminded of this prior to and at different points throughout the study. Your identity will remain confidential during the study; you will be referred to by subject number only (not by name). The personnel of the Biomechanics/Sports Medicine Lab will have the only access to subject information and data reports. All video tapes, data disks, and subject information will be kept in a locked file cabinet in Room 337, HPER.

Your voluntary signature below indicates that you have read and understand the preceding statements and agree to participate according to the procedures described above.

Signature ________________________________ Date ____________

Witness ________________________________ Date ____________

VITA

Steven R. Casto was born in Charleston, WV on August 7, 1973. After graduating from Scott High School in Madison, WV in June of 1991, he went on to attend the University of Charleston (Charleston, WV). While at UC, Steve studied Sports Medicine/Athletic Training, and received his Bachelor of Science degree in May of 1996. In August of 1996 he entered the University of Tennessee, Knoxville, and began study in Exercise Science (emphasis in Kinesiology/Sports Medicine). During this time he was able to serve as a graduate assistant in the Department of Sport and Physical Activity, and earned his athletic training certification (NATABOC) in April of 1997. Upon completion of his thesis, he received the Master of Science degree in Human Performance and Sport Studies in May of 1999. In August of 1999, Steve will begin the Ph. D. program in Exercise Science (Biomechanics) at the University of Georgia, where he will also be working as a graduate research assistant.