4-1995

The Effect of the Active Ankle Brace on Ground Reaction Forces

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by

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A Thesis
Submitted to the
Faculty of The Graduate College
in partial fulfillment of the
requirements for the
Degree of Master of Arts
Department of Health, Physical Education,
and Recreation

Western Michigan University
Kalamazoo, Michigan
April 1995

ACKNOWLEDGMENTS

I would like to express my sincere appreciation to my committee chair, Dr. Mary Dawson, and my committee members Dr. Roger Zabik, Dr. Robert Moss, and Dr. Patricia Frye. Their guidance and understanding helped maintain my motivation throughout the completion of this study, and more importantly for "not giving up on me". Without their hard work and dedication to quality, this study would not have been completed as smoothly as it was.

The subjects deserve some gratitude as well. They were all dependable and dedicated to making the data collection phase of this study as successful as possible, no matter how many trials it took to get everything in working order.

I would also like to thank my family and friends who continuously provided support and encouragement when I was ready to throw in the towel. I mention a few and they include: my mother, Martha P. Howell, who always told me I could do what ever was asked, no matter what bounds might restrict my motions; and my younger brother, Christopher D. Howell, who has been a larger influence on my success than he might ever imagine. I thank Lou, RC, and Darrell for all those times they spent listening and offering solutions that in the end were right.

Finally, I am dedicating this thesis to the memory of my father, Dr. David G. Howell, who raised me to be an achiever, to push myself to the limits, and be dedicated to finish the job through determination and hard work. I now understand the theory behind the moving woodpile and stacking bark side up.

Andrew D. Howell

THE EFFECT OF THE ACTIVE ANKLE BRACE ON GROUND REACTION FORCES

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The problem of this study was to determine the effect an ankle brace worn for stability had on ground reaction forces. Ground reaction forces present when performing a step down from a height of 8.0 in. while wearing an Active Ankle Brace were compared to the ground reaction forces when not wearing the brace. Subjects (N=50) were randomly assigned to a testing condition. A metronome set at a rate of 100 bpm controlled the walking cadence of the subjects. Subjects were told to walk with a normal gait pattern, at the required cadence, and to use a heel strike landing. Each subject completed 20 trials, 10 with the ankle brace and 10 without the brace. Dependent variables measured were peak impact force, vertical loading rate, maximum medial force, maximum lateral force, and time to peak force. Significant differences were found between subjects across the dependent variables, between the 10 trials in vertical loading rate, between the brace and no-brace conditions in maximum lateral force, and in time to peak force between the brace and no-brace conditions. The researcher concluded that the ankle brace: (a) did not affect the peak impact force, (b) affected the vertical loading rate across the trials for subjects, (c) did not affect the maximum medial force, (d) affected the maximum lateral force by decreasing it for the brace condition, and (e) affected the time to peak force by increasing it for the brace condition. Recommendations for further study include replicating the study and investigating the effect of the brace on ground reaction forces in a variety of movement patterns.

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CHAPTER I

INTRODUCTION

An athlete has great demands placed upon his body during every performance. Activities requiring bipedal motion place an even greater demand on the foot and ankle. The athlete's performance often depends on the ability of the foot and ankle to maintain a base of support, absorb shock, and act as a lever through which the forces for mobility can be produced (Hunt, 1990). Cailiet (1969) described four criteria for "normalcy" in the foot and ankle: (1) absence of pain, (2) normal muscle balance, (3) central heel, and (4) straight and mobile toes. Donatelli (1990) stated that an even distribution of weightbearing forces during the stance phase of gait is also important. The ankle has been recognized as one of the most vulnerable areas of the body during athletic participation and is highly susceptible to injury. Because the foot and ankle are sometimes subjected to forces 1.25 to 8.0 times greater than the weight of the body, they are at risk of injury (Pratt, 1989). Focus of past research has been on prevention of injury by increasing the stability of the ankle joint while decreasing mobility (Gehlsen, Pearson, & Bahamonde, 1991).

Lately, researchers have shown increased interest in ground reaction forces because of the injuries caused by excessive shock to the bones and soft tissues of the lower extremities (Dufek & Bates, 1991). Many lower extremity injuries have been associated with overuse phenomena resulting from repetitive impact loading on the foot (James, Bates, & Osternig, 1978). The reduction of excessive shock has been dealt with primarily in the design of shoes and shock absorbing inserts, but nothing has been done to incorporate shock reduction into braces, orthoses, and devices designed to provide stability.

Statement of the Problem

The problem of this study was to determine the effect an ankle brace worn for stability has on ground reaction forces. More specifically, it was the problem of this study to compare ground reaction forces present when performing a step down from a height of 8.0 in. while wearing an ankle brace to the ground reaction forces present when not wearing the brace.

Purpose of the Study

The purpose of the study was to provide athletic trainers with information to aid in understanding the effect an ankle brace will have on ground reaction forces. If ground reaction forces are greater when wearing a brace than when not wearing a brace, trainers and coaches may want to: (a) recommend footwear with greater shock absorbing qualities, or (b) alter the mechanics of sport technique to better disperse the forces associated with landing, or (c) restrict activity. If ground reaction forces are less when wearing a brace than when not wearing a brace, trainers will know that the brace or an alteration in the mechanics of the movement pattern due to wearing the brace, reduced or dissipated the ground reaction forces. Ground reaction forces may not be different between a braced and unbraced ankle, in which case the trainer would not be concerned about the brace changing ground reaction forces or possibly the movement technique.

Delimitations

This study was delimited by the following factors:

1. The subjects were sport active college students, aged 18 to 24 years, both males and females.

2. Only the Active Ankle Brace (Active Ankle Systems, Inc., Louisville, Kentucky) was used.

3. Ground reaction forces were measured as the subject stepped down from a height of 8.0 in.

4. Subjects were free of any ankle, knee, hip, and low back injury or abnormality during the last 6 months.

Limitations

This study was limited by the following factors:

1. The activities took place in a controlled laboratory setting, so the results may not be representative of sport-type activities.

2. Each participant wore his or her own style of shoe.

3. No adjustments were made in landing style used by the participants between the two conditions.

4. The sample used was an opportunistic sample.

Basic Assumptions

The following assumptions were made for this study:

1. The subjects chosen were representative of sport-active college-age students.

2. The Active Ankle braces used in this study were all made and designed equally.

Hypotheses

The following hypotheses were examined:

1. A significant difference was expected for peak impact force between the brace and no-brace conditions.

2. A significant difference was expected for vertical loading rate between the brace and no-brace conditions.

3. A significant difference was expected for maximum medial force between the brace and no-brace conditions.

4. A significant difference was expected for maximum lateral force between the brace and no-brace conditions.

5. A significant difference was expected for time to peak impact force between the brace and no-brace conditions.

Definition of Terms

Terms relevant to the understanding of this study are listed below:

1. Drop step: The action of stepping off a surface at any given height and landing on one foot.

2. Ground reaction force: The action of the ground pushing back toward the athlete in an equal and opposite direction to which the athlete is moving.

3. Peak impact force: In heel-toe walking, the maximum vertical force which occurs within the first 50 ms after touchdown (Nigg, Bahlsen, Luethi, & Stokes, 1987).

4. Shock attenuation: The reduction or absorption of forces in the body by active or passive mechanisms. Active mechanisms are proprioception, joint position, and muscle tone. Passive mechanisms are elasticity of bone, cartilage, soft

tissue, and synovial fluid (Gross & Nelson, 1988).

5. Sport active: Subjects who participated in a sport activity at least three times a week for 30 min per session.

6. Vertical loading rate: The rate at which maximum force occurs, measured in N/s.

CHAPTER II

REVIEW OF RELATED LITERATURE

The ankle is exposed to many forces during normal walking and sporting activities. The importance of the foot and ankle as a functional unit is obvious when one realizes that this unit allows man to walk upright in bipedal motion. Due to exposure to a variety of forces, it is also one of the most commonly injured joints in sport (Magee, 1987). The most common cause for ankle injuries is excessive inversion. This cause has been the focus of the majority of research on prevention of ankle injuries. Another cause for injury to the ankle are excessive ground reaction forces. However, these forces have not been extensively researched (Frederick, Clarke, & Hamill, 1984). In addition to the ankle, ground reaction forces have been noted as a cause of injury to the leg, hip, and lower back (McNitt-Gray, 1991).

Because of the potential for injury caused by ground reaction forces on lower extremity structures, research in the area of prevention is increasing. The key to prevention is to stop or reduce excessive forces before they reach the foot, the ankle, and the rest of the body. Current research has focused on shock-absorbing inserts for shoes, but nothing has been done to determine the ability of an ankle brace to provide shock attenuation properties to help reduce the influence of ground reaction forces on the body (MacLellan, 1984).

Anatomy of the Ankle

The ankle joint is commonly thought of as two separate joints functioning together: (1) the actual ankle joint, or talocrual joint, which can be classified as an uniaxial, modified-hinge, synovial joint formed by the medial malleolus, the lateral malleolus, and the talus; and (2) the subtalar joint, which is formed by the talus and calcaneus. At the ankle joint, the malleoli extend distally forming a mortise into which the talus is housed. The medial malleolus is slightly more proximal than the lateral malleolus, projecting halfway down the talus. The lateral malleolus projects the entire length of the talus, thus providing greater bony stability on the lateral side (Magee, 1987). The talocrual joint is designed for stability, not for mobility, allowing movement only in the sagital plane. Plantarflexion and dorsiflexion occur with normal ranges of motion of 50^{o} and 20^{o}, respectively (Magee, 1987).

At the subtalar joint, movement between the calcaneus and the talus occurs around an oblique axis that extends anteromedially from the neck of the talus to the posterolateral portion of the calcaneus (Donatelli, 1990). According to Bates (1979), in a non-weightbearing situation with the talus remaining stationary in the mortise, the subtalar joint is a simple, single-axis joint that acts as a mitered, oblique hinge in triplanar motion: dorsiflexion, abduction, and eversion for pronation, and plantar flexion, adduction, and inversion for supination. Gray (1993) advocates that the calcaneus is fixed in a closed kinetic chain, and that the talus slides over the calcaneus as it tilts in the ankle mortise from the weight of the body being placed upon it. Subtalar joint pronation and supination are measured clinically by the amount of calcaneal inversion and eversion in an open kinetic chain. Brocato and McPoil (1990) identified ranges of motion between 45^{o} to 60^{o} for inversion and 15^{o} to 30^{o} for eversion. They believed that the calcaneus becomes fixed in weightbearing and that the motion comes from the talus moving within the mortise. It is important to note that even though the mechanical systems necessary for supination and pronation are altered from a weightbearing to non-weightbearing conditions, calcaneal inversion and eversion are not affected.

Stability and Biomechanics

Stormont (1985) determined that stability of the weightbearing ankle is dependent on several factors, including the relationship of articulating surfaces, the orientation of ligaments, and the position of the ankle in time of stress. McCullough and Burge (1980) stated that muscle forces added to the dynamic stability of the weightbearing ankle.

During the initial phases of closed kinetic chain motion, or initial stance weightbearing, the talus tilts medially within the mortise as a result of calcaneal eversion to adapt to the surface below (Gray, 1993). The talus is restricted by many structures, such as the distal end of the lateral malleolus, the strong deltoid ligament on the medial aspect, and the many muscular structures, both medial and posterior, that cross the talus and eccentrically control pronation.

When the calcaneus is inverted during the latter phase of closed kinetic chain motion, supination occurs and the talus tilts laterally in the mortise. Because the medial malleolus does not extend distally as far as the lateral malleolus, the bony stability on the medial side is decreased, which encourages lateral talar tilt. The three collateral ligaments on the lateral side control talar tilt but are not as strong as the medial deltoid ligaments. The lateral side lacks the muscular stability that the medial side has, with only the peroneus longus and brevis to help eccentrically control calcaneal supination (Gray, 1993). A person in motion is more prone to an inversion type of injury than an eversion injury because the bony aspects are tight and the forces tend to go toward the weakest link, the lateral ligaments. In pronation the bones are in a loose pack position, and injuries are more to tendons as they become the primary stabilizers. This excessive inversion is the most common injury to the ankle in sports. There are many methods to prevent this from happening, but most prefer to use taping and bracing of the ankle to provide support.

Ground Reaction Forces

Because the number of injuries connected with sport activities is increasing, a better understanding of the importance of ground reaction forces may have a positive effect on the prevention of further injuries. Assuming that external forces are the cause of pain and injuries during sporting activities, one variable to investigate is the vertical ground reaction force being applied to the body. Research by Voloshin and Wosk (1982) provided circumstantial evidence that vertical ground reaction forces can have injurious effects on the body. Logic follows that research should be focused on trying to delay and/or attenuate the application of the vertical ground reaction force and help prevent the body's own natural shock absorbers from being overloaded.

Vertical ground reaction forces, commonly measured with force plates, often display two components, when recored graphically, a high frequency component in the first quarter of foot contact and a low frequency component in the latter three quarters of foot contact. Frederick, Hagy, and Mann (1981) used the name impact force to describe the high frequency component of the vertical ground reaction force, based on the consideration that it is an impulsive force resulting from the impact of the foot and ground. Nigg (1983) labeled this force as passive because peak force occurred 20 to 30 ms after foot strike, and its duration was shorter than the time required by the muscles for a reflex action. When a force occurs before the muscle can fire the force absorption must take place within the elastic component of the muscles, tendons, cartilage, and bone. Impact forces, or landing forces, can be expressed in mechanical terms in the following ways. The downward momentum of the body must be reduced to 0, and the change in momentum is related to Newton's second law. Impact forces are determined by what the subject did before contact with the ground. Three variables determine these impact forces: (1) the velocity at contact, (2) the effective mass at contact, and (3) the material properties of the damping elements (soft tissue, shoes, surface). The second, or low-frequency, component is referred to as the active force because it recruits active involvement of the musculo-skeletal system that caused these forces (Nigg, Denoth, & Neukomn, 1982). Active forces are mainly determined by the movement of the subject during and after foot contact.

Impulse waves are created in the lower extremity during walking due to impact of the heel with the ground. The impact generates transient stress waves that travel up the lower extremity and result in transient peaks in the forces across the articular surfaces at the ankle, knee, hip, and joints of the spine. Radin and Paul (1971) demonstrated that a relationship existed between excessive loading levels of these impulsive forces and articular cartilage damage and joint degeneration in the lower extremity. Morphological studies by Seireg and Gerath (1975) on animals seem to be compatible with the clinical experience with osteoarthritis. These studies have shown that it is the transient nature of impulsive forces that is degenerating the joints, and that cartilage may have a threshold in which damage is irreparable and progressive.

With an understanding of vertical ground reaction forces, it is worthwhile to try and understand what type of pain and/or injury one might expect due to impact or active forces. Nigg (1983) proposed a possible systematic grouping of forces and their related types of injury. Impact forces were thought to be responsible for chronic injuries such as: (a) fatigue fractures in bones, (b) cartilage damage in joints, (c) insertion problems in ligaments and capsules, (d) insertion tendonitis and shin splints in tendons, and (e) contusions in soft tissues. Active forces were thought to be responsible for the more acute injuries: (a) fractures in bones, (b) ruptures in ligaments, tendons, and muscles; and (c) blisters in soft tissue.

Clinical findings by Radin, Paul, and Rose (1972) suggested that a relationship existed between some changes in bones and joints with repetitive impulsive loading. The results support the idea that the repetitive loading during gait generates intermittent force waves that are propagated through the human locomotion system and attenuated by the shock absorbers. It was found that fatigue fractures, partial or complete, resulted from an inherent inability of a bone with normal elastic resistance to withstand stress applied in a rhythmical, subthreshold manner, without trauma (Radin et al., 1978).

Shock Attenuation

Because of the mechanics of the ankle, an inversion ankle sprain is not the only injury possible at the ankle joint. At contact the calcaneus begins to evert, unlocking the midfoot and allowing it to become a loose adapter and shock absorber (Donatelli, 1990). Relative increases in ground reaction forces may be present that could potentially cause trauma at the ankle joint and proximally up the kinetic chain or lower extremity. Dufek and Bates (1991) stated that if the joint cannot accomodate the load, an injury situation arises.

This path of force distribution exposes many structures to possible injury situations. Subcondral bone, cartilage, and soft tissue have all been identified as potential attenuators to these transient forces (Radin & Paul, 1970). If any of these structures lack the needed shock attenuation, then a potential injury situation arises. Research has shown that lack of sufficient shock attenuation is linked to degenerative changes in joints and low back pain (Voloshin & Wosk, 1982). MacLellan (1984) suggested the use of shock-absorbing inserts for the athletes' shoes to reduce the excessive force present at the foot. With braces on the market that help protect from inversion injuries and/or those that help to prevent the amount of shock, the amount of room in the shoe is greatly reduced if there is an attempt to use both simultaneously.

Stability Braces

There are several braces on the market today that are designed to oppose excessive motion in the frontal plane and provide stability for the ankle joint. Some of those reported in research are the Swede-O-Universal, the Aircast, and the Active Ankle brace. The Aircast and the Active Ankle are of similar design.

Aircast and Active Ankle

Gehlsen et al. (1991) conducted research using both the Active Ankle and the Aircast braces. They reported that the Aircast and Active Ankle are both designed to contour to the medial and lateral aspects of the ankle, with strapping that connects the two sides around the anterior and posterior aspects as well as across the bottom of the foot. Both braces are made of rigid material that acts much like a splint to the ankle joint limiting excessive inversion and eversion. The braces are open in the anterior and posterior aspects to allow for movement of the ankle in plantar and dorsiflexion. A difference was found between these two braces in that the Aircast had significantly reduced the amount of dorsiflexion, the Active Ankle was found to have a range of motion (ROM) close to that of wearing no brace, in both plantar and dorsiflexion (Gehlsen et al., 1991).

Swede-O-Universal

The Swede-O-Universal brace is very similar to a tape support. When fitted properly, it surrounds the ankle, locking the calcaneus in place. It looks like a sock, but it has more restrictive material on the medial and lateral sides. Although the Swede-O-Universal brace gives support and control of the calcaneus, it is also the most restrictive in terms of plantar and dorsiflexion (Gehlsen et al., 1991). Decreasing plantar and dorsiflexion can possibly hinder performance (Robinson, Frederick, & Cooper, 1986).

CHAPTER III

PROCEDURES

Introduction

The purpose of this study was to determine the effect an ankle brace, the Active Ankle brace, worn for medial/lateral stability has on ground reaction forces. Specifically ground reaction forces present while performing a step down from a height of 8.0 in. with and without the ankle brace were compared. This chapter was organized into five areas: (1) subjects, (2) instrumentation, (3) pilot studies, (4) testing procedures, and (5) data analysis.

Subjects

Male and female students (N=50) from Western Michigan University were used in this study. The subjects were opportunistically selected from a group of volunteers meeting the following criteria:

1. The subjects were sport active and participated in activity at least three times a week for a duration of 30 min or more each time.

2. Subjects were also free of any ankle, knee, hip, or low back injury, or any orthopedic abnormality during the past 6 months.

All subjects received oral and written instructions explaining the extent of their participation prior to signing an informed consent statement. Subjects' rights were protected as required by the WMU Human Subjects Institutional Review Board (HSIRB). Appendix A contains the letter of approval from the HSIRB. Appendix B contains a copy of the consent form.

Instrumentation

Active Ankle Brace

The Active Ankle brace was used during testing. Sizes ranged from extra small to large depending on the subject's shoe size. The Active Ankle brace is described as an ankle support that has semi-rigid supports on the medial and lateral sides of the ankle and lower leg and is hinged at the ankle joint to allow for full range of plantar and dorsiflexion. The brace is secured around the ankle and lower leg by three velcro straps, which allows for a custom fit to the individual.

Step Box

The step box was 8.0 in. high, 12 in. wide, and 18 in. long. This was to reproduce the normal stepping motion that occurs while descending stairs. The step box provided a method that allowed the investigator to accurately control the drop step height. The step box was placed 4.0 in. from the force plate to reproduce a natural step and to eliminate the tendency of the subject to reach.

Force Plate

The Kistler Type 9281B force plate was used to collect the ground reaction force data (Kistler Instrument Corp., Amherst, NY). A Kistler 9861A amplifier provided the appropriate signal amplification and range setting. The ADIU 16 (analog to digital unit) was interfaced to a DT 2821 analog to digital board (ATDB). The ATDB was connected to an ESU 4000D event synchronization unit. The ESU 4000D was used to trigger the above interfaced equipment in data collection. The interfaced equipment was connected to a Tenex 486 DX-2 computer running the Peak 5.2 Analog Sampling Module Software (Peak Performance, Inc., 1994).

Pilot Studies

The first pilot study involved 1 subject starting from a stationary position atop an 8.0 in. step, stepping with the dominant foot onto the force plate and then stepping with the nondominant foot off the force plate. This study consisted of 10 trials for two conditions: (1) brace and (2) no-brace. The results from this study indicated the inability of a subject to step down from a stationary starting position with consistency in the degree of muscle control in the stance limb. Descriptive data for this pilot study can be found in Appendix C.

In the second pilot study, the ability of a subject to walk with an approach, step with the dominant foot onto the force plate, and continue walking was investigated. Procedures manipulated in this study included: (a) the number of steps prior to stepping onto the force plate, (b) the length of a step, and (c) the cadence or rhythm of the gait. The results from this study were:

1. 10 steps prior to stepping onto the force plate allowed an appropriate time for the subject to establish the correct cadence.
2. The length of the step was approximately 2.5 ft.
3. Cadence was best maintained by using a metronome.

In the third pilot study the researcher investigated the following procedures: (a) stepping onto an 8.0 in. step on the 11th step and then stepping onto the force plate with the dominant limb, (b) using a metronome set at 100 beats per minute (bpm) and 110 bpm, (c) the number of steps past the force plate in proper cadence, and (d) the effect of selecting 10 out of the 15 trials for analysis. This pilot study was conducted under two conditions: (1) brace and (2) no-brace. The results from this pilot study were:

1. The subject was able to step onto an 8.0 in. step and down onto the force plate in cadence with the metronome.

2. Within-subject variability for maximum peak force was less for 100 bpm compared to 110 bpm.

3. Five steps past the force plate assured the best maintenance of cadence as the subject passed over the force plate.

4. The selection of the 10 trials most alike with respect to maximal peak force represented the true gait pattern.

Descriptive data from this pilot study can be found in Appendix D.

Testing Procedures

Subjects reported to the biomechanics laboratory in the University Recreation Center, Western Michigan University, dressed in shorts/sweat pants and shirt of their choice. Socks were worn with low cut athletic shoes of their choice. Subjects were asked to "walk the hall" for a period of 5.0 min as a warm up. The 1st min was at a pace of 90 bpm, the 2nd min at 95 bpm, and the last 3 min at 100 bpm. When they returned, the subjects were randomly assigned to a testing condition, brace or no-brace.

For the brace condition the investigator applied the brace over the subject's sock on the dominant ankle. After the subject replaced his or her shoes, the investigator checked the fit and placement of the brace.

A metronome was set at a rate of 100 bpm, the cadence at which the subjects walked. Ten strides were estimated and marked on the floor to set up an approach for the subjects to follow to the step box. After the 10th stride the next two steps would be up onto the step box with the nondominant foot, and landing heel first on the force plate with the dominant foot. The subject then followed through and off the plate with at least five strides, keeping a pace of 100 bpm throughout. Subjects were told to walk with: (a) a normal gait pattern, (b) the metronome, and (c) a heel-strike landing. If the subject or the investigator noticed any reaching for the step box or force plate, the trial was repeated. Each subject completed 30 trials, 15 with the ankle brace and 15 without the ankle brace.

Data Analysis

Ground reaction forces were collected on each trial for all subjects. Force data on 10 of the 15 trials for each condition for each subject were used in the analysis. Any trial with an extreme peak loading force at either the high or low end of a subject's distribution of trials was eliminated. The rationale for eliminating these trials was to control for the extreme variance found in any human motion. Thus, the 10 remaining trials were a better representation of the subject's typical gait pattern.

The statistical analysis performed was an intraclass correlation measuring reliability of the trials for each dependent variable. Calculation of the correlation was done using the results of a randomized block factorial ANOVA design with brace and no-brace conditions, across 10 trials for each dependent variable (Kirk, 1968). The ANOVA's were then used to determine the significance of the findings at the .05 level. The dependent variables measured were: (a) the peak impact force, defined as the maximum force in the vertical direction (Fz); (b) vertical loading rate, defined as the rate at which maximum force occurred; (c) maximum medial force; (d) maximum lateral force; and (e) time to peak force, defined as the point at which vertical impact is maximum or the accumulation of time from heel strike to when maximum vertical impact was reached. Software used for the statistical analysis was BMDP (Dixon, 1990).

CHAPTER IV

RESULTS AND DISCUSSION

This study was undertaken to determine if the ground reaction forces present while wearing an Active Ankle brace were less than while not wearing the brace during a drop step from an 8.0 in. step. The purpose was to provide health care professionals with information to aid in understanding the effects an ankle brace will have on ground reaction forces.

Recently researchers have failed to investigate the effects braces have on ground reaction forces. With a current increase in the use of ankle braces to support a weakened or injured joint, informing the user and practitioner of these effects will allow for a more educated use of the braces on the market today.

Results

A randomized block factorial ANOVA design brace/no-brace and trials, were calculated for each dependent variable (Kirk, 1968). The dependent variables measured were: (a) peak impact force, (b) vertical loading rate, (c) maximum medial force, (d) maximum lateral force, and (e) time to peak force. Male and female ($\underline{N}$=50) subjects participated in this study. Descriptive data for the means and standard deviations for the dependent variables can be seen in Table 1.

Table 1

Means and Standard Deviations for Dependent Variables

Dependent Variable	M	SD
Peak Impact Force (N)		
Brace	637.052	321.480
No-brace	647.172	312.089
Vertical Loading Rate (N/s)		
Brace	205.032	111.161
No-brace	212.502	115.285
Maximum Medial Force (N)		
Brace	59.929	31.888
No-brace	57.761	30.908
Maximum Lateral Force (N)		
Brace	65.166	31.104
No-brace	67.996	35.918
Time to Peak Force (ms)		
Brace	129.838	164.323
No-brace	112.142	146.970

Reliability

A two-way nested ANOVA-random model developed by Feldt and McKee in 1958 was used for determining the intraclass correlation coefficients or R (Safrit, 1976). The estimation of reliability by the use of the ANOVA procedure is preferable to the estimation using the product-moment correlation technique. Two major advantages exist with this approach:

1. The magnitude of the sources of variability that are of interest to the investigator can be examined.

2. Several intraclass correlation coefficients, used to estimate reliability and objectivity, can be computed from the same set of data on the basis of wanted and unwanted sources of variance (Safrit, 1976). The R has been shown to be a biased estimator unless the number of subjects is substantial. The R for the five dependent variables peak impact force, maximum lateral force, maximum medial force, vertical loading rate, and time to peak force, were greater all than .70. The specific intraclass R for each variable can be found in Table 2.

Table 2

R for Dependent Variables

Dependent Variable	R
Peak Impact Force (N)	.76
Vertical Loading Rate (N/s)	.86
Maximum Medial Force (N)	.95
Maximum Lateral Force (N)	.90
Time to Peak Force (s)	.72

ANOVA

Repeated measures design ANOVA was calculated for the main effect.

Peak Impact Force

The peak impact force in heel-toe walking was defined as the maximum vertical force that occurs within the first 50 ms after touchdown. The ANOVA for peak impact force (see Table 3) indicated the following:

1. A significant difference was found among subjects for peak impact force, ($F[49, 931] = 36.34$, $p < .05$).

2. No significant difference in peak impact force was found between the brace and no-brace conditions, ($F[1, 931] = 0.70$, $p < .05$).

3. No significant difference in peak impact force was found across the 10 trials, ($F[9, 931] = 0.84$, $p < .05$).

4. No significant interaction effect, brace by trial, was found, ($F[9, 931] = 0.86$, $p < .05$).

Table 3

ANOVA Summary Table for Peak Impact Force

Source	SS	df	MS	F
Subjects	64,744,068.96	49	1,321,307.53	36.34*
Treatments				
Braced	25,603.60	1	25,603.60	0.70
Trials	275,902.64	9	30,655.85	0.84
B x T	280,832.74	9	31,203.64	0.86
Residual	33,847,185.52	931	36,355.73	

*Significant at $p < .05$.

Vertical Loading Rate

Vertical loading rate was defined as the rate at which maximum force occurred, measured in N/s. The ANOVA for vertical loading rate (see Table 4) indicated the following:

1. A significant difference was found among subjects for vertical loading rate, (F[49, 931] = 22.73, p < .05).

2. No significant difference in vertical loading rate was found between the brace and no-brace conditions, (F[1, 931] = 2.04, p < .05).

3. A significant difference in vertical loading rate was found across the 10 trials, (F[9, 931] = 18.01, p < .05).

4. No significant interaction effect, brace by trial, was found, (F[9, 931] = 1.28, p < .05).

Table 4

ANOVA Summary Table for Vertical Loading Rate

Source	SS	df	MS	F
Subjects	7,450,227.39	49	152,045.46	22.73*
Treatments				
Braced	13,669.17	1	13,669.17	2.04
Trials	108,441.769	1	2,049.08	18.01*
B x T	77,318.24	9	8,590.92	1.28
Residual	6,228,954.72	931	6,690.61	

*Significant at p < .05.

Maximum Medial Force

The maximum medial force was defined as the force acting towards the midline of the body during gait. The ANOVA for maximum medial force (see Table 5) indicated the following:

1. A significant difference was found among subjects for maximum medial force, (F[49, 931] = 32.52, p < .05).

2. No significant difference in maximum medial force was found between the brace and no-brace conditions, (F[1, 931] = 3.05, p < .05).

3. No significant difference in maximum medial force was found across the 10 trials, (F[9, 931] = 1.67, p < .05).

4. No significant interaction effect, brace by trial, was found, (F[9, 931] = 0.82, p < .05).

Table 5

ANOVA Summary Table for Maximum Medial Force

Source	SS	df	MS	F
Subjects	601,810.82	49	122,810.85	32.52*
Treatments				
Braced	1,150.86	1	1,150.86	3.05
Trials	3,963.59	9	440.40	1.67
B x T	2,796.79	9	310.75	0.82
Residual	351,620.36	931	377.68	

*Significant at p < .05.

Maximum Lateral Force

Maximum lateral force was defined as the force that is acting away from the midline of the body during gait. The ANOVA for maximum lateral force (see Table 6) indicated the following:

1. A significant difference was found among subjects for maximum lateral force, ($F[49, 931] = 29.16$, $p < .05$).

2. A significant difference in maximum lateral force was found between the brace and no-brace conditions, ($F[1, 931] = 4.22$, $p < .05$).

3. No significant difference in maximum lateral force was found across the 10 trials, ($F[9, 931] = 0.72$, $p < .05$).

4. No significant interaction effect, brace by trial, was found, ($F[9, 931] = 0.96$, $p < .05$).

Table 6

ANOVA Summary Table for Maximum Lateral Force

Source	SS	df	MS	F
Subjects	678,117.49	49	13,839.13	29.16*
Treatments				
Braced	2,002.23	1	2,002.23	4.22*
Trials	3,078.29	9	342.03	0.72
B x T	4,110.23	9	456.69	0.96
Residual	441,877.22	931	474.63	

*Significant at $p < .05$.

<u>Time to Peak Force</u>

The ANOVA for time to peak force (see Table 7) indicated the following:

1. A significant difference was found among subjects for time to peak force, ($F[49, 931] = 12.60$, $p < .05$).

2. A significant difference in time to peak force was found between the brace and no-brace conditions, ($F[1, 931] = 5.00$, $p < .05$).

3. No significant difference in time to peak force was found across the 10 trials, ($F[9, 931] = 0.93$, $p < .05$).

4. No significant interaction effect, brace by trial, was found, ($F[9, 931] = 0.67$, $p < .05$).

Table 7

ANOVA Summary Table for Time to Peak Force

Source	SS	df	MS	F
Subject	9.280	49	.189	12.60*
Treatments				
Braced	.075	1	.075	5.00*
Trials	.127	9	.014	0.93
B x T	.089	9	.010	0.67
Residual	13.750	931	.015	

*Significant at p <.05.

Discussion

The purpose of this study was to determine whether a difference in selected ground reaction force parameters existed during brace and no-brace conditions. Significant differences were found between trials for vertical loading rate, and between the brace and no-brace conditions for maximum lateral force and time to peak force.

Significant differences were also found in the force data among subjects. This result is consistent with the results of a study done by Hamill, Knutzen, Bates, and Kirkpatrick (1986). Hamill et al. noted that individual subjects reacted somewhat differently to the application of a brace. No patterns across conditions for any of the variables were obvious. Therefore, difficulty arises when trying to describe a "normal" reaction of an individual to the ankle brace. This can be explained by the fact that no two people have the same gait pattern.

The mediolateral ground reaction component was defined by Hamill et al. (1986) as the shear force exerted parallel to the running surface and perpendicular to the direction of the movement. Liberia (1972) reported the purpose of the ankle brace was to support the ankle in inversion-eversion of mediolateral movement. Results from this investigation indicated that neither the brace nor the no-brace condition had any significant effect on the task except in the lateral direction. These data were in disagreement with data in studies by both Hamill et al. (1986) and McIntyre, Smith, and Denniston (1983); they found no significant differences. In this study, the researcher noted a statistically significant difference in maximum lateral force, with the mean for the brace condition being less than the mean for the no-brace condition, 65.166 N and 67.996 N, respectively. This was to be expected because the purpose of the brace was to limit or control the mediolateral component of movement. The brace would restrict the pronation of the foot during the contact phase, thus the lateral force would be less with the brace than without it.

A statistically significant difference in time to peak force was also noted between the brace and no-brace conditions. The time to peak force was associated with key events in the footfall that correspond to the collision of the foot on the ground. An explanation of this can be related to the idea that the ankle brace was a foreign body on the foot that altered the subjects' gait pattern. The mean time for the brace condition was greater than the mean time for the no-brace condition, 0.129 s and 0.112 s, respectively. This shows that there was an altering of the subjects' gait pattern. Possible reasons for this occurrence can be explained by the idea that subjects were keeping their center of gravity farther back with the brace condition, thus increasing the overall time to peak force. An explanation for the center of gravity being farther back was that the ankle brace was restricting in nature, not allowing for full ROM, and in turn altering gait. A second explanation could be the relationship between impulse and the force of impact. Without the brace, the force of impact is reduced by transferring the kinetic energy over a greater distance or ROM in the lower extremity: hip, knee, and ankle joints. When the ankle is braced, the ROM is restricted, requiring the subject to alter the gait pattern by slowing down (reducing velocity) and increasing the time to peak force. Thus, the subject uses impulse or lengthens the time over which a force is absorbed to reduce or maintain a consistent peak force. This explanation would justify why there was no significant difference in peak force between the brace and no-brace conditions. Another more functional reason for this occurrence was that the brace did not force the center of gravity back, but also did not allow for the foot to pronate at the proper time, thus delaying foot flat and peak impact force until a later time in the stance phase. These explanations also account for the significant difference found in vertical loading rate. Vertical loading rate was defined as the rate at which maximum force occurs, peak force divided by time to peak force. If time is one of the components of vertical loading rate, and if time is increased for the braced condition, while peak force remains the same for the brace and no-brace conditions, vertical loading rate is going to be less for the brace condition. The mean for the brace condition was less than the mean for the no-brace condition, 205.03 N/s and 212.50 N/s, respectively.

The investigator examined the test-retest reliability for performance on each variable using an intraclass correlation coefficient (Safrit, 1976). The intraclass correlation coefficient R for test-retest reliability for all dependent variables ranged between .72 and .95. These results suggest acceptable reliability for the following factors: maximum lateral force (.90), maximum medial force (.95), and vertical loading rate (0.86). The R for peak impact force (.76) and time to peak force (.72) were interpreted as moderate values. These values were probably affected more by a change of gait (time to peak force) than by any of the other dependent variables.

CHAPTER V

SUMMARY, FINDINGS, CONCLUSIONS, AND RECOMMENDATIONS

Summary

This study was undertaken to determine the effect an ankle brace worn for stability has on ground reaction forces. More specifically, it was the problem of this study to compare the ground reaction forces present when performing a step down from a height of 8.0 in. while wearing an ankle brace to the ground reaction forces present when not wearing the brace.

Male and female students (N=50) from Western Michigan University were used in this study. The subjects were opportunistically selected from a group of volunteers meeting the following criteria:

1. The subjects were sport active and participated in activity at least three times a week for a duration of 30 min or more each exercise session, and

2. Subjects were free of any ankle, knee, hip, low back injury, or other orthopedic abnormality during the past 6 months.

The brace investigated was the Active Ankle Brace, ranging in size from extra small to large depending on the subject's shoe size. The step box was 8.0 in. high to reproduce the normal stepping motion that occurs while descending stairs. The box provided a method that allowed the investigator to accurately control the drop step height. A Kistler Type 9281B force plate was used to collect the ground reaction force data. The interfaced equipment was connected to a Tenex 486 DX-2 computer running the Peak 5.2 Analog Sampling Module Software (Peak Performance, Inc., 1994).

Pilot studies were conducted to determine the correct use of equipment along with determining the proper cadence for the approach to the step box. A metronome was set at a rate of 100 bpm, the cadence at which the subject walked. Ten strides were estimated and marked on the floor to set up an approach for the subjects to follow. Each subject completed 30 trials, 15 with the ankle brace and 15 without the brace.

Force data on 10 of the 15 trials for each condition for each subject were used in the analysis. Any trial with extreme peak loading force at either the high or low end of the subject's distribution of the trials was eliminated. The statistical analysis performed was an intraclass correlation estimating reliability of the trials for each dependent variable. Calculation of the reliability coeffiecient was done using the results of a randomized block factorial ANOVA design with two independent variables, brace/no-brace, and trials (10) for each dependent variable. Dependent variables measured were (a) peak impact force, (b) vertical loading rate, (c) maximum medial force, (d) maximum lateral force, and (e) time to peak force. Software used for statistical analysis was BMDP (Dixon, 1990).

Findings

Significance for all findings of this study was determined at the .05 level. The ANOVA calculations indicated the following:

1. A significant difference was found among subjects for peak impact force, ($F[49, 931] = 36.34$, $p < .05$).

2. A significant difference was found among subjects for vertical loading rate, ($F[49, 931] = 22.73$, $p < .05$).

3. A significant difference was found among subjects for maximum medial

force, ($F[49, 931] = 32.52$, $p < .05$).

4. A significant difference was found among subjects for maximum lateral force, ($F[49, 931] = 29.16$, $p < .05$).

5. A significant difference was found among subjects for time to peak force, ($F[49, 931] = 12.60$, $p < .05$).

6. A significant difference in vertical loading rate was found across the 10 trials, ($F[9, 931] = 18.01$, $p <. 05$).

7. A significant difference in maximum lateral force was found between the brace and no-brace conditions, ($F[1, 931] = 4.22$, $p < .05$).

8. A significant difference in time to peak force was found between the brace and no-brace conditions, ($F[1, 931] = 5.00$, $p < .05$).

Conclusions

The following conclusions were made as a result of this investigation:

1. The ankle brace did not affect the peak impact force.

2. The ankle brace affected the vertical loading rate across the trials for the subjects.

3. The ankle brace had no affect on the maximum medial force.

4. The ankle brace affected the maximum lateral force by decreasing it for the brace condition.

5. The ankle brace increased the time to peak force.

Recommendations

Based on this investigation's research design and findings the following recommendations for further study were apparent:

1. Replicate the study, but control the type of shoes used by the subjects; test one brand and model for all subjects.

2. Include in the research design several different activities (i.e., running, jumping, falling from a controlled height, etc.) and compare ground reaction forces.

3. Use film analysis of the subject's gait to determine alterations in technique between conditions.

4. Use EMG to compare the muscle activity of the ankle's ROM for all conditions.

5. Look at impulse in all directions: vertical, medial, and lateral.

Appendix A

Human Subjects Institutional Review Board Acceptance Letter

Human Subjects Institutional Review Board

Kalamazoo, Michigan 49008-3899
616 387-8293

WESTERN MICHIGAN UNIVERSITY

Date: November 4, 1994

To: Andrew D. Howell

From: Richard Wright, Interim Chair Richard Wright

Re: HSIRB Project Number 94-10-01

This letter will serve as confirmation that your research project entitled "Shock attenuation in the active ankle brace" has been **approved** under the **expedited** category of review by the Human Subjects Institutional Review Board. The conditions and duration of this approval are specified in the Policies of Western Michigan University. You may now begin to implement the research as described in the application.

Please note that you must seek specific approval for any changes in this design. You must also seek reapproval if the project extends beyond the termination date. In addition if there are any unanticipated adverse or unanticipated events associated with the conduct of this research, you should immediately suspend the project and contact the Chair of the HSIRB for consultation.

The Board wishes you success in the pursuit of your research goals.

Approval Termination: Nov. 4, 1995

xc: Dawson, HPER

Appendix B

Informed Consent

APPENDIX B

Informed Consent

Western Michigan University
Department of Health, Physical Education, and Recreation

Principal Investigator: Dr. Mary Dawson
Research Associate: Andrew Howell, ATC

I understand that the purpose of this study titled "Shock Attenuation in the Active Ankle Brace" is to determine if the amount of shock absorbing ability present while wearing the Active Ankle Brace is less than when not wearing the brace. I understand that there are many braces which stabilize the ankle, however, most braces which increase stability sacrifice shock absorbing ability. The rationale for this study is to determine if the Active Ankle Brace provides shock absorbing ability when compared to the no-brace condition. I understand that this study is a requirement for Andy Howell to complete his master's thesis.

I understand that there are no direct benefits from participating in this study.

I understand that I will be wearing the brace while stepping from an 8 inch step and I will perform the same procedures without the brace. The research will involve about 60 minutes of my time. My participation time will include (a) a warm up and stretching period, (b) stepping down 10 trials with a brace on my dominant limb, and (c) stepping down 10 trials without a brace.

I also understand that there are no expected risks to my participation other than those associated with normal walking activities or with stepping down from a 8 in. high step. If any injury should occur, an athletic trainer will be present. As in all research, there may be unforeseen risk to the participant. If an accidental injury occurs, appropriate emergency measures will be taken; however, no compensation or treatment will be made available to the subject except as otherwise stated in this consent form.

I realize that during data collection I will be assigned a number to ensure that no one can associate my name with the results. I will be identified strictly by my number during the study and publications resulting from my data.

I understand that Andy Howell is the student investigator in this research and may be contacted at 387-2690 through the Health, Physical Education, and Recreation Department or at his home (372-2126). I can also reach Dr. Mary Dawson at 387-2711 in the Health, Physical Education, and Recreation Department if I have any questions regarding this study. The participant may also contact the Chair, Human Subjects Institutional Review Board (387-8293) or the Vise President for Research (387-8298) if questions or problems arise during the course of the study.

I understand that my participation in this study is voluntary and I may withdraw at any time without penalty.

I am covered by my own medical insurance, or failing that, accept full responsibility for any and all medical expenses I incur as a result of my participation in the activities involved in this study.

I, by signature, verify that I have freely consented to be a participant in this study. I have read this consent form and agree to its terms.

Signature Date

Appendix C

Descriptive Data for Stationary Step Down Pilot Test from 8.0 in. Step

APPENDIX C

Descriptive Data for Stationary Step Down (Impact)

	Braced	No-braced
Sample Size	10.00	10.00
Mean	642.40	649.00
Median	611.00	605.50
Mode	524.00	577.00
Geometric Mean	627.47	639.99
Variance	22728.90	13891.30
Standard Deviation	150.76	117.86
Standard Error	47.67	37.27
Range	420.00	377.00

Descriptive Data for Stationary Step Down (Time)

	Braced	No-braced
Sample Size	10.00	10.00
Mean	0.46	0.49
Median	0.45	0.54
Mode	0.45	0.37
Geometric Mean	0.41	0.43
Variance	0.04	0.05
Standard Deviation	0.21	0.22
Standard Error	0.06	0.07
Range	0.59	0.59

Appendix D

Descriptive Data for 100 bpm vs. 110 bpm

APPENDIX D

Descriptive Data for 100 bpm vs. 110 bpm

Variable	Mean	SE	SD	Variance
BIMP100	834.93	57.46	222.55	49,528.07
BIMP110	845.73	59.16	229.11	52,490.78
BTIME100	0.08	0.03	0.10	0.01
BTIME110	0.12	0.04	0.15	0.02
BVL100	26,640.07	1,285.09	4,977.13	24,771,775.50
BVL110	29,399.13	715.32	2,770.40	7,675,135.84
NBIMP100	664.47	62.82	243.30	59,194.84
NBIMP110	739.60	64.35	249.23	62,114.54
NBTIME100	0.24	0.05	0.20	0.04
NBTIME110	0.18	0.05	0.18	0.03
NBVL100	23,484.33	971.85	3,763.97	14,167,503.80
NBVL110	30,969.07	1,033.79	4,003.86	16,030,881.60

B = braced

NB = no brace

IMP = impact

TIME = time

VL = vertical loading

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