EFFECTS OF TWO MARKER PLACEMENT AND DATA ANALYSIS METHODS ON RUNNING GAIT ANALYSIS

by

JAMES N.M. BECKER

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Dr. Li-Shan Chou, Chair of the Examining Committee

3/8/2010

Date

Committee in Charge:

Dr. Li-Shan Chou, Chair
Dr. Louis Osternig
Dr. Stanley James

Accepted by:

Dean of the Graduate School

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Approved: ______________________________
Dr. Li-Shan Chou

This study evaluated the effects of two marker placement methods and two data analysis methods on running gait analysis. Markers placed on the shoe heel counter were compared with markers placed directly on the calcaneous and visible through heel windows cut into the shoe. When analyzed using a traditional group design no significant differences were found between marker conditions for rear foot eversion excursion, percent stance at which peak eversion occurred, maximal instantaneous eversion velocity, or maximal instantaneous vertical loading rate. Ankle frontal plane variability was significantly different between conditions. When analyzed with a single subject design some individuals demonstrated significant differences between conditions while others did not. In some individuals the heel windows condition revealed previously masked coupling parameters thought to be related to injury. The results of this study suggest the heel windows method and single subject analysis should be used for a longitudinal study of runners.

CURRICULUM VITAE

NAME OF AUTHOR: James Nicholas Macksoud Becker

PLACE OF BIRTH: Boston, MA

DATE OF BIRTH: 07/17/1979

GRADUATE AND UNDERGRADUATE SCHOOLS ATTENDED:

University of Oregon, Eugene, Oregon
Middlebury College, Middlebury, Vermont

DEGREES AWARDED:

Master of Science, Human Physiology, March 2010, University of Oregon
Bachelor of Arts, Geography, June 2002, Middlebury College

AREAS OF SPECIAL INTEREST:

Biomechanical contributors to running related injuries
Applying biomechanical analyses to improve sports performance

PROFESSIONAL EXPERIENCE:

Graduate Teaching Fellow, University of Oregon, Eugene, Oregon, 2009-2010

High School Science Teacher, Vergennes Union High School, Vergennes, Vermont, 2003-2007

Head Coach Track & Field and Cross Country Teams, Vergennes Union High School, Vergennes, Vermont, 2002-2007

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CHAPTER I

INTRODUCTION

Running Injuries: Incidence and Costs

Runners get injured at an incredibly high rate. Several review articles have reported injury rates of 24% to 75% for recreational and competitive runners over a one year period (Hreljac, 2005; Jacobs & Berson, 1986; Marti, Vader, Minder, & Abelin, 1988; Milani & Hennig, 2000; Van Ghent et al., 2007; Van Mechelen, 1992; Walter, Hart, McIntosh, & Sutton, 1989). A 1986 study of running injury incidence suggested there were approximately ten million regular runners in the U.S. (Jacobs & Berson, 1986). A similar number was reported in a 2004 sports participation report by American Sports Data, Inc. which concluded there were approximately 10.3 million runners in the U.S., defining a runner as someone who ran one hundred or more days per year. With injury rates between 24% - 75%, this suggests anywhere from approximately 2.5 to 7.5 million individuals suffer a running injury each year.

A study on the etiology of almost two thousand running injuries found that 31% of injured runners sought medical treatment, and that most cases required an average of 3.8 medical consultations (Marti et al., 1988). The same study also reported that 5% of injuries were serious enough to lead to an absence from work, with the average duration of absence being 10.1 days. Another prospective cohort study on running injuries found that 40.6% of injured runners sought medical treatment, and that 25% of these individuals had persistent symptoms up to three months after their first consultation (van Middelkoop, Kolkman, van Ochten, Bierma-Zeinstra, & Koes, 2006). Jacobs and Burton (1986) reported that 70% of the injured runners in their study sought medical attention for their injuries, with treatments ranging from muscle strengthening to orthotics to surgical intervention in several cases. Based on this epidemiologic data, while the exact cost of running injuries is unknown, the sheer number of individuals involved suggests it is not insignificant.

One of the more troubling conclusions from the epidemiologic studies study is that previously injured runners carried a high risk of sustaining another injury. For instance, Marti et al. (1988) found previously injured runners had 74% risk of sustaining a second running injury while Walter et al. (1989) found that following an initial injury, men and women were 1.69 and 2.35 times more likely to sustain another injury, respectively. Even with all the advancements in sports medicine over the intervening years these numbers have not changed as a prospective study on injury incidence among 844 recreational runners conducted by Taunton et al. (2003) found that 50% of subjects who reported an injury during their study also had sustained some form of prior running injury. Powell, Kohl, Caspersen, & Blair (1986) suggested three main reasons why a previous history of injury may increase the likely hood of a second injury. These included the previous injury not healing completely, the repaired tissue not functioning as well or having the strength of the original tissue, or the original fundamental cause of the injury was not addressed leading to re-injury upon resumption of activity. It is in addressing this third reason where a biomechanical analysis may prove particularly useful for the clinician and patient.

Retrospective Studies on Possible Causes of Running Injuries

In 1978 James, Bates, and Osternig (1978) published what has become one of the landmark papers on overuse injuries in runners. Based on injuries observed in Dr. James's clinic, the authors concluded that the causes of overuse running injuries fell into three broad categories: training errors, anatomic and biomechanical factors, and external factors such as shoe selection or training surface. Training errors and external errors are, in theory, fairly easy to control or modify since they can be identified through a careful examination of an individual's training plan. However, the anatomical and biomechanical factors contributing to injuries are much more difficult to identify. Often these injuries result from interactions between numerous parameters with no one injury directly caused by a specific anatomic or biomechanical factor. In the thirty years following James et al.'s article there has been a huge volume of literature produced investigating the contributions of anatomic and biomechanical factors to overuse running injuries. These studies have generally focused on three areas: anatomic and anthropometric variables, kinematic variables, and kinetic variables.

Anatomic and anthropometric variables that have been linked to running injuries include leg length discrepancies, femoral neck anteversion, varus or valgus alignment of the calcaneous relative to the fore foot and tibia, pes planus or pes cavus foot structure under static or dynamic conditions, squinting patellae, Q angle at the knee, and genu varus or valgus alignment at the knee (Bandholm, Boysen, Haugaard, Kreutzfeldt-Zebis, & Bencke, 2008; Bennett et al., 2001; James et al., 1978; Kaufman, Brodine, Shaffer, Johnson, & Cullison, 1999; Korpelainen, Orava, Karpakka, Siira, & Hulkko, 2001; Lysholm & Wiklander, 1987; Rauh, Koepsell, & Rivara, 2006; Reinking, 2006; Ryan, MacLean, & Taunton, 2006; Van Mechelen, 1992). However, there is no clear agreement in the literature since other studies have found no link between femoral anteversion, patella alignment, or a rear foot valgus alignment and running injuries (Walter et al., 1989). Further complicating the issue are several studies which have suggested there may not be any relationship between measures of static lower limb alignment and running injuries (Lun, Meeuwisse, Stergiou, & Stefanyshyn, 2003; Wen, Puffer, & Schmalzried, 1997, 1998). The differing findings in these studies leave no clear picture on exactly how specific anatomic or anthropometric variables may predispose and individual to overuse running injuries.

A lack of flexibility or range of motion has been suggested to contribute to overuse running injuries (James et al., 1978; Kaufman et al., 1999). However, as with the anthropometric variables, this too does not find unanimous support in the literaure. Some authors have found that runners who stretch regularly actually experience more injuries than those who do not (Jacobs & Berson, 1986). Other authors have found that individuals who stretch intermittently or infrequently to be at a higher risk than those who either stretch regularly or never (Walter et al., 1989). Still other authors have found that individuals in the most and least flexible quartiles to be more injury prone than those in the middle quartiles (Jones & Knapik, 1999). Even though stretching before and after running is commonly suggested for injury prevention, at this time there exists no empirical experimental studies to support this claim (Hreljac, 2005).

In addition to anatomic and anthropometric factors, numerous studies have focused on the potential relationship between kinetic variables overuse injuries in running. These studies have generally focused on the impact forces imparted to the runner as the foot makes contact with the ground. For a traditional heel striking runner, a plot of the vertical ground reaction force contains two peaks, an impact peak corresponding to the impact of the foot and ground, and an absolute peak generated by muscular activity during push off. The impact peak is usually 1.5 to 2 times body weight while the active peak can range up to 3 or more times body weight, depending on the velocity of the runner (Cavanagh & Lafortune, 1980). Despite being the smaller of the two, the impact peak has received the bulk of the attention in studies relating kinetic parameters and overuse injuries.

As with the anthropometric and alignment parameters, there is disagreement over exactly how, or if, specific kinetic parameters are related to running injuries. Some studies have reported higher instantaneous vertical loading rates in individuals who have previously sustained at least one tibial stress fracuture (Hreljac, Marshall, & Hume, 2000; Milner, Ferber, Pollard, Hamill, & Davis, 2006). Other studies have implicated the twisting torque between the foot and the ground, a parameter which is reportedly higher in individuals who have sustained at least one stress fracture as a potential mechanism. (Holden & Cavanagh, 1991; Milner, Davis, & Hamill, 2006). This higher twisting torque in previously injured individuals was reported in a study examining retrospective predictors of tibial stress fracture in female runners (Pohl, Mullineaux, Milner, Hamill, & Davis, 2008). However, this same study also found that maximal instantaneous vertical loading rates were not a significant predictor for previous stress fracture history. A retrospective study by Pohl, Hamill, & Davis (2009) found individuals with a history of plantar fasciitis demonstrated greater peak vertical loading rates than individuals who had never experienced that injury, suggesting kinetic factors can affect soft tissues as well as bone.

However, there are also studies suggesting impact forces are not related to running injuries. For instance two retrospective studies, one in men and one in women, found that higher peak impact forces was not predictive of a history of stress fracture for either sex (Bennell et al., 2004; Crossley, Bennell, Wrigley, & Oakes, 1999). Further complicating the issue are two studies by Nigg (1997; 2001) which found that not only were impact forces not related to the incidence of running injuries, but that runners with larger vertical loading rates tended to have fewer injuries than those with lower vertical loading rates. So, as with the anthropometric parameters, the exact relationship of impact forces to injury in running is not clear.

Lower Limb Kinematics During Running

Even though anthropometrics and kinetics in running have received significant attention, it seems as if the volume of studies on kinematics during running is considerably larger. Out of all the studies on limb kinematics, by far the most heavily investigated parameter has been rear foot motion. During running, the typical impact force with the ground is around 1.5 to 2 times body weight, though this increases with increased running speed. (Cavanagh & Lafortune, 1980). One of the functions of the foot during this time is to absorb and dissipate some of that impact force. A runner generally strikes the ground with the rear foot in a slightly inverted position. The calcaneous then everts, a motion which "unlocks" the transverse tarsal joints allowing the foot to become flexible and act as a shock absorber (Hintermann & Nigg, 1998; Novacheck, 1988). Due to the tight articulation between bones of the foot this motion cannot happen in isolation and is therefore accompanied by dorsiflexion and abduction of the foot. This combination of rear foot eversion, forefoot abduction, and talocrural dorsi flexion is known as pronation.

While playing an important role, the foot is not the only shock absorbing mechanism in the body. The knees also flex during weight acceptance to help absorb some of the impact shock. However, the shape of the femoral condyles means the tibia must rotate internally with knee flexion and externally with knee extension. Since the foot is fixed to the ground and cannot abduct relative to the line of progression, the abduction component of pronation is accomplished through tibia internal rotation.

In theory therefore, maximal rear foot eversion and maximal knee flexion should occur at the same time. This synchrony appears to be supported by numerous studies of lower limb kinematics during running, a summary of which can be found in review article by DeLeo, Dierks, Ferber, & Davis (2004). The authors cited twelve studies which examined the timing of rear foot eversion and knee flexion, among other variables, and found maximum rear foot eversion occuring between 39.3% and 53.9% of stance and maximal knee flexion occuring between 36.0% and 45.3% of stance (DeLeo et al., 2004).

Relationship Between Rear Foot Kinematics and Running Injuries

Disruptions to this natural foot motion are thought to be related to numerous overuse running injuries. Generally, studies relating foot motion to injury have focused on three different aspects of rear foot motion including the total amount of rear foot motion, the velocity over which this motion takes place, and the relative timing of the motion in relation to other limb segments. Different overuse running injuries appear to be sensitive to different combinations of these three aspects. For instance, too much eversion has been cited as a contributing factor to Achilles tendon injuries due to the whipping nature imposed on the tendon by excessive calcaneal eversion (Donoghue, Harrison, Laxton, & Jones, 2008; Paavola et al., 2002). Too much eversion has also been cited as a contributing factor to plantar fasciitis for the stress it places on that tissue (Warren, 1984, 1990). Perhaps most commonly, excessive eversion is often cited as a contributing factor in the development of medial tibial stress syndrome, as it is thought that greater eversion places increased strain on the soft tissue structures of the lower limb. (Messier & Pittala, 1988; Reinking & Hayes, 2006; Tweed, Campbell, & Avil, 2008; Willems, Witvrouw, De Cock, & De Clercq, 2007).

The injuries described above occur in the soft tissue structures of the limb, with excessive eversion often cited as a contributing factor. Traditionally higher arches and smaller amounts of rear foot eversion have been associated with bony injuries such as stress fractures (Korpelainen et al., 2001; Williams, McClay, & Hamill, 2001). However, this too is not always true. For instance, a retrospective study on tibial stress fractures in female runners found rear foot eversion excursion to be one of the strongest predictors of injury, with injured subjects having greater eversion excursions than non-injured subjects (Pohl et al., 2008). Based on this body of literature it appears excessive rear foot eversion can contribute to injuries to both the soft tissue and bony structures of the lower limb.

Other studies investigating the relationship between rear foot eversion and injury have concluded the actual amount of rear foot eversion may not be as important as the velocity of rear foot eversion. For instance, a study examining potential biomechanical factors contributing to patellar tendiopathy in female runners found that, compared to controls, injured runners had a higher maximal eversion velocity, but similar overall eversion excursions (Grau et al., 2008).

Other authors have suggested that it is not just the amount or velocity of rear foot eversion which matters, but how rear foot motion is coupled with motion at other joints and segments, specifically the mid-tarsal joints and tibial rotation. As the rear foot everts the mid-tarsal joints unlock and the foot becomes more flexible. However, beyond mid-stance through the push off phase the foot needs to be a rigid level for an effective push off. If rear foot eversion is prolonged then this will not happen and the result will be a push off with a soft and flexible foot. Achieving an effective push off in this scenario requires more effort from the musculature of the lower limb placing greater stresses on the soft tissue structures, and potentially cause injury (James et al., 1978; McClay & Manal, 1998a; Novacheck, 1988).

An alternative hypothesis involves the relative timing of maximal eversion and maximal knee flexion. Due to the tight coupling between the rear foot, talus, and tibia, prolonged rearfoot eversion leads to prolonged internal rotation of the tibia. If the knee starts extending while the rear foot is still everted then a torsion stress will be placed on the tibia. This mistiming also increases stress on the soft tissue structures at the knee and has been hypothesized as a potential mechanism for knee injuries in runners (Stergiou, Bates, & James, 1999; Tiberio, 1987).

Measuring Rear Foot Motion

While it is not clear whether it is too much rear foot eversion, too great an eversion velocity, or the timing of rear foot eversion relative to the movement of other joints and segments, that contributes the most to injury, it is clear that rear foot eversion is an important component of any running gait analysis. Therefore, there have been numerous attempts to determine the most optimal measures to quantify rear foot eversion. Historically, with two dimensional analysis methods, rear foot eversion was quantified by two markers on the vertical bisection of the shoe and two markers on the vertical bisection of the lower limb (Edington, Frederick, & Cavanagh, 1990; McClay, 1995;

Novacheck, 1988). The relative movements and angles created between these lines provided an idea of rear foot motion. However, a two dimensional approach may not be optimal since it has been shown that camera alignment can seriously affect the resulting measurements (Areblad, Nigg, Ekstrand, Olsson, & Ekstron, 1990). For instance, changing the camera alignment only 2° can result in a 1° change in the measured parameter.

Additionally, a two dimensional approach would be susceptible to errors resulting from out of plane motions, which is especially concerning given the tri-planar nature of foot motion during running. One study mathematically estimated that inclining a segment 9° from the projected plane of motion could change measured joint angles by as much as 40% (Soutas-Little, Beavis, Verstraete, & Markus, 1987). One study on joint angles during walking suggested out of plane motion was only an issue during toe off, as rear foot motion was not significantly different during the first 60% of stance when measured with two dimensional or three dimensional approaches (Cornwall & McPoil, 1995). However, this is not true in all cases. For instance, some individuals may lack dorsiflexion at the ankle joint. In order to place their foot flat on the ground these individuals must have compensatory pronation. The main component of compensatory pronation will be forefoot abduction, resulting in increased out of plane motion. One study specifically examining differences in joint angles resulting from different levels of foot abduction found that as the abduction angles increased so did the differences between the two dimensional and three dimensional joint angles (McClay & Manal, 1998b). Taken as a whole these studies suggest that, when possible, a three dimensional approach should be used to minimize errors associated with out of plane motions occurring at the ankle joint.

Even when using a three dimensional approach, researchers are still left with a large obstacle in that the foot is enclosed within the shoe and therefore directly measuring the motion of the rear foot is difficult. There are numerous ways researchers have attempted to deal with this problem. One option is to place the markers directly on the heel counter of the shoe, and indeed this is what many authors have done. (McClay & Manal, 1998a, 1998b; Noehren, Davis, & Hamill, 2007; Pohl et al., 2008; Souza & Powers, 2009; Stefanyshyn, Stergiou, Lun, Meeuwisse, & Worobets, 2006; Walter et al., 1989). However, it is recognized that this marker set up is, by necessity, tracking the motion of the shoe, which may or may not be representative of foot bone motion. To explore these differences several studies have used intracortical bone pins to compare differences between rear foot motions as measured with shoe markers and with the bone pins (Reinschmidt, van den Bogert, Murphy, Lundberg, & Nigg, 1997; Stacoff et al., 2001). The authors found that while the patterns of motion were similar between the two marker methods, the magnitudes of motion were slightly different. While the differences were unique for each individual, the authors concluded that, in general, the shoe markers overestimated the true skeletal motion.

While intracortical bone pins may provide the truest measures of bone motion, they are not a practical method for conducting large studies on running kinematics. Therefore, researchers have explored other non-invasive methods which may provide reasonable estimates of rear foot motions. One such method would be using electrogoniometers to record changes in joint angles during running. While each study may use slight variations, the basic premise of this method involves fashioning a heel cup over the fat pad of the calcaneous and attaching it to a fixed anchor point on the lower limb. Eversion and inversion of the calcaneous turns a potentiometer located between the two attachment points, yielding the rear foot angle across the gait cycle. Several studies have used this method to examine rear foot kinematics during running. (Derrick, Dereu, & McLean, 2002; Milani & Hennig, 2000). While the results of these studies suggest this may be a reliable method for measuring rear foot eversion, the method cannot simultaneously measure angles in other planes of motion. Recording motion other than eversion-inversion requires additional electrogoniometers, which can quickly become bulky and cumbersome for the subject, potentially affecting their normal running mechanics.

In an effort to avoid external attachments yet still record accurate foot motion, some authors have used modified running shoes or sandals in their studies. For instance, O'Connor and Hamill (2004) used a specially constructed shoe with an intact sole and no upper or heel counter to examine the role of various foot muscles during running. The shoe was held in place with strong elastic over the dorsum of the foot and an attachment to a band around the tibia which ran over the medial and lateral malleoli. This custom shoe design has been used in other studies as well (MacLean, McClay Davis, & Hamill, 2006; Snyder, Earl, O'Connor, & Ebersole, 2009). Other authors have used specially designed running sandals which allow marker attachment directly the foot to examine kinematics, kinetics, and forefoot-rearfoot coupling patterns during running gait (Eslami, Begon, Farahpour, & Allard, 2007; Morio, Lake, Guegen, Rao, & Baly, 2009; Mundermann, Nigg, Humble, & Stefanyshyn, 2003; Nawoczenski, Saltzman, & Cook, 1998).

While these methods allow easy placement of markers directly on the foot, there are several issues preventing their widespread adoption. Firstly, custom shoes are both costly and not readily available, and their specificity precludes their adoption on a wide scale, especially for a laboratory which may work with a variety of subjects. Given the variation in foot types, foot sizes, and running shoe history, a complete set of custom shoes seems prohibitive. Secondly, there are issues concerning the external validity of any study which examines the mechanics of running without using actual running shoes, specifically the extent to which the observed mechanics in these studies are representative of the subject's actual mechanics. The authors of all these studies reported that subjects ran "comfortably" or "normally" in the sandals or specialized shoes however, no actual assessments were made of the subjects running mechanics in standard running shoes. This is an especially important point considering both the custom shoes and lacked a heel counter. Though a study by Van Gheluwe, Tielemans, & Roosen (1995) suggested rear foot motion may be independent of heel counter stiffness, it is traditionally thought the heel counter is an important component of running shoes which aids in rear foot movement control (Edington et al., 1990; Stacoff & Luethi, 1986).

In an attempt at trying to match standard enclosed running shoes with a more accurate marker placement, some authors have cut holes in the heel counter of the shoe, effectively enabling them to use a traditional gait analysis marker set up but place the markers directly on the rear foot. It is thought this method is one way to get more accurate kinematic data. Initially, this method was done first with two dimensional measuring techniques (Nigg, 1986; Reinschmidt, Stacoff, & Stussi, 1992; Stacoff, Reinchmidt, & Stussi, 1992). More recently, it is has been adopted for use with three dimensional analysis and has been used extensively in the Motion Analysis Laboratory and Running Injury Clinic at the University of Delaware (Butler, Davis, & Hamill, 2006; Dierks, Manal, Hamill, & Davis, 2008; Stackhouse, McClay Davis, & Hamill, 2004; Williams, McClay Davis, & Baitch, 2003; Williams, McClay Davis, Scholz, Hamill, & Buchanan, 2004; Williams, McClay, Hamill, & Buchanan, 2001). Given the relative importance of the heel counter in stabilizing the shoe and controlling foot motion, it is unclear exactly how holes in the shoe would affect the measured parameters. If the holes do not unduly influence rear foot motion, then this may be a method which yields a more accurate estimate of rear foot motion during running and potentially helps clarify the differences between shoe motion and rear foot motion during running.

So far only two studies have examined this issue in any depth. Both studies compared rear foot eversion as measured with both heel windows and shoe mounted markers (Nigg, 1986; Stacoff et al., 1992). Their results were similar to the bone pin studies, and the authors concluded that while the patterns of motion are similar, the shoe markers overestimate rear foot motion compared to the heel markers. Interestingly, the difference in motion between the two marker conditions was related to the size of the holes. As hole size increased, the differences between shoe motion and rear foot motion increased as well. The only three dimensional study examining the effects of hole windows looked at their effect on heel counter stability, concluding that small windows resulted in only a 10% decrease in heel counter stiffness (Butler et al., 2006). However, to date, no study has used a three dimensional approach to examine the effects of heel counter windows on rear foot kinematics and kinetics.

Purposes and Hypotheses of the Study

The current disagreement in the literature over the relationship between various anthropometric, kinematic, and kinetic parameters and running injuries presents several important considerations for designing a prospective, longitudinal study. First, it is clear any one piece of information by itself most likely will yield an incomplete picture or fail as an assessment of injury potential. Therefore, a prospective, longitudinal study should attempt to include as many of these puzzle pieces as possible. Secondly, the disagreement among studies suggests it may be important to consider each runner as an individual, rather than as a member of a group. The variability inherent in the group may wash out the significance of potential injury markers which can be identified in a single individual. Lastly, the lack of agreement reinforces the importance of using methods which provide valid, reliable measures of the parameters of interest.

In light of these suggestions, the purpose of this study was to use three dimensional analysis techniques to examine the effects of heel counter windows on kinematic and kinetic measurements during running gait. Specifically, this study sought to examine differences between shoe mounted markers and heel window markers on common measurements thought to be related to overuse running injuries. The measurements chosen for analysis were the time to maximum rear foot eversion, the percent stance at which maximum rear foot eversion occurs, maximal rear foot eversion excursion, maximal instantaneous eversion velocity, and the maximal vertical loading rate during heel impact in the stance phase of the gait cycle. An additional measurement included the variability of joint angle curves under each marker condition. A secondary purpose of this study was to be able to make a conclusive recommendation on which marker system should be used for a planned future longitudinal study on overuse running injuries.

Given the bone pin studies and previous two dimensional studies which indicated shoe markers overestimated true rear foot motion, it was hypothesized in this study that the maximal rear foot eversion excursion would be less with the heel windows marker set than with the shoe mounted marker set. It was also hypothesized that the time to maximum eversion excursion and the percent stance at which maximum eversion excursion occurred would remain unchanged between the two marker conditions. If the amount of eversion excursion decreases, but the amount of time and percent stance at maximal eversion does not change, then rear foot eversion velocity should be smaller in the heel windows condition. Therefore it was hypothesized that the heel windows marker condition would indicate a higher rear foot eversion velocity compared to the shoe based marker condition. Since the amount of rear foot eversion excursion should be less in the heel windows condition, it was hypothesized that the maximal instantaneous vertical loading rate would be larger in the heel windows condition compared to the shoe mounted marker condition. Lastly, it was hypothesized that foot motion would be more variable in the heel windows condition compared to the shoe markers condition.

CHAPTER II

METHODS

Subjects

Thirteen recreational runners from the University and local communities were recruited for this study. Individuals were recruited through personal correspondence, or referral from individuals familiar with the goals of the study. All subjects reviewed and signed the informed consent form which had been approved by the University of Oregon institutional review board prior to participating in the study. A copy of this form can be found in Appendix A.

The main inclusion criterion for this study was status as an active runner, which for this study was classified as running 20 or more miles per week. The second main inclusion criterion was that the individual was currently healthy and not suffering from any musculoskeletal injuries. A previous history of musculoskeletal injury did not preclude a subject from participating in the study.

Experimental Instruments

Dynamometer

Isometric maximal voluntary torque generation was measured using a BioDex System 3 dynamometer (Biodex Medical Systems, Inc., Shirley, NY).

Force Plates

Three AMTI (Advanced Mechanical Technologies Inc., Watertown, MA) force plates were used to collect ground reaction forces and moments. The force plates were located in series in the center of the 5 meter runway, sampling at a rate of 1000 Hz.

Motion Capture System

An eight camera motion capture system was used to record three dimensional (3D) marker trajectories, with the sampling rate set to 200 Hz. The motion capture system was calibrated prior to each subject's testing session as per the manufacturer's instructions.

Data Collection and Experimental Procedures

Muscle Strength Measurements

Subjects were tested bilaterally for maximal isometric torque generated in hip flexion, hip abduction, ankle plantar and dorsi flexion, and ankle inversion and eversion. It is recognized that there were multiple muscles involved in each contraction and no attempt was made to separate out individual muscles since it was thought that the muscles would act as a group during running scenarios. These specific muscles and motions were selected based on their importance for running and their potential link to running injuries. For instance, several studies have highlighted the importance of the ankle musculature in controlling the motion of the ankle and foot, their role in stabilizing the foot during stance, and their contributions to generating force. (Christina, White, & Gilchrist, 2001; Kibler, Goldber, & Chandler, 1991; Lun et al., 2003; Reber, Perry, & Pink, 1993; Scott & Winter, 1990). There has also been a large volume of literature published investigating the relationship between hip muscle strength and overuse injuries, especially at the knee (Grau et al., 2008; Ireland, Willson, Ballantyne, & McClay Davis, 2003; Noehren et al., 2007; Powers, 2003; Souza & Powers, 2009).

Isometric torque of the hip flexors was measured with the subject standing next to the dynamometer. The resistance pad was placed approximately in the middle of the thigh and center of rotation on the dynamometer was aligned with the greater trochanter of the test leg when the subject was standing in an upright position. This upright position was considered to be 0° of hip flexion and the isometric test was performed at 30° of hip flexion. The subject was allowed to place their hands on the dynamometer and an additional supporting structure for balance but was instructed to maintain an upright posture, not bend at the waist, and attempt to isolate the hip flexors.

Isometric torque of the hip abductors was measured with the subject standing facing the dynamometer. The resistance pad was placed over the lateral aspect of the thigh, approximately mid-way between the lateral femoral condyle and the greater trochanter. The center of rotation on the dynamometer was aligned with the hip joint center estimated to be several centimeters lower than the anterior superior iliac spine in the sagittal plane. This upright position was considered to be 0° of hip abduction and the isometric test was performed at 10° of hip abduction. The subject was allowed to place their hands on the dynamometer for balance but was instructed to maintain an upright posture, not internally or externally rotate the leg, and attempt to isolate the hip abductors.

Isometric torque of the ankle dorsi and plantar flexors was measured with the subject in a seated position and the knee flexed approximately 20°. A supporting pad was placed under the popliteal area and a restraining strap was placed across the subject's waist. The isometric torque was tested with the ankle joint in a neutral position, as indicated by an estimated 90° angle between the foot and tibia. The center of rotation of the dynamometer was aligned with the talocrural joint as estimated by the center of the lateral malleolus.

Isometric torque of the ankle invertors and evertors was measured with the subject in a seated position and the knee flexed approximately 20°. A supporting pad was placed under the popliteal area and restraining straps were placed across the subject's waist and across the thigh. The subject was instructed to perform the motions only at the ankle joint and avoid any rotation at the knee. The center of rotation of the dynamometer was aligned with the approximated subtalar joint axis.

For all tests the subject was allowed to perform a familiarization trial. For the actual tests the subjects were asked to push as hard as possible against the resistance pad for 5 seconds. This procedure was repeated two more times for a total of three trials. The subject rested for 5 seconds between the trials.

Clinical Exam

A clinical exam was performed by Dr. Stan James, an orthopedic surgeon with extensive experience in evaluating running injuries. The exam is a modified version of one he uses in current practice as part of his examination of an injured runner, and has also been previously used in research studies on runners (Stergiou et al., 1999). The exam focuses on general structural alignment, joint mobility, and flexibility of the lower extremity. The complete examination can be found in Appendix B.

One portion of the exam classified the arch structure of the subject using a ratio of the height of the dorsum of the foot at 50% of the full foot length divided by the truncated foot length. Figure 1 illustrates where these measurements were taken. This method has been shown to have high reliability and validity in both 10% and 90% weight bearing positions (Williams & McClay, 2000). In a previous study sampling 200 individuals, the mean arch ratio was 0.316. Ratios lower than 0.274 are considered "low" arches, and ratios higher than 0.356 are considered "high" arches (Williams, McClay, Hamill et al., 2001).

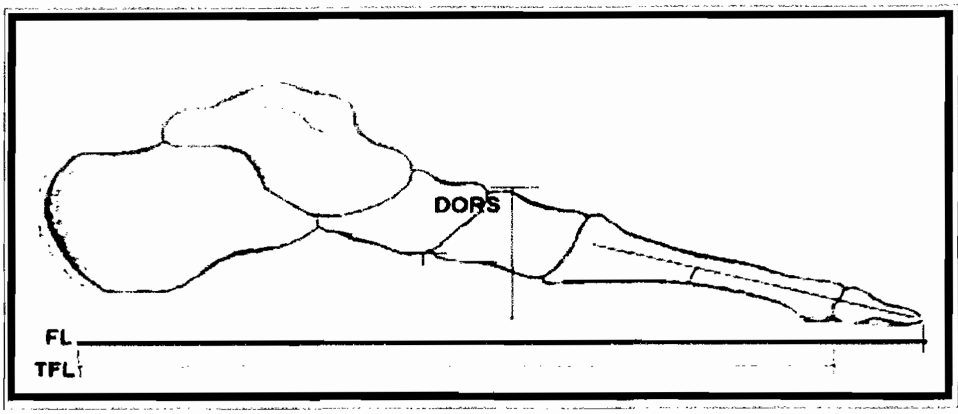


Figure 1. Schematic illustration of where the measurements were taken for the arch height index. The ratio is the height of the dorsum of the foot at 50% full foot length (FL) divided by the truncated foot length (TFL). The truncated foot length is measured from the most posterior aspect of the calcaneous to the 1st metatarsal-phalangeal joint. Image adapted from Willams & McClay (2000).

Motion Analysis

A total of 34 reflective markers were placed on bony landmarks of the subject, using a modified Helen Hayes marker set (Kadaba, Ramakrishnan, & Wootten, 1990). For the pelvis segment markers were placed on the sacrum midway between the posterior superior iliac spines, and bilaterally on the anterior superior iliac spines. Marker placement for the thigh, shank, and foot were similar bilaterally. For the thigh markers were placed on the medial and lateral femoral epicondyles and on the thigh collinear between the lateral femoral epicondyle and the greater trochanter. The anatomic coordinate systems for the thigh and shank were defined as per International Society of Biomechanics (ISB) recommendations (Wu, 2002). The thigh anatomic coordinate system was defined using the two femoral epicondyle markers, the hip joint center and the thigh marker, as per the ISB recommendations. The hip joint center was defined based on the anthropometric measurements of the subject (Vaughan, Davis, & O'Connor, 1999). The anatomic coordinate system for the shank shared the medial and lateral femoral epicondyle markers and had additional markers on the medial and lateral malleoli and a marker on the medial shank collinear with the medial malleolus and the medial femoral epicondyle markers. A static calibration trial was collected and the medial malleoli and medial femoral condyle markers were removed for the actual gait trials.

Since the subject wore shoes for both conditions, the foot markers were placed on the shoe. Subjects wore their own running shoes that they ran in every day. A toe marker was placed approximately between the first and second metatarsal shafts. For the markers on the shoe condition, three markers were placed on the heel counter of the shoe, one on the lateral aspect, and the other two along the vertical bisection of the heel counter, with the midpoint of these two markers level with the toe marker. This marker placement on the heel has previously been used to collect rear foot kinematic data (McClay & Manal, 1997, 1998a, 1998b; Noehren et al., 2007). The anatomic coordinate system was defined using the midpoint of the two heel counter markers, the toe marker, and the ankle joint center, as defined as the midpoint between the medial and lateral malleoli.

For the heel windows condition holes were cut into the heel counter of the shoe and the marker bases were attached directly to the shoe, in the same locations as the markers on the shoe condition. The markers were then screwed into the based from outside the shoe. The size of the holes was kept small, approximately 1.5 – 2.0 cm. in diameter since holes larger than this have been shown to affect rear foot movement (Stacoff et al., 1992). The same drill bit was used to cut each hole in an attempt to make them as similar as possible, however, small adjustments were made on a subject by subject basis. Previous studies using similar methodology have used an Instron materials testing device (Instron Corp., Norwood, MA) to examine the effect on heel counter stability and have found that the holes resulted in only a 10% decrement in heel counter rigidity (Butler et al., 2006; Stackhouse et al., 2004).

The placement of the anatomic and tracking markers are shown in Figure 2 and the heel windows marker set up can be seen in Figure 3. A detailed description of the methods used to define the anatomic and tracking coordinate systems for each segment can be found in Appendix C.

The running protocol involved the subject running laps in the Motion Analysis laboratory. Each lap was approximately 25 meters long. Data were collected as subjects passed through a 5 meter region in the center of the capture volume. Each subject ran at a self selected speed, though their speed was recorded using two photocells placed 5 meters apart. The three force plates were located in series in the center of the capture volume. The subject was instructed to return to approximately the same starting position after each lap to maintain consistency, however they were also instructed not to alter their stride to hit the force plates. Therefore, passes resulting in a clean force plate strike were used for both kinematic and kinetic data analysis, while those without a clean force plate strike were only used for kinematic analysis. Each subject completed between 30 and 40 passes under both conditions.

Data Analysis

Raw marker trajectories were identified using the EvaRT 5.0 motion capture software (Motion Analysis Corp., Santa Rosa, CA). The raw trajectories were low pass filtered using a Butterworth filter with a cutoff of 8 Hz. All trials were analyzed only for the stance portion of the gait cycle from heel strike to toe off. In cases where the trial was a clean force place strike these frames were identified by looking at ground reaction force curves. In cases where the trial was not a clean force plate strike these frames were identified based on visual analysis and the vertical coordinates of the heel and toe markers. The heel strike and toe off frames, and which force plate was cleanly hit, were manually recorded for each trial. The filtered marker trajectories and the analog photocell and force plate data were then exported as ASCII files which were used to calculated the joint angles.

Joint angles during the stance phase of gait were calculated using a custom LabView program (National Instruments, Austin TX). A detailed description of the methods used can be found in chapter seven of David Winter's biomechanics book (Winter, 2005). Three dimensional angles for the ankle, knee, and hip joints were determined using a joint coordinate system (Grood & Suntay, 1983). Cardan angles were used to define the joint angles across stance, referencing the movement of the distal segment to the proximal segment. For the hip and knee joints the order of rotations was ZXY, corresponding to flexion-extension, ab/adduction, and internal rotation. For the ankle joint the order of rotations was ZYX, corresponding to dorsi/plantar flexion, inversion-eversion, and ab/adduction. Zero degrees for all joint angles was assumed to be when the coordinate systems of the proximal and distal segments were aligned.

Once the joint angles across stance were calculated the discreete variables of interest were then identified, again using a custom LabView program (National Instruments, Austin, TX). These included several parameters which are often reported in studies on running kinematics including maximum rear foot eversion excursion, time to and percent stance at which maximum rear foot eversion occurs, the maximal instantaneous eversion velocity, the maximal instantaneous vertical loading rate from the ground reaction force, and a measure of the variability or repeatability of the joint angle curves between trials. Maximum rear foot eversion excursion was defined as the absolute difference between the rear foot angle at heel strike and the point of maximum rear foot eversion. The time to peak eversion referred to the time, in seconds, from heel strike to peak rear foot eversion while the percent stance of peak eversion referred to the percent stance at which peak eversion occurred. Maximal instantaneous rear foot eversion velocity was identified as the maximal point on the plot of the first derivative of the rear foot eversion curve. Maximal vertical instantaneous loading rate was determined as the maximal loading rate between ten and ninety percent of the vertical ground reaction force impact peak (Milner, Ferber et al., 2006; Williams et al., 2004; Williams, McClay, Hamill et al., 2001). The variability and repeatability of the joint angle curves were assessed using a coefficient of multiple correlation (CMC) (Kadaba et al., 1989). A CMC value of one indicates the joint angle curves are identical each trial and the lower the CMC value the more variability in the kinematic patterns.

For each subject, the left and right feet were analyzed separately. Though the limbs belong to the same subject, several studies have demonstrated some level of kinematic and kinetic asymmetry during running is common in most individuals (Vagenas & Hoshizake, 1992; Zifchock, Davis, & Hamill, 2006). Thus, it was thought each the left and right feet of each subject may not respond in the same manner to the two marker conditions. Additionally, since overuse injuries tend to occur in one limb, not both at the same time, it has been hypothesized larger levels of asymmetry might predict or predispose an individual to injuries on one side of their body and not the other (Zifchock, Davis, Higginson, McCaw, & Royer, 2008). Therefore, this study choose to examine analyze each subject's left and right foot independently.

Statistical Analysis

A minimum of five trials per subject were averaged to create an ensemble average for that individual for all the parameters of interest. Dependent observations *t* tests were used to compare differences between marker conditions for all dependent variables. To investigate the possibility of changes being running speed dependent a dependent observations *t* test was used to compare average running speed from the shoe markers condition and the heel windows markers condition. The significance level for all statistical tests was set to $\alpha = 0.05$ a priori. All statistical tests were performed in Statistical Package for the Social Sciences (SPSS) (SPSS Inc., Chicago IL).

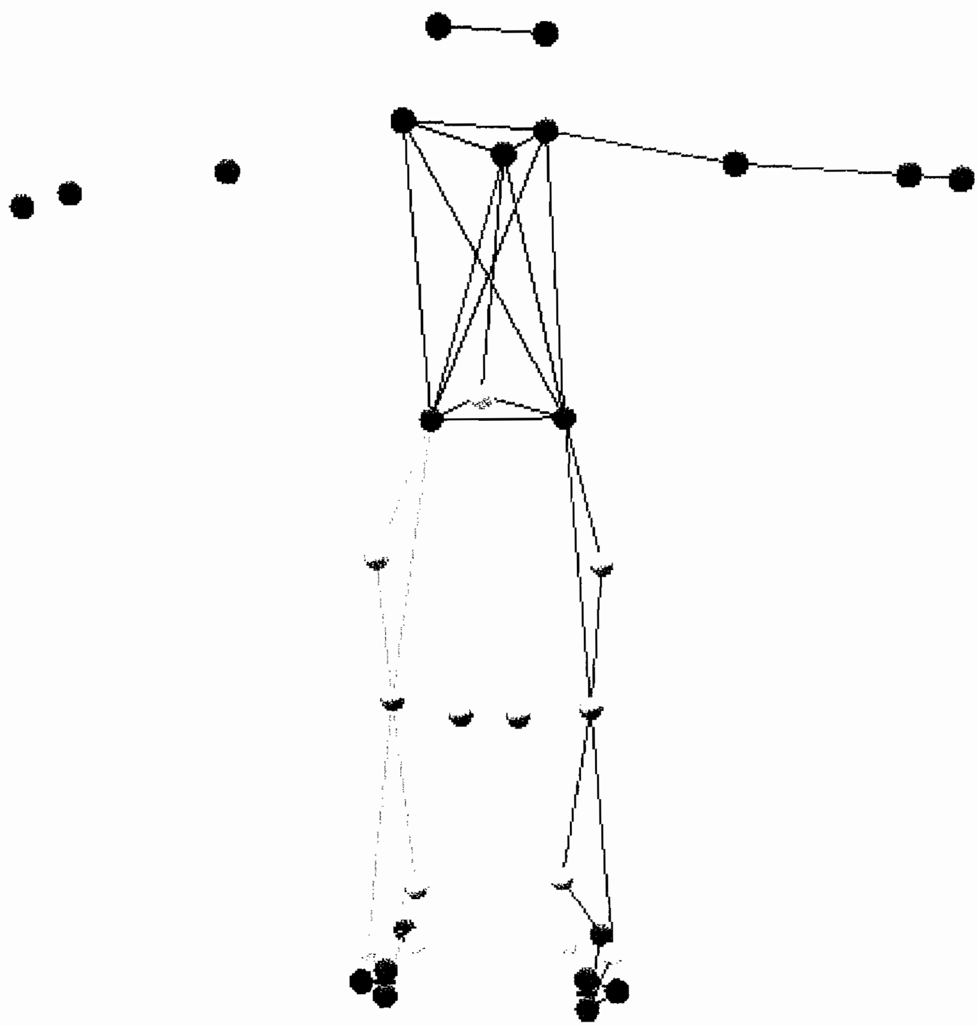


Figure 2. The markers used to develop the anatomic coordinate system. For the tracking marker set the only difference was the medial femoral condyle and medial malleolus markers were removed.

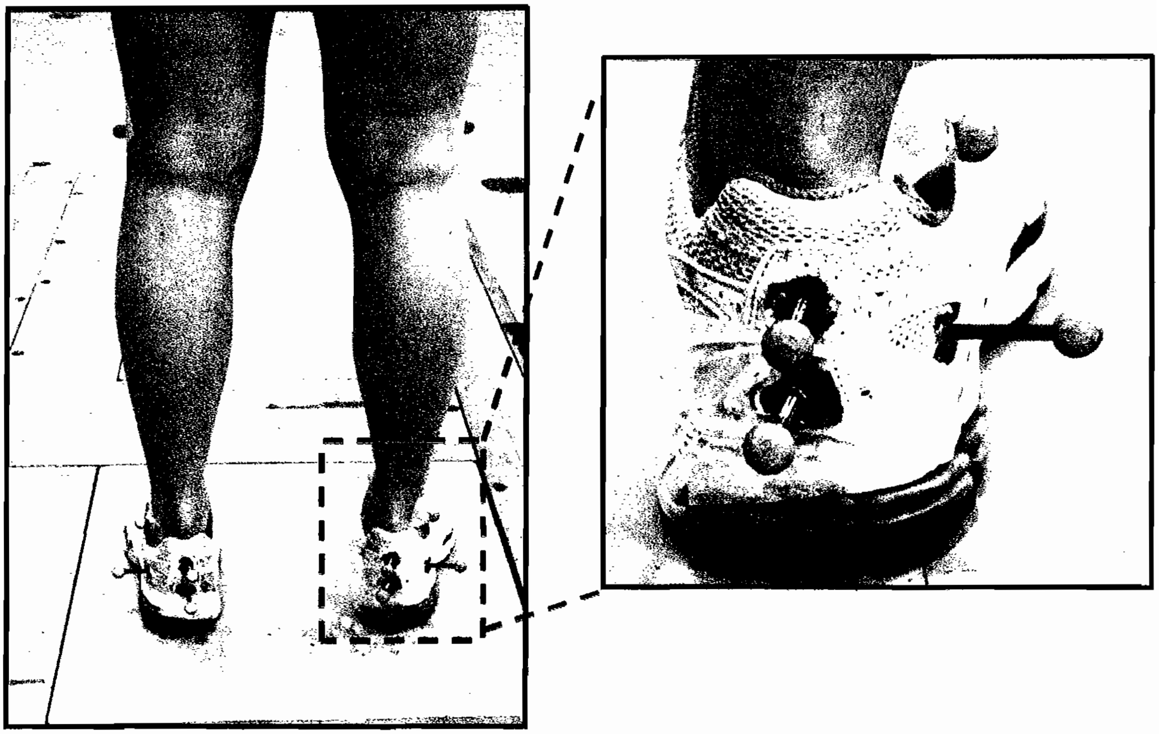


Figure 3. Picture of the heel windows marker placement and set up of the heel windows. The marker bases were attached directly to the calcaneous and the markers were screwed in from the outside once the subject had put the shoe on their foot.

CHAPTER III

RESULTS

Subject Data

Subject data including age, number of years running, weekly mileage, arch height ratios, and injury histories can be seen in Table 1. The average age of the subjects was 25.46 years (± 8.86 years) and the average number of years running was 7 years (± 3.92 years). All runners met the inclusion criteria for weekly mileage, with an average weekly mileage of 40.31 miles (± 11.03 miles). The average arch height ratios for the runner's feet were 0.321 (± 0.022) for the left and 0.317 (± 0.425) for the right, however, it should be noted that there was a wide range of arch heights. Most of the subjects had sustained at least one of the more common running injuries (Jacobs & Berson, 1986; James et al., 1978; Rauh et al., 2006; Taunton, 2002, 2003; Van Ghent et al., 2007).

Though thirteen subjects (twenty six feet) initially participated, only eleven subjects were used in the final analysis. One subject had to leave in the middle of the testing session and did not return to complete the testing. During testing it was discovered that a second subject was currently injured so their data were not used in the study. Additionally, there were errors and difficulties with the marker tracking and data collection on two subjects. For these two only one of feet was used for data analysis. Therefore the final number of feet analyzed for this study was 20 (n = 20).

Heel Windows Dimensions

Table 2 shows the average size of the heel windows for the entire group of subjects. Individual sizes of heel window holes, brand, and type of shoe for each subject can be seen in Table 3. The average heel window dimensions were 2.13 $cm.^2$ (± 0.31 cm.) for the bottom hole and 1.87 $cm.^2$ (± 0.28 cm.) for the top hole. Since the subjects wore their own shoes there was a wide variety of shoe brands and types. The type of shoe was a general classification based on the sole construction. Shoes with dual density midsoles or visible plastic stabilizing devices were classified as stability shoes. Shoes without these devices were classified as cushioning shoes. One subject used a light weight training or racing flat as their everyday training shoe, and this is reflected in the shoe classification.

Table 1. Age, number of years running, approximate weekly mileage, arch height ratios, and injury history for the original thirteen recruited subjects.

Subject	Age	Years Running	Weekly Mileage	Arch Height (L)	Arch Height (R)	Injury History
Mean	25.46	7.0 (± 3.92)	40.38 (± 11.17)	.322 (± 0.02)	.318 (± 0.04)	
R03	25	7	35 - 40	.352	.337	None
R04	20	5	60	.323	.313	L femoral neck stress fracture, L fibula stress fracture
R05	27	15	50	.330	.360	R ankle sprain, L hamstring strain
R06	42	5	35	.352	.385	none
R07	46	8	40	.353	.373	L piriformis strain
R08	28	6	30	.300	.297	R patellar tendonitis, L groin strain
R09	25	5	40	.306	.292	R tibia stress fracture, R foot generic sprain
R10	21	7	35 - 40	.321	.301	R tibia stress fracture
R11	20	3	25 - 30	.301	.256	R IT band strain, R patellar tendonitis
R12	19	8	60	.280	.244	R IT band strain, L 3rd metatarsal stress fracture
R13	18	4	25	.325	.325	L IT band syndrome
R14	19	3	30 - 40	.313	.300	L tibia stress fracture
R 15	21	15	45 - 50	.324	.349	2 x R tibia stress fracture, R hamstring strain, R IT band syndrome

Note. Subjects 4 and 9 were not used in the final analysis. Only the right foot was used for subject 10 and only the left foot was used for subject 12.

Running Speed

Due to unknown alignment issues, the photocells did not record running speed for every subject. In these situations the average forward velocity of the point representing the center of mass for each subject's pelvic segment was used to indicate the running velocity for that trial. This point was selected since it was thought to approximate the location of the subject's center of mass. Appendix C specifies how this point was determined.

Running speeds for both the shoe markers condition and heel windows marker conditions are shown in Table 4. Group mean speeds and speeds for each individual subject are shown. There was no significant difference in running speeds between the shoe markers and heel windows marker conditions, $t(10) = 0.717, p = .490$.

Table 2. Summary of heel window holes for the final feet used in the study.

Variable	*N*	*M*	*SD*
Upper Heel Window Area (cm^2)	20	1.87	± 0.28
Lower Heel Window Area (cm^2)	20	2.13	± 0.31

Table 3. Heel window areas, shoe brand, and shoe type for the subjects used in the final analysis.

Subject	Bottom Hole (cm)	Top Hole (cm)	Brand	Type
R03 L	2.00	1.77	Asics	Stability
R03 R	2.00	1.77		
R05 L	1.88	1.77	Nike	Cushioning
R05 R	1.88	2.14		
R06 L	1.77	1.77	Brooks	Cushioning
R06 R	1.65	1.77		
R07 L	2.00	2.00	Nike	Racing Flat
R07 R	2.36	2.12		
R08 L	2.59	2.36	Addidas	Stability
R08 R	2.27	2.51		
R10 R	2.01	2.12	Nike	Cushioning
R11 L	2.27	1.47	Asics	Stability
R11 R	2.20	1.76		
R12 L	2.83	1.65	Mizuno	Stability
R13 R	2.51	1.77	Nike	Cushioning
R13 L	2.51	1.77		
R14 L	2.14	2.00	Saucony	Stability
R14 R	1.77	1.32		
R15 L	1.88	1.77	Nike	Stability
R15 R	1.98	1.76		

Table 4. Running speed for shoe markers and heel windows conditions trials.

	Shoe Markers			Heel Windows		
Subject	*n*	*M*	*SD*	*n*	*M*	*SD*
Mean	11	7.37	0.62	11	7.32	0.82
	Running Velocity			Running Velocity		
R03	7.17			6.73		
R05	6.85			6.80		
R06	7.75			8.23		
R07	7.12			7.12		
R08	6.59			6.16		
R10	6.81			6.61		
R11	6.77			6.52		
R12	7.58			7.58		
R13	8.05			7.94		
R14	8.13			8.46		
R15	8.32			8.32		

Note: Running speed presented as minutes per mile pace.

Kinematic Variables Results

Figures 4 through 21 show an example of joint angles across the stance phase for one subject (R06) for the shoe marker condition. The general movement patterns were similar across subjects and between conditions, with slight variations. All individuals in the study demonstrated some asynchrony between the joint angle patterns on their left and right feet, which supports the decision to analyze each foot individually. Figures 22 and 23 show an example of one subject's (R06) vertical ground reaction forces for the left and right limbs, respectively. All subjects demonstrated similar magnitude vertical ground reaction forces with impact peaks between 1.5 and 2 times body weight and active force peaks of 2.5 to 3 times body weight.

Mean eversion excursion for all feet under both conditions, eversion excursion for each individual foot under both conditions, and the percent change between the shoe markers and heel windows markers conditions is shown in Table 5. There was not a significant difference in eversion excursion from the shoe markers condition to the heel windows condition, $t(19) = -0.296, p = .770$. Some feet demonstrated greater eversion excursion in the heel windows condition, while some feet demonstrated less eversion excursion. To more clearly illustrate this mixed response, a graph showing the changes in eversion excursion from the shoe markers to the heel windows markers can be seen in Figure 22.

Mean time to peak eversion for all feet under both conditions, time to peak eversion for each individual foot under both conditions, and the percent change in time to peak eversion between the shoe markers and heel windows markers conditions is shown in Table 6. There was not a significant difference in time to peak eversion between the two marker conditions, $t(19) = -0.291, p = .774$. Some feet demonstrated a shorter time to peak eversion in the heel windows conditions while others had longer times to peak excursion. The difference in times from the shoe markers to the heel windows markers conditions can be seen in Figure 25.

The mean percent stance at peak eversion for all feet under both conditions, the percent stance at peak eversion for each individual foot for both conditions, and the percent change in percent stance at which peak eversion occurred from the shoe markers to the heel windows marker conditions can be seen in Table 7. The difference in the percent stance at which peak eversion occurred in the shoe markers and the heel windows conditions was not significant, $t(19) = -1.089, p = .290$. For some feet peak eversion occurred earlier in stance with the shoe markers compared to the heel windows while other subjects had peak eversion occur later with the shoe markers compared to the heel windows. The changes in percent stance at which peak eversion occurred between the two conditions are shown in Figure 26.

The mean maximal instantaneous eversion velocity for all feet under both conditions, the maximal instantaneous eversion velocity for each individual foot under both conditions, and the percent change in maximal instantaneous eversion velocity between the two conditions can be seen in Table 8. There was not a significant difference in the mean maximal instantaneous eversion velocities between the shoe markers and heel windows conditions, $t(19) = 0.837, p = .413$. For some feet the maximal instantaneous eversion velocity increased in the heel windows conditions while for other subjects it decreased. Changes in maximal instantaneous eversion velocity between the shoe markers and heel windows markers are shown in Figure 27.

The mean maximal instantaneous vertical loading rate for all feet under both conditions, the maximal instantaneous vertical loading rate for each individual foot under both conditions, and the percent change in maximal instantaneous vertical loading rate between the shoe markers and heel windows markers conditions is shown in Table 9. There was not a significant difference in the mean maximal instantaneous loading rates between conditions, $t(16) = 0.780$, $p = .447$. Some feet experienced higher maximal instantaneous loading rates in the heel windows condition compared to the shoe markers condition while other experienced lower ones. Changes in maximal instantaneous loading rate from the shoe markers condition to the heel windows conditions are shown in Figure 28.

The mean ankle frontal plane CMC for all feet under both conditions, the ankle frontal plane CMC for each individual foot under both conditions, and the percent change in ankle frontal plane CMC values from the shoe markers to the heel windows markers for each subject is shown in Table 10. Overall, the heel windows marker condition resulted in significantly lower CMC values than the shoe marker conditions, indicating higher variability in the ankle joint frontal plane angle curves, $t(19) = 3.56$, $p = 0.002$. While almost all feet demonstrated greater variability in the heel windows markers condition the magnitude of the change varied from foot to foot. A graphic illustration of the changes in CMC values from the shoe markers condition to the heel windows marker conditions is shown in Figure 29.

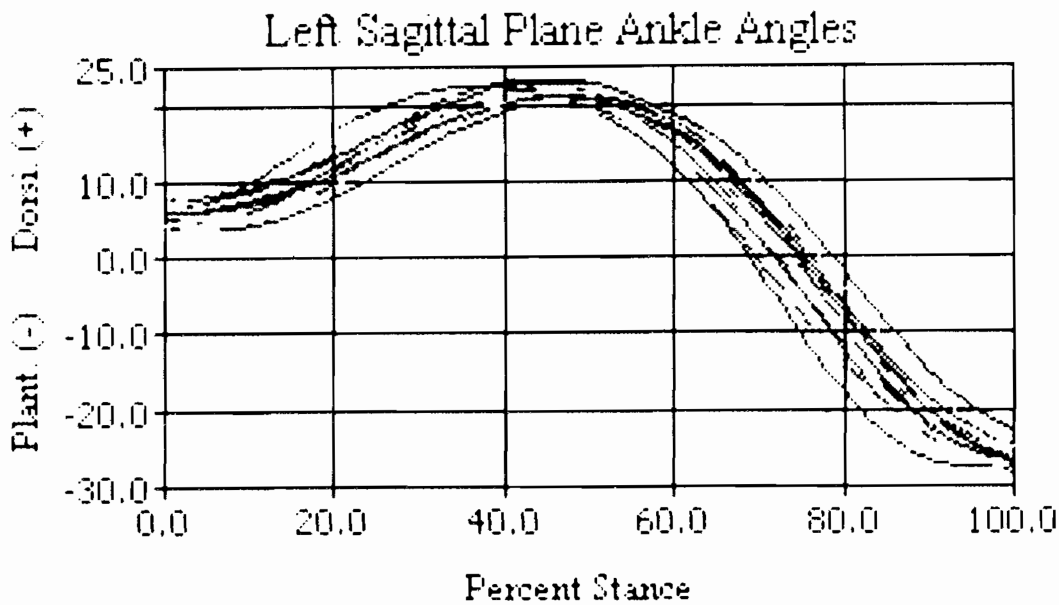


Figure 4. Example left sagittal plane ankle angles. Dorsiflexion is positive, plantar flexion is negative

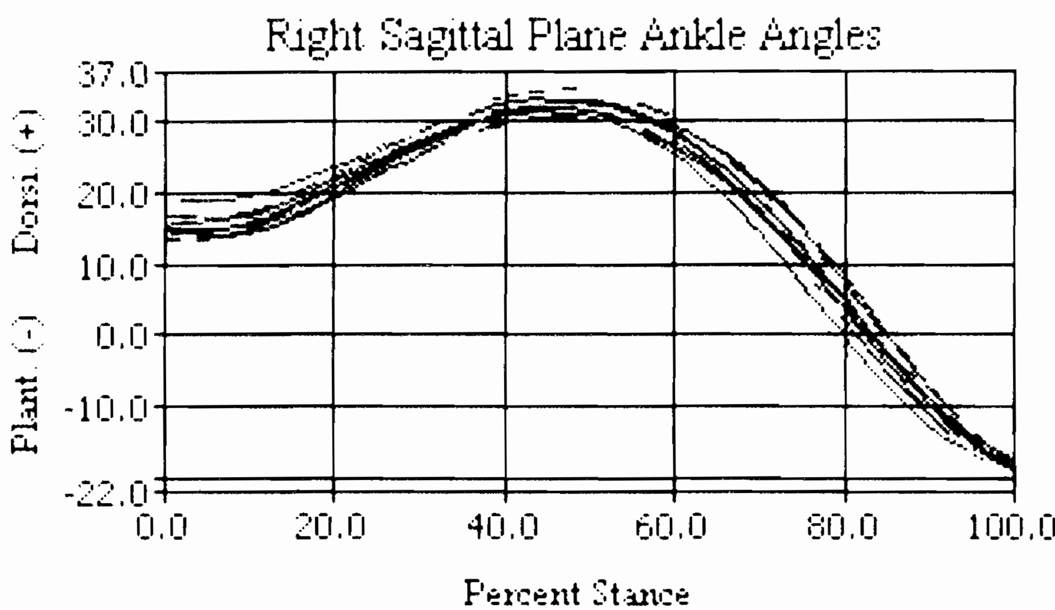


Figure 5. Example right sagittal plane ankle angles. Dorsiflexion is positive, plantar flexion is negative.

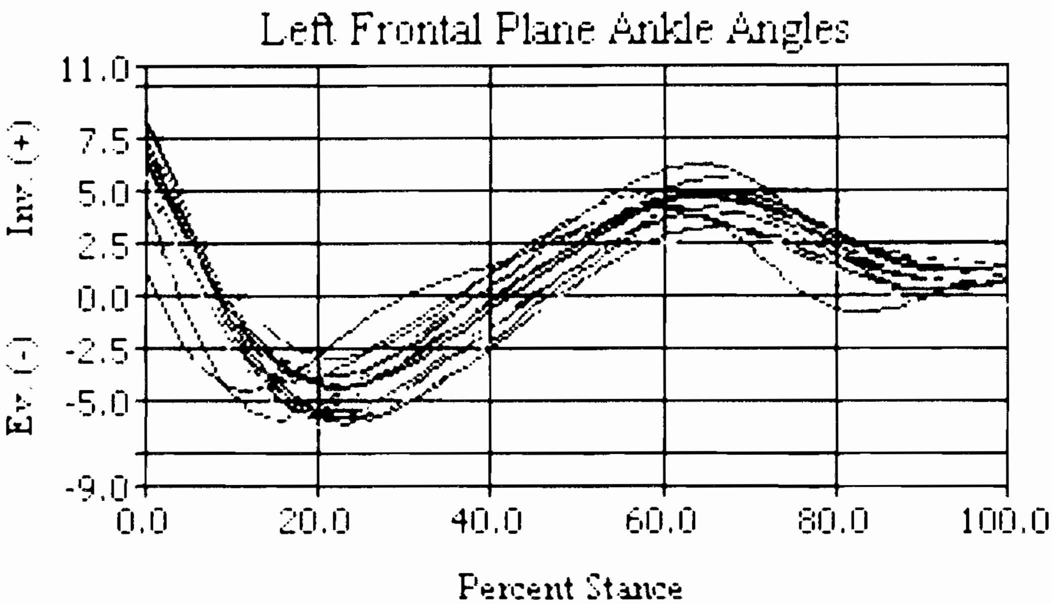


Figure 6. Example frontal plane ankle angles. Rear foot eversion is negative, inversion is positive.

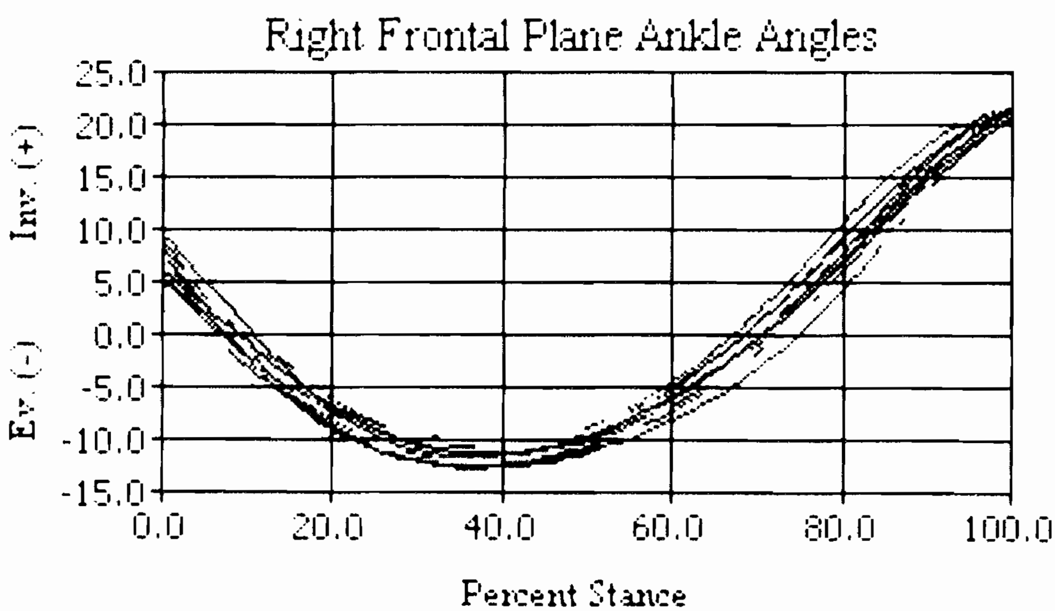


Figure 7. Example right frontal plane ankle angles. Rear foot eversion is negative, inversion is positive.

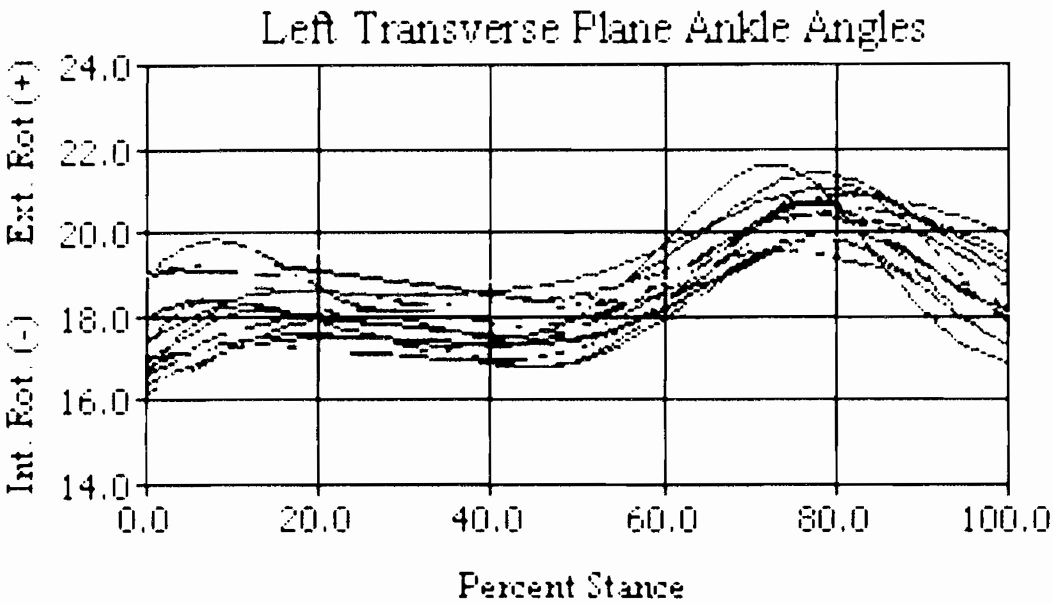


Figure 8. Example left transverse plane ankle angles. Internal rotation is negative,external rotation is positive.

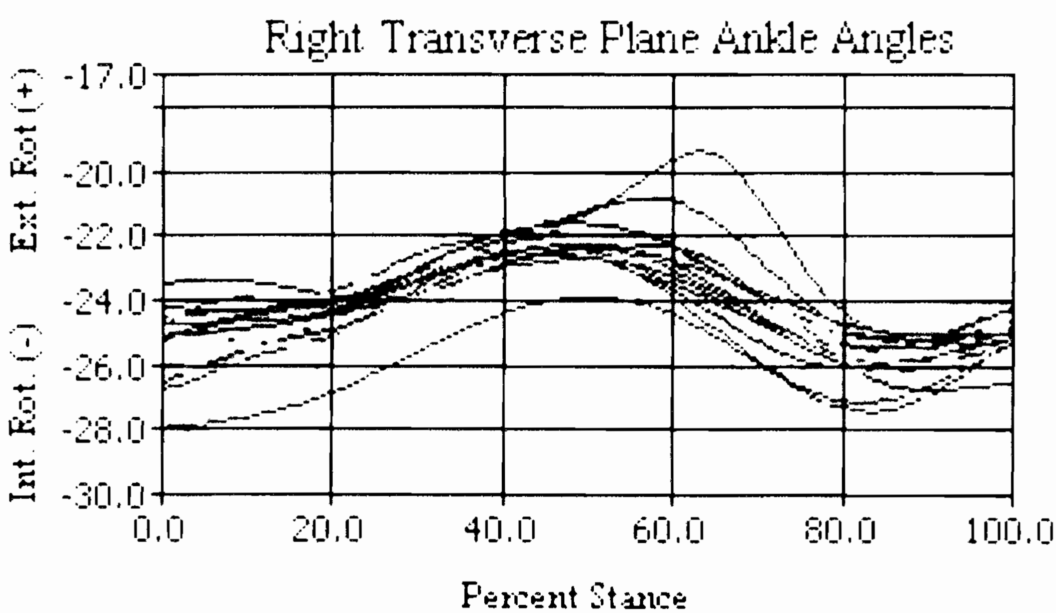


Figure 9. Example right transverse plane ankle angles. Internal rotation is negative and external rotation is positive.

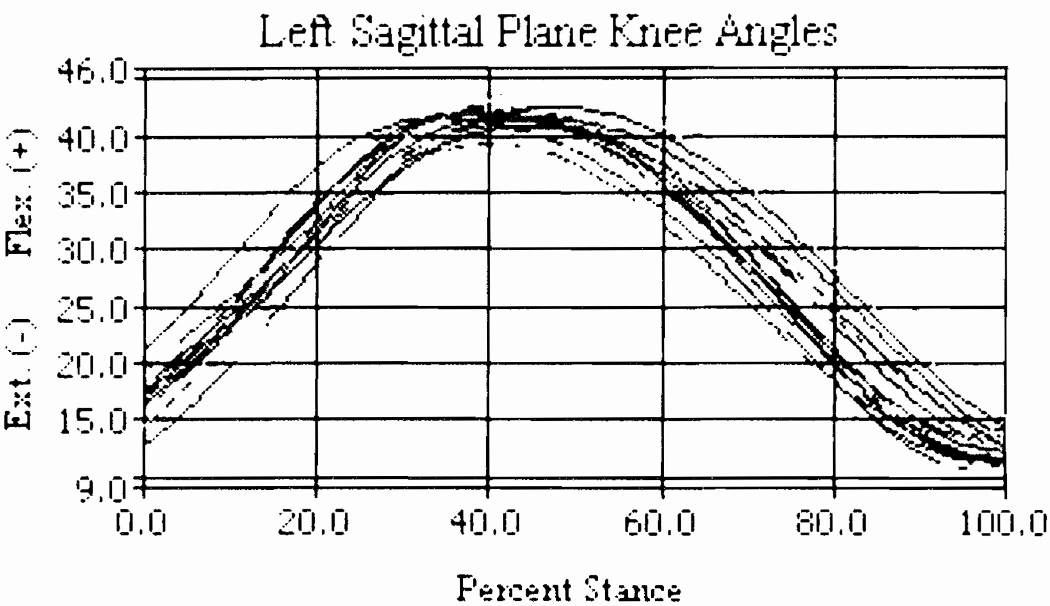


Figure 10. Example left sagittal plane knee angles. Flexion is positive and extension is negative.

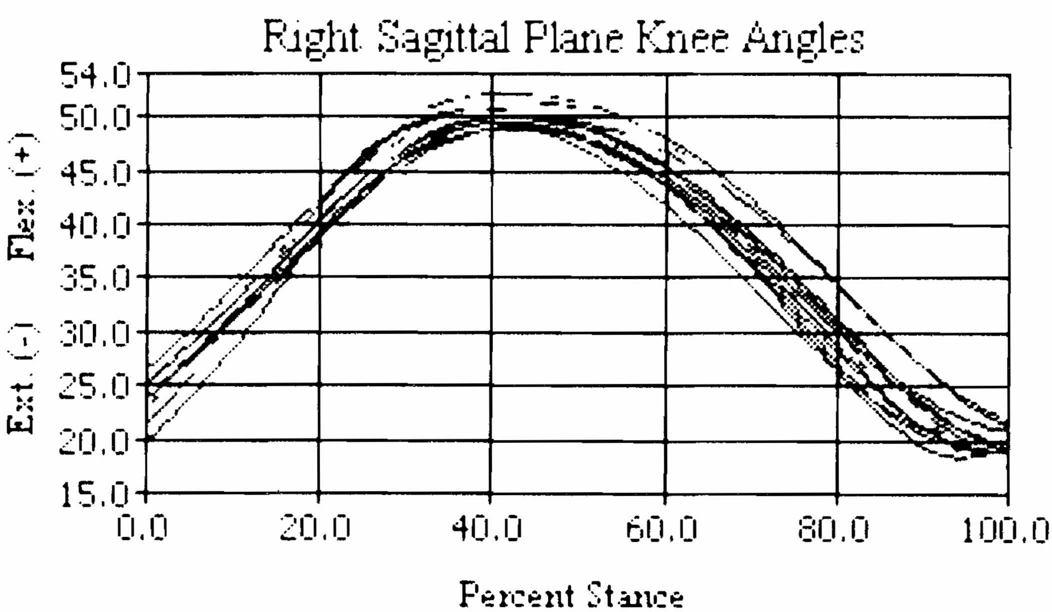


Figure 11. Example right sagittal plane knee angles. Flexion is positive and extension is negative.

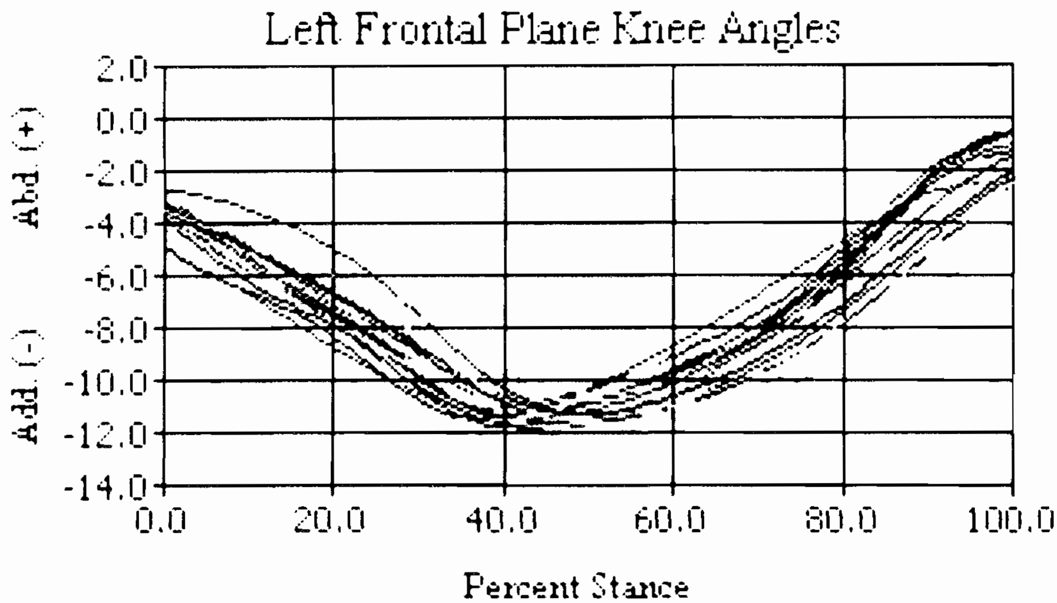


Figure 12. Example left frontal plane knee angles. Abduction is postitive and adduction is negative.

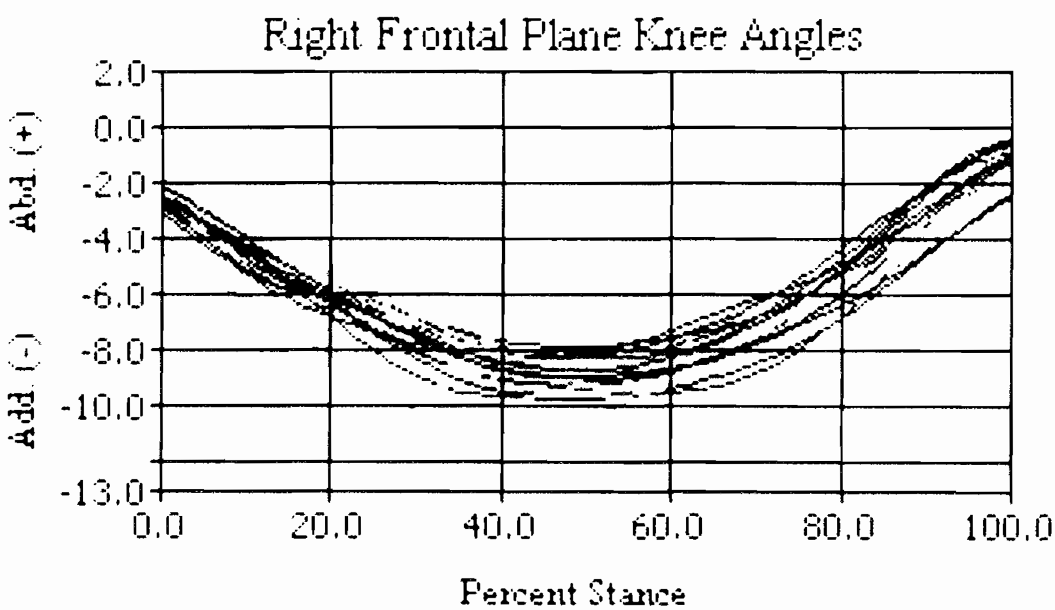


Figure 13. Example right frontal plane knee angles. Abduction is positive and adduction is negative.

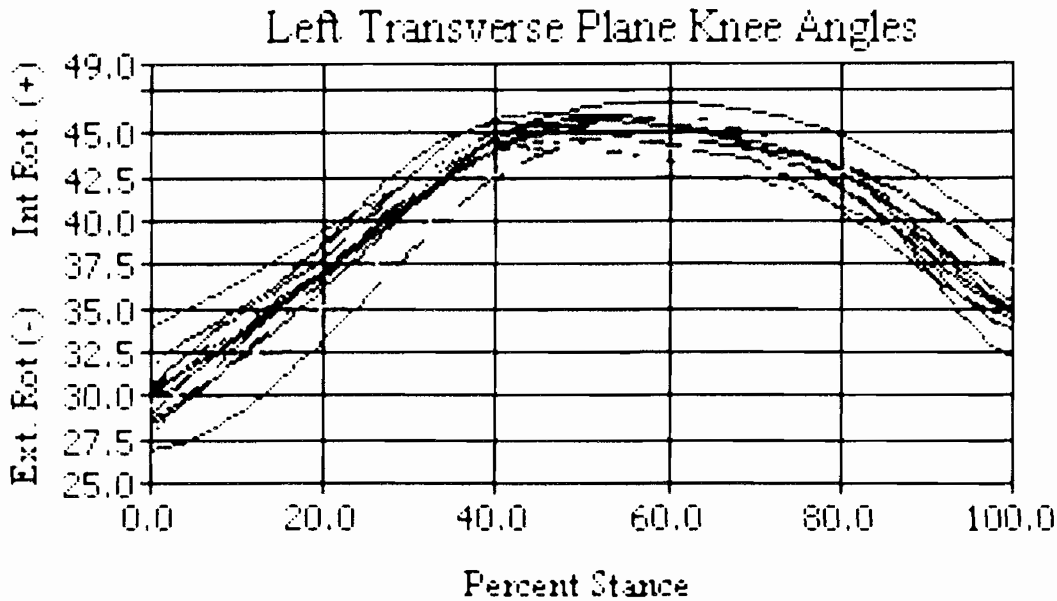


Figure 14. Example left transverse plane knee angles. Internal rotation is positive and external rotation is negative.

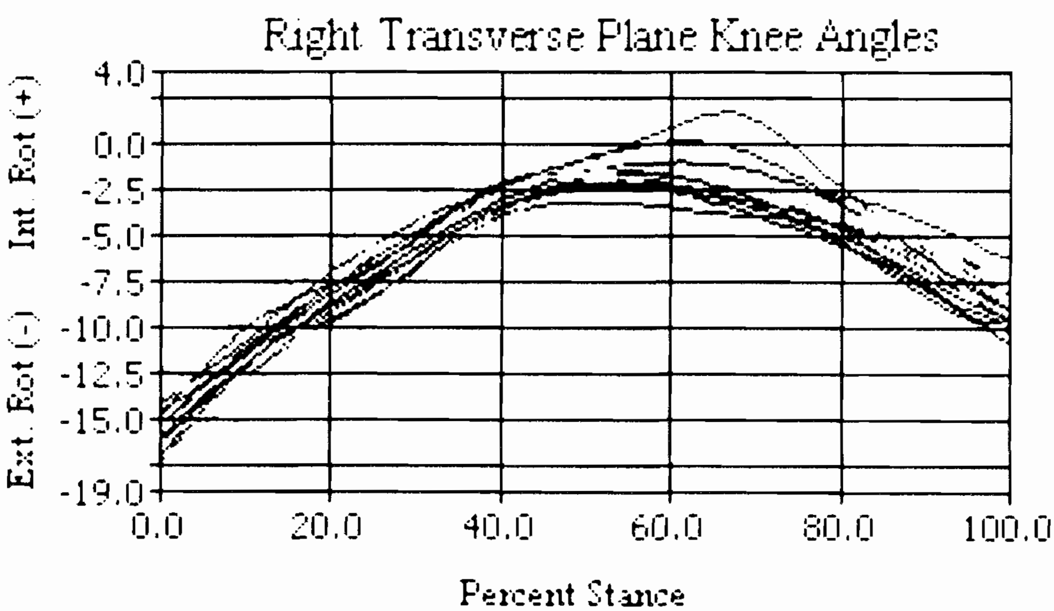


Figure 15. Example right transverse plane knee angles. Internal rotation is positive and external rotation is negative.

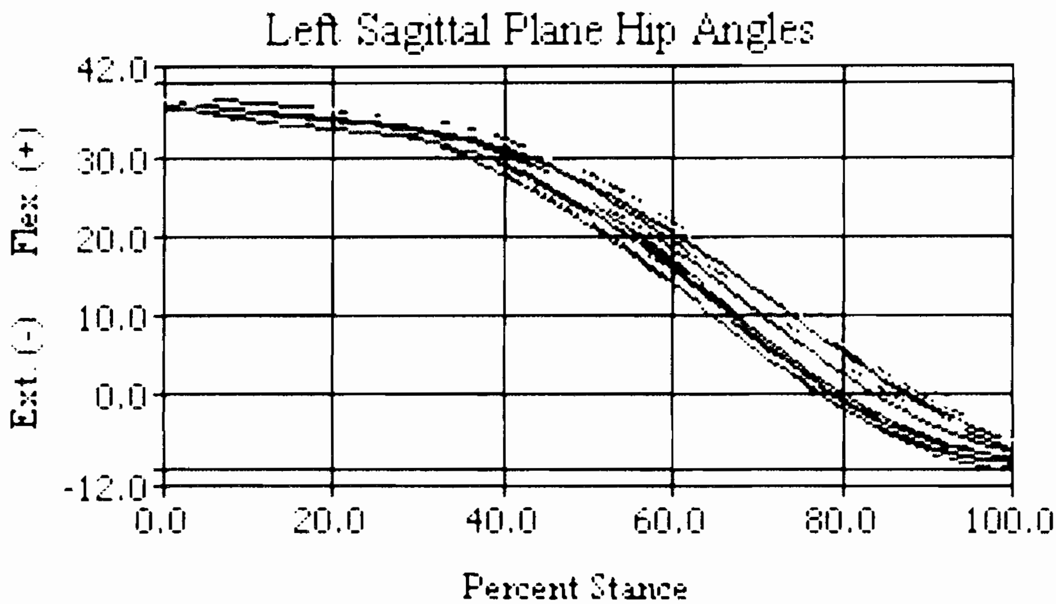


Figure 16. Example left sagittal plane hip angles. Flexion is positive and extension is negative.

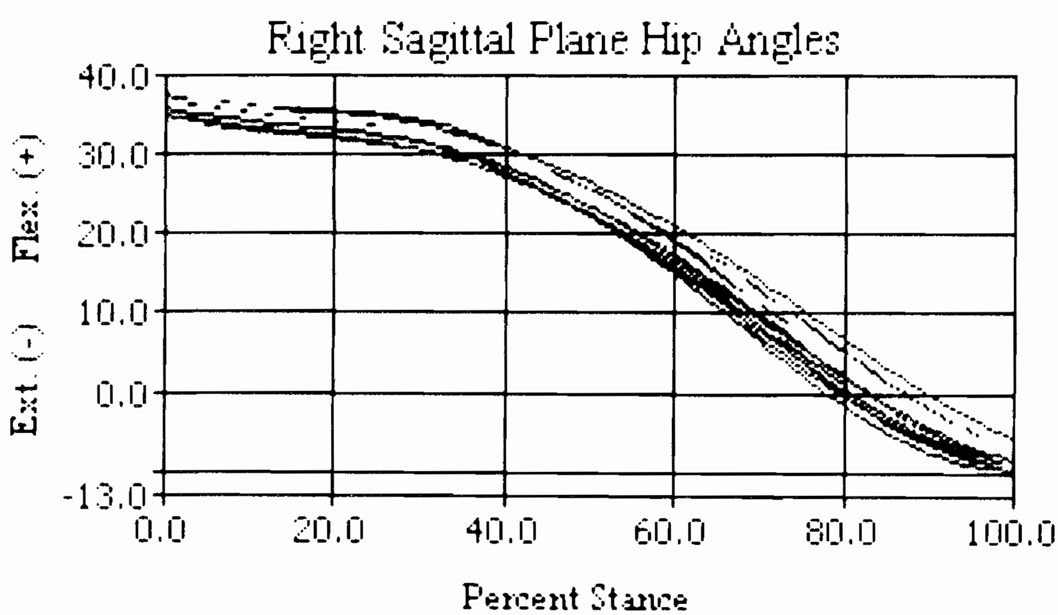


Figure 17. Example right sagittal plane hip angles. Flexion is positive and extension is negative.

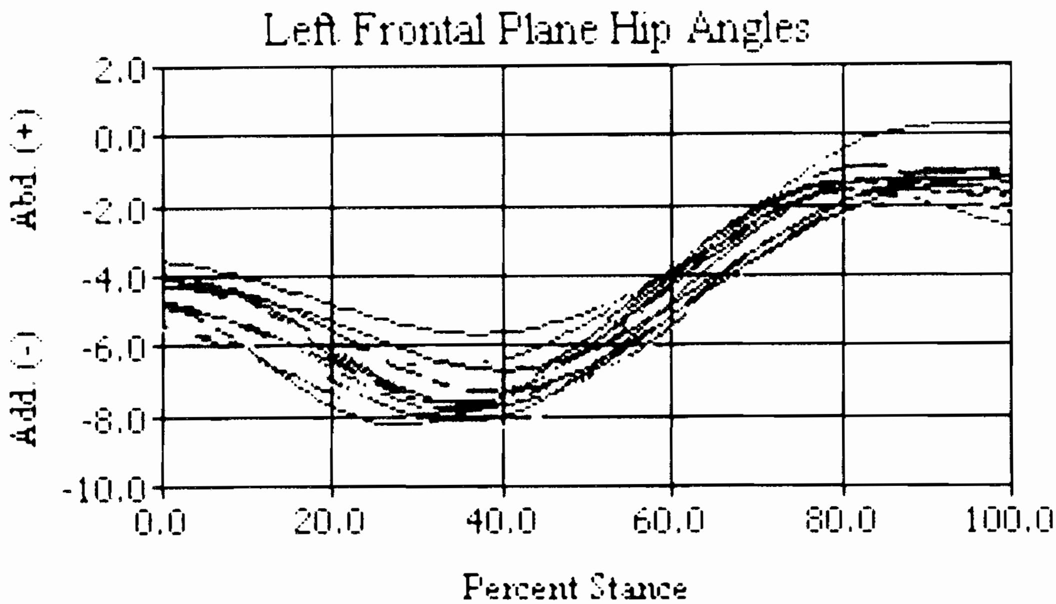


Figure 18. Example left frontal plane hip angles. Abduction is positive and adduction is negative.

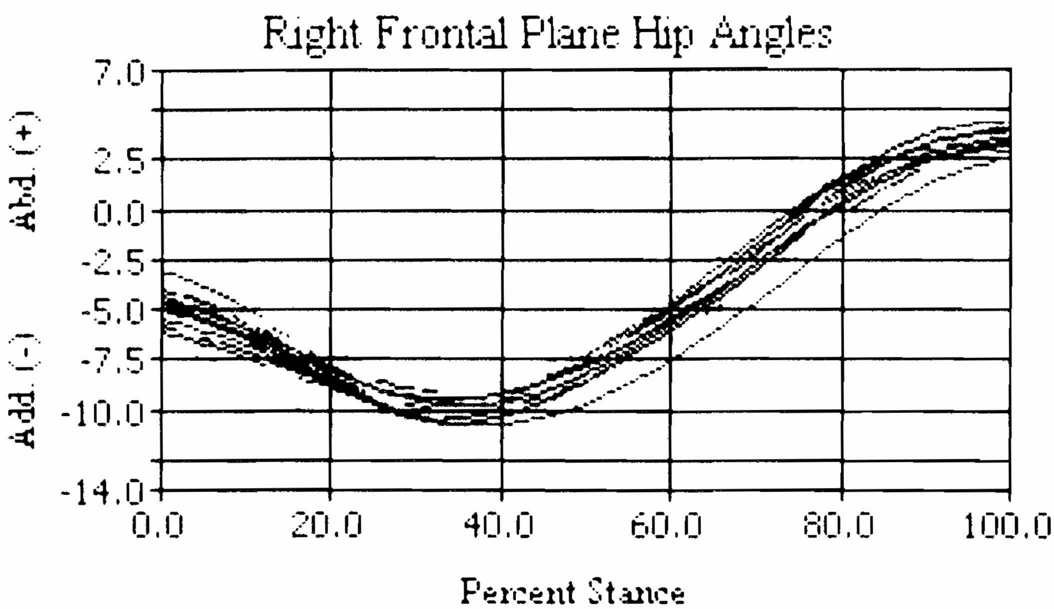


Figure 19. Example right frontal plane hip angles. Abduction is positive and adduction is negative.

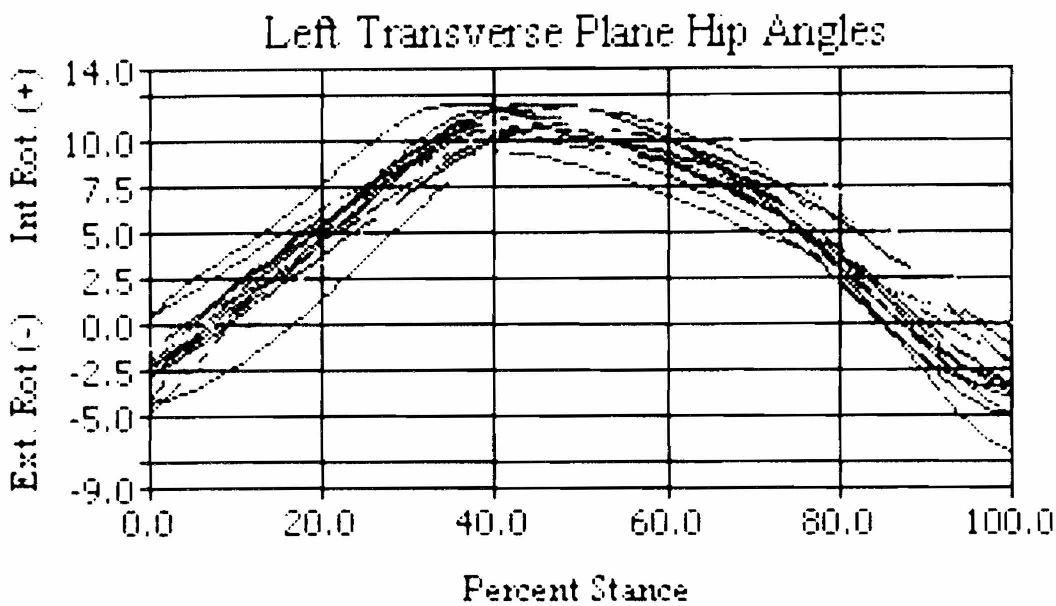


Figure 20. Example left transverse plane hip angles. Internal rotation is positive and external rotation is negative.

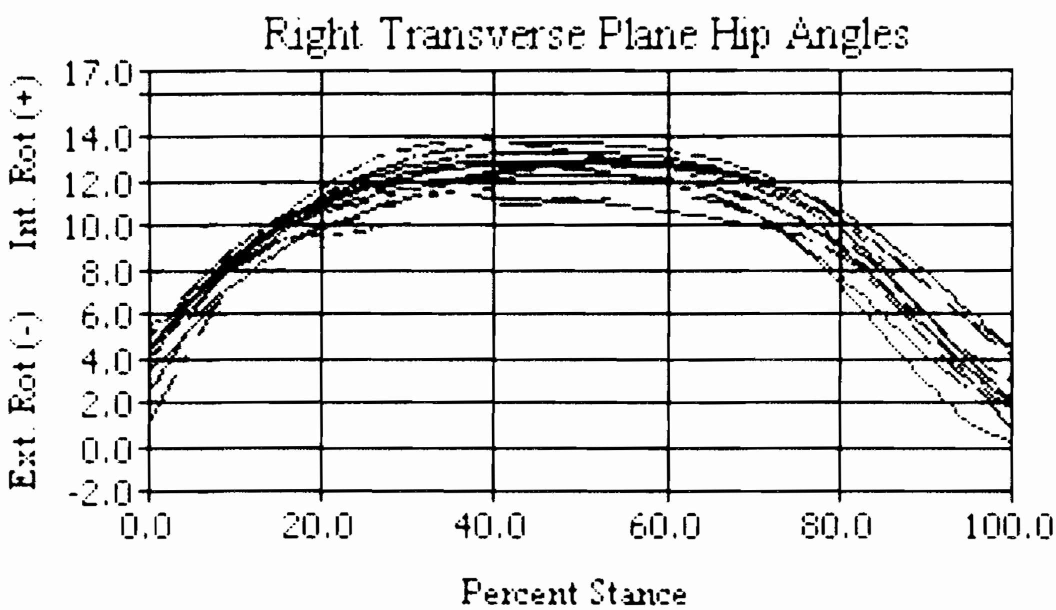


Figure 21. Example right transverse plane hip angles. Internal rotation is positive and external rotation is negative.

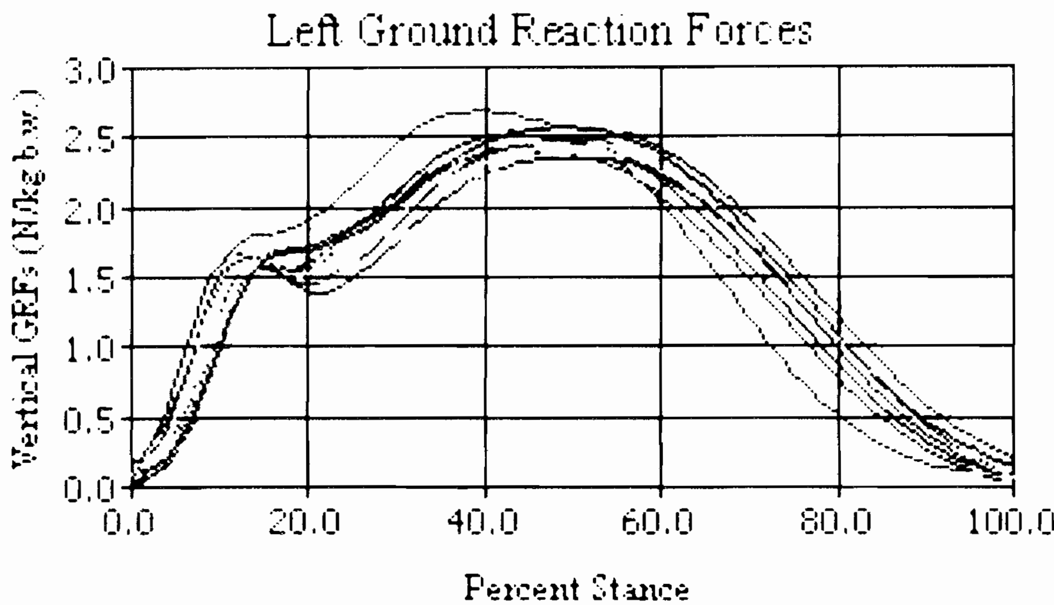


Figure 22. Example left limb vertical ground reaction forces normalized to body weight.

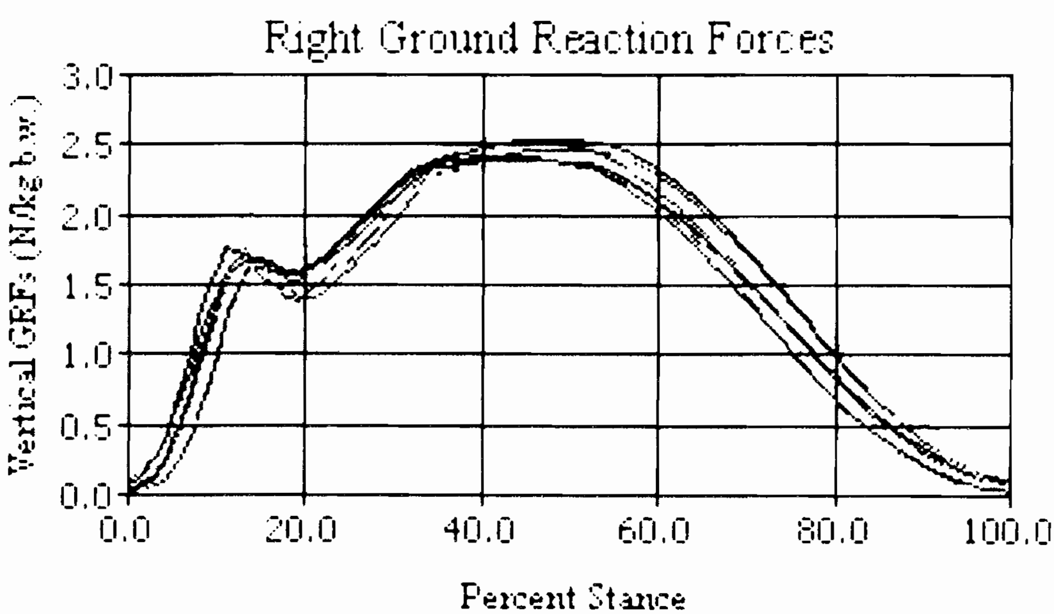


Figure 23. Example right limb vertical ground reaction forces normalized to body weight.

Table 5. Eversion excursion results for both the shoe markers and heel window markers conditions.

	Shoe Markers			Heel Windows			Percent Change		
Subject	*n*	*M*	*SD*	*n*	*M*	*SD*	*n*	*M*	*SD*
Mean	20	13.66	± 6.76	20	13.83	± 6.91	20	3.37	± 6.91
	Eversion Excursion			Eversion Excursion			Percent Change		
R03 L		9.42			9.81			4.09	
R03 R		19.94			20.06			0.60	
R05 L		9.56			6.87			-28.15	
R05 R		12.32			11.41			-7.33	
R06 L		10.87			11.36			4.54	
R06 R		17.92			19.87			10.88	
R07 L		10.33			8.57			-17.06	
R07 R		15.96			15.28			-4.22	
R08 L		7.73			11.09			43.44	
R08 R		9.34			10.48			12.21	
R10 R		17.18			21.23			23.57	
R11 L		10.89			10.05			-7.71	
R11 R		17.42			13.43			-22.95	
R12 L		10.25			13.55			32.26	
R13 L		14.82			12.89			-13.05	
R13 R		33.14			35.56			7.32	
R14 L		8.05			7.09			-11.87	
R14 R		25.21			20.97			-16.85	
R15 L		5.88			6.08			3.38	
R15 R		7.06			10.90			54.31	

Note: Eversion excursion measured in degrees (°), from heel strike to point of maximal eversion. The mean row shows mean for the group, while each individual foot's data are displayed below. Positive percent change indicates greater eversion excursion in the heel windows condition while a negative percent change indicates less eversion excursion in the heel windows condition.

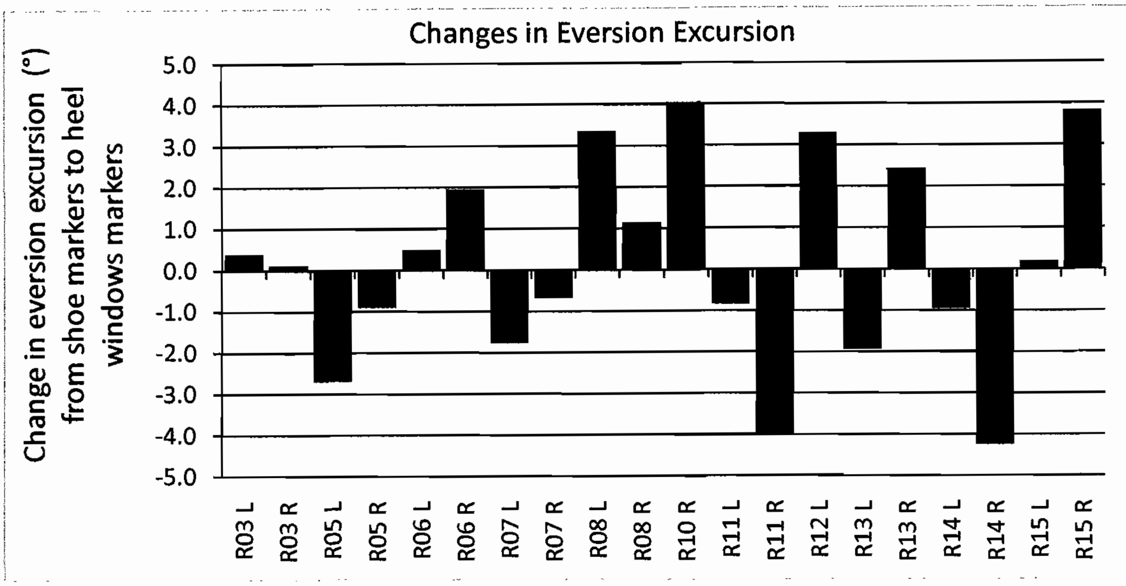


Figure 24. Changes in eversion excursion from shoe markers to heel windows markers condition for each individual subject. Eversion excursion measured in degrees from heel strike to maximum rear foot eversion.

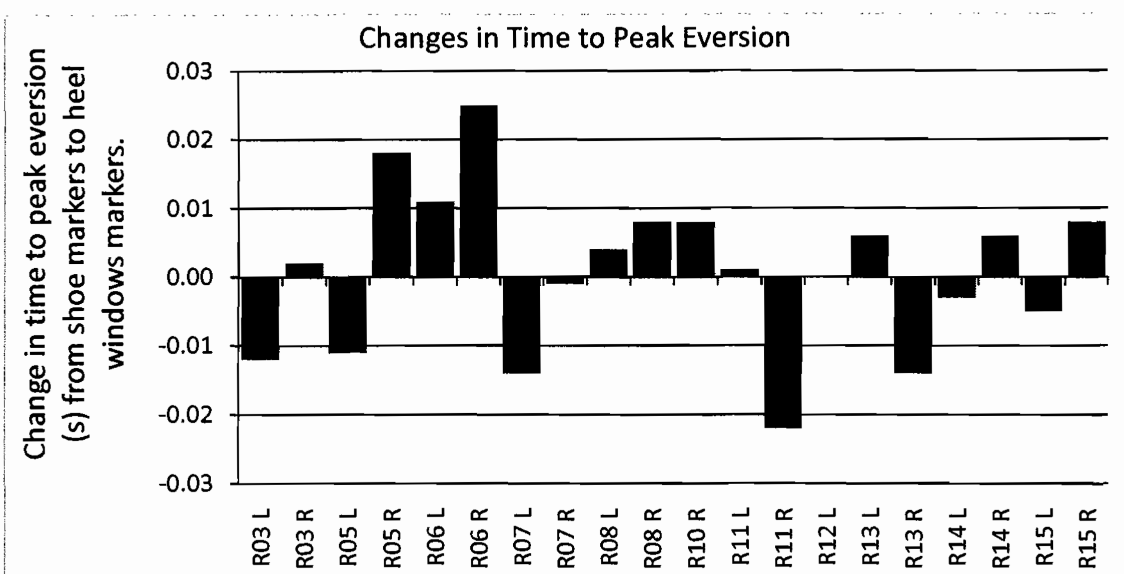


Figure 25. Changes in time to peak eversion from shoe markers to heel windows condition for each individual subject. Time measured in seconds.

Table 6. Time to peak eversion results for both the shoe markers and heel windows markers conditions.

	Shoe Markers			Heel Windows			Percent Change		
Subject	*n*	*M*	*SD*	*n*	*M*	*SD*	*n*	*M*	*SD*
Mean	20	0.088	± 0.029	20	0.089	± 0.032	20	0.45	± 13.11
	Time to Peak Eversion			Time to Peak Eversion			Percent Change		
R03 L	0.075			0.063			-16.00		
R03 R	0.108			0.110			1.85		
R05 L	0.061			0.050			-18.03		
R05 R	0.114			0.132			15.79		
R06 L	0.052			0.063			21.15		
R06 R	0.096			0.121			26.04		
R07 L	0.076			0.062			-18.42		
R07 R	0.119			0.118			-0.84		
R08 L	0.068			0.072			5.88		
R08 R	0.117			0.125			6.84		
R10 R	0.122			0.130			6.56		
R11 L	0.053			0.054			1.89		
R11 R	0.110			0.088			-20.00		
R12 L	0.073			0.073			0.00		
R13 L	0.055			0.061			10.91		
R13 R	0.125			0.111			-11.20		
R14 L	0.045			0.042			-6.67		
R14 R	0.101			0.107			5.94		
R15 L	0.056			0.051			-8.93		
R15 R	0.129			0.137			6.20		

Note: Time to peak eversion measured in seconds from heel strike to maximal eversion. A positive percent change indicates a longer time to peak eversion in the heel windows condition while a negative percent change indicates a shorter time to peak eversion in the heel windows condition.

Table 7. Percent stance at which peak eversion occurred for both the shoe markers and heel windows markers conditions.

	Shoe Markers			Heel Windows			Percent Change		
Subject	*n*	*M*	*SD*	*n*	*M*	*SD*	*n*	*M*	*SD*
Mean	20	36.76	± 12.0	20	36.77	± 14.87	20	- 1.63	± 13.11

	Percent Stance at Peak Eversion	Percent Stance at Peak Eversion	Percent Change
R03 L	28.83	28.17	-2.31
R03 R	46.69	47.46	1.63
R05 L	25.80	21.88	-15.21
R05 R	49.95	55.64	11.40
R06 L	23.70	27.60	16.47
R06 R	41.26	49.90	20.93
R07 L	30.21	24.85	-17.75
R07 R	47.14	46.21	-1.97
R08 L	26.74	29.08	8.76
R08 R	46.14	49.83	8.00
R10 R	51.20	54.63	6.69
R11 L	23.08	25.27	9.49
R11 R	48.56	42.67	-12.13
R12 L	29.46	28.78	-2.30
R13 L	22.45	24.81	10.51
R13 R	49.89	43.96	-11.88
R14 L	20.68	19.15	-7.43
R14 R	46.10	49.03	6.38
R15 L	23.92	25.92	8.36
R15 R	53.40	60.50	13.30

Note: Percent stance indicates the percent of the stance phase of the gait cycle at which peak eversion occurred. A positive percent change indicates peak eversion occurred later in stance in the heel windows condition while a negative percent change indicates peak eversion occurred earlier in stance in the heel windows condition.

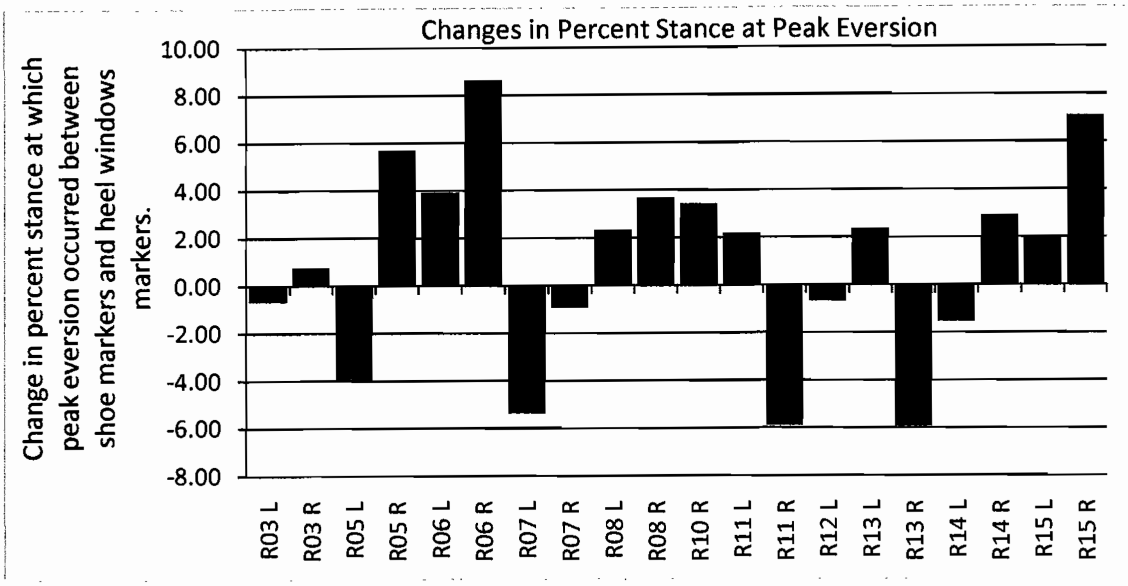


Figure 26. Changes in percent stance at which peak eversion occurred between shoe markers and heel windows markers.

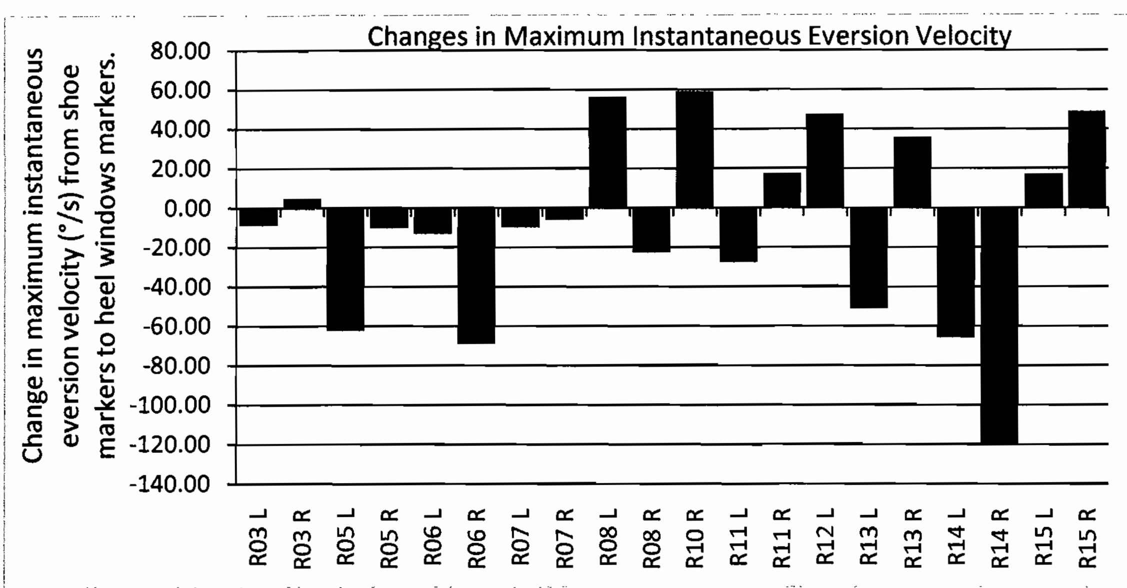


Figure 27. Change in maximum instantaneous eversion velocity from shoe markers to heel windows markers. Velocity measured in degrees per second (°/s).

Table 8. Maximal instantaneous eversion velocity for shoe markers and heel windows markers conditions.

	Shoe Markers			Heel Windows			Percent Change		
Subject	*n*	*M*	*SD*	*n*	*M*	*SD*	*n*	*M*	*SD*
Mean	20	269.5	± 117.6	20	260.6	± 115.1	20	- 0.28	± 20.47
	Max.Instant. Eversion Velocity			Max. Instant. Eversion Velocity			Percent Change		
R03 L	225.50			216.85			-3.83		
R03 R	285.62			290.44			1.69		
R05 L	252.82			190.59			-24.61		
R05 R	178.59			168.38			-5.71		
R06 L	296.59			283.60			-4.38		
R06 R	315.45			246.57			-21.84		
R07 L	209.39			199.51			-4.72		
R07 R	246.69			240.69			-2.43		
R08 L	179.29			235.58			31.39		
R08 R	152.03			129.43			-14.86		
R10 R	257.99			316.92			22.84		
R11 L	296.23			268.70			-9.29		
R11 R	267.96			285.31			6.48		
R12 L	247.65			295.09			19.16		
R13 L	381.55			330.56			-13.36		
R13 R	645.29			681.20			5.56		
R14 L	276.14			210.39			-23.81		
R14 R	424.24			304.21			-28.29		
R15 L	159.28			176.59			10.87		
R15 R	91.42			140.34			53.51		

Note: Maximal instantaneous eversion velocity measured in degrees per second (°/s). A positive percent change indicates a greater eversion maximal instantaneous eversion velocity in the heel windows condition while a negative percent change indicates smaller maximal instantaneous eversion velocity in the heel windows condition.

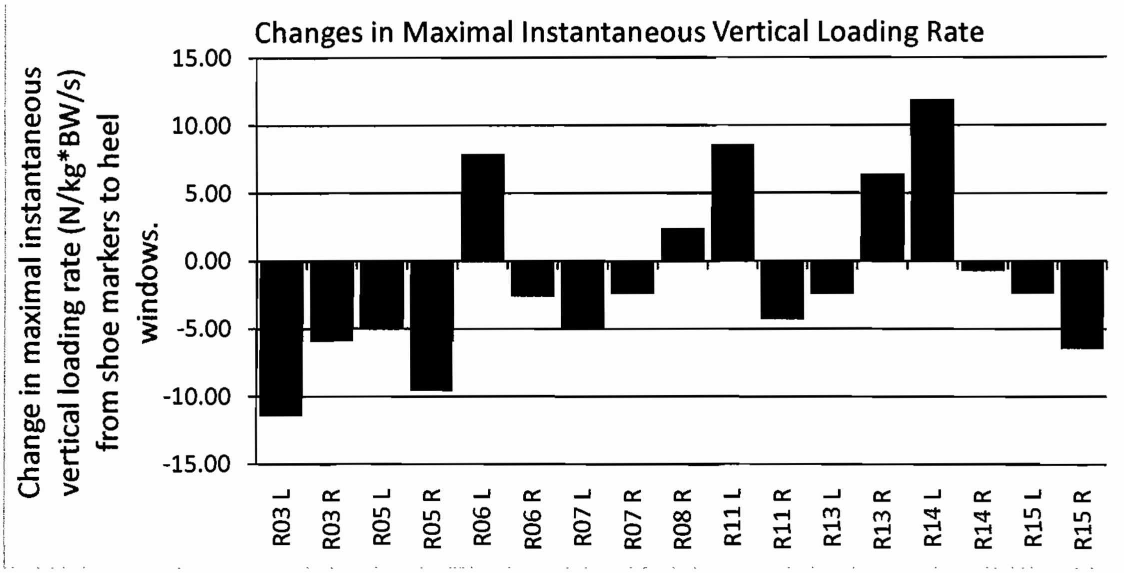


Figure 28. Changes in maximal instantaneous vertical loading rate from shoe markers to heel windows marker conditions. Loading rate measured in Newtons per kilogram of body weight per second (N/kg*BW/s).

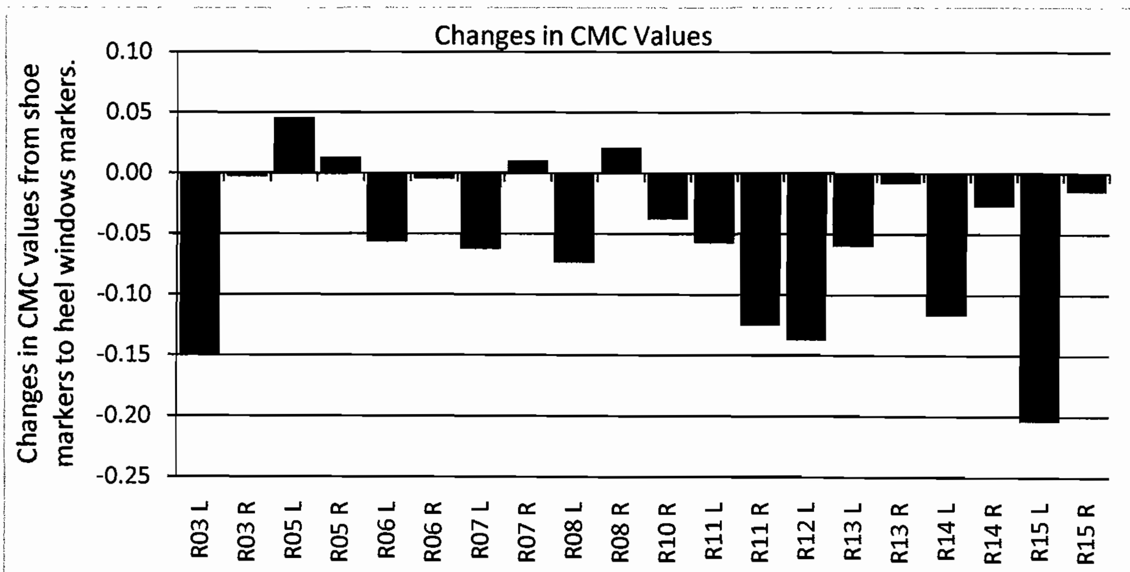


Figure 29. Changes in CMC values from shoe markers to heel windows markers.

Table 9. Maximal instantaneous vertical loading rates for shoe markers and heel windows markers.

	Shoe Markers			Heel Windows			Percent Change		
Subject	*n*	*M*	*SD*	*n*	*M*	*SD*	*n*	*M*	*SD*
Mean	20	84.47	± 18.48	20	82.23	± 17.16	20	- 2.01	± 7.52
	Max. Instant. Vertical Loading Rate			Max. Instant Vertical Loading Rate			Percent Change		
R03 L	108.41			96.96			-10.56		
R03 R	102.76			96.84			-5.76		
R05 L	97.91			92.85			-5.17		
R05 R	81.37			71.79			-11.78		
R06 L	84.16			92.05			9.38		
R06 R	97.34			94.73			-2.68		
R07 L	76.15			71.13			-6.59		
R07 R	65.42			62.99			-3.71		
R08 L	-			-			-		
R08 R	112.92			115.35			2.15		
R10 R	-			-			-		
R11 L	74.96			83.54			11.45		
R11 R	80.88			76.56			-5.34		
R12 L	-			-			-		
R13 L	54.16			51.73			-4.49		
R13 R	41.98			48.41			15.31		
R14 L	84.78			79.80			-5.87		
R14 R	91.18			90.44			-0.81		
R15 L	89.38			86.98			-2.69		
R15 R	92.25			85.78			-7.02		

Note: Maximal instantaneous vertical loading rate is measured in Newtons per kilogram of body weight per second (N/Kg BW/s). A positive percent change indicates a higher loading rate in the heel windows markers condition compared to the shoe markers condition while a negative percent change indicates a lower loading rate in the heel windows condition.

Table 10. CMC values for the shoe markers and heel windows markers conditions.

	Shoe Markers			Heel Windows			Percent Change		
Subject	*n*	*M*	*SD*	*n*	*M*	*SD*	*n*	*M*	*SD*
Mean	20	0.924	± 0.070	20	0.872	± 0.120	20	- 6.05	± 7.95
	CMC Value			CMC Value			Percent Change		
R03 L	0.846			0.698			-17.57		
R03 R	0.985			0.982			-0.28		
R05 L	0.870			0.917			5.30		
R05 R	0.972			0.986			1.38		
R06 L	0.924			0.867			-6.12		
R06 R	0.992			0.987			-0.47		
R07 L	0.886			0.823			-7.08		
R07 R	0.980			0.990			1.04		
R08 L	0.897			0.823			-8.21		
R08 R	0.969			0.990			2.18		
R10 R	0.986			0.948			-3.83		
R11 L	0.894			0.836			-6.42		
R11 R	0.984			0.858			-12.77		
R12 L	0.877			0.739			-15.68		
R13 L	0.845			0.785			-7.10		
R13 R	0.979			0.971			-0.82		
R14 L	0.886			0.769			-13.21		
R14 R	0.988			0.961			-2.73		
R15 L	0.754			0.550			-27.06		
R15 R	0.967			0.952			-1.55		

Note: CMC indicates the repeatability or variability of the angle curve across stance. A higher CMC value indicates less variability while a lower CMC value indicates greater variability. A positive percent change indicates a higher CMC value in the heel windows markers condition compared to the shoe markers while a negative percent change indicates a lower CMC value in the heel windows condition compared to the shoe markers condition.

CHAPTER IV

DISCUSSION

The purpose of this study was to examine differences in commonly measured parameters thought to be related to overuse injury in runners when measured with markers mounted on the shoe and with markers mounted directly on the rear foot using heel windows. Specifically, this study examined differences between the two marker conditions in the time to peak rear foot eversion, the percent stance at which maximal eversion occurred, the maximal instantaneous eversion velocity, and the maximal instantaneous vertical loading rate. Additionally, this study also measured the variability and repeatability of the ankle joint inversion-eversion curve. A secondary purpose of this study was to make a recommendation on marker methods and protocols to use in a longitudinal study.

The hypotheses for this study were that maximal rear foot eversion would be less in the heel windows condition compared to the shoe marker condition, that time to peak eversion excursion and the percent stance at which peak eversion excursion occurred would not be different between the two marker systems, the maximal instantaneous eversion velocity would increase in the heel windows condition compared to the shoe markers condition, and that the maximal instantaneous vertical loading rate would decrease from the shoe marker condition to the heel window marker condition. It was also hypothesized that the variability of the rear foot inversion-eversion curve would increase in the heel windows condition compared to the shoe markers conditions.

The only hypothesis which was supported in the results of this study was that the variability of the ankle joint inversion-eversion curve was higher in the heel windows condition compared to the shoe markers condition. For all the other variables the results were not statistically significant, failing to provide support for most hypotheses of the study.

There are numerous possible explanations for why there were no significant results among the kinematic parameters. Firstly, the subjects themselves could be a poor sample, not representative of an average cross section of the running population. Secondly, even if this sample of subjects does provide a reliable estimate of the general running population, they may have different or unique kinematic or kinetic patterns which led to the lack of statistical significance. Thirdly, there could have been differences between running trials under the two marker conditions which were not adequately controlled and resulted in the lack of significance. Lastly, perhaps the analysis methods employed in this study were not robust enough to reveal subtle differences between the two conditions. The remainder of this chapter will focus on discussing the observed changes in variability and addressing each of the potential confounding effects among the other parameters observed.

Variability of Ankle Joint Inversion Eversion Angles

The only statistically significant findings of this study was that the heel windows condition resulted in greater variability in the ankle joint inversion eversion curve when compared to the shoe markers conditions. There are several potential explanations for this change. Perhaps the most obvious explanation is the cutting of the heel windows weakened the mechanical properties of the shoe in a way which allowed greater movement of the foot. While no materials testing was performed in this experiment to quantify the effects of cutting the heel windows, this explanation seems unlikely for several reasons.

First, the heel windows method for placing markers on the rear foot has been frequently used in previous studies where materials testing was performed (Butler et al., 2006; Ferber, 2005; Williams et al., 2003; Williams et al., 2004). These studies indicated that the heel windows resulted in only a 10% decrement in heel counter stability. Whether such a small change is enough to seriously influence foot motion is unknown, however it appears unlikely. Secondly, the heel windows employed in this study were, on average, 2.13 cm^2 for the upper hole and 1.87 cm^2 for the lower hole. Stacoff et al. (1992) found significant changes in rear foot motion did not occur until the heel windows were 4.12 cm^2 for the lower window and 3.61 cm^2 for the upper window, values substantially larger than the ones used in this study. Lastly, it is debatable the extent to which the rigidity of the heel counter is actually responsible for controlling rear foot motion, as Van Gheluwe et al. (1995) demonstrated that rear foot motion occurred independently from the stiffness of the heel counter. When taken together, the results of these studies suggest it is unlikely any changes in rear foot motion were due only to the cutting of the holes in the heel counter of the shoe. A more likely explanation for the increased variability observed in the heel windows markers condition involves the relationship between the number of movement patterns available within the central nervous system and how the marker placement may reveal or mask these various options.

Though the exact mechanisms are not completely understood, it is thought that the general motor firing patterns responsible for producing human gait are established by central pattern generators located in the spinal cord (Duysens & Van de Crommert, 1998; MacKay-Lyons, 2002). These generators are heavily influenced by both supraspinal and afferent sensory nerve input, resulting in a wide range of individual movement patterns within the context of the larger central pattern (MacKay-Lyons, 2002; Van de Crommert, Mulder, & Duysens, 1998). It is thought that this variety of movement patterns allows for flexibility and adaptability within the system.

When the markers are placed on the exterior of the shoe they are all attached to the heel counter, which is essentially a rigid object. This means the markers have little ability to move independently from each other. Additionally, in theory, the deformations of the shoe during the running stance phase should generally be repeatable from stride to stride. When analyzed with markers placed in this fashion, the combination of these two conditions will suggest foot motion with very little variability between trials. However, this is most likely not an accurate measure of the actual amount of variability present in a given individual.

Using the heel windows method allows placement of the markers directly on the skin of the rear foot. This alleviates the restrictions of placing the markers on the shoe in that the three markers can move independently from each other, and are not constrained to move in the same pattern each stride. As seen in the results, the measured variability is increased when the markers are placed in this fashion. This marker placement allows a more accurate assessment of the variety of movement patterns available to an individual, a measurement with potentially important clinical implications

As previously discussed, the variety of available movement patterns allows for flexibility and adaptability within the system. Additionally, for cyclical motions such as running, this variety of movement patterns also helps avoid loading biological tissues in exactly the same manner every cycle. Some authors have suggested that there is an optimum amount of variability and common running injuries may be related to a decrease in the variability of movement patterns available to an individual (Hamill, van Emmerik, Heiderscheit, & Li, 1999; Heiderscheit, Hamill, & Van Emmerik, 2002; Miller, Meardon, Derrick, & Gillette, 2008; Stergiou, Harbourne, & Cavanaugh, 2006). Placing the markers on the shoe would mask any changes in variability, thereby potentially masking potential injury causing mechanisms.

The Subject Sample as a Potential Source of Non-Significance

There is the potential that the lack of significant results in this study resulted from a sample which is not representative of the general running population. Subjects were recruited into this study based on their desire or ability to participate, and their membership in local running clubs or groups, making the subjects in this study a sample of convenience and not truly a random sample. Additionally, most of the subjects were recruited from a university campus, thereby potentially biasing the sample in regards to age, number of year running, or injury history. Lastly, the subjects in this study were all recreational runners or recreationally competitive and none of them were highly trained or elite athletes.

Though these issues raise concerns, it is felt they had little influence on the external validity of the current study and are similar methods to what has previously been reported in the literature. For instance, in a study validating the use of the arch height index Williams and McClay (2000) examined fifty one recreational runners who volunteered from the surrounding community. The mean age of these subjects was 27.1 years and their mean arch height index was 0.316. Another study specifically using this arch height index method to establish reference data among recreational runners found a mean arch height index of 0.340 (Butler, Hillstrom, Song, Richards, & Davis, 2008). This method and the reference data presented in these two studies has since been used in numerous studies on runners as an established method and mean for comparing arch heights among individuals (Chang, Van Emmerik, & Hamill, 2008; Molloy et al., 2009; Pohl et al., 2009; Williams et al., 2003; Williams et al., 2004; Williams, McClay, & Hamill, 2001; Williams, McClay, Hamill et al., 2001; Zifchock, Davis, Higginson, McCaw, & Royer, 2009). In the current study the mean age of the subjects was 25.46 and the mean arch height index was 0.30, suggesting these subjects were representative of the mean foot structure found in the broader population of recreational runners examined by Butler et al. (2008) and Willams and McClay (2000).

It has been shown by Williams et al. (2001) that there are significant kinematic and kinetic differences between high arched and low arched runners. The subjects in this study had arch height indices ranging from 0.244 to 0.390, suggesting they represented a cross section of individuals with low, medium, and high arches. If only high or low arched runners had been used in this study it would have both potentially biased the results and limited the potential applications of the findings. However, this does not appear to have happened and the foot structure of the subjects does not appear to have played a role in the results observed in this study.

Kinematic and Kinetic Parameters

In general, reviews on the biomechanics of running suggest an individual lands with their rear foot in a slightly inverted position then everts until approximately midstance. The amount of eversion excursion usually falls within a range of approximately 10° to 20° with maximal eversion occurring between 20% and 40% of the stance phase (Edington et al., 1990; McClay, 1995; Novacheck, 1988). Experimental evidence supports these numbers. Summaries of several studies which reported the same kinematic and kinetic parameters as this study are shown in Table 11. Most of these studies compared these parameters in an injured and a healthy control group. When that scenario occurred, these values were taken from the healthy control group.

From this information it appears the results reported in this study are well within the ranges observed in other studies. Therefore, the lack of statistical significance between the shoe markers and heel windows marker methods is not due to kinematic or kinetic values unique to the subjects in this study and other reasons must be considered.

Uncontrolled Factors

As a result of the methods used in this particular study there are some uncontrolled factors which could potentially affect the internal validity of the study and help explain the lack of statistical significance between the shoe markers and heel windows marker conditions. For instance, the subjects ran at a self selected pace for both trials. Most of the parameters examined in this study have been shown to vary with running speed (Cavanagh & Lafortune, 1980; Pohl, Messenger, & Buckley, 2007). Therefore, large differences in running speed between the shoe markers and heel windows markers conditions might cause significant differences in the kinematic and kinetic parameters. However, as shown in Table 4 running speeds were not significantly different between the two marker conditions, suggesting differences or lack of differences between the two conditions cannot be simply attributed to running speed.

Table 11. Representative kinematic and kinetic results reported from several studies

Authors	N	Eversion Excursion (°)	Time to Peak Eversion (s) or (% stance)	Maximal Instantaneous Eversion Velocity (°/s)	Maximal Instantaneous Vertical Loading Rate (BW/s)
(McClay & Manal, 1998a)	9	12.70 (± 4.1)	0.09 s (± 0.026)	-	-
(Pohl et al., 2009)	25	14.90 (± 4.0)	52.6 % (± 6.0)	-	82.9 (± 18.7)
(Willems et al., 2007)	334	18.06 (± 4.53)	47.18% (± 12.14)	447.27 (± 131.71)	133.30 (± 45.98)
(Pohl et al., 2008)	30	8.80 (± 4.1)	-	-	83.80 (± 23.20)
(Milner, Ferber et al., 2006)	20	-	-	-	79.65 (± 18.81)
(Mundermann et al., 2003)	20	16.0 (± 2.3)	-	464.7 (± 155.2)	-
(McClay & Manal, 1997)	9	12.7 (± 3.5)	0.11 s (± 0.05)	-	-

Note: Some studies report percent stance at maximal eversion in seconds (s) and some report it as a percent of stance phase (%). This convention has been followed in the table above.

Another uncontrolled factor in this study was the fact subjects wore their own running shoes instead of a standardized laboratory shoe. Some of the subjects in this study were wearing stability shoes while others wore cushioning shoes. In theory stability shoes are designed to control rear foot motion during running, suggesting the integrity of these shoes could potentially be more susceptible to modifications such as cutting holes in the heel windows. While this may affect the internal validity of the study it was felt this was an important point since the shoes worn for this study are the shoes the individual wears on a daily basis while running.

It has been shown that due to the repetitive compressive forces over time the foam used to construct a running shoe slowly breaks down, losing its cushioning properties and changing the pressure distribution under the foot (Verdejo & Mills, 2004). Additionally, it has been shown that hardness of the shoe midsole has significant effects on the kinematics and kinetics of the foot during the stance phase of running gait (De Wit, De Clercq, & Lenoir, 1995; Hamill, Bates, & Holt, 1992; Stergiou & Bates, 1997). Given this information it was felt that placing a subject in a controlled laboratory shoe with different foam characteristics and hardness then what they were used to had the potential to artificially alter their kinematics and kinetics and lead to artificial conclusions about differences in foot motion between the two marker conditions.

However, it appears having subjects wear their own shoes did not have any effect on the results of this study. There was no consistent pattern between changes in any of the parameters measured in this study and the type of shoe the subjects wore. For instance, all subjects wearing stability shoes did not respond the same for any the parameters measured in this study when switching from the shoe markers to the heel windows markers. This further supports the argument presented earlier in this chapter that the increase in variability observed in the ankle joint inversion eversion curve in the heel windows condition was not a result of the modifications to the shoe.

The last uncontrolled factor between the subjects in this study was the actual size of the heel windows cut into the shoes. While every attempt was made to make the heel windows the same size on each shoe, the windows were cut by hand resulting in small, unavoidable variations between shoes. However, the impact of these small variations appears negligible. As previously mentioned, the heel windows used in this study were substantially smaller than the large windows used by Stacoff et al. (1992) and therefore should not have affected the motion of the rear foot. Though it was not tested statistically, a visual examination of the data presented in Tables 3, 5, 6, 7, 8, and 9 shows no clear relationship between either the magnitude or direction of changes in any of the measured parameters and the size of the heel windows. Therefore it is unlikely that any small variations in heel window sizes resulted in the lack of statistical significance observed in this study.

Data Analysis Methods

The original hypothesis of this study suggested that the changes from the shoe markers to heel windows marker conditions should be similar across both subjects and feet within a subject. Based on bone pin studies by Stacoff et al. (2000; 2001) and Reinschmidt et al. (1997) it was hypothesized the shoe markers would overestimate the amount of rear foot motion compared to the heel windows markers, a claim commonly found in studies citing differences between shoe and foot motion (Butler et al., 2006; Butler, Hamill, & Davis, 2007; Dierks et al., 2008; Milani & Hennig, 2000; Pohl, Messenger, & Buckley, 2006). However, as the information in Tables 4 through 9 and Figures 24 through 28 illustrate, this clearly was not the case in this study. For most variables there is an even split with approximately half the feet responding one way and the other half responding in the opposite direction. For instance, Figure 24 shows that half the feet demonstrated an increase in eversion excursion with the heel windows compared to the shoe markers while the other half of the feet demonstrated a decrease. Similar patterns are shown in Figures 25, 26, 27, and 28 for time to peak eversion, percent stance at peak eversion, maximal instantaneous eversion velocity, and maximal instantaneous vertical loading rates.

This individualized response to the two marker conditions can also be seen in the results of the original bone pin studies. For example, the eversion results from the Reinschmidt (1997) study are shown in Figure 30. While most of the subjects do demonstrate a decrease in eversion when measured with the bone pins, subject three does not. For subject three the bone pins actually resulted in greater eversion. The authors have suggested this may be due to the shoe markers being placed in a more inverted position, thereby artificially shifting the shoe marker curve, however there is the possibility that this subject simply responded differently than the other subjects.

Regardless of the marker alignment issues, a more detailed examination of Figure 30 shows that the differences between the shoe markers and bone pin markers were unique to each individual and not standardized across subjects. Similar individualized differences were reported in a study which also used bone pins to compare differences between barefoot and shod running, whose rear foot eversion curves can be seen in Figure 31 (Stacoff et al., 2000). While not directly comparing the differences in rear foot motion between shoe markers and bone pin markers, these results also reinforce the idea that each subject or foot may respond differently to the different marker conditions.

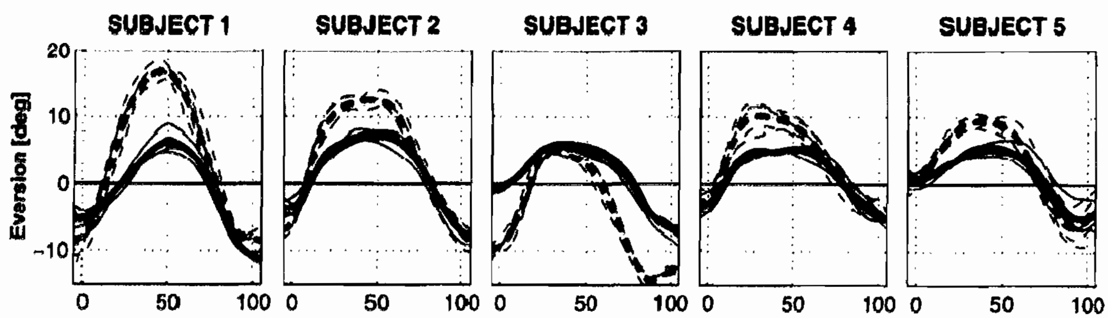


Figure 30. Rear foot eversion curves from a study by Reinschmidt et al. (1997) examining differences in rear foot motion during running as measured with intracortical bone pins and external shoe based markers. The dashes lines represent the shoe based markers and the solid lines represent the bone pin markers.

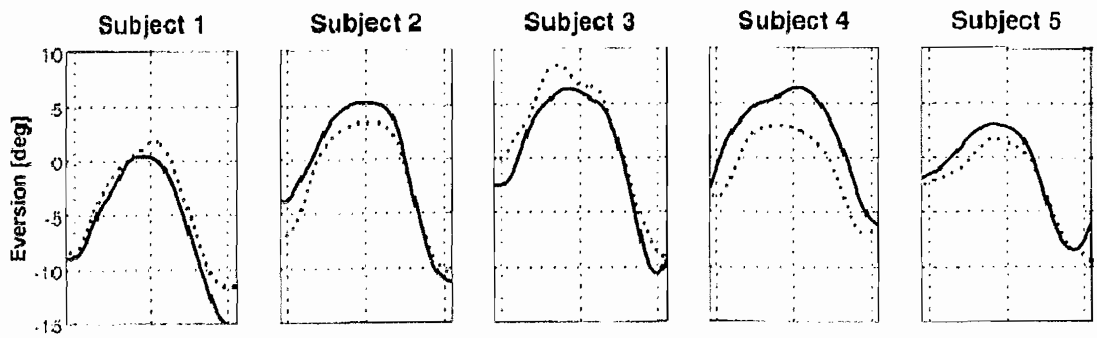


Figure 31. Rear foot eversion curves from a study by Stacoff et al. (2000) examining differences in rear foot motion between barefoot and shod running. The dotted lines represent the shod condition while the solid lines represent the barefoot condition.

The results of the current study are similar to the two bone pin studies in that each foot demonstrated a unique, individualized change in the heel windows marker condition compared to the shoe markers. For instance, subject R05's left foot demonstrated a 28.5% decrease in eversion excursion between the two conditions while subject R12's left foot demonstrated a 32% increase in the same parameter. Similar magnitude percent changes can be seen in other subjects and parameters such as R06's right foot demonstrating a 26% increase in the time to peak eversion while subject R11's right foot showed a 20% decrease in the same parameter. However, the dependent observations *t* test used for statistical analysis in this study assesses differences between the mean scores of the group in the shoe markers and heel windows markers conditions. With some feet demonstrating increases in the parameters measured while others demonstrated decreases, the overall differences wash out and the result indicates there are not statistically significant differences between the two marker conditions.

Yet, as discussed above, some feet demonstrated changes of up to 32% between the two marker conditions, a number which suggests significant differences are present between the two marker conditions. If there are individually significant changes which get washed out with a group analysis then the particular method of data analysis employed in this study may not be suitable for this purpose. A more useful method would examine each subject individually and determine on a case by case basis whether there were statistically significant differences between the two marker conditions. This approach is called a single subject analysis and the theory and methods behind its application are presented in the next chapter.

CHAPTER V

SINGLE SUBJECT DESIGN

Background and Rationale

The results and discussion detailed in Chapters III and IV indicated that besides variability in the ankle frontal plane joint angles, there were no significant differences between the shoe markers and the heel windows markers. As discussed in Chapter IV, this lack of statistically significant differences may be due to the methods utilized to analyze the data. The methods used in the previous analysis were a traditional group analysis where the data from multiple trials of each subject was averaged to indicate an average performance for that individual. These representative individual responses were then averaged to yield an average group response. The statistical analysis was performed on these average group responses to the two marker conditions and found no significant differences between conditions.

This approach is wide spread and common in both running studies and biomechanics research in general, and a prime example of a traditional group approach. Review articles and book chapters suggest traditional group design evolved from a desire to compare individuals to some "average" level, which was assumed to be desirable (Bates, 1996; Bates, James, & Dufek, 2000). From a research perspective this is enticing since it is thought to allow the generalization of the results to a larger population.

Furthermore, any errors in measurement or individual variation can more or less be overcome with a large enough sample size.

However, as Bates (1996) points out, with the wide range of natural variation present in each individual, rarely does any one subject ever perfectly conform to the "average" parameters. Over the years, many studies on running have confirmed this, finding wide interindividual variations in the various parameters they measured (Bates, Osternig, Mason, & James, 1979; De Wit, De Clercq, & Aerts, 2000; McClay & Manal, 1999; Van Gheluwe et al., 1995). This same analysis holds true within the individual since the numerous movement patterns available to an individual's neuromuscular system mean rarely will any single trial perfectly match their "average" performance. As with the interindividual variability, several studies have also cited high levels of intraindividual variability in runners for both kinematic and kinetic parameters (Bates, Osternig, Sawhill, & James, 1983; Cavanagh & Lafortune, 1980; Devita & Skelly, 1990).

With a group statistics approach variability in a sample can be dealt with by incorporating a larger sample size. However, this approach will not work when the variability is a result of each individual using different, unique neuromuscular strategies to accomplish the given task (Bates et al., 2000). These different strategies are a result of the numerous degrees of freedom in the neuromuscular system and how the body responds to any constraints which may influence the movement (Bates et al., 2000). The constraints could be from environmental sources, due to anatomical variations, or influenced by sensory feedback during the movement task, however the end result is that rarely will any two trials be identical, both within and between individuals (Bates et al., 2000). Evidence of the usage of different movement strategies have been previously observed in athletic tasks such as running and landing (Caster & Bates, 1995; Dufek, Bates, Stergiou, & James, 1995; Reinschmidt & Nigg, 1995). In these cases, when individuals are using different strategies to solve similar movement problems, the variability will remain, no matter how large the sample size.

This interindividual and intraindividual variability is of particular concern from the viewpoint of trying to assess factors related to overuse injuries in runners. For instance, imagine a prospective scenario where one wants to predict which individuals are at risk for different injuries. If an individual resembles the normal "average" one might conclude they are healthy and have a relatively low risk for and injury. A similar scenario could be envisioned for a retrospective approach where one wants to identify biomechanical factors which may have caused a particular injury. If all the subject's kinematic and kinetic measurements are within "average" ranges one might not be able to make any conclusions about contributing factors to the injury. In both scenarios, when using a group average approach, interindividual and intraindividual variability would mask any subtle variations in the subject's anatomic alignment or changes in their kinematics and kinetics which may be related to an injury. In these situations a single subject design approach with its focus on changes and variability within the individual rather than the group average would be a more effective approach.

Single subject designs traditionally have been used in the social sciences to examine topics such as behavioral interventions in education settings or effects of different teaching strategies (Richards, Taylor, Ramasamy, & Richards, 1999). In these studies multiple baseline measurements are taken on a single subject, some intervention is then imposed, and multiple measures are recorded after the intervention. The researchers treat each measurement as an independent observation and examine if there were significant changes in the pre and post intervention observations. This particular approach is commonly referred to as an AB design, however there are numerous other approaches which are also commonly used such as ABA, or ABABA (Richards et al., 1999). For behavioral studies these repeated measurements may take place over the course over several days, however, for application to biomechanics studies, one could construe multiple consecutive foot strikes during running as the repeated measurements (Bates et al., 2000). For instance, if one was evaluating the effects of an orthotic on rear foot motion during running, a subject could perform twenty trials in a normal shoe, have the orthotic inserted, and then perform twenty additional trials. The researchers would then examine differences between the trials to determine the effects of the orthotic intervention.

As mentioned above, variability is an important consideration in any study, whether a traditional group or a single subject design. However, one of the appeals of the single subject methodology is how it accommodates intersubject and intrasubject variability compared to traditional group designs. Bates et al. (1992) and Dufek et al. (1995) have performed several computer modeling studies to explore how variability affects the results of a statistical analysis with both group and single subject designs. Their results suggest that, compared to single subject designs, group design methods are more susceptible to interindividual and intraindividual variability. When such variability is present there is a greater chance of failing to appropriately reject the null hypothesis. However, their results also indicate that, compared to single subject analysis methods, group designs require far fewer trials per subject to achieve high levels of statistical power. This has important implications for the design of any study considering the use of single subject analysis methods.

One of the biggest considerations, and arguments against the use of single subject methods, is that it violates several important guiding principals of statistical analysis, including the independence of observations and the normality of the distribution of the sample. In theory, for traditional group statistical analysis the observations of the dependent variable are considered to be independent. In single subject analysis the observations are repeatedly taken from the same individual. In a situation such as running they are repeated foot strikes that occurred within a close time span. Being from one individual, potentially in a short time span, there is the potential that one observation influences the next and they are not truly independent observations.

Bates et al. (1996; 2000) argue, based on both computer simulations and actual data collected in their lab, that the assumption of independence is not usually violated in single subject analysis and these authors suggest calculating an autocorrelation coefficient for the data to demonstrate this fact. In the autocorrelation coefficient one calculates the correlation coefficient between consecutive observations in a sample, such as between 1-2, 2-3, 3-4, etc. If the autocorrelation coefficient is high, or significant, then the data should not be considered as independent observations (Bates, 1996; Bates et al., 2000; Richards et al., 1999). The auto correlation coefficient can easily be calculated by most commercially available statistical programs, or can be calculated by hand using the proper formulas and spread sheet software such as Microsoft Excel.

The second argument against the use of statistical analysis with single subject design, that it violates assumptions of normality, is also handled in a fairly easy fashion. Bates(1996; 2000) suggests this assumption is not an issue since many statistical tests, such as the *t* test are robust to violations of normality. However, a researcher can easily get an idea of the normality of the data set using a statistical test such as the Shapiro-Wilk test (Bates et al., 2000).

When the two conditions of independence of observations and normality of data have been confirmed, most of the standard statistical tests can then be applied in a single subject design setting. In addition to statistical analysis, single subject design also lends itself to graphical analysis of the data (Richards et al., 1999). In this approach a researcher might plot the data and note any apparent changes in magnitude or trends between the conditions tested. This can be an especially valid method when there are numerous trials to examine and the intraindividual variability is low (Richards et al., 1999).

As this discussion has indicated, single subject analysis can be a powerful tool when examining data where subjects may present with individualized motor strategies to a common movement task. The group analysis discussed in Chapters II, III, and IV suggested there were not significant differences between the shoe markers and heel windows marker conditions. However, as previously mentioned, several individuals demonstrated rather large percent changes between the two conditions. Therefore, the purpose of this next chapter is to reanalyze the data presented in Chapter III using a single subject approach. Again, each foot was analyzed separately. The hypothesis for this single subject approach is that feet with large percent changes from the shoe markers condition to the heel windows marker condition in the group analysis will show significant differences under a single subject approach. Furthermore, it is hypothesized that, under a single subject approach each foot would respond uniquely to the two marker conditions, with some demonstrating significant differences between conditions while others demonstrate no significant differences between conditions.

Single Subject Design Methods

As the name implies, a single subject design will consider all the trials done by one individual. Therefore, a large number of trials are needed. Bates et al. (1979) suggested that, for a traditional group analysis, a minimum of eight trials were required to produce a reasonable picture of an individuals kinematics and kinetics while running. Given the statistical concerns raised by Bates et al. (1992) and Dufek et al. (1995), it seems the number of trials is potential concern for single subject analysis, however in a discussion about single subject analysis methods Richards et al. (1999) indicate adequate results can be achieved with a minimum of eight trials per condition. Therefore, only feet which had at least eight trials for both the shoe markers and heel windows markers conditions were included in the single subject analysis.

Richards et al. (1999) suggest graphical analysis as a potentially powerful tool for single subject design. In this method the results from every individual trial both conditions are plotted against time. For instance, if the particular value being examined was eversion excursion, then the eversion excursion would be plotted for each consecutive trial the subject completed. This scatter plot will have eversion excursion on the Y axis with the trial number on the X axis. Any significant differences between the two conditions might be observed as changes in the level or trend of the trials from one condition to the next (Richards et al., 1999). This may allow researchers to identify changes which may be meaningful, yet not statistically significant, and can be a valuable option when certain conditions required for statistical analysis are not met in the data (Richards et al., 1999).

Graphical evaluation was undertaken for all trials for all feet meeting the minimum number of trials condition. The variables analyzed for the single subject design were eversion excursion, percent stance at peak eversion, maximal instantaneous eversion velocity, and maximal instantaneous vertical loading rate. For each foot, all these values from each trial, under both conditions, were plotted on the Y axis, with the trial number of on the X axis. The graphs were inspected visually for apparent changes in trend or levels of the particular variable between the shoe markers and the heel windows markers.

However, graphical analysis does have several limitations, with the foremost limitation being the subjective nature of the analysis (Richards et al., 1999). Two different researchers may not agree on what level of difference between the two conditions constitutes a meaningful change. Potentially worse, they may not even agree if there is a difference between the two conditions, especially if there is substantial variability present in the data, as the variability may mask changes in trend or level between conditions (Richards et al., 1999). Therefore, in this study, all the data were also analyzed using statistical analysis.

Before statistical analyses were carried out the data were examined to ensure adherence to standard assumptions of normality and independence of observations. Evaluation of normality of the distributions for both the shoe markers trials and heel windows trials were done with a Shapiro-Wilk test (Bates et al., 2000). Distributions were assumed to be normal when the resulting W value was greater than 0.05, as this indicates a 95% confidence that the data is normally distributed (Bates et al., 2000). Independence of observations was examined by calculating autocorrelation coefficients for all the trials of each foot for each variable of interest. Coefficients for the shoe markers trials and heel windows marker trials were calculated separately. Autocorrelation coefficients were calculated using Statistical Package for the Social Sciences (SPSS), version 18.0 (SPSS, Inc., Chicago IL), with $\alpha = .05$. In this manner a p value less than .05 indicates statistically significant autocorrelation was present. For all variables only autocorrelation of the first lag was calculated as it was suggested this should be sufficient to detect autocorrelation in the data (Bates et al., 2000; Richards et al., 1999).

An independent samples *t* test was used to compare the trials with the shoe markers to the trials with the heel windows markers. Though these measurements were repeated measurements on the same foot, the whole theoretical basis for single subject design assumes they were independent observations, thereby contraindicating the use of repeated measures statistical methods such as a paired observations *t* test (Bates, 1996). All independent *t* tests were conducted using SPSS (SPSS, Inc., Chicago, IL) statistical analysis software and were performed with $\alpha = .05$, set a priori.

Statistical significance of results was also evaluated using a method specifically designed to work with single subject analysis called Model Statistics. This method was developed in the University of Oregon Biomechanics Laboratory by Dr. Barry Bates and has been frequently used in analysis of single subject design (Bates et al., 1992; Bates et al., 2000; Dufek & Bates, 1991; Dufek, Bates, & Davis, 1995; Dufek, Bates, Stergiou et al., 1995; Stergiou & Bates, 1997). In this method the absolute difference of the means of the trials under both conditions are compared to a critical value. If the absolute difference in means is larger than the critical value then there is a statistically significant difference between the two conditions and if the absolute difference in means is less than the critical value there is not a statistically significant difference between the two conditions. The critical value is calculated based on a test statistic multiplied by the weighted standard deviation of the trials from the two conditions. The weighted standard deviation is calculated as:

$$\mathrm{SD_w} = \sqrt{\frac{\mathrm{SD_1}^2 + \mathrm{SD_2}^2}{2}} \tag{1}$$

SD_1 is the standard deviation of all the trials from the shoe markers condition and SD_2 is the standard deviation of all the trials from the heel windows markers condition. The critical statistic was selected based on the table of critical statistics found in Bates et al. (1992). This table is reproduced in Table 12. Critical statistics are provided for various sample sizes and for α levels of 0.1, 0.05, and 0.01, respectively. The sample size was selected based on the smallest number of trials for either the shoe markers or the heel windows markers. For instance if a foot had fifteen good trials in the shoe markers and only eleven good trials in the heel windows condition, then the sample size used for the Model Statistics analysis was eleven. This statistical analysis was carried out for each of the four variables which had enough trials under both conditions to identify statistically significant differences between the shoe markers condition and heel windows marker conditions on an individual subject basis.

Table 12. Critical statistic values for the Model Statistics analysis.

Sample Size	α = 0.10	α = 0.05	α = 0.01
3	1.3733	1.6533	2.2133
4	1.2643	1.5058	1.9867
5	1.1597	1.3662	1.7788
6	1.0629	1.2408	1.6044
7	0.9751	1.1306	1.4623
8	0.8964	1.0351	1.3473
9	0.8270	0.9536	1.2542
10	0.7673	0.8857	1.1776
11	0.7172	0.8307	1.1129
12	0.6757	0.7867	1.0581
13	0.6415	0.7516	1.0117
14	0.6132	0.7234	0.9720
15	0.5896	0.7001	0.9375
16	0.5695	0.6798	0.9070
17	0.5522	0.6618	0.8796
18	0.5371	0.6458	0.8548
19	0.5237	0.6311	0.8318
20	0.5114	0.6175	0.8102

Note: Table of critical statistic values for Model Statistics from sample sizes of three to sample sizes of twenty. From Bates et al. (1992).

CHAPTER VI

SINGLE SUBJECT RESULTS

Subjects

A criterion of at least eight good trials for both the shoe markers and heel windows markers conditions was required for a foot to be included in the single subject analysis. Most of the feet achieved this for the kinematic variables of eversion excursion, percent stance at peak eversion, and maximal instantaneous eversion velocity. Much fewer achieved this benchmark for the kinetic parameter of maximal instantaneous vertical loading rate. However, when this value was met, most feet had more trials than the required minimum.

Overall, of the original twenty feet used in the group analysis, seventeen were used for the single subject analysis for the kinematics and ten were used for the single subject analysis for the kinetics. Seven feet had enough trials to be used to analyze the kinematics but not enough trials to analyze the kinetics, so only the kinematics were analyzed in these feet. These data are summarized in Table 13.

Table 13. Number of trials used in the single subject analysis for both the kinematic and kinetic variables.

Subject	Number of Kinematic Trials	Number of Kinetic Trials
R03 R	11	9
R05 R	14	8
R06 L	20	-
R06 R	19	9
R07 L	20	11
R07 R	19	11
R08 L	21	-
R08 R	21	10
R10 R	8	-
R11 L	11	-
R11 R	9	-
R12 L	11	-
R13 L	20	14
R13 R	20	18
R14 L	19	12
R14 R	19	16
R15 L	11	-

Note: The number of trials refers to the smaller of the shoe markers or heel windows markers conditions. For instance, if the shoe markers condition had 15 good trials and the heel windows condition only had 11 good trials then the number of trials used for the analysis was 11.

Graphical Results

As indicated above, seventeen feet had enough trials in both the shoe markers and heel windows markers to analyze their kinematics. The three kinematic variables being evaluated were eversion excursion, percent stance at peak eversion, and maximal instantaneous eversion velocity. Seventeen feet with three graphs each means there were fifty one graphs for the single subject analysis of the kinematic variables. There were only ten feet with enough trials under both marker conditions to analyze the kinetic parameter of maximal instantaneous vertical loading rate. This means, in total, the graphical analysis produced sixty one graphs.

Of these sixty one, seventeen graphs indicated significant differences between the shoe markers and the heel windows markers conditions. Two examples of these graphs are shown in Figures 32 and 33. As discussed previously, one of the issues with a graphical analysis is that the combination of the subjective nature of their interpretation and variability in the parameters measured can make them difficult to interpret. An example of this difficulty can be seen in Figures 34 and 35 which show examples of situations where the statistical analysis indicated significant differences between conditions while the graphical analysis may or may not demonstrate significant differences between conditions.

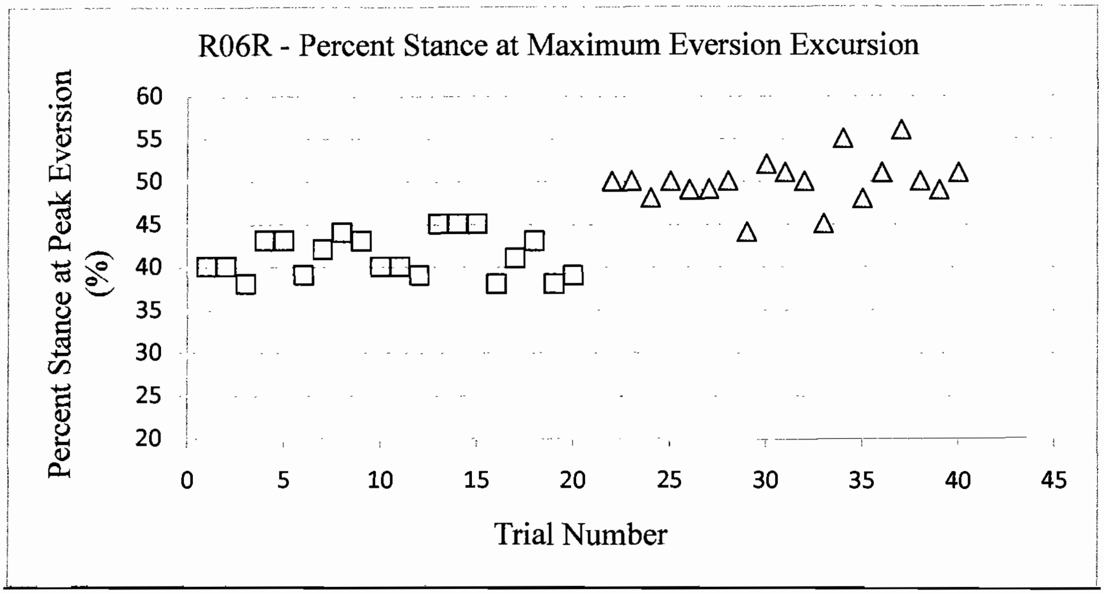


Figure 32. Example of a graphical analysis which demonstrated significant differences between shoe markers and heel windows markers for the percent stance at maximum eversion excursion. Graph shows the percent stance at which peak eversion occurred for subject R06R. The squares are the shoe markers trials and the triangles are the heel windows trials.

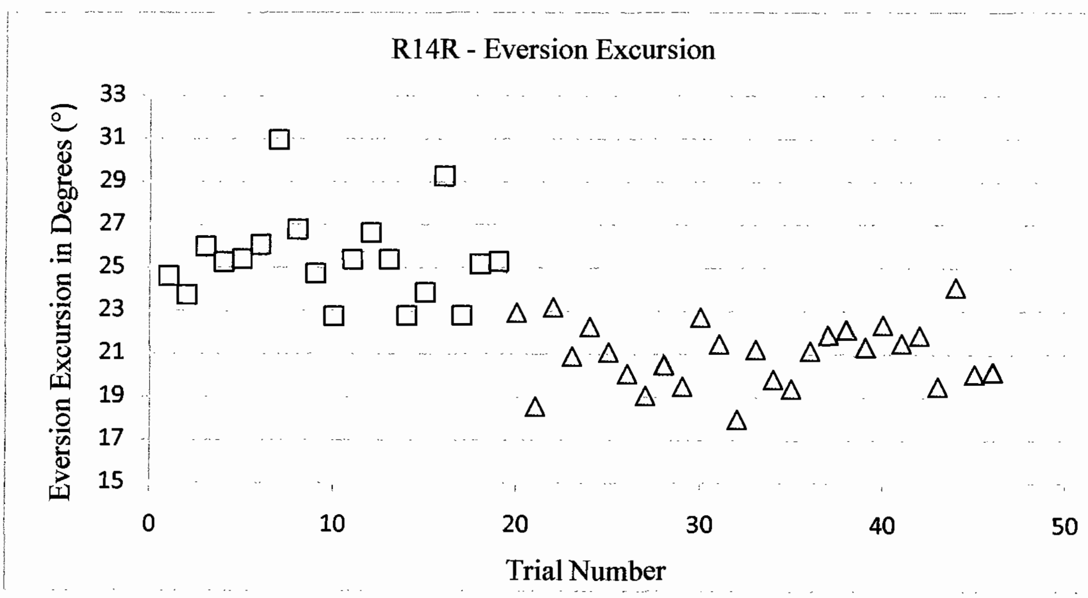


Figure 33. Example of a graphical analysis which demonstrated significant differences between shoe markers and heel windows markers for eversion excursion. Graph shows the eversion excursion for subject R14R. The squares are the shoe marker trials and the triangles ad the heel windows trials.

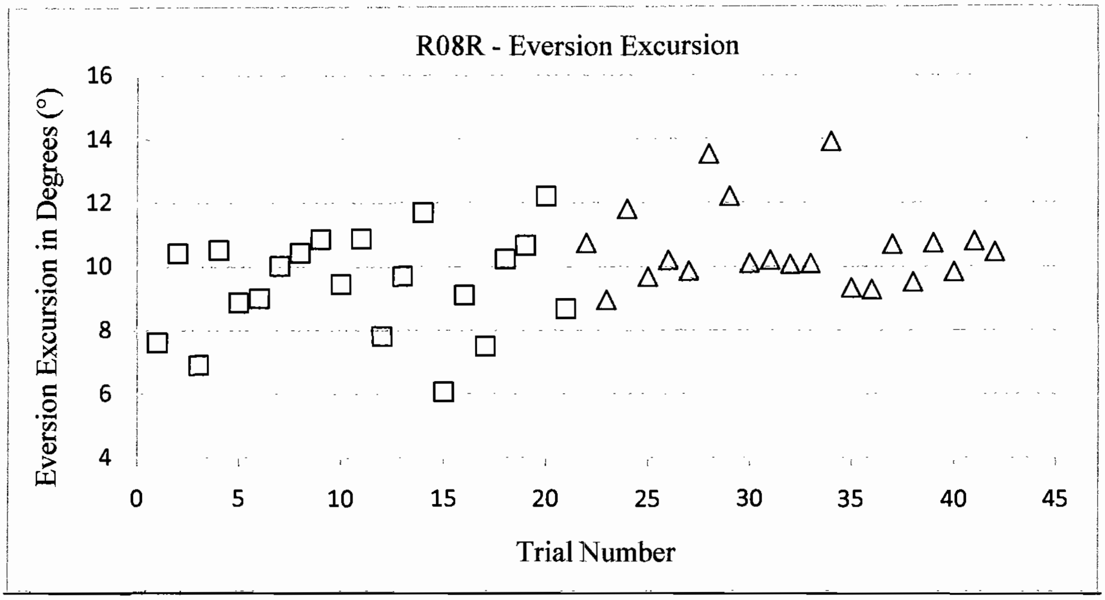


Figure 34. A graphical example where there may or may not be a significant difference between the two marker conditions. This graph shows eversion excursion in degrees for subject R08R. The squares represent the shoe marker trials and the triangles represent the heel windows trials.

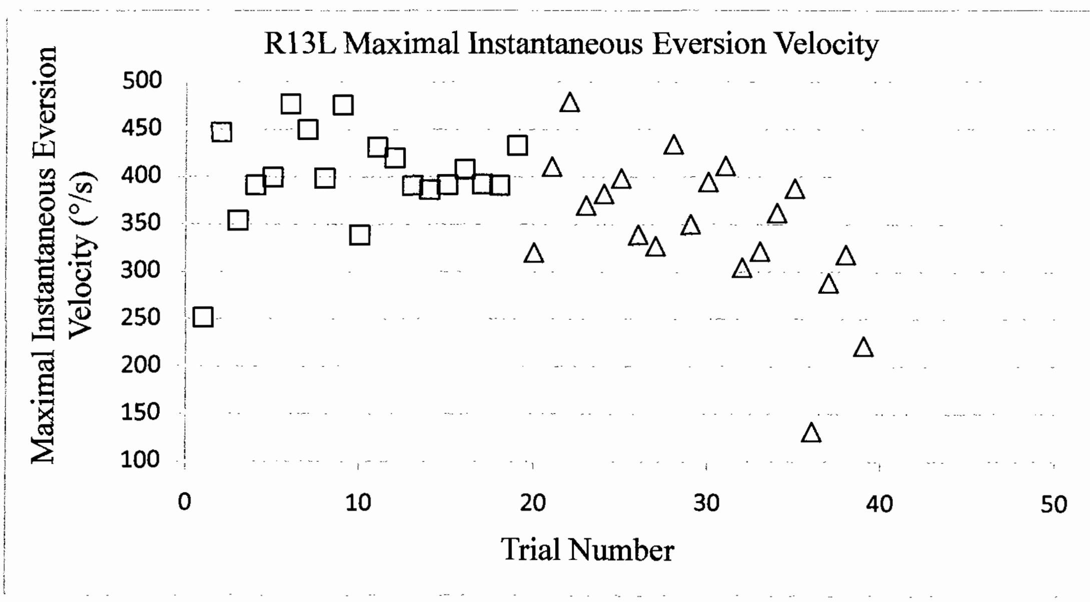


Figure 35. A second graphical example where there may or may not be a significant difference between the two marker conditions. The graph shows maximal instantaneous eversion velocity in degrees per seconds (°/s) for subject R13L. The squares represent the shoe markers trials and the triangles represent the heel windows markers trials.

Statistical Analysis Results

Seventeen feet had autocorrelations assessed for all trials under the shoe markers condition and the heel windows marker condition for the three kinematic variables. This means there were one hundred and two autocorrelation analyses performed for the kinematic variables. Ten feet had autocorrelation analyses for both the shoe markers and heel windows markers conditions for the one kinetic parameter. This added twenty additional autocorrelation analyses, bringing the total to one hundred twenty two autocorrelation analyses.

The results of the tests for autocorrelation are shown in Tables 14 and 15. Of the one hundred twenty two autocorrelation analyses performed only five, or four percent demonstrated statistically significant autocorrelations. Of the one hundred twenty two Shapiro-Wilks tests for normality, eleven, or nine percent, returned results with $p < .05$, suggesting they were not a normal distribution.

The results of the individual *t* tests and the Model Statistics Analysis can be seen in Tables 16 through 19. According to both the independent *t* test and Model Statistics methods, ten of the seventeen feet showed significant differences between the shoe markers and heel windows markers for eversion excursion (Table 16). Ten feet also showed significant differences between conditions for the percent stance at which peak eversion occurred (Table 17). For maximal instantaneous eversion velocity only seven feet demonstrated significant differences between conditions (Table 18) and for maximal instantaneous vertical loading rate only two feet demonstrated significant differences between conditions (Table 19). In all instances where the Model Statistics method indicated there was a significant difference between conditions for that foot, the independent *t* test also yielded a *p* value less than .05, indicating good agreement between these two statistical analysis methods.

Table 14. Autocorrelation coefficients for the three kinematic variables for both the shoe markers and the heel windows markers conditions.

	Eversion Excursion		Percent Stance at Peak Eversion		Maximal Instantaneous Eversion Velocity	
Subject	Shoe Markers	Heel Windows	Shoe Markers	Heel Windows	Shoe Markers	Heel Windows
R03 R	.215	.186	-.011	-.441	.234	.117
R05 R	-.553 *	.372	.172	.017	-.355	.306
R06 L	.160	-.330	-.167	.219	-.211	.321
R06 R	.252	.012	.100	-.296	.066	-.169
R07 L	-.232	-.243	.215	-.224	-.181	.047
R07 R	.095	-.143	-.259	-.337	.223	-.266
R08 L	.039	.036	.303	.201	-.072	-.118
R08 R	-.251	-.156	.318	-.174	.174	-.096
R10 R	-.080	-.355	.426	.226	-.086	.551
R11 L	-.327	.267	-.180	-.011	-.316	.293
R11 R	.212	-.104	-.081	.057	.048	-.099
R12 L	-.006	-.119	-.241	.651 *	.235	.408
R13 L	.392	.430 *	.420 *	.001	-.246	.191
R13 R	-.020	.050	-.159	.300	.020	.111
R14 L	.092	.108	-.139	-.152	-.069	-.043
R14 R	.014	-.244	-.074	-.001	.083	.154
R15 L	-.347	.404	.263	-.253	-.352	.397

Note: Autocorrelations which were significant at the $p < .05$ level are indicated with an asterisk (*).

Table 15. Autocorrelation coefficients for the maximal instantaneous vertical loading rate for both shoe markers and heel windows markers conditions.

	Maximal Instantaneous Vertical Loading Rate	
Subject	Shoe Markers	Heel Windows
R03 R	.283	.122
R05 R	.078	.349
R06 L	-	-
R06 R	.149	-.380
R07 L	.333	-.177
R07 R	.048	.047
R08 L	-	-
R08 R	.351	.153
R10 R	-	-
R11 L	-	-
R11 R	-	-
R12 L	-	-
R13 L	.591 *	.130
R13 R	.119	.116
R14 L	.064	.018
R14 R	-.021	-.027
R15 L	-	-

Note: Autocorrelations which were significant at the $p < .05$ level are marked with an asterisks (*). Feet which did not have enough trials to analyze the kinetic parameters in a single subject design are indicated with a dash (-).

Table 16. Independent *t* test and Model Statistics results for eversion excursion.

Subject	Percent change from group design	*t* test *p* value	Critical Difference	Absolute Mean Difference	Significant?
R03 R	0.60	.793	1.560	0.183	NO
R05 R	-7.33	.256	1.685	0.904	NO
R06 L	4.54	.850	1.056	0.101	NO
R06 R	**10.88**	**.005 ***	**1.346**	**2.021**	**YES**
R07 L	**-17.06**	**.001 ***	**0.617**	**2.027**	**YES**
R07 R	-4.22	.367	1.184	0.541	NO
R08 L	**43.44**	**< .001 ***	**1.151**	**3.226**	**YES**
R08 R	**12.21**	**.018 ***	**0.901**	**1.115**	**YES**
R10 R	**23.57**	**.001 ***	**2.097**	**4.048**	**YES**
R11 L	-7.71	.459	2.210	0.840	NO
R11 R	**-22.95**	**.016 ***	**2.876**	**3.998**	**YES**
R12 L	**32.26**	**< .001 ***	**1.375**	**3.304**	**YES**
R13 L	-13.05	.108	2.068	1.743	NO
R13 R	**7.32**	**< .001 ***	**2.141**	**2.516**	**YES**
R14 L	**-11.87**	**.026 ***	**0.975**	**1.253**	**YES**
R14 R	**-16.85**	**<.001 ***	**1.141**	**5.547**	**YES**
R15 L	3.38	.792	0.001	0.005	NO

Note: Feet where the independent *t* test indicated there were significant differences between shoe and heel windows markers are indicated with an asterisks (*). Trials where the Model Statistics indicated a significant difference between the two conditions are highlighted in bold text.

Table 17. Independent *t* test and Model Statistics results for percent stance at which maximal eversion occurs.

Subject	Percent Change from Group Design	*t* test *p* value	Critical Difference	Absolute Mean Difference	Significant?
R03 R	1.63	.318	2.551	1.764	NO
R05 R	**11.40**	**< .001 ***	**1.593**	**5.695**	**YES**
R06 L	**16.47**	**.002 ***	**2.219**	**3.838**	**YES**
R06 R	**20.93**	**< .001 ***	**1.651**	**8.561**	**YES**
R07 L	**-17.75**	**< .001 ***	**0.617**	**3.251**	**YES**
R07 R	-1.97	.265	2.551	1.456	NO
R08 L	**8.76**	**< .001 ***	**19.641**	**53.542**	**YES**
R08 R	**8.00**	**.003 ***	**17.687**	**28.222**	**YES**
R10 R	**6.69**	**.018 ***	**2.708**	**3.425**	**YES**
R11 L	9.49	.099	2.505	2.189	NO
R11 R	-12.13	.059	5.572	5.589	NO
R12 L	-2.30	.814	5.614	0.616	NO
R13 L	10.51	.109	2.883	2.453	NO
R13 R	**-11.88**	**< .001 ***	**1.795**	**4.907**	**YES**
R14 L	**-7.43**	**.018 ***	**1.418**	**1.421**	**YES**
R14 R	**6.38**	**.012 ***	**2.723**	**3.175**	**YES**
R15 L	8.36	.289	2.354	1.280	NO

Note: Feet where the independent *t* test indicated there were significant differences between shoe and heel windows markers are indicated with an asterisks (*). Trials where the Model Statistics indicated a significant difference between the two conditions are highlighted in bold text.

Table 18. Independent *t* test and Model Statistics results for maximum instantaneous eversion velocity.

Subject	Percent Change from Group Design	*t* test *p* value	Critical Difference	Absolute Mean Difference	Significant?
R03 R	-3.83	.960	35.941	0.777	NO
R05 R	-5.71	.283	19.623	10.200	NO
R06 L	-4.38	.431	50.103	20.803	NO
R06 R	**-21.84**	**< .001 ***	**30.147**	**70.383**	**YES**
R07 L	-4.72	.189	27.008	18.390	NO
R07 R	-2.43	.708	15.833	2.993	NO
R08 L	**31.39**	**.009 ***	**1.742**	**3.281**	**YES**
R08 R	**-14.86**	**.004 ***	**3.016**	**4.762**	**YES**
R10 R	22.84	.063	59.877	58.935	NO
R11 L	-9.29	.243	44.657	22.527	NO
R11 R	6.48	.599	65.037	17.352	NO
R12 L	**19.16**	**.012 ***	**38.467**	**47.437**	**YES**
R13 L	**-13.36**	**.0146 ***	**41.172**	**53.542**	**YES**
R13 R	5.56	.101	39.477	33.648	NO
R14 L	**-23.81**	**.004 ***	**48.531**	**65.743**	**YES**
R14 R	**-28.29**	**< .001 ***	**21.367**	**122.630**	**YES**
R15 L	10.87	.472	31.159	11.552	NO

Note: Feet where the independent *t* test indicated there were significant differences between shoe and heel windows markers are indicated with an asterisks (*). Trials where the Model Statistics indicated a significant difference between the two conditions are highlighted in bold text.

Table 19. Independent *t* test and Model Statistics results for maximal instantaneous vertical loading rate

Subject	Percent Change from Group Design	*t* test *p* value	Critical Difference	Absolute Mean Difference	Significant?
R03 R	-5.76	.291	12.714	5.918	NO
R05 R	**-11.78**	**< .001***	**4.940**	**9.585**	**YES**
R06 R	-2.68	.369	7.012	2.606	NO
R07 L	-6.59	.282	8.776	3.923	NO
R07 R	-3.71	.397	5.549	2.427	NO
R08 R	2.15	.537	8.176	2.425	NO
R13 L	-4.49	.536	6.543	2.249	NO
R13 R	**15.31**	**.001 ***	**4.059**	**7.767**	**YES**
R14 L	-5.87	.1067	6.275	4.977	NO
R14 R	-0.81	.052	7.019	6.431	NO

Note: Feet where the independent *t* test indicated there were significant differences between shoe and heel windows markers are indicated with an asterisks (*). Trials where the Model Statistics indicated a significant difference between the two conditions are highlighted in bold text.

CHAPTER VII

SINGLE SUBJECT ANALYSIS DISCUSSION

The purpose of this single subject analysis was to determine if individual feet demonstrated significant differences in kinematic and kinetic parameters between the two shoe conditions which may have been masked by the group design analysis method. Since there were no significant differences between conditions in the group analysis, the hypothesis for this portion of the study was that the feet which demonstrated larger percent differences in the group design would demonstrate significant differences between conditions for the kinematic and kinetic parameters. The results support this hypothesis.

Due to the requirement of a higher number of trials for the single subject analysis, only seventeen of the original twenty feet were analyzed for the kinematic variables, and only ten were analyzed for the kinetic variable. For the kinematic variables, ten feet demonstrated significant differences in eversion excursion between the shoe markers and heel windows markers conditions. Ten feet also demonstrated significant differences between conditions in the percent stance at which peak eversion occurred and seven subjects demonstrated significant differences between conditions in maximal instantaneous eversion velocity. Of the ten feet analyzed for the maximal instantaneous vertical loading rate, only two demonstrated significant differences between conditions. In all cases, the feet which demonstrated significant differences demonstrated larger absolute percent changes in the group design. However, while large in magnitude, these changes were not always in the same direction, suggesting a single subject analysis might better detect differences between the two marker conditions, differences which were masked when using a traditional group analysis. The remainder of this chapter will discuss the appropriateness of using statistical analyses for the single subject design in this particular study, differences revealed in the single subject analysis which were not apparent in the group analysis, and discuss why the number of feet demonstrating a significant difference between conditions decreased for the kinetic parameters.

Appropriateness of Using Statistical Analysis

When performing a statistical analysis on single subject data, there are two main criteria which must be met, including the data not being auto-correlated and the single subject data having normal distributions (Bates, 1996; Bates et al., 2000; Richards et al., 1999). Bates (1996) indicates results of his investigations suggest that autocorrelation is not a significant problem in running gait data, a statement supported by the findings of this study. Only five of the one hundred twenty two autocorrelations demonstrated statistical significance. This suggests that autocorrelation was not an issue in applying statistical analyses to this data.

On the normality of single subject data Bates (1996) suggests the nature of data produced when studying human gait means this assumption is often violated, even in traditional group designs. However he also suggests that statistical analyses such as the *t* test are robust against minor violations of this assumption, and therefore minor violations of normalcy should not be considered grounds against using statistical analysis with single subject data (Bates et al., 2000). This statement was also supported by the results of this study. Eleven of the Shapiro-Wilks tests performed to assess the normality of the data indicated non-normal distributions, meaning the bulk of the data was approximately normal in distribution. When examined individually, the eleven results which indicated non-normality were most likely the result of low variability with a large cluster of scores at a particular value, and not a distribution which was wildly different than normal. These results suggest that a statistical analysis was appropriate for the single subject data used in this study.

The statistical significance of the single subject results was analyzed by both a traditional independent samples *t* test and a unique analysis method called Model Statistics. Model Statistics was developed specifically for analyzing single subject data and has been used in numerous studies (Bates et al., 1992; Bates et al., 2000; Dufek & Bates, 1991; Dufek, Bates, & Davis, 1995; Dufek, Bates, Stergiou et al., 1995; Stergiou & Bates, 1997). While not a main focus of this particular study, the results of these two different statistical analyses were always in agreement on the statistical significance or lack of significance of any particular variable. This suggests Model Statistics is a robust method for incorporating statistical analysis into single subject design data.

Kinematic Differences Revealed with Heel Windows and Single Subject Analysis

The single subject analysis suggested that ten of the seventeen feet analyzed demonstrated significant differences in kinematic parameters between the shoe markers and heel windows marker conditions. This has important implications when considering the most optimal method to measure rear foot motion during running. For instance, eversion excursion and maximal instantaneous eversion velocity are two of the most commonly assessed biomechanical parameters and are widely thought to be related to a variety of overuse running injuries (Donoghue et al., 2008; Grau et al., 2008; Paavola et al., 2002; Pohl et al., 2008; Reinking & Hayes, 2006; Tweed et al., 2008; Warren, 1984, 1990; T. Willems et al., 2007). If one assessed these parameters using the shoe markers one might conclude that the parameters are within normal physiologic ranges and not potential sources for injury. However, as the results of the single subject analysis demonstrate, the motion reported by the shoe based markers may or may not reflect the true motion of the rear foot, as the measurement for one variable may differ by as much as 40% between the two conditions. In such a case a clinician might wrongly rule out potentially injurious biomechanical markers. An example of such a situation using the single subject data from this study is discussed below.

As the heel makes contact with the ground the rear foot everts and the knee flexes to absorb some of the impact forces. At the end of stance, during push off, the rear foot inverts and the knee extends. Caught between the subtalar joint and the knee is the tibia, which, given the shape of the distal and proximal articulations, must rotate internally with rear foot eversion and knee flexion and then rotate externally with rear foot inversion and knee extension. If rear foot eversion is prolonged so that peak eversion and peak knee flexion do not occur at the same time, then the knee will start extending while the foot is still everted. This will rotate the ends of the tibia in opposite directions, inducing torsion stress within the bone and soft tissue structures of the knee. Several authors have suggested this mal-coupling could be a major contributor to running injuries at the knee (Dierks & Davis, 2007; Stergiou & Bates, 1997; Stergiou et al., 1999; Tiberio, 1987).

Figures 36 and 37 shows the rear foot eversion curves plotted with the knee flexion curves for two sample feet. These curves are averages from all the trials used in the single subject analysis for these feet and results from both marker conditions are shown. The solid lines show results from the shoe markers trials while the dashed lines show results from the heel windows markers trials. When examining the data from the shoe markers condition, maximum rear foot eversion and maximum knee flexion appear to be fairly synchronous and one might conclude there are no coupling related issues with these parameters. However, when the same parameters are examined using data from the heel windows markers a noticeable shift in the coupling pattern is apparent, with maximal eversion occurring later in the stance phase. This means these subjects will have a time period when their knee is extending while their foot is still everting, a scenario which, as described above, might potentially contribute to injury of the soft tissue structures at the knee.

Figures 36 and 37 are only two examples of how the shoe markers might mask potentially injurious biomechanics. Overall five of the seventeen feet analyzed in the single subject analysis demonstrated similar patterns, with the timing of peak eversion excursion and knee flexion being noticeably different when measured with the heel windows markers instead of the shoe markers. This many individual feet showing differences suggests this observation is not a fluke due to the unique kinematics of one or two individuals. However, these results were not found in every individual foot examined, suggesting this mal-coupling occurs on an individual basis. This example both reinforces how potentially injurious biomechanical patterns could be masked when assessed using both the shoe markers and a group analysis method and provides more evidence for the importance of using the heel windows method for a truer measure of rear foot motion.

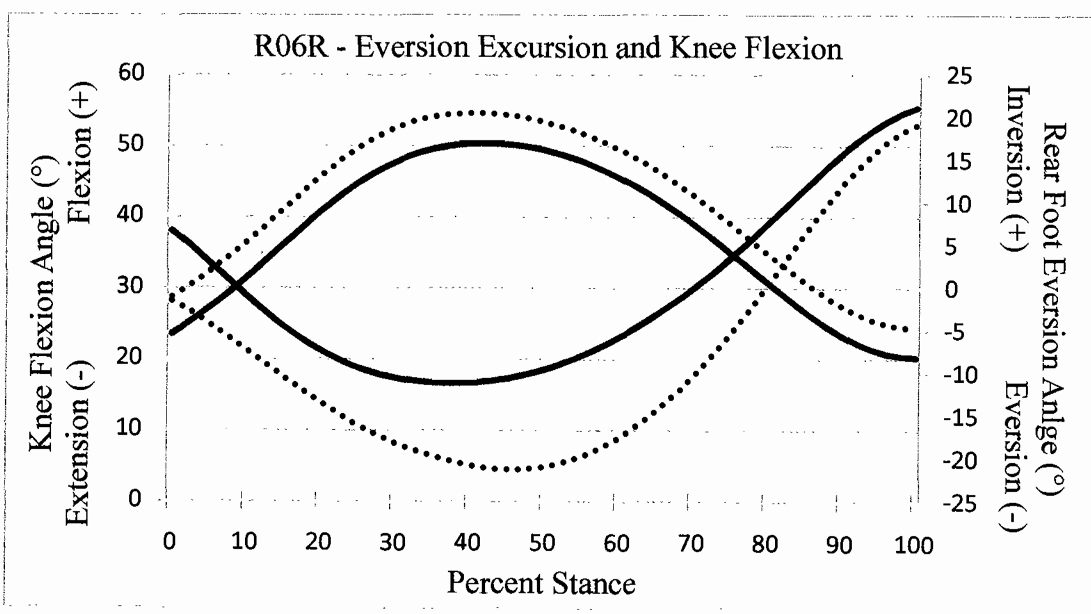


Figure 36. Sample rear foot eversion and knee flexion curves for both the shoe markers and heel windows marker conditions. The solid lines are from the shoe markers condition while the dashed lines are from the heel windows marker condition. Knee flexion is on the left vertical axis while rear foot eversion is on the right vertical axis.

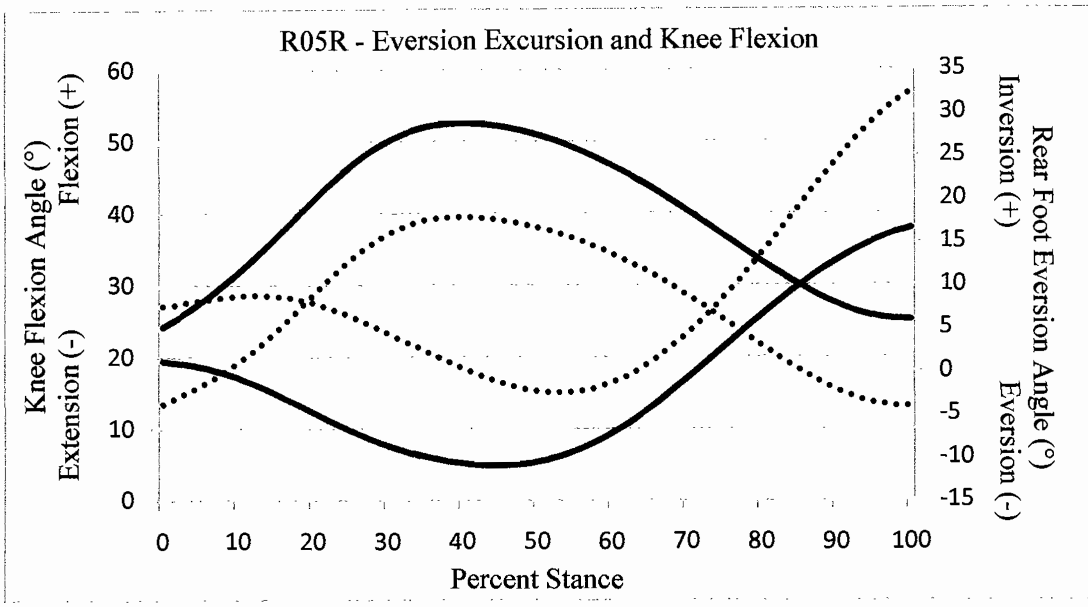


Figure 37. A second example of rear foot eversion and knee flexion curves for both the shoe markers and heel windows marker conditions. The solid lines are from the shoe markers condition while the dashed lines are form the heel windows marker condition. Knee flexion is shown on the left vertical axis while rear foot eversion is shown on the right vertical axis.

Differences in Maximal Instantaneous Vertical Loading Rates

An interesting finding in the single subject analysis was that while a substantial number of feet demonstrated significant differences between marker conditions for the kinematic variables, only two feet demonstrated a significant difference for the kinetic variable. One potential reason could be that, due to the number of trials required for the single subject analysis, fewer feet were analyzed. If more feet had been analyzed, perhaps more would have demonstrated significant differences between conditions. However, it is interesting to note that the percent changes from the group design showed, on average, this particulate parameter had the smallest percent change between the two marker conditions. The two feet with significant differences in the single subject analysis were the exceptions to this trend, suggesting that even if more feet had been added, they most likely would not have shown significant differences between conditions.

One potential reason for the lack of differences between conditions involves a reconsideration of the role of the impact forces experienced during running. In general, the vertical ground reaction force experienced by heel striking runners has two peaks, an impact peak and an active peak. The impact peak is often around 1.5 times body weight while the active peak is around 2.5 to 3 times body weight, though these magnitudes vary with running speed (Cavanagh & Lafortune, 1980). Traditionally, it has been thought that excessively large ground reaction forces, especially during the impact peak, play a role in the development of running related injuries (Cavanagh & Lafortune, 1980; Hreljac et al., 2000; Messier & Pittala, 1988; Milner, Ferber et al., 2006). However, Nigg (2001) has proposed new way of looking at ground reaction forces and their role in running injuries.

According to Nigg's model, impact forces in running can be thought of as input signals to the body which contain both amplitude and frequency components. This input signal induces vibrations in both the soft and bony tissues of the lower limb. The vibrations within the soft tissue are not only uncomfortable, but also inefficient from a metabolic standpoint since they cost energy and require larger oxygen consumption from the individual. Therefore, based on the input signal from one foot strike, the central nervous system will tune the leg muscles to reduce muscle vibrations on the next foot strike. In essence, the central nervous system is anticipating a certain level of impact force and will tune the leg muscles appropriately. Subtle kinematic differences from stride to stride are a result of this tuning.

There is a growing body of evidence supporting this theory, though it should be noted that the bulk of this research was carried out in the same laboratory. Studies examining muscle activity during standing have shown that muscle activity changes in response to vibrations applied to the lower limb (Wakeling & Nigg, 2001; Wakeling, Nigg, & Rozitis, 2002). Studies using a pendulums to apply impact forces to the plantar surface of a foot have found similar results (Wakeling, Von Tscharner, Nigg, & Stergiou, 2001). Studies using actual running have also indicated that muscle activity changes from foot strike to foot strike (Boyer & Nigg, 2004; Wakeling, Pascual, & Nigg, 2002). Lastly, computer modeling simulations have also shown support for this model, suggesting that muscle tuning could result from attempts to keep either soft tissue vibrations or the amount force applied to the body consistent from foot strike to foot strike (Nigg & Liu, 1999; Zadpoor & Nikooyan, 2010).

Though the concept of muscle tuning is fairly new, it could potentially explain the lack of statistically significant differences in loading rates between the two marker conditions. While this study calculated loading rates not impact forces, the loading rates could easily be influenced by the magnitude of the impact force. Therefore, if muscle tuning is happening, either the vibrations applied to the body or the forces applied to the body are held constant. Under this model one would not expect to see differences in loading rates between the two marker conditions.

CHAPTER VIII

CONCLUSIONS, LIMITATIONS, FUTURE DIRECTIONS

Conclusions

The purpose of this study was to assess the effects of two marker systems on kinematic and kinetic measurements commonly made during a running gait analysis. A secondary purpose was to recommend a marker placement method for a planned longitudinal study on overuse injuries in runners. As the data analysis for the study evolved it became apparent that the originally planned analysis may not have been suitable. As such, a tertiary purpose of this study was to explore any differences when the data was analyzed using a single subject analysis instead of a traditional group analysis. From the results of this study the following conclusions are drawn.

1) It is recommended that the heel windows marker placement method be used when conducting a running gait analysis, especially when rear foot motion is one of the parameters of interest. The heel windows method may yield insights into potential injury related mechanisms such as changes in movement variability or joint and segment coupling in the lower extremity that are not evident when markers are placed directly on the shoes. It is suggested that using the heel windows method and placing markers

directly on the skin provides a more accurate measurement of the rear foot motion during running than does placing markers on the exterior of the shoe.

2) For a longitudinal study on overuse running injuries, it is recommended that each subject obtain a baseline measurement and then a single subject analysis methodology be used to compare each subject to their own baseline. The results of this study suggest group designs mask individual changes in kinematics which could be important biomechanical markers in regards to overuse running injuries. A similar result will occur if subjects are compared to some normative average value for a given parameter. This average will not pick up small changes in an individual which may be precursors to or results of, an injury. Therefore this study highlights the importance of a single subject design for a longitudinal evaluation of injury risk in runners.

This has important implications for study design and planning. Studies intent on using single subject design should consider how the number of subjects affects the statistical power of the results and plan accordingly. Most likely, this suggests more subjects will be needed than would be needed for a traditional group design and more trials per subject will have to be collected to ensure sufficient statistical power in the analysis. Additionally, researchers should consider the expected nature of their data for adherence to the requirements of autocorrelation and normalcy required for performing statistical analysis on single subject design data.

Limitations

Perhaps the most visible limitation to this study is the extent to which the running the subjects performed in this study actually reflects their true running kinematics. The subjects were running indoors in a laboratory, not outdoors over ground where they normally would run. Several subjects indicated the pace they ran was similar to their daily run pace, while others thought it was a bit slower. Additionally, data was collected over the course of a 10 meter straight in the laboratory and even though subjects were instructed to run at an even pace they may have inadvertently accelerated through the data collection zone.

Though subjects could run continuous laps, the track around which they ran was fairly small with tight turns. Though it did not appear this way from watching the trials, there is the possibility the subjects were not completely upright and using their normal running gait when passing through the data collection zone. If they were still slightly at an angle this would artificially enhance their leg varus with respect to the floor. It has been suggested that increased tibial varus leads to excessive compensatory pronation (James et al., 1978). If this happened more during the heel windows trials it could potentially suggest kinematic differences which were simply a result of the running course. However, visual inspection of the trials suggests this did not occur and the results found in this study match those presented in the literature, suggesting this should not be considered a significant limitation.

Another, potentially more serious limitation to this study was the fact that the markers in the heel windows condition had to be taped to the heels of the individual, the effects of which are not known. Though all subjects indicated the shoe felt normal even with the marker bases taped to their calcaneous, the effect to which this extra material subconsciously modified their kinematics is unknown. There is the possibility that the presence of the tape against the skin provided the subject with additional sensory feedback during the heel windows marker trials which allowed them to subtly adjust their gait from stride to stride. This phenomenon has previously been observed in other situations where enhanced cutaneous feedback has been used to modify kinematics such as placing tape over the vastus medialis muscle in individuals with patellofemoral pain syndrome (Christou, 2004; MacGregor, Gerlach, Mellor, & Hodges, 2005; Tiberio, 1987)).

A final limitation to this study involves the model used to represent the motion of the foot. This model focused on rear foot motion, and thus three markers were placed on the rear foot, a method which, in reality, describes motion of the calcaneous relative to the tibia. Motion between the calcaneous, talus, and navicular, and motion between the mid-tarsal joints play an important role in the motion of the entire foot. By improving the model to include a multi segmented foot one might be more able to accurately model the true motion of the entire foot.

Future Directions

This study found significant differences in kinematics between the shoe markers and heel windows markers conditions and it was suggested that using the heel windows markers has the potential to mask injury related biomechanical markers. While this study did show some preliminary work comparing the timing of peak knee flexion and rear foot eversion which suggested this might be true, an in depth investigation of this possibility was not the main focus of this study. Future studies could reinforce the conclusions of this study that heel windows markers should be used for running motion capture, when possible, by exploring the effects of the two marker systems on coordination patterns between the lower limbs. This would be especially important considering the limited nature of the coordination assessments in the current study.

Another direction which should be explored in future studies is the applicability of the heel windows concept to the front of the shoe in addition to the heel counter. This method has been used for assessing foot motion while walking (Wolf et al., 2008). However, to the best of this author's knowledge this has never been done for a running gait analysis using standard running shoes. Exploring this possibility might allow the use of a multi-segmented foot model and, by extension, more accurate quantification of the motion of the entire foot, not just the calcaneous.

APPENDIX A

SUBJECT INFORMED CONSENT FORM

CONSENT FORM

You are invited to participate in a research study conducted by Drs. Li-Shan Chou, Louis Osternig, Stan James, and graduate student James Becker. We hope to develop a protocol for a thorough clinical and biomechanical assessment of runners, and using this protocol, track the runners over time to see if there are any changes in these parameters prior to, during, or post injury. At this point we are testing, refining, and trying to validate the protocol.

If you decide to participate, you will be tested in the Motion Analysis Laboratory at the University of Oregon.

TESTING PROCEDURES: The assessments in the Motion Analysis Lab will include both clinical and biomechanical evaluations. The clinical evaluation will include measures of your body alignment, joint range of motion, and muscle strength. The alignment and range of motion assessments will be made by a trained clinician while the strength measures will be tested. For the running gait analysis reflective markers will be placed on your at selected bony landmarks and muscle surfaces to record the motion of each individual body segment. You will run laps around the laboratory space and your body movement (indicated by motion of reflective makers) during running will be recorded by our optoelectronic cameras for further analysis. With your approval we may also record your running with traditional video cameras and/or take photographs of the marker set up placed on your body. We will record your running under several different conditions. In the first condition the markers for your feet will be placed directly on the outside of your shoe. For the second condition we will drill holes in your running shoe and the markers will be directly attached to your heel through the shoe. You will be asked to wear a pair of paper physical therapy shorts and sleeveless shirt (tank top) during testing. The testing session will require a maximum of 3 hours of your time.

COMPENSATION: You will be compensated $75 for participating in this study as reimbursement for a new pair of running shoes. You should understand that your old shoes will no longer be usable after your participation in the study.

RISKS AND DISCOMFORTS: We expect that there will be no more risk for you during these tests than there normally is for you when outside of the laboratory. However, running in the laboratory is different than running outside. You will be asked to speed up then slow down over a 20 meter distance. Running laps in the laboratory will require negotiating tight corners. We will do our best to arrange the lab equipment and furniture to minimize any discomforts and provide as much room as possible. If you are not comfortable you may stop the trials at any time. You may feel fatigue during or after the testing. Our staff member will check with you frequently and provide any required assistance. You will be given frequent breaks as requested. Drilling the holes in your running shoes will require the removal of the inner lining so there is the possibility of rubbing or discomfort on your feet. We will do our best to reduce these effects, and should they still be present you may request additional modifications or stop the trials at any time. There is also the possibility of discomfort involved in removing adhesive tape (used for marker placement) from skin at the end of the experiment.

ADDITIONAL INFORMATION: Any information that is obtained in connection with this study and that can be identified with you will remain confidential and will not be shared without your permission. Subject identities will be kept confidential by coding the data as to study, subject pseudonyms, and collection date. The code list will be kept separate and secure from the actual data files.

Your participation is voluntary. Your decision whether or not to participate will not affect your relationship with the Department of Human Physiology or University of Oregon. You do not waive any liability rights for personal injury by signing this form. In spite of all precautions, you might develop medical complications from participating in this study. If such complications arise, the researchers will assist you in obtaining appropriate medical treatment. In addition, if you are physically injured because of the project, you and your insurance company will have to pay your doctor bills. If you are a University of Oregon student or employee and are covered by a University of Oregon medical plan, that plan might have terms that apply to your injury. If you have any questions about your rights as a research subject, you can contact the Office for Protection of Human Subjects, 5219 University of Oregon, Eugene, OR 97403, (541) 346-2510. This office oversees the review of the research to protect your rights and is not involved with this study.

If you decide to participate, you are free to withdraw your consent and discontinue participation at any time without penalty.

If you have any questions, please feel free to contact Dr. Li-Shan Chou, (541) 346-3391, Department of Human Physiology, 112C Esslinger Hall, University of Oregon, Eugene OR, 97403-1240. You will be given a copy of this form to keep. Your signature indicates that you have read and understand the information provided above, that you willingly agree to participate, that you may withdraw your consent at any time and discontinue participation without penalty, that you will receive a copy of this form, and that you are not waiving any legal claims, rights or remedies.

Name: ______________________________

Signature: __________________________________

Date: ________________________

APPENDIX B

CLINICAL EVALUATION FORM

Running Study Subject Questionnaire and Clinical Evaluation Form

Subject Code:______________ Date:______________

Age: ________

Year in college, if applicable: _________

Number of Years Running: ___________

Approximate Mileage Run per Week: ______________

Over the course of your running career, have you sustained any running related injuries? Y N

If Yes then please describe the nature of the injury, diagnosis by a physician, extent or duration of the injury, and treatment protocols you underwent to relieve symptoms:

____

____

____

____

____

____

Other Comments of History Information:

General Lower Body Alignment and Mobility Assessment

Angle of Gait				**Subtalar Joint Motion**		
Toe In	L	R		Inversion	L:	R:
Straight	L	R		Eversion	L:	R:
Toe Out	L	R		**Forefoot Alignment**		
Tibial Torsion				Neutral	L	R
Leg Varus to Floor	L:	R:		Varus	L	R
Extremity Length	L:	R:		Valgus	L	R
Standing Arch Type				**Position 1st Ray**		
High	L	R		Plantar Flexed	L	R
Medium	L	R		Dorsiflexed	L	R
Ankle Dorsi Flexion				Neutral	L	R
Knee Extended	L:	R:		**Motion 1st Ray**		
Knee Flexed	L:	R:		Normal	L	R
Ankle Plantar Flexion				Mod. Restricted	L	R
Knee Extended	L:	R:		Restricted	L	R
Knee Flexed	L:	R:		**1st MPJ Joint**		
Prone Hip Rotation				Restricted	L	R
Internal	L:	R:		Dosiflexion	L:	R:
External	L:	R:		**Toe Position**		
Foot Motion				Straight	L	R
Loose	L	R		Heel Varus @ STN	L:	R:
Tight	L	R				
Normal	L	R				

<u>Measurement of Arch Height</u>

(from Williams, McClay, Hammill, and Buchanan (2001). Lower Extremity Kinetic and Kinematic Differences in High and Low Arched Runners. *J. Applied Biomechanics.* Vol. 17, pp. 153-161)

1. Height of Dorsum of foot @ 50% foot length: __________L ____________R

2. Truncated Foot Length ____________L _____________R

(measured from most posterior point of calcaneous to medial joint space of first metatarsal phalangeal joint).

3. Arch Height Ratio: __________ L __________R

(measurement 1 divided by measurement 2)

<u>Measurements of Strength and Range of Motion</u>

Hip Rotation Strength (manual testing since cannot do this test on the Biodex)

Int.: ________L ________ R

Ext.: _______ L ________ R

<u>General Flexibility and Range of Motion</u>

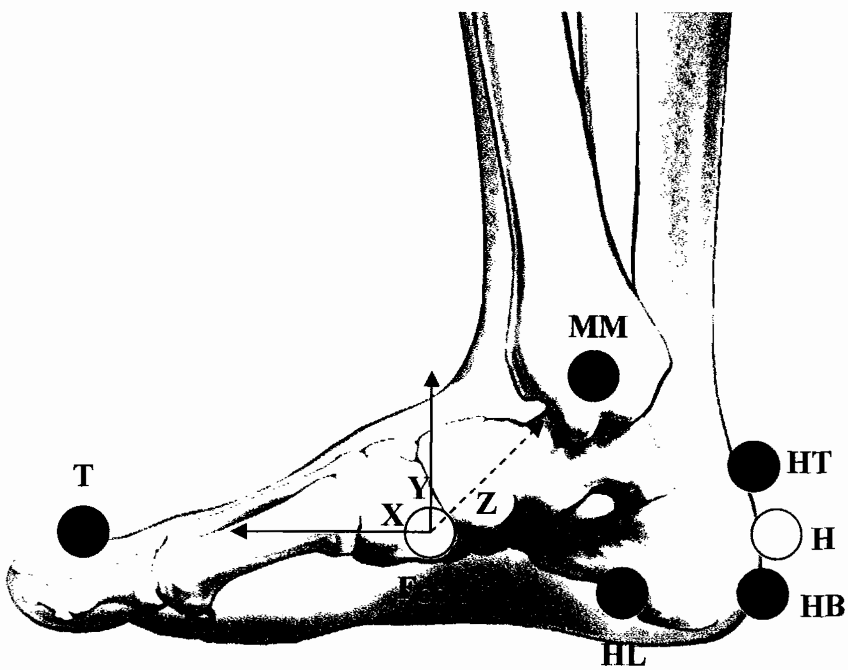
Gastroc (w/STJ in neutral)			**IT Band (Ober's)**	
Left			Left	
Right			Right	
Hamstrings			**Inversion**	
Left			Left	
Right			Right	
Quadriceps			**Eversion**	
Left			Left	
Right			Right	

APPENDIX C

METHODOLGY FOR ESTABLISHING THE ANATOMIC AND TRACKING MARKER COORDINATE SYSTEMS

C.1.1 – Markers used to establish the anatomic coordinate system and tracking marker coordinate system for the foot.

MM:	Medial Malleolus	On the most prominent part of the medial malleolus.
LM:	Lateral Malleolus	On the most prominent part of the lateral malleolus.
AJC:	Ankle Joint Center	The midpoint between MM and LM.
TM:	Toe Marker	On the shoe approximately over the space between the first and second metatarsals.
HT:	Heel Top	The upper marker on the vertical bisection of the heel counter.
HB:	Heel Bottom	The lower marker on the vertical bisection of the heel counter.
H:	Heel	Virtual marker located at the midpoint between HT and HB, collinear with TM
F_{COM}	Foot Center of mass	Virtual marker located at the foot center of mass.



C.1.2 – Definition of the XYZ anatomic coordinate system for the foot.

The origin of the foot segment is located at the foot center of mass (F_{COM}) marker. This point is defined by the heel marker (H) and the toe marker (TM).

$$COM_{Foot} = \vec{H} + 0.44*(\overrightarrow{TM} - \vec{H}) \tag{2}$$

The X axis for the foot segment pointed anteriorly and the markers were placed so that it was parallel with the floor in a normal standing position. It was the normalized unit vector generated from the vector running from the virtual heel marker (H) to the toe marker (TM) and defined as:

$$\hat{x} = \frac{\overrightarrow{TM} - \vec{H}}{|\overrightarrow{TM} - \vec{H}|} \tag{3}$$

The Z axis for the foot segment pointed to the right. It was the normalized vector generated from the cross product of the vector running from the virtual heel marker (H) to the toe marker (TM) and the vector running from the virtual heel marker (H) to the virtual ankle joint center marker (AJC). This vector was defined as:

$$\hat{k} = \frac{(\overrightarrow{TM} - \vec{H}) x (\overrightarrow{AJC} - \vec{H})}{|(\overrightarrow{TM} - \vec{H}) x (\overrightarrow{AJC} - \vec{H})|} \tag{4}$$

The Y axis for the foot segment pointed superiorly and was defined as the cross product of the vectors defining the z and x axes. It was defined as:

$$\hat{j} = \hat{k} \, x \, \hat{\imath} \tag{5}$$

C.1.3 – Definition of the xyz coordinate systems for the tracking markers of the foot.

The y axis of the tracking marker coordinate system was the normalized vector generated from the vector running from the heel bottom marker (HB) to the heel top marker (HT). It was defined as:

$$\vec{y} = \frac{\overrightarrow{HT} - \overrightarrow{HB}}{\left|\overrightarrow{HT} - \overrightarrow{HB}\right|} \tag{6}$$

The x axis of the tracking marker coordinate system was the normalized vector generated from the cross product of the vector running from the heel bottom marker (HB) to the heel top marker (HT) and the vector running from the heel bottom marker (HB) to the lateral heel marker (HL). It was defined as:

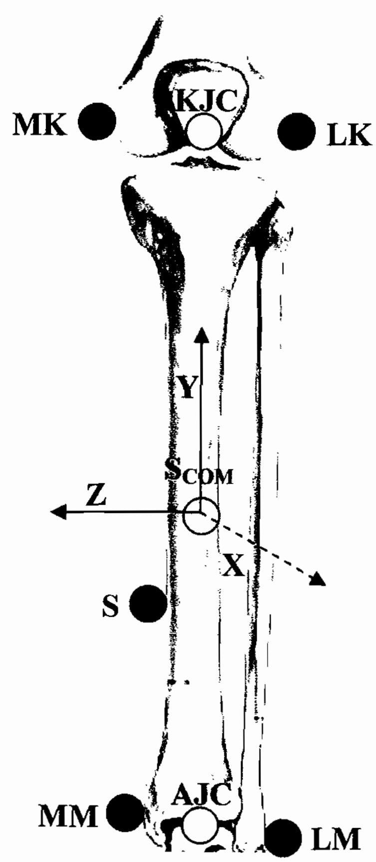
$$\vec{x} = \frac{\left(\overrightarrow{HL} - \overrightarrow{HB}\right) \mathrm{x} \left(\overrightarrow{HT} - \overrightarrow{HB}\right)}{\left|\left(\overrightarrow{HL} - \overrightarrow{HB}\right) \mathrm{x} \left(\overrightarrow{HT} - \overrightarrow{HB}\right)\right|} \quad . \tag{7}$$

The z axis of the tracking marker coordinate system was the cross product of the vectors defining the x and y axes of the tracking marker coordinate system. It was defined as:

$$\vec{z} = \vec{x}\,\mathrm{x}\,\vec{y} \tag{8}$$

C.2.1 – Markers used to establish the anatomic and tracking marker coordinate systems of the shank

MM:	Medial Malleolus	On the most prominent part of the medial malleolus.
LM:	Lateral Malleolus	On the most prominent part of the lateral malleolus.
AJC:	Ankle Joint Center	The midpoint between MM and LM.
LK:	Lateral Knee	On the most prominent part of the lateral femoral epicondyle.
MK:	Medial Knee	On the most prominent part of the medial femoral epicondyle.
KJC:	Knee Joint Center	The midpoint between MK and LK.
S:	Shank	On the inferior 1/3 of the medial portion of the shank collinear with MK and MM.
S_{COM}	Shank Center of Mass	Virtual marker located at the shank center of mass



C.2.2 – Definition of the XYZ anatomic coordinate system for the shank.

The origin of the shank segment is located at the shank center of mass (S_{COM}) marker. This point is defined by the virtual ankle joint center marker (AJC) and the virtual knee joint center marker (KJC).

$$COM_{Shank} = \overrightarrow{KJC} + 0.42*(\overrightarrow{KJC} - \overrightarrow{AJC}) \tag{9}$$

The Y axis for the shank segment pointed superiorly. It was the normalized unit vector generated from the vector running from the virtual ankle joint center (AJC) to the knee joint center (KJC) and defined as:

$$\hat{j} = \frac{\overrightarrow{KJC} - \overrightarrow{AJC}}{|\overrightarrow{KJC} - \overrightarrow{AJC}|} \tag{10}$$

The X axis for the shank segment pointed anteriorly. It was the normalized vector generated from the cross product of the vector running from the virtual ankle joint center (AJC) to the knee joint center (KJC) and the vector running from the virtual ankle joint center (AJC) to the medial malleolus marker (MM). This vector was defined as:

$$\hat{\imath} = \frac{(\overrightarrow{KJC} - \overrightarrow{AJC}) \text{ x } (\overrightarrow{MM} - \overrightarrow{AJC})}{|(\overrightarrow{KJC} - \mathrm{AJC}) \text{ x } (\mathrm{MM} - \overrightarrow{AJC})|} \tag{11}$$

The Z axis for the shank segment pointed to the right and was defined as the cross product of the vectors defining the X and Y axes. It was defined as:

$$\hat{k} = \hat{\imath} \text{ x } \hat{\jmath} \tag{12}$$

C.2.3 – Definition of the xyz coordinate systems for the tracking markers of the shank.

The y axis of the tracking marker coordinate system was the normalized vector generated from the vector running from the lateral malleolus marker (LM) to the lateral knee marker (LK). It was defined as:

$$\vec{y} = \frac{\overrightarrow{LK} - \overrightarrow{LM}}{|\overrightarrow{LK} - \overrightarrow{LM}|} \tag{13}$$

The x axis of the tracking marker coordinate system was the normalized vector generated from the cross product of the vector running from the lateral malleolus marker (LM) to the shank marker (S) and the vector running from the lateral malleolus marker (LM) to the lateral knee marker (LK). It was defined as:

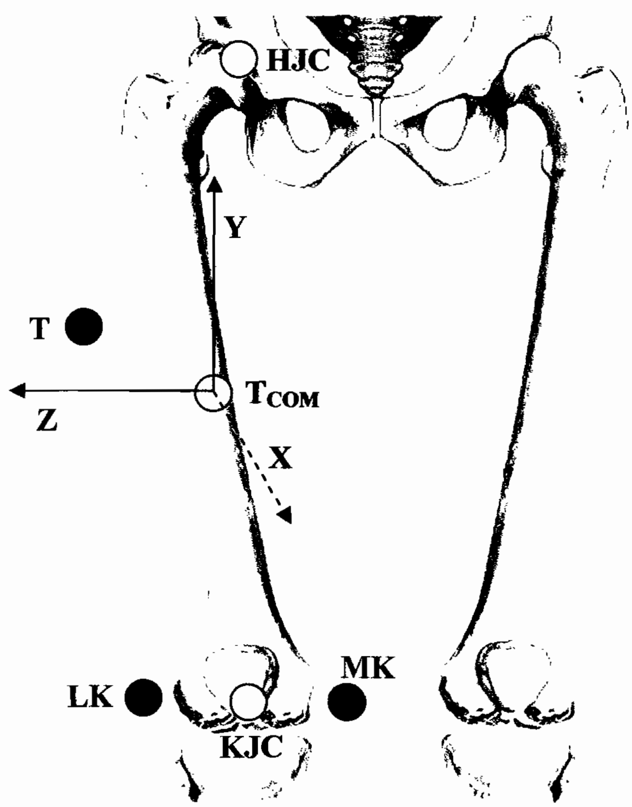
$$\vec{x} = \frac{(\vec{S} - \overrightarrow{LM}) \text{ x } (\overrightarrow{LK} - \overrightarrow{LM})}{|(\vec{S} - \overrightarrow{LM}) \text{ x } (\overrightarrow{LK} - \overrightarrow{LM})|} \tag{14}$$

The z axis of the tracking marker coordinate system for the shank was the cross product of the vectors defining the x and y axes. It was defined as:

$$\vec{z} = \vec{x} \text{ x } \vec{y} \tag{15}$$

C.3.1 – Markers used to establish the anatomic and tracking marker coordinate systems of the thigh

T:	Thigh Marker	Thigh marker placed collinear with LK and the greater trochanter.
HJC:	Hip Joint Center	The hip joint center defined based on anthropometric measurements of the ASIS as explained in Vaughan, Davis, & O'Connor (1999).
LK:	Lateral Knee	On the most prominent part of the lateral femoral epicondyle.
MK:	Medial Knee	On the most prominent part of the medial femoral epicondyle.
KJC:	Knee Joint Center	The midpoint between MK and LK.
T_{COM}	Thigh Center of Mass	Virtual marker located at the thigh center of mass



C.3.2 – Definition of the XYZ anatomic coordinate system for the thigh.

The origin of the thigh segment is located at the thigh center of mass (T_{COM}) marker. This point is defined by the virtual hip joint center marker (HJC) and the virtual knee joint center marker (KJC).

$$COM_{Thigh} = \overrightarrow{HJC} + 0.39*(\overrightarrow{KJC} - \overrightarrow{HJC}) \tag{16}$$

The Y axis for the thigh segment pointed superiorly. It was the normalized unit vector generated from the vector running from the virtual knee joint center (KJC) to the virtual hip joint center (HJC) and defined as:

$$\hat{j} = \frac{\overrightarrow{HJC} - \overrightarrow{KJC}}{|\overrightarrow{HJC} - \overrightarrow{KJC}|} \tag{17}$$

The X axis for the thigh segment pointed anteriorly. It was the normalized vector generated from the cross product of the vector running from the virtual knee joint center (KJC) to the thigh marker (T) and the vector running from the virtual knee joint center (KJC) to the virtual hip joint center marker (HJC). This vector was defined as:

$$\hat{\imath} = \frac{(\vec{T} - \overrightarrow{KJC}) \text{ x } (\vec{T} - \overrightarrow{KJC})}{|(\vec{T} - \overrightarrow{KJC}) \text{ x } (\vec{T} - \overrightarrow{KJC})|} \tag{18}$$

The Z axis for the shank segment pointed to the right and was defined as the cross product of the vectors defining the X and Y axes. It was defined as:

$$\hat{k} = \hat{\imath} \text{ x } \hat{\jmath} \tag{19}$$

C.2.3 – Definition of the xyz coordinate systems for the tracking markers of the thigh.

The y axis of the tracking marker coordinate system was the normalized vector generated from the vector running from the lateral knee marker (LK) to the thigh marker (T). It was defined as:

$$\vec{\mathrm{y}} = \frac{\vec{\mathrm{T}} - \overrightarrow{\mathrm{LK}}}{\left|\vec{\mathrm{T}} - \overrightarrow{\mathrm{LK}}\right|} \tag{20}$$

The x axis of the tracking marker coordinate system was the normalized vector generated from the cross product of the vector running from the lateral knee marker (LK) to the thigh marker (T) and the vector running from the lateral knee marker (LK) to the virtual hip joint center marker (HJC). It was defined as:

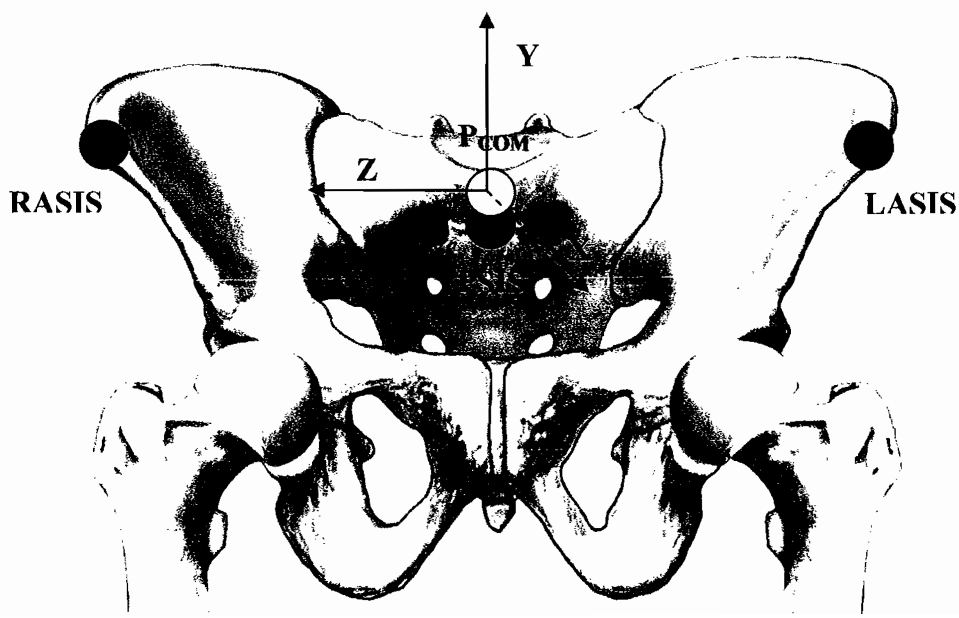
$$\vec{\mathrm{x}} = \frac{\left(\vec{\mathrm{T}} - \overrightarrow{\mathrm{LK}}\right) \mathrm{x} \left(\vec{\mathrm{T}} - \overrightarrow{\mathrm{LK}}\right)}{\left|\left(\vec{\mathrm{T}} - \overrightarrow{\mathrm{LK}}\right) \mathrm{x} \left(\vec{\mathrm{T}} - \overrightarrow{\mathrm{LK}}\right)\right|} \tag{21}$$

The z axis of the tracking marker coordinate system for the thigh was the cross product of the vectors defining the x and y axes. It was defined as:

$$\vec{\mathrm{z}} = \vec{\mathrm{x}}\,\mathrm{x}\,\vec{\mathrm{y}} \tag{22}$$

C.4.1 – Markers used to establish the anatomic and tracking marker coordinate systems of the pelvis

LASIS:	Left ASIS	Placed on the left anterior superior iliac spine.
RASIS:	Right ASIS	Placed on the right anterior superior iliac spine.
SC:	Sacral	Place midway between the two posterior superior iliac spines.
P_{COM}	Pelvis Center of Mass	Calculated as the point half way along the vector connecting the PSIS marker with the midpoint of the vector connecting the two ASIS markers.



C.3.2 – Definition of the XYZ anatomic coordinate system and the xyz tracking marker coordinate system for the pelvis.

The anatomic and tracking marker coordinate systems are the same for the pelvis segment since only three markers are being placed on this segment.

The origin of the pelvis segment is located at the pelvic center of mass marker (P_{COM}). This point is defined by the midpoint along a vector connecting the PSIS marker with the midpoint of the vector between the right and left ASIS markers

$$COM_{Pelvis} = \frac{\left(\frac{LASIS-RASIS}{2}\right)-PSIS}{2} \tag{23}$$

The Z and z axes for the pelvis segment pointed to the right. It was the normalized unit vector generated from the vector running from the left ASIS (LASIS) to the right ASIS (RASIS) and is defined as:

$$\hat{k} = \frac{\overrightarrow{\text{RASIS}} - \overrightarrow{\text{LASIS}}}{\left|\overrightarrow{\text{RASIS}} - \overrightarrow{\text{LASIS}}\right|} \tag{24}$$

The Y and y axes for the pelvic segment pointed superiorly. It was the normalized vector generated from the cross product of the vector connecting the two ASIS markers and the vector running from the left ASIS marker to the PSIS marker. This vector was defined as:

$$\hat{j} = \frac{(\overrightarrow{\text{RASIS}} - \overrightarrow{\text{LASIS}}) \text{ x } (\overrightarrow{\text{PSIS}} - \overrightarrow{\text{LASIS}})}{\left|(\overrightarrow{\text{RASIS}} - \overrightarrow{\text{LASIS}}) \text{ x } (\overrightarrow{\text{PSIS}} - \overrightarrow{\text{LASIS}})\right|} \tag{25}$$

The X and x axes for the pelvic segment pointed anteriorly and was defined as the cross product of the vectors defining the Z and Y axes. It was defined as:

$$\hat{\imath} = \hat{\jmath} \text{ x } \hat{k} \tag{26}$$

BIBLIOGRAPHY

Areblad, M., Nigg, B. M., Ekstrand, J., Olsson, K. O., & Ekstron, H. (1990). Three Dimensional Measurement of Rearfoot Motion During Running. *Journal of Biomechanics, 23*(9), 933 - 940.

Bandholm, T., Boysen, L., Haugaard, S., Kreutzfeldt-Zebis, M., & Bencke, J. (2008). Medial Longitudinal Arch Deformation During Quiet Standing and Gait in Subjects with Medial Tibial Stress Syndrome. *Journal of Foot and Ankle Surgery, 47*(2), 89 - 95.

Bates, B. (1996). Single Subject methodology: an alternative approach. *Medicine and Science in Sports and Exercise, 28*(5), 631 - 638.

Bates, B., Dufek, J. S., & Davis, H. (1992). The effect of trial size on statistical power. *Medicine and Science in Sports and Exercise, 24*(9), 1059 - 1068.

Bates, B., James, C. R., & Dufek, J. S. (2000). Singe Subject Analysis. In N. Stergiou (Ed.), *Innovative Analyses of Human Movement*. Champaign, Ill: Human Kinetics.

Bates, B., Osternig, L., Sawhill, J., & James, S. (1983). An assessment of subject variability, subject-shoe interaction, and the evaluation of running shoes using ground reaction force data. *Journal of BIomechanics, 16*(3), 181 - 191.

Bates, B., Osternig, L. R., Mason, B. R., & James, S. L. (1979). Functional variability of the lower extremity during the support phase of running. *Medicine and Science in Sports and Exercise, 11*(4), 328 - 331.

Bennell, K., Crossley, K., Jayarajan, J., Walton, E., Warden, S., Kiss, Z., et al. (2004). Ground Reaction Forces and Bone Parameters with Tibial Stress Fracture. *Medicine and Science in Sports and Exercise, 36*(3), 397 - 404.

Bennett, J., Reinking, M., Pluemer, B., Pentel, A., Seaton, M., & Killian, C. (2001). Factors Contributing to the Development of Medial Tibial Stress Syndrome in High School Runners. *Journal of Orthopaedic & Sports Physical Therapy, 31*(9), 504 - 510.

Boyer, K., & Nigg, B. (2004). Muscle activity in the leg is tuned in response to impact force characteristics. *Journal of Biomechanics, 37*, 1583 - 1588.

Butler, R., Davis, I., & Hamill, J. (2006). Interaction of Arch Type and Footwear on Running Mechanics. *American Journal of Sports Medicine, 34*(12), 1998 - 2005.

Butler, R., Hamill, J., & Davis, I. (2007). Effect of footwear on high and low arched runners' mechanics during a prolonged run. *Gait and Posture, 26*, 219 - 225.

Butler, R., Hillstrom, H., Song, J., Richards, C., & Davis, I. S. (2008). Arch height index measurement system - Establishment of reliability and normative values. *Journal of the American Podiatric Medical Association, 98*(2), 102-106.

Caster, B. L., & Bates, B. T. (1995). The assessment of mechanical and neuromuscular response strategies during landing. *Medicine and Science in Sports and Exercise, 27*(5), 736 - 744.

Cavanagh, P. R., & Lafortune, M. (1980). Ground reaction forces in distance running. *Journal of Biomechanics, 13*, 397 - 406.

Chang, R., Van Emmerik, R., & Hamill, J. (2008). Quantifying rearfoot-forefoot coordination in human walking. *Journal of Biomechanics, 41*, 3101 - 3105.

Christina, K., White, S., & Gilchrist, L. (2001). Effect of localized muscle fatigue on vertical ground reaction forces and ankle joint motion in running. *Human Movement Sciences, 20*(3), 257 - 276.

Christou, E. (2004). Patella taping increases vastus medialis oblique activity in the presence of patellofemoral pain. *Journal of Electromyography and Kinesiology, 14*, 495 - 504.

Cornwall, M. W., & McPoil, T. G. (1995). Comparison of 2 dimensional and 3 dimensional rearfoot motion during walking. *Clinical Biomechanics, 10*(1), 36 - 40.

Crossley, K., Bennell, K. L., Wrigley, T., & Oakes, B. W. (1999). Ground reaction forces, bone characteristics and tibial stress fractures in male runners. *Medicine and Science in Sports and Exercise, 31*, 1088 - 1093.

De Wit, B., De Clercq, D., & Aerts, P. (2000). Biomechanical analysis of the stance phase during barefoot and shod running. *Journal of BIomechanics, 33*, 269 - 278.

De Wit, B., De Clercq, D., & Lenoir, M. (1995). The effect of varying midsole hardness on impact forces and foot motion during foot contact in running. *Journal of Applied Biomechanics, 11*(4), 395 - 406.

DeLeo, A., Dierks, T., Ferber, R., & Davis, I. (2004). Lower Extremity Joint Coupling During Running: A Current Update. *Clinical Biomechanics, 19*, 983 - 991.

Derrick, T., Dereu, D., & McLean, S. (2002). Impacts and kinematic adjustments during an exhaustive run. *Medicine and Science in Sports and Exercise, 34*(6), 998 - 1002.

Devita, P., & Skelly, W. (1990). Intrasubject variability of lower extremity joint moments of force during the stance phase of running. *Human Movement Science, 9*(2), 99 - 115.

Dierks, T., & Davis, I. (2007). Discrete and continuous joint coupling relationships in uninjured recreational runners. *Clinical Biomechanics, 22*, 581 - 591.

Dierks, T., Manal, K., Hamill, J., & Davis, I. (2008). Proximal and Distal Influences on Hip and Knee Kinematics in Runners with Patellofemoral Pain During a Prolonged Run. *Journal of Orthopaedic & Sports Physical Therapy, 38*(8), 448 - 456.

Donoghue, O. A., Harrison, A. J., Laxton, P., & Jones, R. K. (2008). Lower limb kinematics of subjects with chronic achilles tendon injury during running. *Print, 16*(1), 23-38.

Dufek, J. S., & Bates, B. (1991). Dynamic performance assessment of selected sport shoes on impact forces. *Medicine and Science in Sports and Exercise, 23*(9), 1062 - 1067.

Dufek, J. S., Bates, B., & Davis, H. (1995). The effect of trial size and variability on statistical power. *Medicine and Science in Sports and Exercise, 27*(2), 288 - 295.

Dufek, J. S., Bates, B. T., Stergiou, N., & James, C. R. (1995). Interactive effects between group and single subject response patterns. *Human Movement Science, 14*, 301 - 323.

Duysens, J., & Van de Crommert, H. (1998). Neural control of locomotion; Part 1: The central pattern generator from cats to humans. *Gait and Posture, 7*, 131 - 141.

Edington, C., Frederick, E. C., & Cavanagh, P. (1990). Rear Foot Motion in Distance Running. In P. R. Cavanagh (Ed.), *Biomechanics of Distance Running*. Champaign, IL: Human Kinetics.

Eslami, M., Begon, M., Farahpour, N., & Allard, P. (2007). Forefoot-rearfoot coupling patterns and tibial internal rotation during stance phase of barefoot versus shod running. *Clinical Biomechanics, 22*, 74 - 80.

Ferber, R., McClay Davis, I., and Williams, D. (2005). Effect of foot orthotics on rearfoot and tibia joint coupling patterns and variability. *Journal of Biomechanics, 38*, 477 - 483.

Grau, S., Maiwald, C., Krauss, I., Axmann, D., Janssen, P., & Horstmann, T. (2008). What are causes and treatment strategies for patellar-tendinopathy in female runners? *J Biomech, 41*(9), 2042-2046.

Grood, E., & Suntay, W. (1983). A joint coordinate system for the clinical description of three dimensional motions: applications to the knee. *Journal of Biomechanical Engineering, 105*, 136 - 144.

Hamill, J., Bates, B., & Holt, K. (1992). Timing of lower extremity joint actions during treadmill running. *Medicine and Science in Sports and Exercise, 24*(7), 807 - 813.

Hamill, J., van Emmerik, R., Heiderscheit, B., & Li, L. (1999). A dynamical systems approach to lower extremity running injuries. *Clinical Biomechanics, 14*, 297 - 308.

Heiderscheit, B., Hamill, J., & Van Emmerik, R. (2002). Variability of Stride Characteristics and Joint Coordination Among Individuals with Unilateral Patellofemoral Pain. *Journal of Applied Biomechanics, 18*, 110 - 121.

Hintermann, B., & Nigg, B. (1998). Pronation in Runners: Implications for Injuries. *Sports Medicine, 26*(3), 169 - 176.

Holden, J. P., & Cavanagh, P. R. (1991). The free moment of ground reaction in distance running and its changes with pronation. *J Biomech, 24*(10), 887-897.

Hreljac, A. (2005). Etiology, prevention, and early intervention of overuse injuries in runners: a biomechanical perspective. *Physical medicine and Rehabilitation Clinics of North America, 16*, 651 - 667.

Hreljac, A., Marshall, R. N., & Hume, P. A. (2000). Evaluation of lower extremity overuse injury potential in runners. *Medicine and Science in Sports and Exercise, 32*(9), 1635 - 1641.

Ireland, M., Willson, J., Ballantyne, B., & McClay Davis, I. (2003). Hip Muscle Strength in Females With and Without Patellofemoral Pain. *Journal of Orthopaedic & Sports Physical Therapy, 33*(11), 671 - 676.

Jacobs, S., & Berson, B. (1986). Injuries to Runners: A study of entrants to a 10,000 meter race. *American Journal of Sports Medicine, 14*(2), 151 - 155.

James, S., Bates, B., & Osternig, L. (1978). Injuries to Runners. *American Journal of Sports Medicine, 6*(2), 40 - 50.

Jones, B., & Knapik, J. (1999). Physical Training and Exercise Related Injuries: Surveillance, Research, and Injury Prevention in Military Populations. *Sports Medicine, 27*(2), 111 - 125.

Kadaba, M., Ramakrishnan, H., & Wootten, M. (1990). Measurement of Lower Extremity Kinematics During Level Walking. *Journal of Orthopaedic Research, 8*, 383 - 392.

Kadaba, M., Ramakrishnan, H., Wootten, M., Gainer, J., Gorton, G., & Cochran, G. (1989). Repeatability of Kinematic, Kinetic, and Electromyographic Data in Normal Adult Gait. *Journal of Orthopaedic Research, 7*(6), 849 - 860.

Kaufman, K., Brodine, S., Shaffer, R., Johnson, C., & Cullison, T. (1999). The Effect of Foot Structure and Range of Motion on Musculoskeletal Overuse Injuries. *American Journal of Sports Medicine, 27*(5), 585 - 593.

Kibler, W., Goldber, C., & Chandler, T. (1991). Functional biomechanical deficits in running athletes with plantar fasciitis. *American Journal of Sports Medicine, 19*(6), 66 - 71.

Korpelainen, R., Orava, S., Karpakka, J., Siira, P., & Hulkko, A. (2001). Risk Factors for Recurrent Stress Fractures in Athletes. *American Journal of Sports Medicine, 29*(3), 304 - 310.

Lun, V., Meeuwisse, W. H., Stergiou, P., & Stefanyshyn, D. (2003). Relation between running injury and static lower limb alignment in recreational runners. *British Journal of Sports Medicine, 38*(9), 576 - 580.

Lysholm, J., & Wiklander, J. (1987). Injuries in Runners. *American Journal of Sports Medicine, 15*(2), 168 - 170.

MacGregor, K., Gerlach, S., Mellor, R., & Hodges, P. (2005). Cutaneous stimulation from patella tape causes a differential increase in vasti muscle activity in people with patellofemoral pain. *Journal of Orthopaedic Research, 23*(2), 351 - 358.

MacKay-Lyons, M. (2002). Central Pattern Generation of Locomotion: A Review of the Evidence. *Physical Therapy, 82*(1), 69 - 83.

MacLean, C., McClay Davis, I., & Hamill, J. (2006). Influence of a custom foot orthotic intervention on lower extremity dynamics in healthy runners. *Clinical Biomechanics, 21*, 623 - 630.

Marti, B., Vader, J. P., Minder, C., & Abelin, T. (1988). On the etiology of running injuries The 1984 Bern Grand-Prix Study. *American Journal of Sports Medicine, 16*(3), 285 - 293.

McClay, I. (1995). The Use of Gait Analysis to Enhance the Understanding of Running Injuries. In R. L. Craik, and Oatis, CA (Ed.), *Gait Analysis: Theory and Application*. Champaign, Il: Human Kinetics.

McClay, I., & Manal, K. (1997). Coupling Parameters in Runners with Normal and Excessive Pronation. *Journal of Applied Biomechanics, 13*, 109 - 124.

McClay, I., & Manal, K. (1998a). A Comparison of three dimensional lower extremity kinematics during running between excessive pronators and normals. *Clinical Biomechanics, 13*(3), 195 - 203.

McClay, I., & Manal, K. (1998b). The Influence of Foot Abduction on Difference between Two Dimensional and Three Dimensional Rearfoot Motion. *Foot and Ankle International, 19*(1), 26 - 31.

McClay, I., & Manal, K. (1999). Three dimensional kinetic analysis of running: significance of secondary planes of motion. *Medicine and Science in Sports and Exercise, 31*(11), 1629.

Messier, S. P., & Pittala, K. A. (1988). Etiologic factors associated with selected running injuries. *Med Sci Sports Exerc, 20*(5), 501-505.

Milani, T. L., & Hennig, E. M. (2000). Measurements of Rearfoot Motion during Running. *Sportverl Sportschad, 14*, 115 - 120.

Miller, R., Meardon, S., Derrick, T., & Gillette, J. (2008). Continuous Relative Phase Variability During and Exhaustive Run in Runners with a History of Iliotibial Band Syndrome. *Journal of Applied Biomechanics, 24*, 262 - 270.

Milner, C., Davis, I., & Hamill, J. (2006). Free moment as a predictor of tibial stress fracture in distance runners. *Journal of Biomechanics, 39*, 2819 - 2825.

Milner, C., Ferber, R., Pollard, C., Hamill, J., & Davis, I. (2006). Biomechanical Factors Associated with Tibial Stress Fracture in Female Runners. *Medicine and Science in Sports and Exercise, 38*(2), 323 - 328.

Molloy, J. M., Christie, D. D., Teyhen, D. S., Yeykal, N. S., Tragord, B. S., Neal, M. S., Nelson, E.S., et al. (2009). Effect of Running Shoe Type on the Distribution and Magnitude of Plantar Pressures in Individuals with Low-or High Arched Feet. *Journal of the American Podiatric Medical Association, 99*(4), 330 - 338.

Morio, C., Lake, M., Guegen, N., Rao, G., & Baly, L. (2009). The influence of footwear on foot motion during walking and running. *Journal of Biomechanics, 42*, 2081 - 2088.

Mundermann, A., Nigg, B. M., Humble, R., & Stefanyshyn, D. (2003). Foot orthotics affect lower extremity kinematics and kinetics during running. *Clinical Biomechanics, 18*, 254 - 262.

Nawoczenski, D., Saltzman, C., & Cook, T. (1998). The Effect of Foot Structure on the Three Dimensional Kinematic Coupling Behavior of the Leg and Rear Foot. *Physical Therapy, 78*(4), 404 - 416.

Nigg, B. (1986). Experimental Techniques used in running shoe research. In B. Nigg (Ed.), *Biomechanics of Running Shoes*. Champaign, Il: Human Kinetics.

Nigg, B., & Liu, W. (1999). The effect of muscle stiffness and damping on simulated impact force peaks during running. *Journal of Biomechanics, 32*, 849 - 856.

Nigg, B., (1997). Impact forces in running. *Current Opinion in Orthopaedics, 8*(6), 43 - 47.

Nigg, B. M. (2001). The Role of Impact Forces and Foot Pronation: A New Paradigm. *Clinical Journal of Sports Medicine, 11*(1), 2 - 9.

Noehren, B., Davis, I., & Hamill, J. (2007). Prospective Study of the biomechanical factors associated with iliotibial band syndrome. *Clinical Biomechanics, 22*, 951 - 956.

Novacheck, T. (1988). The Biomechanics of Running. *Gait and Posture, 7*, 77 - 95.

O'Connor, K., & Hamill, J. (2004). The role of selected extrinsic foot muscles during running. *Clinical Biomechanics, 19*, 71 - 77.

Paavola, M., Kannus, P., Jarvinen, T., Khan, K., Jozsa, L., & Jarvinen, M. (2002). Current Concepts Review: Achilles Tendinopathy. *Journal of Bone and Joint Surgery, 84-A*(11), 2062 - 2076.

Pohl, M., Hamill, J., & Davis, I. (2009). Biomechanical and Anatomic Factors Associates with a History of Plantar Fasciitis in Female Runners. *Clinical Journal of Sports Medicine, 19*(5), 372 - 376.

Pohl, M., Messenger, N., & Buckley, J. (2006). Changes in foot and lower limb coupling due to systematic variations in step width. *Clinical Biomechanics, 21*, 175 - 183.

Pohl, M., Messenger, N., & Buckley, J. (2007). Forefoot, rearfoot and shank coupling: Effect of variations in speed and mode of gait. *Gait and Posture, 25*, 295 - 302.

Pohl, M., Mullineaux, D., Milner, C., Hamill, J., & Davis, I. (2008). Biomechanical Predictors of retrospective tibial stress fracture in runners. *Journal of Biomechanics, 41*, 160 - 1165.

Powell, K., Kohl, H., Caspersen, C., & Blair, S. (1986). An Epidemiological Perspective on the Causes of Running Injuries. *Physician and Sports Medicine, 14*(6), 100 - 114.

Powers, C. (2003). The Influence of Altered Lower Extremity Kinematics on Patellofemoral Joint Dysfunction: A Theoretical Perspective. *Journal of Orthopaedic & Sports Physical Therapy, 33*(11), 639 - 646.

Rauh, M., Koepsell, T., & Rivara, F. (2006). Epidemiology of musculoskeletal injuries among high school cross country runners. *American Journal of Epidemiology, 163*, 151 - 159.

Reber, L., Perry, J., & Pink, M. (1993). Muscular Control of the Ankle in Running. *American Journal of Sports Medicine, 21*(6), 805 - 810.

Reinking, M. (2006). Exercise Related Leg Pain in Female Collegiate Athletes: The Influence of Intrinsic and Extrinsic Factors. *American Journal of Sports Medicine, 34*(9), 1500 - 1507.

Reinking, M., & Hayes, A. (2006). Intrinsic factors associated with exercise-related leg pain in collegiate cross-country runners. *Clin J Sport Med, 16*(1), 10-14.

Reinschmidt, C., & Nigg, B. (1995). Influence of heel height on ankle joint moments in running. *Medicine and Science in Sports and Exercise, 27*(3), 410 - 416.

Reinschmidt, C., Stacoff, A., & Stussi, E. (1992). Heel Movement within a court shoe. *Medicine and Science in Sports and Exercise, 24*(12), 1390 - 1395.

Reinschmidt, C., van den Bogert, A. J., Murphy, N., Lundberg, A., & Nigg, B. M. (1997). Tibiocalcaneal motion during running, measured with external and bone markers. *Clinical Biomechanics, 12*(1), 8 - 16.

Richards, S., Taylor, R., Ramasamy, R., & Richards, R. (1999). *Single Subject Research: Applications in Educational and Clinical Settings*. San Diego, CA: Singular Publishing Group.

Ryan, M. B., MacLean, C. L., & Taunton, J. E. (2006). A review of anthropometric, biomechanical, neuromuscular, and training related factors associated with injuries in runners. *International Journal of Sports Medicine, 7*(2), 120 -137.

Scott, S., & Winter, D. (1990). Internal Forces at Chronic Running Injury Sites. *Medicine and Science in Sports and Exercise, 22*(3), 357 - 369.

Snyder, K., Earl, J., O'Connor, K., & Ebersole, K. (2009). Resistance training is accompanied y increases in hip strength and changes in lower extremity biomechanics during running. *Clinical Biomechanics, 24*, 26 - 34.

Soutas-Little, R. W., Beavis, G. C., Verstraete, M. C., & Markus, T. L. (1987). Analysis of foot motion during running using a joint co-ordinate system. *Medicine and Science in Sports and Exercise, 19*(3), 285 - 293.

Souza, R., & Powers, C. (2009). Predictors of Hip Internal Rotation During Running. *American Journal of Sports Medicine, 37*(3), 579 - 587.

Stackhouse, C., McClay Davis, I., & Hamill, J. (2004). Orthotic intervention in forefoor and rearfoot strike running patterns. *Clinical Biomechanics, 19*, 64 - 70.

Stacoff, A., & Luethi, S. (1986). Special Aspects of shoe contruction and foot anatomy. In B. Nigg (Ed.), *Biomechanics of Running Shoes*. Champaign, Il: Human Kinetics.

Stacoff, A., Nigg, B., Reinchmidt, C., van den Bogert, A. J., & Lundberg, A. (2000). Tibiocalcaneal kinematics of barefoot versus shod running. *Journal of Biomechanics, 33*, 1387 - 1395.

Stacoff, A., Reinchmidt, C., & Stussi, E. (1992). The movement of the heel within a running shoe. *Medicine and Science in Sports and Exercise, 24*(6), 695 - 701.

Stacoff, A., Reinschmidt, C., Nigg, B. M., van den Bogert, A. J., Lundberg, A., Denoth, J., et al. (2001). Effects of shoe sole construction on skeletal motion during running. *Medicine and Science in Sports and Exercise, 33*(2), 311 - 319.

Stefanyshyn, D., Stergiou, P., Lun, V., Meeuwisse, W., & Worobets, J. (2006). Knee Angular Impulse as a Predictor of Patellofemoral Pain in Runners. *American Journal of Sports Medicine, 34*(11), 1844 - 1851.

Stergiou, N., & Bates, B. (1997). The relationship between subtalar and knee joint function as a possible mechanism for running injuries. *Gait and Posture, 6*, 177 - 185.

Stergiou, N., Bates, B., & James, S. (1999). Asynchrony between subtalar and knee joint function during running. *Medicine and Science in Sports and Exercise, 31*(11), 1645 - 1655.

Stergiou, N., Harbourne, R., & Cavanaugh, J. (2006). Optimal Movement Variability: A New Theoretical Perspective for Neurologic Physical Therapy. *Journal of Neurologic Physical Therapy, 30*(3), 120 - 129.

Taunton, J. E., Ryan, M.B., Clement, D.B., McKenzie, D.C., Lloyd-Smith, D.R., and Zumbo, B.D. (2002). A retrospective case control analysis of 2002 running injuries. *British Journal of Sports Medicine, 36*, 95 - 101.

Taunton, J. E., Ryan, M.B., Clement, D.B., McKenzie, D.C., Lloyd-Smith, D.R., and Zumbo, B.D. (2003). A prospective study of running injuries: the Vancouver Sun Run "In Training" Clinics. *British Journal of Sports Medicine, 37*, 239 - 244.

Tiberio, D. (1987). The Effect of Excessive Subtalar Joint Pronation on Patellofemoral Mechanics: A Theoretical Model. *Journal of Orthopaedic & Sports Physical Therapy, 9*(4), 160 - 165.

Tweed, J. L., Campbell, J. A., & Avil, S. J. (2008). Biomechanical risk factors in the development of medial tibial stress syndrome in distance runners. *J Am Podiatr Med Assoc, 98*(6), 436-444.

Vagenas, G., & Hoshizake, B. (1992). A multivariable analysis of lower extremity kinematic asymmetry in running. *Journal of Applied Biomechanics, 8*(1), 11 - 29.

Van de Crommert, H., Mulder, T., & Duysens, J. (1998). Neural control of locomotion; Part 2: sensory control of the central pattern generator and its relation to treadmill training. *Gait and Posture, 7*, 251 - 263.

Van Gheluwe, B., Tielemans, R., & Roosen, P. (1995). The Influence of Heel Counter Rigidity on Rearfoot Motion During Running. *Journal of Applied Biomechanics, 11*, 47 - 67.

Van Ghent, R. N., Siem, D., Middelkoop, v. M., van Os, A. G., Bierma-Zeinstra, S. M. A., & Koes, B. W. (2007). Incidence and determinants of lower extremity running injuries in lond distance runners: a systematic review. *British Journal of Sports Medicine, 41*, 469 - 480.

Van Mechelen, W. (1992). Running Injuries: A review of the epidemiological literature. *Sports Medicine, 14*(5), 320 - 335.

van Middelkoop, M., Kolkman, J., van Ochten, J., Bierma-Zeinstra, S., & Koes, B. (2006). Course and Predicting Factors of Lower Extremity Injuries After Running a Marathon. *Clinical Journal of Sports Medicine, 17*(1), 25 - 30.

Vaughan, C., Davis, B., & O'Connor, J. (1999). *Dynamics of Human Gait* (2 ed.). Howard Place, Western Cape, South Africa: Kiboho Publishers.

Verdejo, R., & Mills, N. J. (2004). Heel-shoe interactions and the durability of EVA foam running-shoe midsoles. *Journal of Biomechanics, 37*(9), 1379 - 1386.

Wakeling, J., & Nigg, B. (2001). Modification of soft tissue vibrations in the leg by muscular activity. *Journal of Applied Physiology, 90*(2), 412 - 420.

Wakeling, J., Nigg, B., & Rozitis, A. (2002). Muscle activity damps the soft tissue resonance that occurs in response to pulsed and continuous vibrations. *Journal of Applied Physiology, 92*(3), 1093 - 1103.

Wakeling, J., Pascual, S. A., & Nigg, B. (2002). Altering muscle activity in the lower extremities by running with different shoes. *Medicine and Science in Sports and Exercise, 34*(9), 1529 - 1532.

Wakeling, J., Von Tscharner, V., Nigg, B., & Stergiou, P. (2001). Muscle activity in the leg is tuned in response to ground reaction forces. *Journal of Applied Physiology, 91*(3), 1307 - 1317.

Walter, S., Hart, L., McIntosh, J., & Sutton, J. (1989). The Ontario Cohort Study on Running Injuries. *Archives of Internal Medicine, 149*(11), 2561 - 2564.

Warren, B. L. (1984). Anatomical factors associated with predicting plantar fasciitis in long-distance runners. *Med Sci Sports Exerc, 16*(1), 60-63.

Warren, B. L. (1990). Plantar fasciitis in runners. Treatment and prevention. *Sports Med, 10*(5), 338-345.

Wen, D. Y., Puffer, J. C., & Schmalzried, T. P. (1997). Lower Extremity alignment and risk of overuse injuries in runners. *Medicine and Science in Sports and Exercise, 29*(10), 1291 - 1298.

Wen, D. Y., Puffer, J. C., & Schmalzried, T. P. (1998). Injuries in Runners: a prospective study of alignment. *Clinical Journal of Sports Medicine, 8*(3), 187 - 194.

Willems, T., Witvrouw, E., De Cock, A., & De Clercq, D. (2007). Gait Related Risk Factors for Exercise-Related Lower Leg Pain During Shod Running. *Medicine and Science in Sports and Exercise, 39*(2), 300 - 339.

Williams, D., McClay Davis, I., & Baitch, S. (2003). Effect of Inverted Orthoses on Lower Extremity Mechanics in Runners. *Medicine and Science in Sports and Exercise, 35*(12), 2060 - 2068.

Williams, D., McClay Davis, I., Scholz, J., Hamill, J., & Buchanan, T. (2004). High arched runners exhibit increased leg stiffness compared to low arched runners. *Gait and Posture, 19*, 263 - 269.

Williams, D., & McClay, I. (2000). Measurements Used to Characterize the Foot and the Medial Longitudinal Arch: Reliability and Validity. *Physical Therapy, 80*(9), 864 - 871.

Williams, D., McClay, I., & Hamill, J. (2001). Arch Structure and injury patterns in runners. *Clinical Biomechanics, 16*, 341 - 347.

Williams, D., McClay, I., Hamill, J., & Buchanan, T. (2001). Lower Extremity Kinematic and Kinetic Differenced in Runners with High and Low Arches. *Journal of Applied Biomechanics, 17*, 153 - 163.

Winter, D. (2005). *Biomechanics and motor control of human movement*. Hoboken, NJ: Wiley & Sons, Inc.

Wolf, S., Simon, J., Patikas, D., Schuster, W., Armbrust, P., & Doderlein, L. (2008). Foot motion in children shoes: a comparison of barefoot walking with shod walking in traditional and flexible shoes. *Gait and Posture, 27*, 51 - 59.

Wu, G. (2002). ISB recommendation on definitions of joint coordinate system of various joints for reporting of human joint motion: - part I: ankle, hip, and spine. *Journal of Biomechanics, 25*, 543 - 548.

Zadpoor, A., & Nikooyan, A. (2010). Modeling muscle activity to study the effects of footwear on the impact forces and vibrations of the human body during running. *Journal of Biomechanics, 43*, 186 - 193.

Zifchock, R., Davis, I., & Hamill, J. (2006). Kinetic Asymmetry in female runners with and without retrospective tibial stress fractures. *Journal of Biomechanics, 31*(15), 2792 - 2797.

Zifchock, R., Davis, I., Higginson, J., McCaw, S., & Royer, T. (2008). Side to side difference in overuse running injury susceptibility: A retrospective study. *Human Movement Sciences, 27*, 888 - 902.

Zifchock, R., Davis, I., Higginson, J., McCaw, S., & Royer, T. (2009). Side to side difference in overuse running injury susceptibility: A retrospective study. *Human Movement Sciences, 27*, 888 - 902.