THE EFFECT OF SHOE DESIGN AND CUSTOM FOOT ORTHOTIC INTERVENTION ON LOWER EXTREMITY DYNAMICS IN FEMALE RUNNERS

by

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Abstract

This study used measures of joint coordination variability, plantar pressure loading profiles, rearfoot dynamics and a shoe comfort questionnaire to determine whether a motion control shoe or a neutral shoe is more appropriate for female runners who wear moderately posted foot orthoses. To date, there has been no research to suggest the prescription of either shoe, over the other, for female runners who present with moderately posted foot orthoses.

Fifteen female runners participated in this study. All subjects had previously been prescribed and currently wear foot orthoses with 3° of rearfoot posting and no forefoot posting. Plantar pressure data and electrogoniometer data were collected simultaneously while the subjects ran at 3.35 m/s (8 minute-mile pace), or at a speed as close to 3.35 m/s as possible. The electrogoniometers were attached to the lateral aspect of the right leg collecting knee angle data in the sagittal plane and to the heel counter of the shoe and the lower leg, collecting rearfoot angle data in the frontal plane. Both the neutral and the motion control shoe were tested while the subjects ran with and without their foot orthoses. A questionnaire was administered to determine shoe satisfaction.

Variability, as it pertains to joint coordination is thought to be characteristic of more flexible gait patterns. Those with increased joint coordination variability are thought to be able to adapt to external perturbations better than those with reduced variability of joint coordination. Differences were deemed statistically significant if $p < 0.05$. For the orthoses condition, the motion control shoe showed decreased variability in the regional analysis as compared to the neutral shoe. When running in the shoes without orthoses (SHOD), the motion control shoe showed increased variability as compared to the neutral shoe. When comparing all conditions, the neutral shoe, when worn with orthoses showed the greatest joint coordination variability and thus is the preferred shoe-orthoses combination with respect to healthy joint coordination patterns.

Differences between the two shoes were exhibited in the SHOD condition for plantar pressure loading profiles on the medial side of the heel and mid-foot region as well as the hallux region ($p<0.01$). There were no statistically significant correlations between navicular drop and shoe that was deemed most comfortable. The research presented in this thesis provides valuable information as to appropriate shoe-orthoses combinations and contributes to the new direction the research of running pathomechanics appears to be taking.

Table of Contents

List of Tables

List of Figures

Acknowledgements

As this chapter of my life comes to a close, I would like to take this opportunity to reflect on the past sixteen months. There have been some ups and downs, some great experiences and a few challenges. I've met some wonderful people who've given up their time to either participate in the study or who've taken the time to discuss different analysis techniques and give me guidance, thank you all. Specifically, I would like to thank David Garcia from the Nike Sport Research Lab for setting me up with the pressure analysis equipment and being my technical support. Thank you to Christian Maiwald and Stefan Grau at *Sportmedizin Tübingen* at the University of Tübingen for allowing us to be the first external institution to use the pressure analysis software and for helping to implement it into this research. And Chris MacLean, thank you for all the support with biomechanics theory, you've been an excellent resource and I appreciate your patience; your diligence and encouragement has been inspiring to me.

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Chapter 1

1.0 Introduction

When it comes to correcting potential pathology of the lower limb due to poor biomechanics, one of the greatest challenges for the physician, physiotherapist, podiatrist or other attending practitioner, is the patient who presents with a mild degree of abnormality (McKenzie *et al.,* 1985). The patient presenting with extreme variations in lower limb biomechanics can be treated with orthotic intervention, however it remains unclear what the best treatment modality is for the patient presenting with only slight variations in their lower limb and foot biomechanics. Can a high quality motion control or stability shoe enhance a runner's lower-limb function enough to prevent running-related complications along the kinetic chain and alleviate any symptoms that may arise due to unhealthy biomechanics? Are foot orthoses necessary to help enhance lower-limb function? Or do we need a motion control shoe and foot orthoses to achieve our goal? Hintermann and Nigg (1998) posed a very significant and pertinent question and one that this study will attempt to answer, "to what extent may shoe modifications and supports reduce [unhealthy] foot pronation and thus, help to prevent injuries?"

Past research has shown that the use of correctly fitting orthoses in combination with the correct shoe can aid in the treatment of injuries sustained along the kinetic chain, as well as prevention of such injuries (Janisse, 1994). D'Ambrosia reported the alleviation of symptoms caused by pes planovalgum, metatarsalgia, calcaneal spurs, plantar fasciitis, and iliotibial band syndrome with the use of correctly molded orthotics (D'Ambrosia, 1985). Much of the research conducted has attempted to find a link between the kinematic findings of the lower limb and clinical findings of foot orthoses and or shoe use. Ambiguity enters the equation when a shoe designed to function similarly to an orthosis is prescribed in addition to foot orthoses. The theory is clear as to what the goal of the shoe and orthosis combination is, but it is unclear as to the level of involvement each should have in preventing and or reducing lower limb pathomechanics.

What is the ideal combination of posting in orthoses and level of control provided by the shoe? The biomechanics, running, and sports medicine industries are filled with opinions as to what type of shoe should be prescribed for a given foot structure, however, there is no quantifiable evidence. Currently many physicians are recommending neutral shoes be worn with orthoses, however this may not offer sufficient rearfoot control, and thus, we are seeking to collect data that indicates the prescription of one design of shoe over another.

1.1 Purpose of the Investigation

The purpose of this study is to establish the effect of two different shoe-orthoses combinations on female runners' joint coordination variability (between the knee and rearfoot joint couple), plantar pressure loading profiles and rearfoot dynamics to determine whether we can "standardize" shoe prescription such that female runners wearing orthoses with 3° of rearfoot posting will consistently be prescribed either a motion control shoe, or a neutral shoe based on quantifiable evidence rather than opinion. As an important component to shoe prescription is fit, our quantitative data will be correlated with qualitative data (See Appendix D), which will aim to establish patient satisfaction with respect to shoe fit and comfort, in order to determine whether there is a relationship between foot structure, and the overall satisfaction with the shoe (Taunton and Moore, 1991).

Recent gait analyses using the F- Scan Insole System (Tekscan, Boston, MA) in conjunction with video analysis have illustrated the power of shoe construction to change the biomechanics of female runners' gait (research submitted for publication). It is our plan to quantify the changes along the kinetic chain (using joint coordination variability, rearfoot dynamics and plantar pressure loading patterns as outcome measures), that occur as a result of different shoe construction.

1.2 Statement of Problem

A good motion control shoe on the market right now is capable of controlling approximately 5° of heel varus, any amount of heel varus greater than this, may require additional support offered by foot orthoses (personal communication with Dr. Taunton). A patient presenting with 9° of heel varus would be prescribed an orthosis with 3° of rearfoot posting by the physicians at the Allan McGavin Sports Medicine Centre (personal communication with Dr. Taunton). Because an orthosis with 3° of rearfoot posting offers only a mild amount of rearfoot control, often the user is fit in either a motion control or a neutral shoe, depending on the opinion of the physician, physiotherapist, podiatrist, and or shoe vendor. It is our aim to seek data that suggests the prescription of either a motion control shoe or a neutral shoe for those wearing moderately posted, functional foot orthoses. Often orthoses will have forefoot posting in addition to rearfoot posting. Our study is only concerned with rearfoot posting as this is the characteristic of an orthosis that is designed to serve a similar function as the rearfoot construction of a motion control shoe.

1.3 Significance of the Study

Abnormally "pronated" feet predispose a runner to overuse running injuries such as posterior tibialis tendonitis, pattellofemoral pain syndrome (PFPS) and tibial stress syndrome (McKenzie *et al.,* 1985; Hintermann & Nigg, 1998). It is believed that the aforementioned chronic injuries are a result of abnormal, or excessive motion of the subtalar joint (McKenzie *et al.,* 1985; Hintermann & Nigg, 1998). Control of subtalar joint movement can be achieved through appropriate footwear selection and appropriate orthosis foot control (McKenzie *et al.,* 1985; Hintermann & Nigg, 1998). In 1967 the selection of shoes on the market was limited to 16 (McKenzie *et al.,* 1985). Today, the consumer has a choice of several hundred shoe types; however, there is yet to be a shoe designed specifically for runners who wear orthoses. This current research will provide insight for shoe designers and developers as to the relationship between rearfoot control offered by the shoe and orthoses while running. As well, a new tool will be used, vector coding, that will allow us to assess the joint coordination variability between the knee and rearfoot coupling continuously through stance. The results will hopefully prove useful to shoe designers and developers, physicians, physiotherapists, shoe vendor and the consumer as to the required shoe-orthoses combination for a patient presenting with mild lower-limb and foot dysfunction. The results will also hopefully lead to the design of a shoe specifically for those who wear orthoses.

1.4 Delimitations

1. Sample selection; female runners who are injury-free and symptom free and who wear orthoses with 3° of rearfoot posting and no forefoot posting.

2. The use of two shoe designs only.
3. Electrogoniometric analysis of knee flexion/extension and rearfoot eversion/inversion for ten complete gait cycles.
4. Pressure analysis of ten stance phases of ten complete gait cycles.
5. Questionnaire results for the orthoses condition only.
6. Running on a treadmill.

1.5 Limitations

1. The Questionnaire might result in recall bias
2. This study might not necessarily apply to male runners who wear orthoses with the same level of rearfoot control.
3. The findings may not necessarily apply to locomotion at a different pace.
4. The findings may not necessarily apply to running over ground.
5. Uni-planar measurement of rearfoot motion.
6. Uni-planar measurement of knee motion.
7. Use of two shoe designs from the same company
8. Attachment of the electrogoniometer to the shoe.
9. Data collection at two different frequencies for the two measurement system.

1.6 Hypotheses

1.6.1 Primary Outcome Measures

It is hypothesized that:

1. There will be a significant difference ($p< .05$) in variability of joint coordination (with respect to the coupling between the knee and rearfoot) between the motion control shoe and the neutral shoe conditions, with and without orthoses (MacLean, 2004: CSB abstract).
2. There will be a significant ($p< .05$) difference in the plantar surface loading patterns between the two shoe conditions, with and without orthoses.
3. There will be a significant ($p<.05$) relationship between the amount of navicular drop and the shoe-orthoses combination that is found to be most comfortable.

1.6.2 Secondary Outcome Measures

It is hypothesized that:

1. The motion control (M/C) shoe, will produce a significantly ($p< .05$) smaller maximum rearfoot angle (β_{max}) than the neutral shoe (Hamill, 1992).
2. The M/C shoe will produce significantly ($p< .05$) less, total rearfoot excursion (β) than the neutral shoe (Taunton *et al.*, 1985; Hamill, 1992).
3. Time to maximum eversion (β_T) during the stance phase will occur significantly ($p< .05$) later for the M/C shoe than for the neutral shoe. (Taunton *et al.*, 1985; Hamill, 1992).

4. Time to maximum eversion velocity ($\beta_{Tmaxvel}$) will occur significantly ($p<.05$) earlier in the stance phase for the M/C shoe (Hamill, 1992).

Chapter 2

2.0 Review of Literature

The science of running and biomechanics shows abundant research regarding the function of orthoses while walking or running, shoes and shoe sole modifications and their effect on biomechanics and locomotion and the use of orthoses and shoes as a modality for injury prevention and treatment. A challenge arises when interpreting this literature and using the findings from previous studies to shape the direction of the present study. For every study that claims one effect on rearfoot motion, change in Q-angle, or lower extremity intra-limb coupling as a result of orthoses or shoe intervention, there is another study claiming the opposite effect. One of the reasons behind the apparent discrepancy in results is due to the methodology adopted by the researchers. Two dimensional versus three dimensional video analysis, markers placed on the skin and shoe versus bone pins, the use of a constant testing shoe versus the subjects' own shoe, the use of standardised foot orthoses versus custom fabricated orthoses, speed of locomotion, treadmill running versus over ground running and method of data analysis and the use of discrete analyses of kinematic data rather than continuous analyses, are all differences in methodology which contribute to the discrepancy in results.

There has been a plethora of studies published claiming the positive clinical outcomes of orthoses in correcting abnormal pronation. There is also a general belief that abnormal pronation is a causative factor for running related injuries. However to date, there is very little evidence to suggest an exact mechanism for the role of orthoses and the effect unhealthy pronation has on the lower limb. There have been a few studies conducted in the past ten years that have moved away from disrete analysis of lower limb kinematics in favour of a dynamical systems approach originally discussed by Sparrow *et al.* (1987)

in an attempt to use continuous joint coordination to better understand running pathomechanics. This review of the literature will include a brief analysis of the epidemiology of running injuries, and some statistics relating to running-associated injuries, a brief review of running gait and the pathomechanics of running gait followed by a review of lower-extremity intra-limb coupling, the biomechanics of the lower limb and a thorough review of orthoses and shoe literature.

2.1 Epidemiology of Running Injuries

It has been suggested that running is one of the most common sports for the occurrence of overuse symptoms and injuries (Taunton *et al.,* 1988; Hrelijac, 2000). The definition of overuse injuries has evolved through the years to the definition of those injuries that occur as a result of repetitive stress placed on the body's structure or system beyond what the body is capable of withstanding, or adapting to (Taunton *et al.,* 1988; McKenzie *et al.,* 1986; Hreljac, 2000). Such overuse injuries include, but are not limited to: stress fractures, medial tibial stress, chondromalacia patellae, plantar fasciitis, and Achilles tendonitis (Hrelijac, 2000). The literature tends to divide the aetiological factors associated with overuse running injuries into five categories: training errors, inadequate strength and/or flexibility, the training surfaces used, running shoes used and biomechanical abnormalities (Taunton *et al.,* 1988; McKenzie *et al.,*1985; McKenzie *et al.,*1986; James *et al.,* 1978; Hreljac *et al.,* 2000; Duffey *et al.*, 2000; Clement *et al.,* 1981; Macintyre *et al.,* 1991; Taunton *et al.,* 2002; Nigg & Hintermann, 1998; Davlin & Evanski, 1991). During running, it is estimated that the foot will strike the ground approximately 1600 times per kilometer, and based on Newton's 3rd law, stating that for every action, there's an equal and opposite reaction, impact stresses at heel strike can reach levels of 275-300% of the runner's body weight, and thus poor biomechanical alignment can result in inordinate amounts of stress placed upon the kinetic chain (Brown & Yavorsky, 1987).

Duffey *et al.* (2000) and Hrelijac *et al.* (2000) were so bold as to suggest that all overuse running injuries were a result of training errors. It is their belief that the body has a predetermined limit for the number of kilometers the body can log running, the intensity at which the runner can train at, the amount of force and impact the body can tolerate and for how long, and the amount of rest required. If the body is pushed beyond this predetermined limit, injuries will ensue (Duffey *et al.,* 2000; Hrelijac *et al.*, 2000). This limit differs from one person to the next and is dependent on the remaining four etiological factors, specifically, strength and/or flexibility, training surface, shoes and biomechanics (Duffey *et al.,* 2000; Hrelijac *et al.*, 2000). With improvements in strength and flexibility, biomechanics, and appropriate shoe and training surface changes, the ability of the body to increase its limit for withstanding the stresses imparted on it through running, will improve (McKenzie *et al.,* 1985; Taunton *et al.,* 1988).

There have been several retrospective studies published in the last twenty five years that examine the incidence and location of running-related overuse injuries, attended to at a given sports medicine clinic. James *et al.* (1978) published the first known, of such studies. They found that in a chart review of 180 runners, 65% of running-related injuries were amongst distance runners and 24% amongst joggers. Of the injuries, 60% were due to training errors, 29% of which were associated with excessive mileage (average of 49 miles per week) (James *et al.* 1978). Other training errors included intense track workouts on a hard track surface wearing a spiked shoe, a rapid and sudden change in training regime and hill running (James *et al.* 1978). The most common overuse injuries amongst this population were knee pain (29%), posterior tibialis syndrome (13%), Achilles tendonitis (11%), plantar fasciitis (7%), and stress fractures (6%) (James *et al.* 1978). Of the problems associated with knee pain, chondramalacia accounted for 25% of diagnoses, iliotibial band friction syndrome (ITBFS) accounted for 17% and peripatellar pain accounted for 15% (James *et al.* 1978). The most effective treatment modalities were rest (47%), orthotics (46%) reduced mileage (26%) and shoe change/modification (19%) (James *et al.* 1978).

A series of three major retrospective reviews at the sports medicine clinic at the University of British Columbia (Vancouver, BC) were conducted over a twenty-year period. The first study was published by Clement *et al.*, in 1981, and included an analysis of 1650 patients seen from 1978 – 1980 at the clinic. The second analysis was published in 1991 by Macintyre *et al.* and included a survey of 4173 patients seen from the same population as the previous survey, between 1980 and 1985. The third study was published in 2002 by Taunton *et al.* and included a survey of the clinical records of 2002 patients with running related injuries seen at the same clinic between 1998 and 2000. In all three surveys of the patients' clinical records, the knee was the most common site of injury, with patellofemoral pain syndrome (PFPS) the most common diagnosis. A change in the pattern of injury was seen between the first and second study, with a greater proportion of injuries occurring at the knee in the 1991 study, and a relatively fewer number of lower leg and foot injuries (Macintyre *et al.*, 1991). This change in injury pattern was attributed to improvements in footwear design (Macintyre *et al.*, 1991). However, footwear was also implicated in the change in pattern of ITBFS from 1981 –

2000. In 1981, ITBFS accounted for 4.3% of running related injuries, in 1991, 7.5% of running related injuries and in 2000, 8.4%. Shoe technology has improved over the last 20 years such that medial posts or varus wedges have been constructed into the shoe sole to help align the skeleton. If worn properly, the shoe sole construction can have a significant positive effect on the lower limb alignment (Nigg *et al.,* 1998), however, it can also have a significant negative impact as witnessed with the increase in ITBFS.

2.2 Biomechanics of healthy running gait

In order to have a clear understanding of the biomechanical factors implicated in overuse running injuries, it is important to be familiar with the sequences of events that occur in the lower extremity in normal gait function. The gait cycle is comprised of a swing phase and a stance phase (*See Figure 2.1*). The stance phase is further divided into three stages, a contact phase a midstance phase and a final propulsive phase. The contact period, accounting for approximately 27% of total stance, begins with foot strike and ends as the forefoot becomes fully weight-bearing (Brown & Yavorsky, 1987; Knutzen & Price, 1994). The midstance phase, accounting for approximately 40% of total stance is characterized by single limb support in the mechanics of normal walking gait (Brown & Yavorsky, 1987; Knutzen & Price, 1994). And finally the propulsive phase, accounting for approximately 33% of total stance, begins with heel lift and ends with toe-off (Brown & Yavorsky, 1987; Knutzen & Price, 1994).

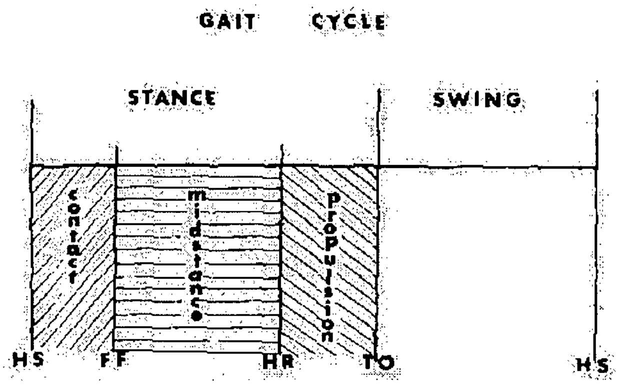


Figure 2.1. Phases of the gait cycle. Heel strike (HS), foot flat (FF), heel rise (HR) and toe off (TO). From: Tiberio, D. The Effect of Excessive Subtalar Joint Pronation on Patellofemoral Mechanics: A Theoretical Model. *Journal of Orthopaedic and Sports Physical Therapy* 9(4): 160 – 165, 1987.

2.2.1 Contact Phase

Just prior to foot strike, the hip is flexed ~30º, the knee is in close to full extension, the dorsiflexors and inverters of the ankle are contracted in preparation for deceleration of plantarflexion and pronation that happens just following foot strike. In addition, the tibia is externally rotated and the foot is supinated (inversion of the subtalar joint, ankle plantar flexion and fore foot adduction) with the midtarsal joint locked (fully pronated about its oblique axis and supinated about its longitudinal axis) to stabilize the forefoot. The midtarsal joint is maintained in this position throughout the contact phase due to the anterior compartment muscles being under tension in order to control plantarflexion. At foot strike, the hip extends, the knee flexes and the tibia begins to internally rotate as controlled pronation (eversion of the calcaneus, ankle dorsiflexion and forefoot abduction) begins as a result of the combined ground reaction and inertial forces. Controlled pronation is a result of the anterior compartment muscles of the lower limb acting eccentrically to resist plantarflexion. The midtarsal joint unlocks and the forefoot becomes unstable allowing the foot to adapt to the surface and absorb and dissipate the ground reaction forces. A healthy foot will reach maximum pronation (roughly 4 – 6° of calcaneal eversion) at the end of the contact phase, just prior to the beginning of midstance. Once the ankle reaches a position of full dorsiflexion (~10°), the forefoot is in contact with the ground such that the ground reaction forces (GRFs) cause the foot to begin dorsiflexion; the GRFs also act to maintain the midtarsal joint in its aforementioned position as the anterior musculature are no longer under tension (James *et al.,* 1978; Cavanagh, 1980; Clarke *et al.,* 1983; McKenzie *et al.,*1985; Brown & Yavorsky, 1987; Tiberio, 1987; Taunton *et al.,* 1988; Lattanza *et al.*, 1988; Hintermann & Nigg, 1998; Knutzen & Price, 1994; Michaud, 1997).

An important clinical consideration of the contact phase of the gait cycle is the absorption of GRFs that occurs (Michaud, 1997). The controlled plantarflexion as a result of the eccentric muscles of the anterior compartment allow for the GRFs to be absorbed over a longer time. Similarily, the plantarflexion of the talus allows the ankle mortise to be lowered allowing the supporting musculature more time to dampen the forces imparted on the body, by the ground (Michaud, 1997). In addition to plantarflexion, as the subtalar joint pronates, the talus also adducts thereby causing tibial internal rotation; the internal rotation of the tibia allows the knee to flex, thereby allowing more forces to be dampened by the surrounding musculature of the knee (Michaud, 1997).

2.2.2 Midstance Phase

Midstance is characterized by a position of forefoot loading. As the body moves over the fixed foot into the midstance phase of the gait cycle, the hip is in a position of full extension. The contralateral swing phase leg causes the pelvis to externally rotate on the stance phase limb which in turn causes the lower limb to externally rotate, the talus abducts and dorsiflexes on the calcaneus, and the calcaneus inverts which results in subtalar joint supination. During the first half of midstance, the rearfoot supinates from a position of maximum pronation, to subtalar joint (STJ) neutral. Once STJ neutral has been reached, the GRFs in the rearfoot decrease and the STJ continues to supinate, the midtarsal joint locks up, converting the foot from a mobile adapter into a rigid lever for the gastroc-soleus complex to push off of. As the rearfoot moves into a vertical position, the midtarsal joint locks as a result of the superior border of the pronating cuboid coming into contact with the dorsal border of the calcaneus representing an osseous locking mechanism. This locking of the forefoot against the rearfoot allows for efficient propulsion as smooth transition of forces along the boney column is facilitated. By the end of midstance, the osseous locking mechanism described earlier will ideally cause the rearfoot to be perpendicular to the ground thus improving the functional alignment of the surrounding tissue and reducing asymmetric stresses (James *et al.,* 1978; Cavanagh, 1980; Clarke *et al.,* 1983; McKenzie *et al.,*1985; Brown & Yavorsky, 1987; Tiberio, 1987; Taunton *et al.,* 1988; Lattanza *et al.*, 1988; Hintermann & Nigg, 1998; Knutzen & Price, 1994; Michaud, 1997).

2.2.3 Propulsion Phase

The propulsive phase begins the instant heel lift occurs. Heel lift occurs as a result of the center of mass being displaced over the forefoot, and the contraction of the soleus and posterior compartment muscles acting to resist plantarfllexion and as a result, decelerating the forward momentum of the proximal tibia. The gastrocnemius muscle simultaneously flexes the knee and plantarflexes the ankle thereby causing the heel to lift (See *Fig 2.2*). During the final stage of the contact phase of gait, the body weight is shifted from a more lateral position on the foot to the medial side and through the hallux. This movement is accomplished by the continuation of external rotation of the leg and rapid knee flexion thereby causing closed chain supination of the foot. In order for effective propulsion to occur, the 1st MTP joint must be stabilised thus promoting dorsiflexion as the weight is transferred from the 1st ray to the hallux and then toe-off with transfer of weight to the contralateral limb (James *et al.,* 1978; Cavanagh, 1980; Clarke *et al.,* 1983; McKenzie *et al.,*1985; Brown & Yavorsky, 1987; Tiberio, 1987; Taunton *et al.,* 1988; Lattanza *et al.*, 1988; Hintermann & Nigg, 1998; Knutzen & Price, 1994; Michaud, 1997).

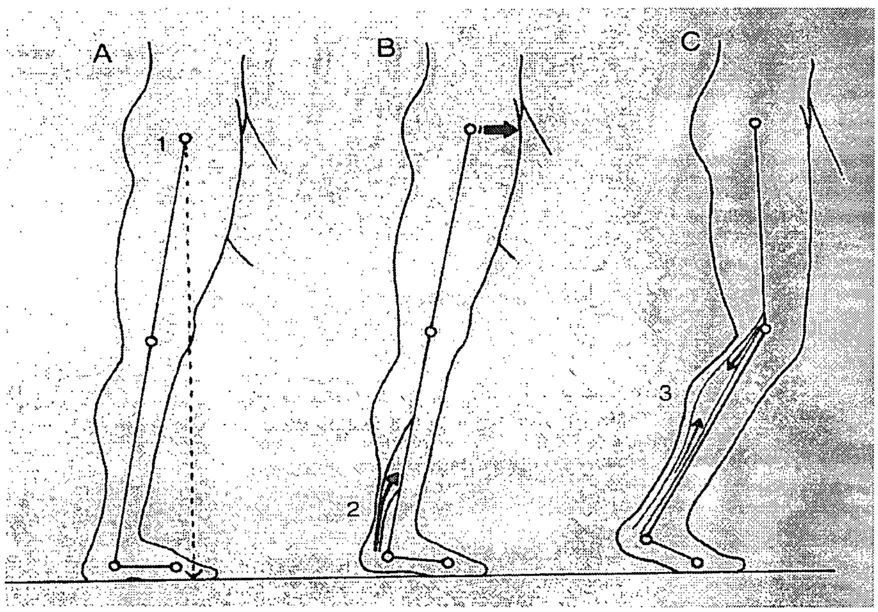


Figure 2.2. Heel lift occurs due to a shift in center of mass over the forefoot (1), contraction of the posterior compartment muscles of the lower leg to decelerate dorsiflexion (2) and contraction of the gastrocnemius muscles that causes rapid knee flexion (3). From: Michaud, TC. (1993). *Foot Orthoses and other forms of conservative foot care.* Thomas C. Michaud. USA.

Brown and Yavorsky propose criteria for normalcy that may prevent overuse injuries or pathological conditions along the kinetic chain (Brown and Yavorsky, 1987). Their criteria for biomechanical "normalcy" must be met prior to heel lift (*See Fig. 2.3*), and suggests that the distal third of the lower leg be vertical, i.e. no tibial vara; the subtalar, talocrural and knee joints lie in the transverse plane and be parallel to the ground; the bisection of the posterior surface of the calcaneus lie vertical or failing that, inverted no more than 4°; the midtarsal joint be locked in its max position of pronation; the metatarsals and plantar surface of the calcaneus should be parallel with one another and there should be no rotational or torsional movements in the lower leg (Brown and Yavorsky, 1987). James *et al.* (1978) suggest that this position of the foot is one in which the foot will function most efficiently and with the least amount of stress imposed on the surrounding joints, ligaments and tendons. Root *et al.* as cited in Neely (1998) are in agreement that the distal third of the tibia be perpendicular to the floor in the frontal plane with the metatarsal heads perpendicular to the heel, however he proposes that an acceptable amount of variance of 2° of varum or valgum, with the STJ in neutral, will not adversely affect the foot.

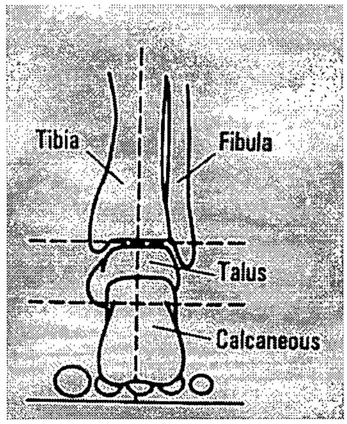


Figure 2.3. The neutral foot and lower leg as it relates to biomechanical normalcy. From: Brown LP & Yavorsky P. Locomotor Biomechanics: A Review. *Journal of Orthopaedic and Sports Physical Therapy* 9(1): 3 – 10, 1987.

During the contact and midstance phases, the coupled relationship between the ankle and the knee, knee flexion with STJ pronation and knee extension with STJ supination, as seen in *Figure 2.4*, are interdependent motions while the tibial rotation that occurs is a necessary requirement for normal kinematics of the knee and ankle (Tiberio, 1987; Michaud, 1997).

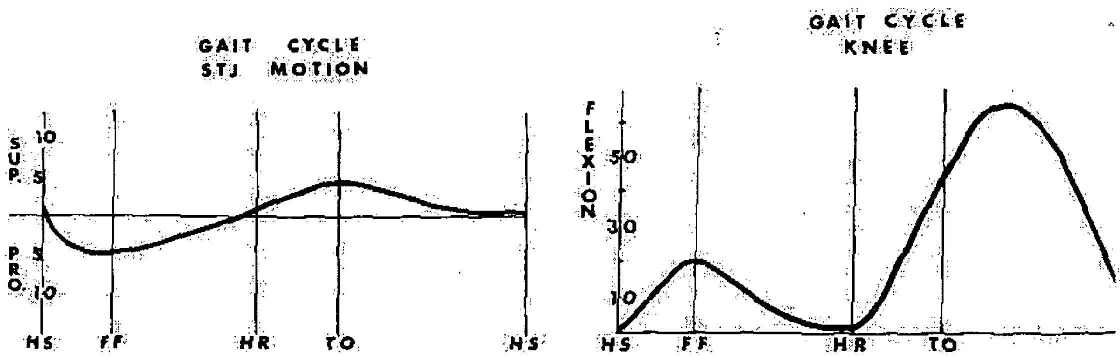


Figure 2.4. Normal motion of the subtalar joint and knee joint during the stance phase of the gait cycle. From: Tiberio, D. The Effect of Excessive Subtalar Joint Pronation on Patellofemoral Mechanics: A Theoretical Model. *Journal of Orthopaedic and Sports Physical Therapy* 9(4): 160 – 165, 1987.

2.3 Pathomechanics of running gait

In the review of 180 clinical charts of patients presenting with overuse injuries as a result of running by James *et al.*, (1978), variance in normal biomechanical alignment of the kinetic chain was seen repeatedly amongst those complaining of knee pain associated with running. Examination of these patients revealed femoral neck anteversion, genu varum, squinting patellae, excessive Q-angle, tibia varum, functional equinus and pronated feet (James *et al.,* 1978). In the retrospective case analysis by Taunton *et al.,* (2002), varus and valgus knee alignment were implicated in PFPS, ITBFS, stress fractures and plantar fasciitis, with varus alignment contributing to PFPS and ITBFS to a greater extent.

Compensation of certain structures of the body in an attempt to adjust to deviations in structure, function or position of another body part is a normal occurrence in locomotion (Lattanza *et al.,* 1988). When this compensation becomes repetitive in nature due to the repetitive demand placed on one structure of the body by another, pathology or overuse injuries may present (Lattanza *et al.,* 1988). For example, pronation is considered to be a compensatory action for knee flexion, however, pronation that exceeds what is required for the task, or occurs when supination should be occurring, is considered unhealthy (Lattanza *et al.,* 1988; Powers, 2003). Intrinsic causes of unhealthy compensatory pronation include subtalar varus, forefoot varus, tibial varum, forefoot supinatus and ankle joint equinus (James *et al.,* 1978; McKenzie *et al.,* 1986; Taunton *et al.,* 1988; Neely, 1998 Hintermann & Nigg, 1998; Hreljac *et al.,* 2000; Powers, 2003). These biomechanical irregularities can lead to unhealthy compensatory action of the involved structures by creating an environment that leads to an unstable foot during the stance phase of gait and can contribute to hypermobility of the foot, subluxation of bones of the foot, and resultant microtrauma and overuse symptoms (Hintermann & Nigg, 1998). Similarly, unhealthy supination is that in excess of what is required for the given task, or occurs out of sequence (Lattanza *et al.,* 1988; Powers, 2003). Causes of unhealthy supination include forefoot valgus, a plantarflexed 1st ray and equinus deformities of the ankle (Hintermann & Nigg, 1998).

Rearfoot control is referred to throughout the literature, and is best described as the ability of a shoe to limit the amount and or rate of foot pronation following heel strike (Clarke *et al*, 1983). Although rearfoot control is an important modality for achieving proper gait function, too much control may cause problems along the kinetic chain (Michaud, 1993). Pronation within normative values is a necessary function of normal gait as it provides shock absorption through the dissipation of forces (Clarke *et al*, 1983).

The prevention of a "normal" degree of subtalar joint pronation, either through orthoses, shoe construction (i.e. a motion control shoe), or a combination of both, results in a cavus type foot that reduces shock absorption and can cause an excessive amount of force to travel up the leg, into the pelvic area and even the spine (Michaud, 1993). Runners with decreased motion at the subtalar joint have a compensatory decrease in internal rotation of the tibia, resulting in a decreased ability of the foot to absorb the GRFs (McKenzie *et al.*, 1985). However, excessive pronation can lead to pathological conditions along the kinetic chain due to antagonism between tibial rotation and foot function, causing torsion at the knee (Clarke *et al*, 1983; Powers, 2003).

Measures of angular displacement of the calcaneus are an indicator of the level of stress imposed on the Achilles tendon. With healthy lower-limb alignment, GRFs are directed vertically and symmetrically up the tendon whereas with calcaneal eversion, the tendon is forced laterally and the pressure is distributed asymmetrically across the tendon (Clement *et al.*, 1981; Stacoff *et al.*, 2000). Stacoff *et al.* (1981) found excessive eversion velocity to be associated with medial tibial stress syndrome, while in a study conducted by Hrelijac *et al.* (2000) with 20 injured and 20 non-injured subjects, there was a trend toward more rapid pronation and greater touchdown supination in the injury free group, however, statistical significance was not achieved. This study was in direct contradiction with previous studies showing that injured subjects exhibited greater calcaneal angular displacement and eversion velocity (Hrelijac *et al.*, 2000). A reduction in the rate and amount of pronation was found by Kuhn *et al.* (2002) to reduce the amount of tibial and femoral internal rotation with a subsequent reduction in Q-angle.

2.3.1 Lower Extremity Intra-Limb Coupling

It has been suggested that the asynchronous actions of the lower-extremity segments lead to anterior knee pain (Tiberio, 1987; Brown and Yavorsky, 1987; Nawoczenski, 1995; Hintermann & Nigg, 1998; Heiderscheit *et al.*, 2000; Duffey *et al.*, 2000). Namely, there is a coupling of rotational actions that occur at the knee, shank and ankle. The research suggests that following heel contact, the subtalar joint pronates, the tibia internally rotates and the knee flexes (*See Fig. 2.5*) (Bates, 1979; Tiberio, 1987; Brown and Yavorsky, 1987; Nawoczenski, 1995; Hintermann & Nigg, 1998; Heiderscheit *et al.*, 2000; Duffey *et al.*, 2000). Conversely, as the knee extends, the tibia externally rotates and the subtalar joint supinates (Bates, 1997; Tiberio, 1987; Brown and Yavorsky, 1987; Nawoczenski, 1995; Hintermann & Nigg, 1998; Heiderscheit *et al.*, 2000; Duffey *et al.*, 2000). It has been investigated and suggested that decoupling of these events occur when maximum rearfoot eversion occurs before midstance, or after midstance (Tiberio, 1987; Brown and Yavorsky, 1987; Nawoczenski, 1995). If maximum knee flexion and rearfoot eversion are reached at different times in the stance phase, knee joint dysfunction may occur due to conflicting requirements of the tibia (Tiberio, 1987; Hintermann & Nigg, 1998; Heiderscheit MS *et al.,* 2000). For example, if the maximum knee flexion angle is achieved at midstance, yet the maximum rearfoot eversion angle is delayed beyond the onset of midstance, the tibia will be caught in an antagonistic environment whereby it will be required to simultaneously externally rotate with knee extension, yet continue to internally rotate to accompany the prolonged rearfoot eversion (Brown and Yavorsky, 1987; Nawoczenski, 1995; Heiderscheit, 2000).

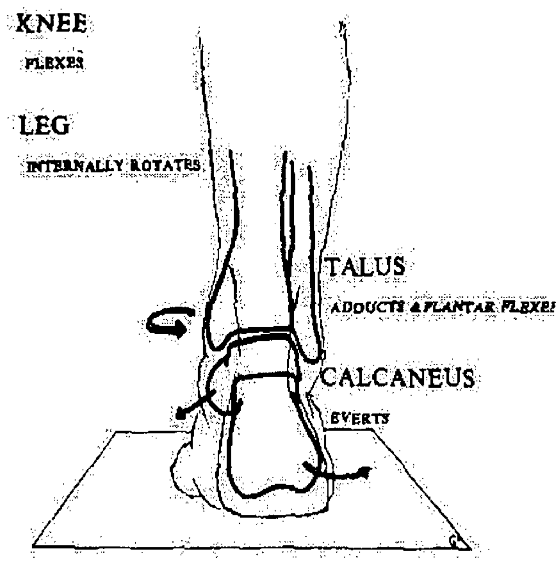


Figure 2.5. Closed chain pronation of the subtalar joint.
From: Tiberio, D. The Effect of Excessive Subtalar Joint Pronation on Patellofemoral Mechanics: A Theoretical Model. *Journal of Orthopaedic and Sports Physical Therapy* 9(4): 160 – 165, 1987.

Tiberio (1987) proposed a theoretical biomechanical model whereby if the tibia cannot externally rotate due to prolonged STJ pronation, in order for knee extension to occur, the locomotor system will compensate through internal rotation of the femur in order to reduce trauma to the tibiofemoral joint. Tiberio (1987) believes that this compensatory internal rotation of the femur (CIRF) will provide the necessary amount of rotation in order for knee extension to occur. CIRF however disrupts the normal mechanics of the gait cycle and in turn, alters patellofemoral tracking (Tiberio, 1987). At the onset of CIRF, the quads are contracting while the patella glides in the femoral trochlear groove (Tiberio, 1987). As the femur internally rotates, the compression between the lateral articular surfaces of the patella and femoral condyle is increased, thus creating anterior knee pain (Tiberio, 1987). The degree of pain experienced is dependent upon the degree of abnormal pronation as this will dictate the amount of patellofemoral compression that occurs (Tiberio, 1987).

2.3.2. Factors affecting the timing of lower extremity joint actions

The concept of lower extremity intra-limb coupling and the believed potential effects of decoupling on the propagation of overuse running injury have sparked further research designed to identify the effect of changing anthropometric variables and equipment on the lower-extremity coupling mechanism. Bates *et al.* (1979) examined the effects of a custom foot orthosis (CFO) on the timing of lower limb events, while Hamill *et al* (1992) examined the effects of varying midsole composition on the timing of lower limb joint actions. Finally, Nigg *et al.* (1993) and Nawoczenski *et al.* (1998) examined the effects of arch height on lower extremity intra-limb coupling.

Bates *et al.* published a study in 1979 that assessed the effects of foot orthotics on foot and leg running mechanics for runners who had been classified as "pronators" and wore foot orthoses. The subjects were filmed using two high speed video cameras while running on a treadmill at their self selected pace. The subjects were filmed under three conditions: barefoot, test shoe, and test shoe with an orthotic. All subjects wore the same standard Nike shoe for testing. This study found that the timing of maximum knee flexion and maximum pronation occurred at approximately the same time, when the center of gravity passed over the base of support (Bates *et al.*, 1979). The researchers found significant differences between the orthotic and shoe conditions when comparing the beginning and ending of pronation, maximum ankle dorsiflexion and period of pronation ($p<0.15$) (Bates *et al.*, 1979). The values for the orthotic conditions indicated that pronation started later, and ended earlier than the barefoot conditions, and the period of pronation was shorter (Bates *et al.*, 1979). Maximum ankle dorsiflexion was increased in the orthotic condition. The shod condition produced intermediate values between the two extreme conditions (Bates *et al.*, 1979). Further analysis reveals that there were significant differences between barefoot and shoe for timing of the beginning of pronation and maximum knee flexion (Bates *et al.*, 1979). The data indicate foot orthoses have the ability to alter joint action timing in the lower limb, and that shoes are able to significantly alter the joint actions involved in running gait. Mean absolute values for the maxima of pronation, knee flexion and ankle dorsiflexion were also collected and compared. The results indicate that orthotics were able to significantly reduce the amount of pronation and knee flexion when compared with the barefoot condition.

Hamill, Bates and Holt conducted a study that examined the effects of varying midsole densities on the timing of lower limb joint action (Hamill *et al.*, 1992). The shoes they used were identical in construction except for their midsole durometer: 70 (C1), 55 (C2) and 45 (C3) on a Shore A scale (Hamill *et al.*, 1992). Subjects ran on a treadmill and the right lower extremity was filmed in the sagittal plane and from the rear (Hamill *et al.*, 1992). No differences were seen between the two harder sole conditions (C1/C2) however differences were observed between C1/C2 and C3 for selected parameters (Hamill *et al.*, 1992). As well, no significant differences were realized for knee angle parameters (*See Fig. 2.6*) (Hamill *et al.*, 1992). The softer midsole (C3) produced a larger rearfoot angle and it occurred earlier in stance phase (*See Fig. 2.7)* ($p<0.05$). Although significant values were not attained, the trend for rearfoot angular velocity was that the harder midsoles produced a larger angular velocity, occurring earlier in the stance phase (Hamill *et al.*, 1992). When comparing the ankle and knee joint coupling, the time to maximum rearfoot angle occurred significantly earlier than the time to maximum knee flexion angle for the C3 condition 39.7% of stance phase vs. 45.9% of stance phase, respectively (*See Fig. 2.8*)(Hamill *et al.*, 1992). The combination of the knee joint action not being significantly effected by changing midsole durometers, with the fact that the soft midsole produced a greater rearfoot angle, occurring earlier in the stance phase, and resupinating later in the stance phase indicates a possible antagonistic environment is present in the lower limb when wearing a shoe with a soft midsole (Hamill *et al.*, 1992). If the rearfoot begins to supinate while the knee is still flexing, then the tibia is forced into external rotation at the distal end and internal rotation and the proximal end, creating torsional stress on the tibia (Hamill *et al.*, 1992).

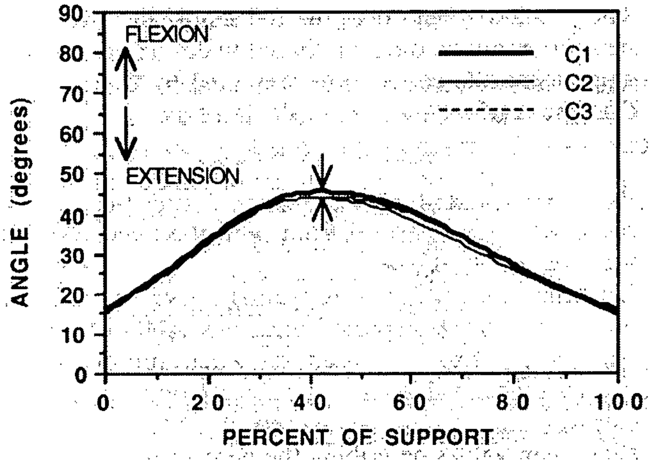


Figure 2.6. Knee flexion and extension patterns for three different midsole durometers: C1 (70), C2 (55) and C3(45) on a Shore A scale. From: Hamill *et al.* Timing of lower extremity joint actions during treadmill running. *Med. Sci. Sports Exerc.* 24(7): 807 – 813, 1992.

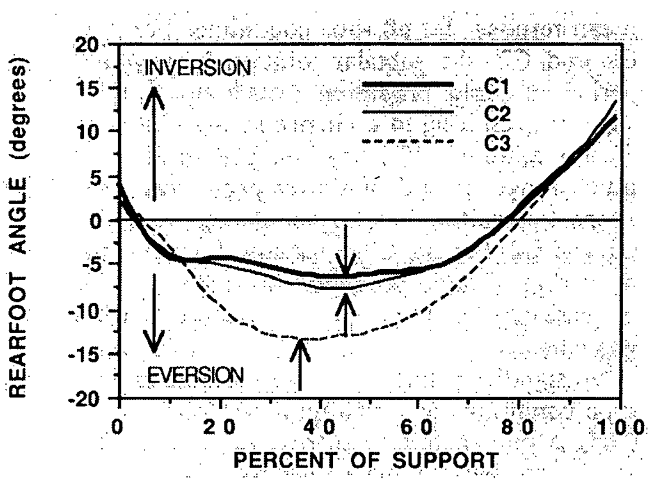


Figure 2.7. Rearfoot angle measures throughout stance phase for subjects wearing three different shoes with three different midsole durometers: C1 (70), C2 (55) and C3(45) On a Shore A scale. From: Hamill *et al.* Timing of lower extremity joint actions during treadmill running. *Med. Sci. Sports Exerc.* 24(7): 807 – 813, 1992.

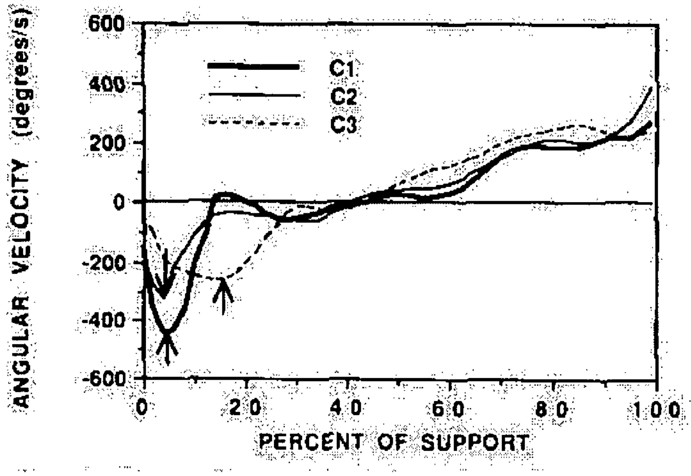


Figure 2.8. Rearfoot angular velocity measures for subjects wearing three different shoes with three different midsole durometers: C1 (70), C2 (55) and C3(45) On a Shore A scale. From: Hamill *et al.* Timing of lower extremity joint actions during treadmill running. *Med. Sci. Sports Exerc.* 24(7): 807 – 813, 1992.

Nigg *et al* (1993) and Nawoczenski *et al* (1998) conducted studies to identify the effect arch structure had on lower extremity motions. The study by Nigg *et al.* was designed to quantify the relationship between the height of the medial longitudinal arch of the foot and (a) the eversion movement of the foot (b) the axial rotation of the leg and (c) the transfer of eversion to internal leg rotation during heel-toe running. The relationship would prove to provide insight into the functional relationship between arch height and injury as a result of intralimb coupling. This study defined arch height as being the highest point along the soft tissue margin of the medial plantar curve, when the subject was in a full weight bearing position (Nigg *et al.*, 1993). Markers were placed on the lower leg and foot with adhesive tape and video analysis was conducted with four cameras sampling at 200 frames s^{-1}. Force data was also collected at 1000 Hz while subjects ran at 4.0 m s^{-1}. A significant correlation was found between maximum internal leg rotation and arch height (r^2=0.152, $p<0.033$) with a trend indicating that as arch height increased, there was increased internal leg rotation with respect to the foot (Nigg *et al.,* 1993). A significant correlation was also found between the transfer coefficient (T_{fl}) and arch height (r^2=0.267, $p<0.0034$), meaning that as arch height increased there was an increase in transfer of foot eversion to internal leg rotation (Nigg, *et al.*, 1993). The researchers found however, only 27% of the variance in transfer coefficient could be explained by arch height, which indicates that there must be other variables involved in the transfer of rearfoot eversion to tibial rotation (Nigg, *et al.,* 1993). The study went further to conduct a regression analysis to determine the relationship between inversion/eversion and leg rotation using the mean values of the two variables. A strong correlation (r^2=0.9991, $p<0.0001$) was found which supports the assumption made by previous researchers that foot eversion is indeed coupled with lower limb axial rotation (Nigg *et al.,* 1993). Inman (1976) wrote a book on the joints of the ankle. He related the inclination of the STJ axis to the amount of tibial rotation that occurs during locomotion. Inman believed that the more vertical the STJ axis was in the sagittal plane, the amount of talar rotation per degree of lower leg rotation would decrease; a higher arched foot would have a more vertical STJ axis while a flatter arch would have a more horizontal STJ axis *(See Fig. 2.9)* (Inman, 1976). His theory suggested that an STJ axis of ~42° in the sagittal plane would produce equal amounts of calcaneal in/eversion and tibial in/external rotation (i.e. for every 1° of calcaneal eversion there would be 1° of tibial internal rotation) (Inman, 1976). However, the STJ axes vary amongst individuals and thus, there must be a relationship between increased/decreased arch height (vs. 45°) in the sagittal plane with calcaneal and tibial rotation.

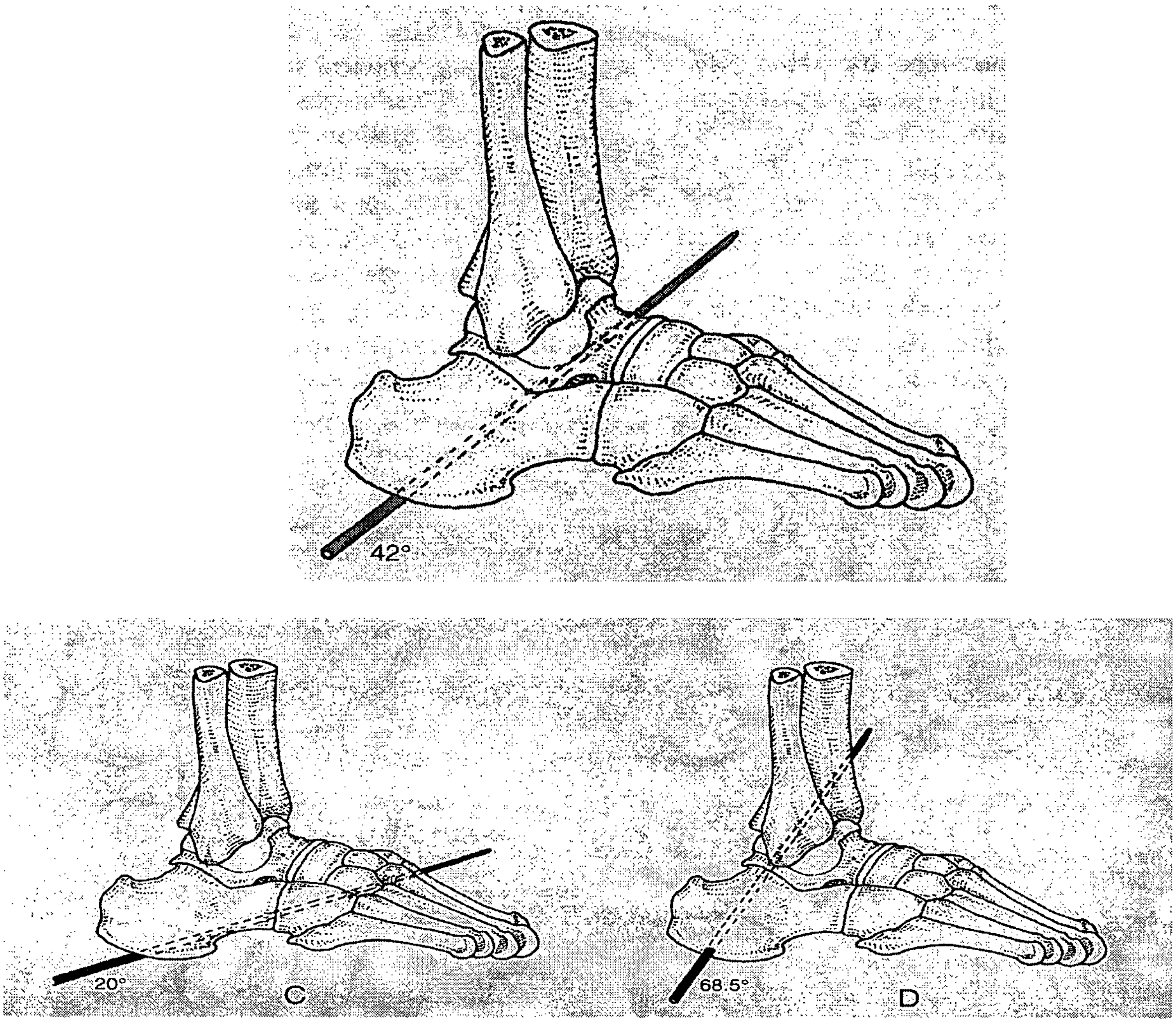


Figure 2.9. Axis of STJ that produces 1° of tibial axial rotation for 1° of calcaneal rotation according to Inman (1976) (Top). Representations of varying STJ axes: 20° (C) to 68.5° (D). From: From: Michaud, TC. (1993). *Foot Orthoses and other forms of conservative foot care.* Thomas C. Michaud. USA.

Nawoczenski and colleagues conducted a study in 1998 that further explored the relationship that Nigg *et al* (1993) identified. This study however classified their subjects into two groups according to radiological evidence: a high arch group (clinically defined as pes cavus) and a low arch group (clinically defined as pes planus) (Nawoczenski *et al.*, 1998). This research focussed on the stance phase tibial internal/external rotation and calcaneal in/eversion as well as their coupled relationships with the STJ and talocalcaneal joints (Nawoczenski *et al.*, 1998). Ten subjects were included in each group

(Nawoczenski *et al.*, 1998). Subjects ran in sandals on a treadmill at their self-selected pace while being filmed at 50 frames s^{-1}. Coupling ratios were determined for the calcaneus and lower leg rotations (Nawoczenski *et al.*, 1998). A coupling relationship of 1.53 for the low rearfoot group and 0.91 for the high rearfoot group was found (Nawoczenski *et al.*, 1998). A coupling relationship >1 indicates that the predominate component of the coupling relationship is calcaneal rotation (Nawoczenski *et al.*, 1998). A coupling relationship <1 indicates that the predominate component of the coupling relationship was axial rotation of the lower limb (Nawoczenski *et al.*, 1998). This study confirms Inman's theory relating the position of STJ and talocalcaneal axes to axial rotation of the lower limb whereby an STJ axis >45° will produce greater tibial rotation and an STJ axis <45° will produce greater internal calcaneal rotation (Inman, 1976; Nawoczenski *et al.*, 1998).

2.3.3 Dynamical Systems approach to understanding joint coupling and joint coordination.

The previous sections, 2.3.1 and 2.3.2 introduced intra-limb coupling and described some of the factors associated with intra-limb coupling. The bulk of previous research conducted on running pathomechanics has focussed on the timing of events that occur with respect to specific instances in time, these methods are called Discrete Relative Phase (DRP) methods. Discrete methods provide information based on the relative timing of two oscillators (i.e. joints) at specific points in time. A joint coupling of interest to many is that between the rearfoot and the knee. If one were concerned with the timing of maximum rearfoot eversion versus the time at which maximum knee flexion occurred, they would employ the following calculation to determine the DRP angle (ϕ):

$$\phi = t_1 - t_2 / T * 360°$$

Where, 't_1' is the time to maximum flexion angle, 't_2' is the time to maximum ankle eversion angle, and T is the period of support. The interpretation of ϕ is simple, if the resultant DRP angle = 360°, the two oscillators are synchronous with one another; angles ranging from 1 – 359° indicate that the two oscillators are out of synchrony (Hamill et al. (2000)). The disadvantage of this system is that only specific instances in time are able to be evaluated; no information about the joint coordination is available for the rest of the cycle.

There are continuous methods of analysis that are available that allow the researcher to analyse a full gait cycle, or break it up into components such as stance or swing. These methods are called Continuous Relative Phase (CRP) methods. Depending on whether the oscillators of interest exhibit sinusoidal-like motion, one method is preferred over the other. The method of interest for this research is a method that was initially proposed by Sparrow *et al.* in 1987 and later modified by Heiderscheit *et al.* in 2002. The method is called vector coding and is used for oscillators that are not sinusoidal in nature. Vector coding is a method that quantifies coordinative pattern between two oscillators (i.e. joint couples) and assess the joint coordination variability (Sparrow *et al.*, 1987; Tepevac and Field-Fote, 2001; Hamill *et al.* 2000; Heiderscheit *et al.* 2000; Heiderscheit *et al.* 2002). Through the use of angle-angle plots and circular statistics, vector coding allows the researcher to quantify joint coordination and variability (Hamill *et al.* 2000; Heiderscheit *et al.* 2000; Heiderscheit *et al.* 2002).

An angle-angle plot as is seen in *Figure 2.10* is produced based on the angular data of two oscillators, in this case it's the knee angle in the sagittal plane (plotted on the abscissa) and rearfoot angle in the frontal plane (plotted on the ordinate) for one entire stance phase. The procedure begins by calculating the coupling angle (γ) between two adjacent

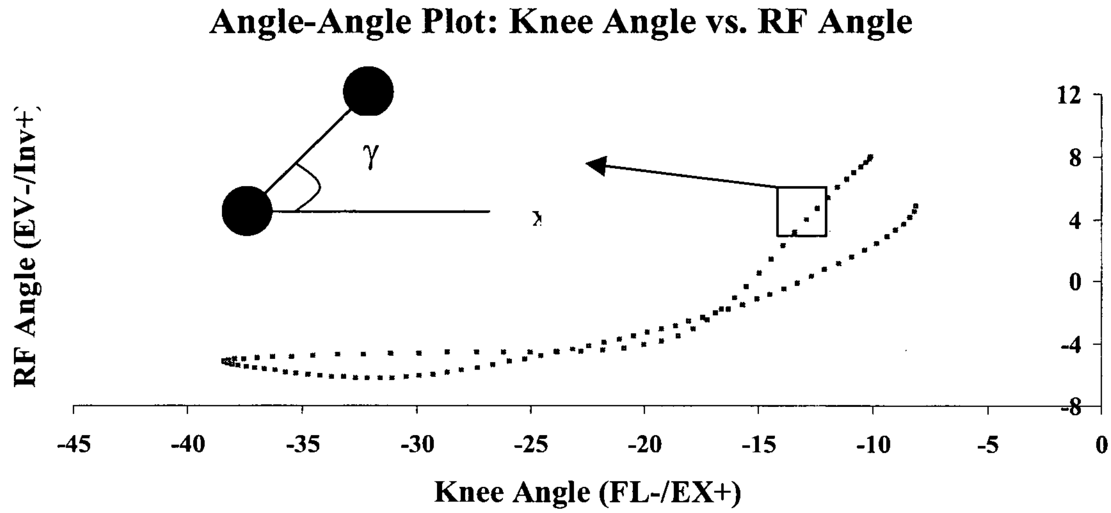


Figure 2.10. Example of vector coding technique to calculate coupling angle.

data points. The coupling angle is defined as the angle between a vector adjoining to adjacent data points and the right horizontal. Coupling angle ranges from 0°-360° and the interpretation of the coupling angle (γ) is as follows (Hamill et al. (2000)):

γ = 0°, 90°, 180°, or 270°: indicates movement of one joint or segment,

γ = 0° or 180°: indicates movement of proximal oscillator, distal oscillator is fixed,

γ = 90°or 270°: indicates movement of the distal oscillator, proximal oscillator is **fixed,**

γ = 45° or 225°: indicates equal relative movement of both oscillators in the same direction, and

γ = 135° or 315°: indicates equal relative motions but in opposite directions.

Circular statistics were thus required in order to convert angles ranging from 0° - 180° to a range between 0° and 90°. The mean direction of the vector (or (x_{mean},y_{mean}) coordinates of the vector) for each instant in time during the movement sequence is determined by calculating the mean cosine and sine of each directional component of the coupling angle (γ):

$$\mathbf{x_{mean} = 1/\mathit{n}\ \Sigma\ cos\gamma}$$

$$\mathbf{y_{mean} = 1/\mathit{n}\ \Sigma\ sin\gamma}$$

The mean direction of γ is then calculated as:

$$\textbf{If } y_{mean} > 0,\ \gamma_{mean} = \tan^{-1}(y_{mean} / x_{mean})$$

$$\textbf{If } y_{mean} < 0,\ \gamma_{mean} = 180 + \tan^{-1}(y_{mean} / x_{mean})$$

The length of mean vector (r) provides an estimate of the spread of data as it reflects the directional concentration of the data (Heiderscheit Dissertation, 2000):

$$r = (x_{mean}{}^{2} + y_{mean}{}^{2})^{1/2}$$

Lastly, the most suitable measure of angular variance is measured as follows (Heiderscheit Dissertation, 2000):

$$S^2 = 2(1 - r)$$

The coupling angle (γ) was then calculated as the angle between a vector connecting two adjacent data points, and the right horizontal (Heidercheit, 2002). In order to quantify joint coordination, a modification of the technique proposed by Sparrow *et al.* (1978), by Heiderscheit *et al.* (2002). The variability of the coordinative pattern was established by calculating the length of mean vector (r):

$$r = (x_{mean}^2 + y_{mean}^2)^{1/2}$$

In 1999, Hamill *et al.* conducted a study that looked at CRP and CRP variability of two groups of subjects, one that was symptomatic with patellofemoral pain and the other group was uninjured. The researchers found the PFP group to exhibit less CRP variability of the lower extremity joint couplings than the healthy subjects exhibited. The researchers hypothesized that the lower variability of the couplings in the symptomatic is a possible indicator of repeatable joint actions within a very narrow range. The discussion of variability breaks away from traditional view points of gait variability being linked to unsteady or unstable gait. When assessing joint coordination, variability shows flexibility and adaptability of the lower limb, allowing it to attenuate shock better and adapt to external perturbations. This study introduced a new assessment tool for analysing the pathomechanics of running injury and quantifying intra-limb couplings and coordination, continuously.

In 2002, Heiderscheit and colleagues took the previous study to the next level. Rather than using CRP methods, which are more appropriate for joints that allow sinusoidal motion, they used a modified version of the vector coding methods originally proposed by Sparrow *et al.* (1987) to assess joint coordination and variability of two groups of subjects who were uni-laterally symptomatic and asymptomatic of PFPS. As was the case in the study by Hamill *et al.* (1999), Heiderscheit *et al.* (2002) found that there was no significant difference in joint coordination over the entire stance phase. They conducted a more sensitive analysis that divided the stance phase into regions of particular interest. The authors found that the region that included heel contact had the most variability. The symptomatic leg exhibited the least coordination variability.

The emergence of methods to assess intra-limb coupling, coordination and coordination variability has provided more insight into underlying mechanisms behind running injury. While traditional methods have looked at discrete timing and kinematics, continuous methods allow a more all-encompassing approach to analysis of the pathomechanics of running injury (MacLean Dissertation, 2003).

2.3.4 Q-Angle

The Q-angle represents the frontal plane angle of the quadriceps' resultant pull force on the patella and tibial tuberosity (Heiderscheit *et al.,* 2000, Mizuno *et al.,* 2001). It is measured as the angle between the line connecting the anterior superior iliac crest (ASIS) with the center of the patella and the line connecting the center of the patella with the tibial tuberosity, *see Figure 2.11* (Heiderscheit *et al.,* 2000, Mizuno *et al.,* 2001; Powers, 2003). This angle is an important component to the biomechanics of the lower limb and diagnosing overuse injuries causing anterior knee pain (Horton MG, 1989; Heiderscheit MS *et al.,* 2000; Mizuno *et al.,* 2001). The greater the Q-angle, the more lateral pull the quadriceps' exert on the patella, causing the patella to track abnormally over the femoral condyles predisposing one to subluxation and or leading to erosion of the patella and femoral cartilage, ultimately causing chondromalacia (Horton MG, 1989; Heiderscheit MS *et al.,* 2000; Mizuno *et al.,* 2001; Kuhn *et al.,* 2002).

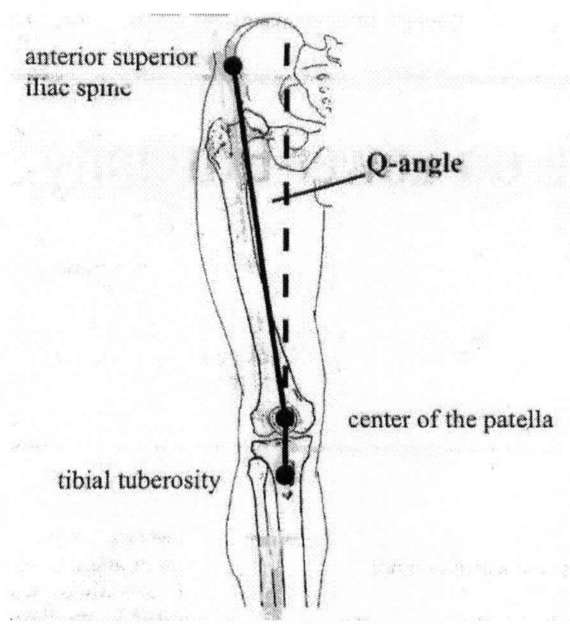


Figure 2.11. Schematic of Q-angle measurement. From: Heiderscheit *et al.* Influence of Q-angle on Lower-Extremity Running Kinematics. *Journal of Orthopaedic & Sports Physical Therapy* 30(5): 271 – 278, 2000.

The study conducted by Mizuno *et al.* (2002) was one involving cadaver knees, simulating a squatting maneuver. As squatting involves different mechanics from those of walking and running, the results cannot directly be applied to running, however, they can provide an indication of the influence the Q-angle has on patellofemoral kinematics. An increase in Q-angle caused the patella to shift laterally and rotate medially, thus following the pull of the quadriceps' muscles. Mizuno *et al.* (2002) found that as the Q-angle increases, the lateral patellofemoral contact pressure increases while decreasing the Q-angle increases the medial patellofemoral contact pressure (Mizuno *et al.,* 2001). However, the changes in contact pressure medially and laterally are not the same in magnitude; there is a smaller medial patellofemoral contact pressure increase for a Q-angle decrease than a lateral patellofemoral contact pressure increase for a Q-angle increase (Mizuno *et al.,* 2001).

By decreasing the Q-angle from 20° to 11°, the propensity for a lateral shift of the patella and a consequential tracking of the patella on the femoral condyles, as well as the propensity for patella subluxation/dislocation is reduced (Mizuno *et al.*, 2002). However, Mizuno *et al.* (2002) found that decreasing the Q-angle beyond 11° did not play as influential role in patellofemoral mechanics. In addition, Heiderscheit *et al.* (2000) found that increased Q-angles resulted in increased time to maximum tibial internal rotation yet no difference in time to maximum angle of rearfoot eversion; however, these results should be viewed with caution as their study had a small effect size resulting in low statistical power.

The Q-angle can be altered via changes in joint kinematics both distally and proximally to the patella (Powers, 2003). The kinematics distal to the patella effect the Q-angle as follows: external rotation of the tibia, a coupled movement with subtalar joint supination, will cause the tibial tuberosity to move laterally, and in turn, increasing the Q-angle (Powers, 2003). The converse is also true whereby internal rotation of the tibia, a coupled movement with subtalar joint pronation, will cause the tibial tuberosity to move medially and in turn decrease the Q-angle. The kinematics proximal to the patella include motions of the femur. If CIRF is indeed what occurs due to abnormal subtalar joint pronation, as probosed by Tiberio (1987), then excessive internal rotation of the femur (with respect to either the ASIS and/or the tibia) results in the patella shifting medially with respect to the ASIS and tibial tuberosity and in turn increasing the Q-angle

(Powers, 2003). The converse is also true, whereby external rotation of the femur allows a more vertical pull on the patella by the extensor mechanism of the knee (Powers, 2003).

2.4 Orthoses

Clinically, the role of orthoses has proven beneficial for runners presenting with overuse injuries as a result of running. Of the 180 runners in the study by James *et al* (1978) presenting with overuse injuries to the lower limb, 46% of injuries were resolved with orthoses intervention, second only to rest (47%) (James *et al.*, 1978). However, rest and or a complete hiatus from running until symptoms resolve is often not a viable option for competition runners, or people who enjoy running as a part of their every day routine. A study by Davlin and Evanski (1991) issued a questionnaire to 347 runners who had been diagnosed with an overuse injury. The respondents had been wearing the orthoses for an average of 23 months post-diagnosis and were averaging 39.6 miles per week during their training regime. More than 75% reported that their overuse syndrome complaints were completely resolved or greatly improved; a further test to their effectiveness was the fact that the orthoses continued to be worn by 90% of the runners after the complaints were completely resolved (Davlin & Evanski, 1991). Hintermann & Nigg (1998) claim that approximately 70% of runners with lower extremity injuries treated with orthotic devices will show improvement.

A study conducted by D'Ambrosia (1985), reviewed the charts of 200 runners prescribed orthoses between 1978 and 1983 at the Louisiana State University Medical Center's Runners clinic. The researchers found 83% of women showed improvement in symptoms associated with posterior tibialis tendonitis, 89% of women with metatarsalgia showed improvement, 76% of women with plantar fasciitis with a calacaneal spur showed improvement, and 66% of women with ITBFS showed improvement with the use of foot orthoses (D'Ambrosia, 1985). Orthoses however are not the panacea for treatment of lower-limb malalignment and thus overuse syndromes; they are however, an important modality to the entire prescription protocol (James *et al.*, 1978).

The hypermobile, pes planus foot is a problematic foot that if left untreated, can create problems along the kinetic chain (Cavanagh, 1980). Excessive eversion has been found to be a reliable indicator for injuries sustained when running (Clarke *et al* 1983; Stacoff, 1991). Features of orthoses such as material composition and posting are effective in controlling excessive eversion during the gait cycle. One goal of a foot orthosis is to place the subtalar joint in the neutral position, allowing for optimum alignment during locomotion, and minimizing stress placed on joints proximal to the subtalar joint, surrounding ligaments and musculature (D'Ambrosia, 1985). By reducing the potential for unhealthy maximum pronation, which tends to occur later in the stance phase than is supposed to, the synchrony between maximum knee flexion and maximum pronation is improved, thus reducing the potential for an antagonistic environment at the knee from occurring (Hintermann & Nigg, 1998). It is interesting to note, that the exact opposite was found by Taunton *et al.* (1985) who conducted a triplanar electrogoniometer investigation on runners exhibiting compensatory overpronation. They found that corrective orthoses decreased the total amount of foot eversion during stance; however, the timing sequence of knee lower-extremity intra-limb coupling was not significantly altered (Taunton *et al.*, 1985).

Historically, the prescription and fabrication of foot orthoses has been done, and still is done today, to align the skeleton (i.e. create biomechanical normalcy), improve cushioning, improve sensory feedback and to improve comfort, not all of which is required of any given orthotic (Nigg *et al.* 1999). Foot orthoses are also prescribed to reduce the frequency of movement related injuries by controlling excessive subtalar and transversal tarsal joint motion during the stance phase of gait through the use of posting and wedges (Eng & Pierrynowski, 1994; Nawoczenski *et al.*, 1995; Nigg *et al.*, 1999).

In order to achieve a neutral subtalar joint when using an orthosis, based on Root theory, the foot must be casted in the neutral position so that when weight bearing, the vertical axis of the heel is parallel to the longitudinal axis of the distal tibia and the plane of the metatarsal heads is perpendicular to the heel (D'Ambrosia, 1985; Novick and Kelley, 1990). It is important however to slightly under-correct pronation in order to provide a comfortable environment for the user (James *et al.* 1978). Slight under correction showed better compliance than overcorrection (James *et al.*, 1978).

In order to reduce unhealthy compensatory pronation of the STJ during stance, the foot orthoses must be designed such that it increases the supination moment (i.e. torque) across the STJ axis (Kirby, 1992; Novick & Kelley, 1990). Recalling Newton's 3rd law, ground reaction forces (GRF) are directed vertically upon the plantar axis of the foot if healthy lower-limb alignment is present during the stance phase (Kirby, 1992; Novick & Kelley, 1990). A supination moment is caused by any GRF acting medial to the STJ while ground reaction forces acting lateral to the STJ produces a pronation moment. By posting an orthosis in the medial aspect of the heel and longitudinal arch, the orthoses reaction force shifts the weight supported by the foot from the lateral aspect of the foot, to a position medial to the STJ axis, thus creating a net supination moment across the

STJ, further decreasing the total pronation moment across the STJ axis (Kirby, 1992; Novick & Kelley, 1990).

Mundermann *et al.* (2003) conducted a study that examined the effects posting, molding, and posting and molding in combination, had on gait correction. The post functions during the first half of the gait phase, from heel contact to midstance (Mundermann *et al*, 2003). The post places the foot in a more inverted position, bringing the ground up to the foot at heel contact and thus, reducing the maximum amount of rearfoot eversion (Mundermann *et al*, 2003). The post does not however function to reduce max foot inversion, which occurs during the latter half of the gait cycle (Mundermann *et al*, 2003). Molding is a technique used in orthoses manufacturing that contributes to the kinematics of the foot during the latter half of the gait cycle (Mundermann *et al*, 2003). Molding is effective in shock absorption as it creates a larger contact area between the foot and orthosis, as a molded orthosis is in contact with the foot over the entire forefoot and midfoot regions (Mundermann *et al*, 2003). Molding also plays another crucial role during the stance phase as it was found to reduce maximum foot inversion and maximum foot inversion velocity (Mundermann *et al.,* 2003). However, molding did have adverse effects on the gait cycle. Its ability to place the foot in a relatively more inverted position during the latter half of the stance phase, creates a larger external rotation moment at the knee joint, creating implications for PFPS (Mundermann *et al.*, 2003). In their study, a combination of posting and molding produced the same effects as did orthoses with only a molded construction, however, the timing of maximum inversion moment at the ankle joint, vertical impact peak and maximum abduction moment at the knee was delayed with the combination post/mold orthosis (Mundermann *et al.*,2003). Mundermann *et al.* (2003) found that maximum rearfoot control through orthoses, is achieved through molding of an orthosis, as opposed to posting, or molding in combination with posting.

Stacoff *et al.* (2000) conducted a study using bone pins to examine the effects of two different orthosis conditions on the rearfoot motion during the stance phase of running. Intracortical Hoffman bone pins were inserted into the calcaneus and tibia of five healthy males, under local anesthetic. The subjects ran in a test shoe that had a dual density midsole, with the medial aspect of the midsole harder than the lateral aspect. The subjects ran with the stock insole and in two different orthoses: an anterior orthosis that supported the foot arch and a posterior orthosis that supported the calcaneus at the sustentaculum tali (Stacoff *et al.*, 2000). The researchers found no significant difference between the SHOD condition and the orthoses conditions with respect to skeletal calcaneal and tibial kinematics. Their methodology however may be flawed in that their test shoe was itself a motion control shoe. The dual density midsole has the ability to change the tibial and calcaneal kinematics on its own, without orthoses. Therefore, it is not surprising that the researchers did not find any differences between SHOD running and running with orthoses in tibial and calcaneal kinematics. This study could have been improved if the researchers tested the runners in a neutral shoe, this would allow them to draw conclusions with respect to the contribution orthoses have to tibial and calcaneal kinematics.

2.5 Shoes

In addition to rearfoot control provided by foot orthoses, shoes have evolved over the years such that they are able to provide a moderate amount of rearfoot control, during the gait cycle (*See Fig. 2.12*) (Taunton and Moore, 1991*)*. A good motion control shoe on the market right now is able to control five degrees of heel varus (discussions with Dr. J. Taunton), and any amount of varus greater than that, may require additional support offered by orthoses.

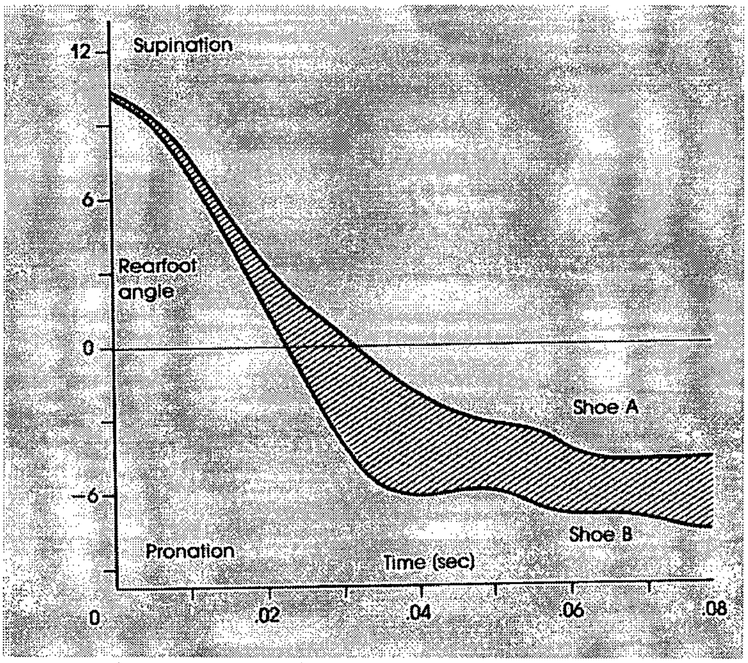


Figure 2.12. Graphical representation of the effect a motion control shoe (A) has on the rearfoot angle during the stance phase. From: Cavanagh, P.R. The Running Shoe Book. Mountain View, CA: Anderson World, Inc., 1980.

As the previous sections have revealed, the biomechanics of walking and running gait are complex in nature, and unfortunately, is understood by very few who are in a position to suggest one shoe over another (McKenzie *et al.*, 1985). In a runner with healthy lower-limb alignment, various structural components of the shoe have the ability to alter the STJ mechanics significantly, increasing the propensity for injury (Hamill *et al.*, 1992). When the person prescribing/suggesting a shoe to a runner doesn't understand how different components of the kinetic chain are affected by the structural components of the shoe, prescription of the wrong shoe often occurs (McKenzie *et al.*, 1985).

The ultimate goal of shoe design is to produce a shoe that considers the physiological ranges of motion of the foot, minimizes the moment arm of the GRF through proper sole geometry and thus avoids excessive rotational movement of the lower limb (Nigg & Segesser, 1992). By constructing a shoe that minimizes the magnitude of inevitable impact forces imposed on the kinetic chain during landing, supports the foot during the stance phase and guides it through propulsion, frequency of running related injuries may be reduced (Nigg & Segesser, 1992). Construction features of shoes implicated in running injuries include inadequate heel wedging, soft and/or loose fitting heel counters, inflexible midsoles (especially under the metatarsal heads), and excessive heel wear (Clement *et al.*, 1981; McKenzie *et al.*, 1986; Taunton *et al.*, 1988).

The term rearfoot control refers to the ability of a shoe to limit the amount and/or rate of foot pronation during the contact phase of gait (Clarke *et al.*, 1983). There are several architectural features of the shoe, which control the rearfoot motion of the foot. Medial support in a shoe provides increased stability for the foot and leg and acts to reduce maximal foot pronation (Hintermann & Nigg, 1998). Medial posting is usually in the form of a dual density sole material, with higher density foam placed medially to control the foot and a softer midsole material placed laterally to aid in shock dissipation (McKenzie *et al.,* 1985; Moore and Taunton, 1991). It is important however that the durometer differential in the midsole not be too pronounced as the softer material tends to break down much quicker than the harder, medial midsole, thus putting the user at an increased risk of ITBS or Achilles tendonopathy (Taunton, 1991). In the retrospective case analysis conducted by Macintyre *et al.* (1991), many of the patients with ITBFS presented upon examination with running shoes that had deteriorated on the lateral aspect due to compression of the lower density foam. A study by Clarke *et al.* (1983) found the midsole hardness of shoes to have a significant effect upon maximum pronation, total angular displacement of the calcaneus and time to max velocity of pronation, *see Figure 2.13.* Absolute values of maximum pronation and total rearfoot angular displacement were larger for shoes with a softer midsole as opposed to hard or medium midsoles (Clarke *et al.*, 1983). As well, the time to maximum velocity of pronation increased as the midsoles became softer (Clarke *et al.*, 1983).

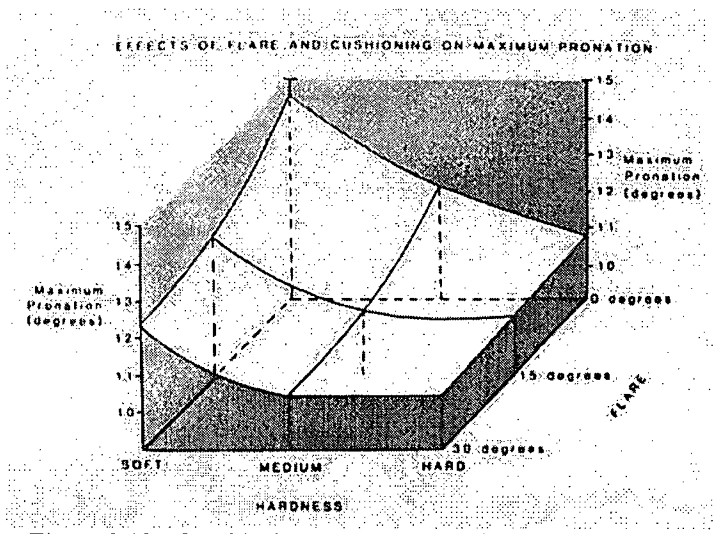


Figure 2.13. Graphical representation of the relationships found among midsole hardness, heel flare, and maximum pronation exhibited by a group of subjects running at 3.8 m/s.
From: Clarke TE, Frederick EC, & Hamill CL. The effects of shoe design parameters on rearfoot control in running. *Medicine and Science in Sports and Exercise* 15(5): 376 – 381, 1983.

Other design components of the shoe that aid in motion control are a straight lasted shoe (McKenzie *et al.*, 1985; Moore & Taunton, 1991) with torsional rigidity (Moore &

Taunton, 1991). The ability of the sole to resist longitudinal torque is traditionally achieved by laying a firm board down between the upper and the midsole, which acts to resist lateral forces produced as the foot moves through the three phases of stance (Taunton & Moore, 1991).

Arguably the most important component of the stability system is the heel counter (*see Figure 2.14*). Often thermoplastic in design, the heel counter acts to stabilize the calcaneus and STJ, helping to control abnormal pronation (Moore & Taunton, 1991).

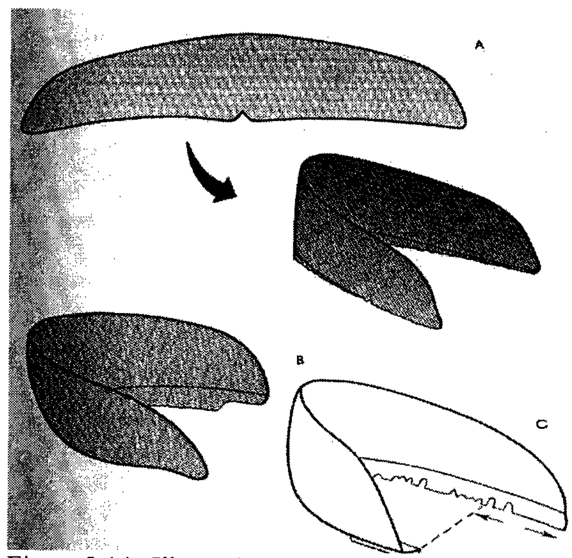


Figure 2.14. Illustration of thermoplastic heel counters. From: Cavanagh, P.R. The Running Shoe Book.
Mountain View, CA: Anderson World, Inc., 1980.

Past research involving shoe design and runners has focussed primarily on the effects of shoe sole and insert construction on the biomechanics of healthy runners (Bauer *et al,* 2002; Gross *et al* 1991; Nigg *et al,* 1998; Stacoff *et al* 2001). The study by Clarke *et al* used 36 different midsole hardness, heel height and heel flare combinations in order to gain a better understanding of the role each component played in controlling rearfoot motion (Clarke *et al*, 1983). Clarke *et al* found that the harder the midsole and the greater the flare, the smaller the absolute value of maximum pronation and total rearfoot movement (Clarke *et al*, 1983). A softer midsole with no heel flare allowed for the greatest amount of rearfoot movement (Clarke *et al*, 1983). There was a lack of significant correlation between maximum velocity of pronation and maximum pronation over the 360 subject/shoe conditions, indicating that there may be two phases of rearfoot control (Clarke *et al.*, 1983). It is the author's belief that the ideal shoe for controlling rearfoot motion would minimize the average rearfoot angular acceleration, maximum velocity of pronation and maximum amount of pronation (Clarke *et al*, 1983).

While a medial post to the foot is an important treatment modality for anterior knee pain and is now an important component of a multifaceted treatment protocol for PFPS diagnosed by the physicians at the AMSMC (Taunton *et al.,* 1998), it has the potential to expose the lower limb musculoskeletal structure to potentially harmful impact loading (Perry & Lafortune, 1995). Excessive stress is placed on the lower limb by increasing the amount of axial rotation of the tibia due to the increased inclination of the STJ axis as a result of wearing M/C shoes (Perry & Lafortune, 1995; Hintermann & Nigg, 1998). And thus, a greater understanding into the effect a motion control shoe has on lower limb biomechanics, and the type of foot it should be prescribed to, will help to prevent overuse running injuries.

Chapter 3

3.0 Methodology

3.1 Sample Description

In order to correctly reject the null hypothesis with a power of 80%, $p < 0.05$ our N was 12, however we used 15 females for analysis. The calculation we used was: $\delta = \gamma \times N^{1/2}$ At an $\alpha = 0.05$, $\delta = 3.40$; γ = mean difference/standard deviation. Values used to conduct the power analysis were based on those collected in the "*Timing of lower extremity joint actions*" study, conducted by Hamill *et al.* (1992). The sample used for this study were female, recreational runners age 23 to 52 years (Avg = 35.78, St.Dev. = 9.52) who had on average 6.75° ± 3.15° of non-weight bearing heel varum and wore semi-rigid foot orthoses (with 3° of rearfoot varus posting and 0° of forefoot posting) manufactured by Paris Orthotics within the last 18 months. The runners were clear rearfoot strikers. The subjects had not currently been seeking medical attention and had not reduced their training pace or mileage due to symptoms or injury. Subjects with a leg length discrepancy >1.0 cm were excluded from this study as the longer limb would produce compensatory pronation while the shorter limb would produce compensatory supination. The subjects were recreational runners all of whom planned on competing in a recreational running competition within 2004/2005 (i.e. the Vancouver Sun Run).

3.2 Sampling Procedure

Our subjects were recruited from the database at Paris Orthotics. An initial Letter of Introduction was mailed to those who fit the criteria for foot orthoses. The criteria pertaining to symptoms, injuries and training pace was further explained in the letter. If the subject fit our criteria, and was interested in participating in the study, they were asked to contact me for further information.

3.3 Data Collection

This study was approved by the UBC Ethics board prior to commencement of data collection. All testing took place at the Allan McGavin Sports Medicine Centre at the University of British Columbia (Vancouver, B.C.). Before testing commenced, the subjects were asked to sign a Letter of Informed Consent. The Letter of Informed Consent described the methods of data collection, potential risks to the subject, compensation for participation, and included contact information for the Principle Investigator and the UBC Office of Research Services.

3.3.1 Anthropometric Data

Following the signing of the Letter of Informed Consent, a brief history was taken and an anthropometric physical exam was conducted on each subject (See Appendix A). The measurement protocol was followed in its entirety, two times, if the angular measurements for each variable were different by $> 0.5°$, a third measurement was taken. For measures of leg length, differences > 0.25 cm were repeated. Navicular drop was measured with calipers by first measuring the height of the navicular while standing in a position of subtalar joint neutral. The subject then stood in the resting calcaneal stance position and navicular height was measured again. The difference between the two measures was deemed navicular drop.

Following the collection of the running history and anthropometric data, the subjects were tested in both the Nike Air Kantara II (a motion control shoe) and Nike Air Pegasus (a neutral shoe) while wearing their foot orthoses (see Appendix B for shoe design information) and without their foot orthoses. The order of test conditions was randomized across all subjects. Prior to running in each test shoe in the orthoses condition, subjects ran for two minutes in a control shoe (Nike Air Free), at their self-selected speed in order to answer the questionnaire following each test shoe. Following running in the control shoe, the first of the two shoes was run in while wearing foot orthoses. Electrogoniometric and pressure analysis was conducted simultaneously with one another.

3.3.2 Pressure Analysis

Insole pressure analysis was used to establish the differences in plantar surface loading patterns with the two different shoe conditions, with and without orthoses. The Pedar Online System (Novel Gmbh – Munich, Germany) was the device used to collect foot pressure data while running at a pace of close to 3.35 m/s (an eight minute mile pace). A MatLab program written by Stefan Grau of the University of Tubingen was used to quantify the plantar pressure loading patterns. The Pedar system for in-shoe pressure measurement uses insoles that have 99 pressure sensors (Novel Gmbh, Novel, Germany). The flexible insole is able to measure both static and dynamic pressure distribution in real time, over the entire plantar surface of the foot. Prior to use, the insoles were calibrated using a calibration chamber that applies a known pressure across the whole insole. Calibration values were used for subsequent pressure analysis. The insole was placed in between the right foot and the orthosis and was connected via a sensor lead to a data acquisition box (180g, 120 x 104 x 44 mm) that was strapped to the handle of the treadmill. The data acquisition box was connected to a computer running

Windows 2000, via a 9.25 m coaxial cable. A BNC cable connects the Pedar system to a trigger connection on an A/D data acquisition board (National Instruments, BNC-2110), that was used to synchronize the two electrogoniometers with the Pedar system.

After the subject ran in the shoe for five minutes on the treadmill at their self-selected speed, the treadmill speed was adjusted to3.35 m/s (8 minute-mile pace) if possible, or to the fastest possible speed close to 3.35 m/s. The subject ran at this pace for two minutes. At the beginning of the third minute, Pressure and Force vs. Time data for both feet were collected at a sampling frequency of (99 Hz) for 10 seconds. The data was then trimmed to ten complete gait cycles for analysis of results and statistics. Following data collection for the first shoe condition, the subject brought the treadmill to a stop. The remaining three test conditions were tested similarly. The insoles were re-calibrated and the subject commenced their five minute familiarization period at the self-selected speed. Following familiarization with the new shoe condition, the treadmill was adjusted to the same pace as the previous condition and the subject ran at this pace for two minutes. At the commencement of the third minute, pressure data and electrogoniometric data was collected as per the previous shoe condition.

Previous research conducted by Taunton, Baker and Pritoula (2003) at the Allan McGavin Sports Medicine Centre identified pressure analysis as being effective and useful in identifying a need for an orthoses/heel lift and or strengthening/stretching regime as a treatment intervention (*Unpublished research*). It was thought then that this tool would also be useful in identifying a healthy shoe-orthotic combination.

3.3.3 Electrogoniometer data acquisition

Electrogoniometers (Biovision – Wehrheim, Germany) were to measure knee flexion and extension in the sagittal plane as well as rearfoot angle in the frontal plane (Figure 3.1). The electrogoniometers sampled at 990 Hz for 10 seconds. The knee electrogoniometer (potentiometer: 43 mm x 15 mm x 20 mm; two arms: 110 mm x 16 mm x 2 mm) and the rearfoot electrogoniometer (potentiometer: 43 mm x 15 mm x 20 mm; one arm: 180 mm x 16 mm x 2 mm) were connected to the A/D data acquisition board (National Instruments, BNC-2110) which connected to a second laptop computer, running Windows XP, via a data acquisition card (DAQCard-6024E). The software used to convert the analog data to digital data is NI – DAQ 7. A program was written in LabView (*Just Collect It)* to collect the electrogoniometers' data (Nike Inc, Beaverton OR).

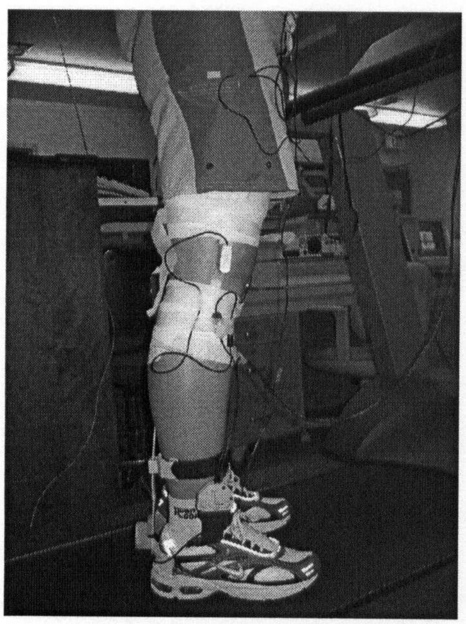
a)  b) 

Figure 3.1. Example of experimental set-up with the electrogoniometer attached to the a) lateral aspect Of the right knee and b) heel counter of the right shoe, with arm of the electrogoniometer extending up the calf inserting into the brace.

The rearfoot electrogoniometer was attached to the heel counter of the right shoe via a brace custom designed and molded out of thermoplastic material that attached to the shoe using Velcro. The axis of the electrogoniometer corresponded to the axis of subtalar joint motion. The arm of the electrogoniometer attached below the belly of the gastrocnemius muscle on the right leg via a Velcro strap. A custom made thermoplastic brace was attached to the Velcro strap to guide the arm of the electrogoniometer without impeding proximal/distal motion as a result of the shoe cushioning heel contact. The brace was attached firmly to the heel counter such that the axis of rotation of the electrogoniometer remained constant. The brace for the arm of the electrogoniometer was designed to minimize noise due to the arm moving back and forth in the brace, irrespective of rearfoot motion.

The axis of the knee goniometer corresponded with the axis of the knee joint. The areas of the upper and lower leg where the goniometers were secured were first sprayed with athletic adhesive spray. Following, the arms were secured with elastic athletic tape, insuring that the axis of the goniometer remained on the knee joint. The wires connecting the electrogoniometers to the power supply were taped along the leg and then secured at the waist along with the power supply. Prior to running on the treadmill, subjects performed exaggerated strides to ensure the wires were not restricted or did not impede their normal running gait.

3.3.4 Questionnaire

Once the running trial finished for the orthoses conditions, the subjects were given a questionnaire to fill out regarding the fit and of the shoes using a 15 cm VAS for each question (see Appendix D).

3.4 Data Reduction

The following methods of data reduction were used to reduce the raw data collected from the electrogoniometers. A program written in Labview by the biomechanists at the Nike Sport Research Lab in Beaverton, OR was used to collect the data from analog-to-digital board. Raw data was filtered using a low-pass fourth-order Butterworth filter with a 15 Hz cutoff frequency. Prior to each subject being tested, the electrogoniometers were calibrated, and values obtained during calibration were used to convert volts to degrees for analysis. Prior to data being collected while running, a static trial was taken in order to establish the relative angular values. We analysed ten stance phases from the position of maximum knee extension at heel contact to maximum knee extension at toe off. This was the best method for determining stance phase using the equipment we had access to. The electrogoniometers were collecting data with a frequency of 990 Hz vs the pressure data that was being collected at 99 Hz. The difference in collection frequency was too large to use it as a tool for identifying when heel contact occurred with the electrogoniometers. Data from ten complete gait cycles was used for analysis. The angle data for each of the ten stance phases were normalized to 101 data points, with each point representing 1% of stance for analysis of coordination variability. For analysis of the secondary outcome measures, all data for each stance phase were preserved.

A program written in MatLab by the biomechanists and tested for reliability was used for the analysis of joint coordination variability. A vector coding system was used to compare the joint coupling between the knee and rear foot. The two oscillators are not sinusoidal in nature, so vector coding was used for the analysis of coupling coordination and variability throughout the stance phase (Heidercheit, 2002). This method quantifies coordination and variability patterns using circular statistics (Heidercheit, 2002). In order to quantify coordination variability, angle-angle plots were created for the coupling of knee angle (in the sagittal plane) and the rear foot (in the frontal plane). Knee angle data were plotted on the X-axis and rear foot data on the Y-axis, as per the conventions required for this analysis.

The coupling angle (γ) was then calculated as the angle between a vector connecting two adjacent data points, and the right horizontal (Heidercheit, 2002). In order to quantify joint coordination, a modification of the technique proposed by Sparrow *et al.* (1978), by Heiderscheit *et al.* (2002) was used. Coupling angles were calculated in relation to the right horizontal. Circular statistics were thus required in order to convert angles ranging from 0° - 180° to a range between 0° and 90°. The variability of the coordinative pattern was established by calculating the length of mean vector (r):

$$r = (\mathbf{x}_{\mathbf{mean}}{}^{2} + \mathbf{y}_{\mathbf{mean}}{}^{2})^{1/2}$$

and then calculating the angular variance using:

$$\mathbf{s^2 = 2(1 - r)}$$

Coordination variability was calculated over the entire stance phase, and specific regions of interest as suggested by Heiderscheit *et al.* (2002) due to the changing functional demands placed on the musculo-skeletal system throughout the stance phase.

The rearfoot dynamic data was tested for reliability. Three subjects were tested two times, and their data was compared between test sessions. The data was not statistically different between test conditions. This data was used for further analysis of coordination variability. Coordination variability was tested for reliability using the data from three subjects, tested over two different occasions. Repeated measures t-tests were used to test the data from each trial for significant differences. The two sets of data were deemed statistically different, the patterns were similar, but of different magnitude. Figure 3.2 shows that although there were significant differences between the two data sets, the pattern of coordination variability is similar; indicating the coordination between the knee and the rear foot is consistent, however at a different magnitude for each trial, depending on the demands that was placed on the body at the time of testing. It is thought that over repeated trials, running in similar conditions, the coupling would be the same between the rearfoot in the frontal plane and knee motion in the sagittal plane, however, it does vary from session to session in order to maintain flexibility and adaptability.

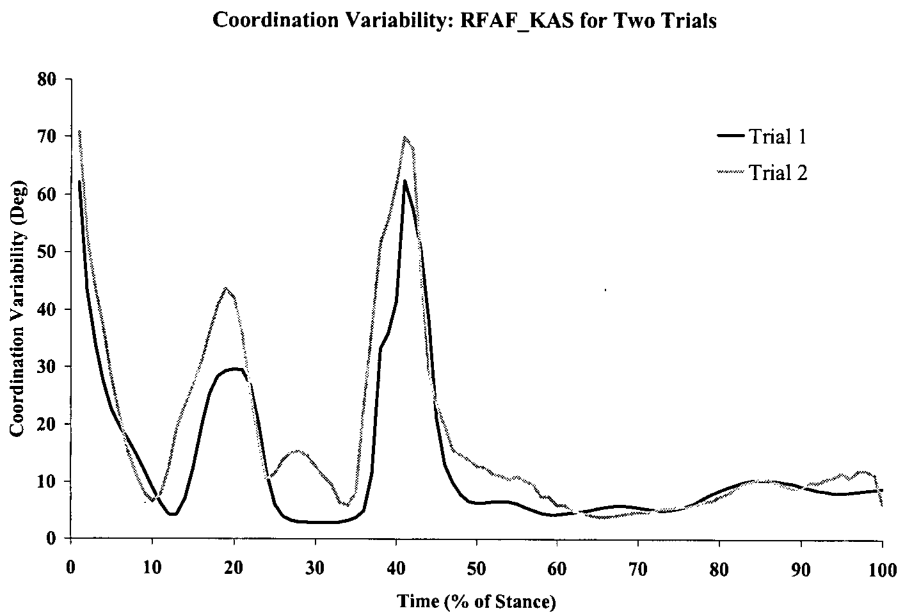


Figure 3.2. Example of coordination variability plot for the rearfoot in the frontal plane and knee angle in the sagittal plane for three subjects averaged over two trials of the same condition

The pressure analysis data was reduced as follows. Ten stance phases for each running condition were used for analysis. The data were imported into a MatLab data analysis program written by the University of Tübingen in Germany. This analysis program was tested for reliability. The outcome measure that was found to be reliable for shoe research was that of the plantar pressure loading patterns.

The foot was divided into seven regions of interest. The regions are referred to as 'masks'. The heel was divided into two masks, each mask comprising 33.33% of normalized foot length and 50% of normalized heel width. The midfoot was similarly divided into two masks, and the forefoot into three masks. Each mask in the forefoot comprised 33.33% of normalized foot length, and 33.33% of normalized foot width. The global Force-time (F*t) impulse was calculated for the entire stance phase, over the entire foot, and averaged among the subjects for each trial, over 10 stance phases. A F*t impulse was also calculated for each mask, over the entire stance phase, and averaged over the 10 stance phases for each subject for each condition.

3.5 Statistical Analysis

Circular statistics were used to determine the coupling angles and further, the coordination variability between the two oscillators. Circular statistics are required because the data that is being analysed is directional and thus conventional statistics are not suitable (Heiderscheit, 2000). Figure 3.3 is a clear illustration of why circular statistics are used.

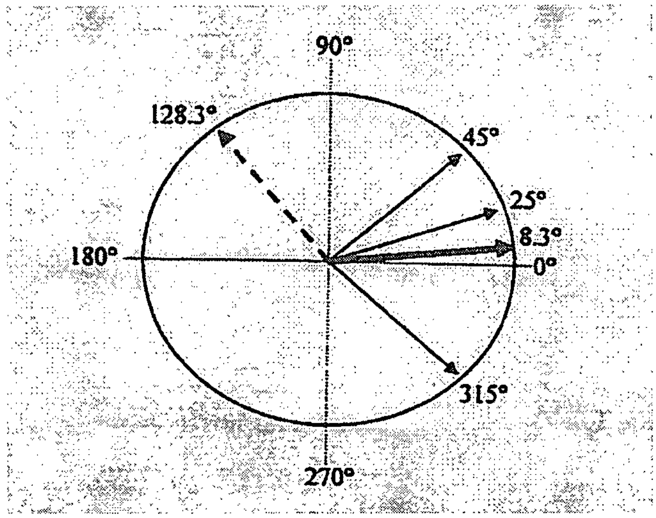


Figure 3.3. Example of data that requires the use of circular statistics rather than conventional statistics due to its directional nature

If the following angles were averaged: 25° and 45° and 315°, the resultant angle would be 128.3°, however, the correct mean angle is 8.3°. Batschelet (1981) proposed the following arithmetic to calculating the mean direction of the vector. The mean direction is first determined by calculating the mean cosine (mean x) and mean sine (mean y) of each direction (Φ).

$$\bar{x} = \frac{1}{n}\sum_{i=1}^{n}\cos\phi_i \quad \text{and} \quad \bar{y} = \frac{1}{n}\sum_{i=1}^{n}\sin\phi_i$$

As long as mean-x and mean-y are not zero, the mean direction (Φ) is calculated by:

$$\bar{\phi} = \tan^{-1}\left(\frac{\bar{y}}{\bar{x}}\right), \text{if } \bar{y}>0 \quad \text{and} \quad \bar{\phi} = 180 + \tan^{-1}\left(\frac{\bar{y}}{\bar{x}}\right), \text{if } \bar{y}<0$$

Using equations introduced earlier, angular variance is calculated as a measure of joint coordination variability (Heiderscheit, 2000).

Following, paired samples t-tests were used to assess the differences between the motion control shoe and the neutral shoe for both the orthoses condition and the shod condition. Differences with $p< 0.01$ were deemed to be statistically significant.

Paired samples t-tests were also used assess the differences between the two shoes, for both the orthoses and shod conditions for the plantar pressure loading pattern data and the questionnaire data. Differences with $p< 0.01$ were deemed to be statistically significant.

Finally the Pearson correlation was calculated to determine the degree and direction of linear relationship between navicular drop and questionnaire results. The difference in responses between each shoe-orthoses combination, for each question was correlated with navicular drop. A perfect linear relationship would indicate that for every increase in questionnaire score there would be a corresponding increase in navicular height. A non-linear relationship between the two would indicate that there is no covariability such that a change in questionnaire results would produce no predictable change in navicular height. Correlations with $p < 0.01$ were deemed to be statistically significant.

Effect size (ES) was calculated where t-tests were used. ES was calculated by dividing the mean difference between each shoe-orthoses/shod combination and the pooled standard deviation for the pair. ES can be evaluated using the standards set out by Cohen in 1988. Cohen suggested that ES values of 0.2 represent small differences, 0.5 represent moderate differences and values of 0.8 and greater represent large differences. The p-values were used as a basis for establishing the difference between the motion control shoe and neutral shoe for the orthoses and shod running trials while the ES was a measure of the strength of the difference between the two conditions.

Chapter 4.0 Results

4.1 Primary Outcome Measures

4.1.1 Vector Coding

The data for the vector coding analysis is presented for comparisons between the motion control shoe (K) and the neutral shoe (P) for the orthoses conditions and the shod conditions. Figures 4.1 and 4.3 are traditional angle-angle plots for rear foot angle in the frontal plane and knee angle in the sagittal plane for the orthoses and shod conditions respectively. In the past angle-angle plots were used to analyze joint couplings. However, it is difficult to analyze the graphs in a quantitative fashion, and thus, vector coding has been adopted by the biomechanics community as a more appropriate way to analyze joint couplings. Figures 4.2 and 4.4 are plots of the mean coordination variability across the entire stance phase for knee rotation/rearfoot rotation couplings during running on a treadmill. All angle data were normalized to 100% of stance.

Tables 4.1 and 4.2 present the group means (±SD) of variability, measured in degrees. Because the demands placed on the lower-limb musculo-skeletal system change throughout the stance phase, it has been suggested by Heiderscheit et al. (2002) that the stance phase is broken up into regions of interest for a more sensitive analysis of between-conditions differences. Therefore, the variability was averaged not only across the entire stance phase, but also across: 1) 0 – 40% of stance 2) 10 – 30% of stance and 3) 31 – 60% of stance.

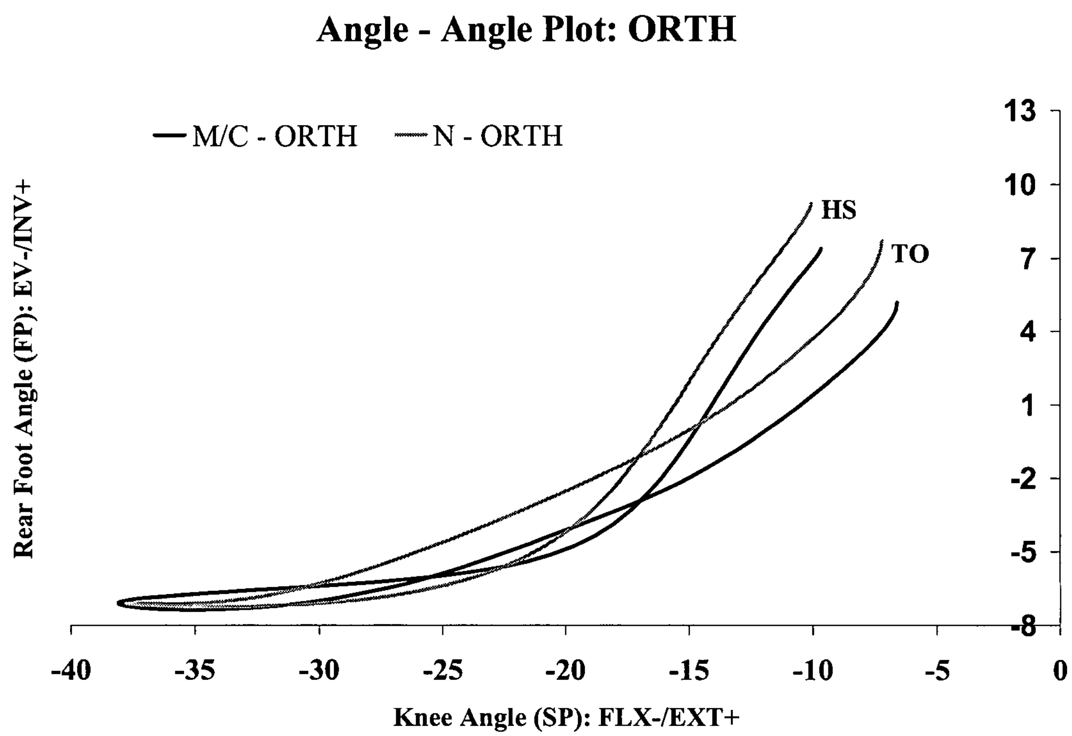


Figure 4.1. Angle-angle plot of the rear foot angle in the frontal plane vs the knee angle in the saggital plane from heel strike (HS) to toe off (TO) while running in the motion control shoe and the neutral shoe with orthoses (ORTH).

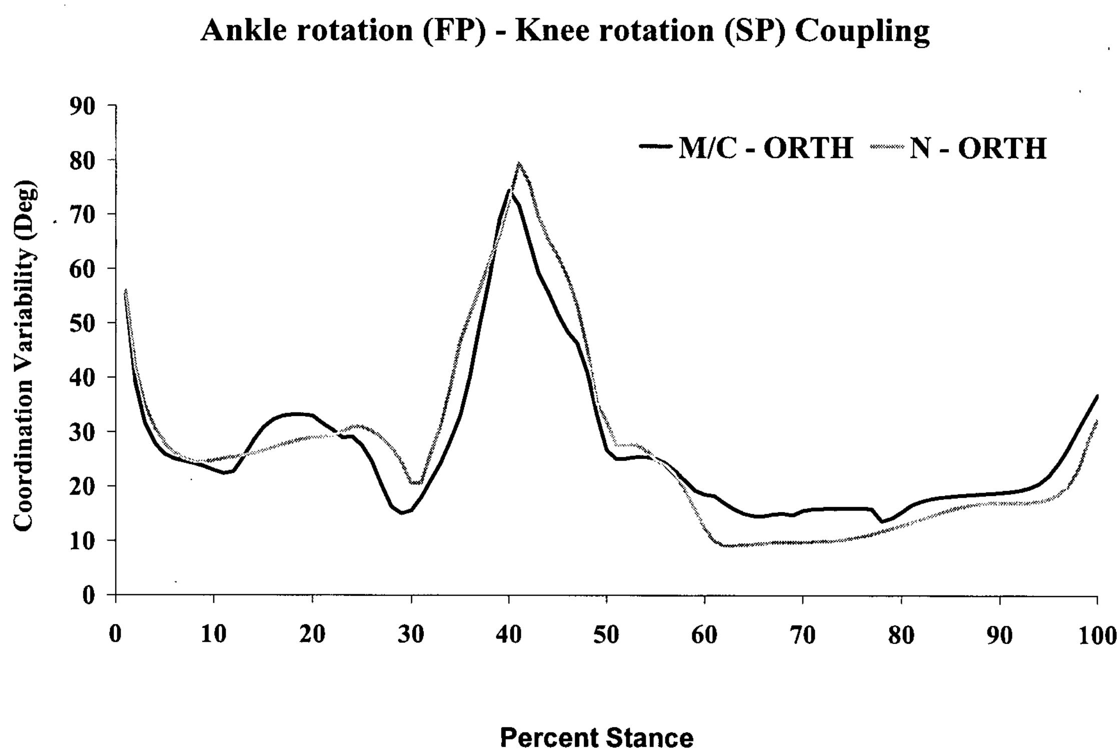


Figure 4.2. Mean coordination variability across the stance phase of a treadmill run in the motion control shoe (K) and neutral shoe (P) while wearing orthoses (ORTH).

Table 4.1 Average variability (°) of intralimb joint coordination across the entire stance phase as well as regions of interest within the stance phase of running on a treadmill in a motion control shoe (K) and neutral shoe (P) while wearing orthoses (ORTH). (*) denotes a significant difference.

Coupling	Stance	0 – 40%	10 - 30%	31 - 60%
M/C - orth	27.27 (13.72)	31.27 (13.28)	26.49 (6.03)	38.25 (17.86)
N - orth	27.15 (16.78)	33.48 (12.86)	27.43 (2.57)	42.54 (20.16)
Difference	0.12	-2.21	-0.93	-4.29
ES	0.023	0.45	0.19	0.83
P-value	0.818	0.007*	0.404	0.00*

It was hypothesized that the motion control and neutral shoes would demonstrate statistically different coordination variability, regardless of whether or not orthoses were worn, however, due to a significant lack of research in this area, it was unclear as to which shoe and orthoses combination would produce the most variability. Overall, the orthoses conditions demonstrated more variability than the SHOD counterparts, however within the orthoses condition, the motion control shoe demonstrated the most variability. The two shoes were not statistically different over the entire stance phase. From 0 – 40% of stance, the region encompassing loading and the beginning of mid-stance, the two shoes were statistically different ($p = 0.007$; ES = 0.45) with the neutral shoe performing with 2.21° more variability than the motion control shoe. Between 10 and 30% of stance there was no significant difference in variability between the two shoes ($p = 0.404$; ES = 0.19). Between 31 and 60% of stance phase, which encompasses almost the entire mid-stance period of the gait cycle the neutral shoe exhibits 4.29° more variability than the motion control shoe ($p=0.00$; ES = 0.83).

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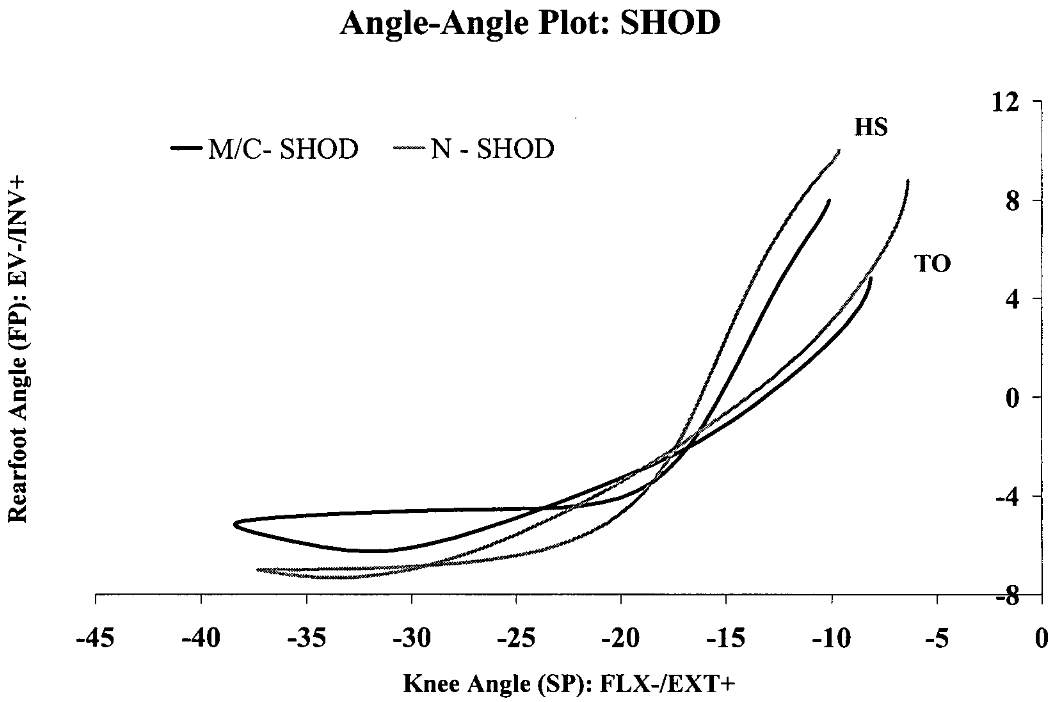

Figure 4.3. Angle-angle plot of the rear foot angle in the frontal plane vs the knee angle in the saggital plane from heel strike (HS) to toe off (TO) while running in the motion control shoe and the neutral shoe without orthoses (SHOD).

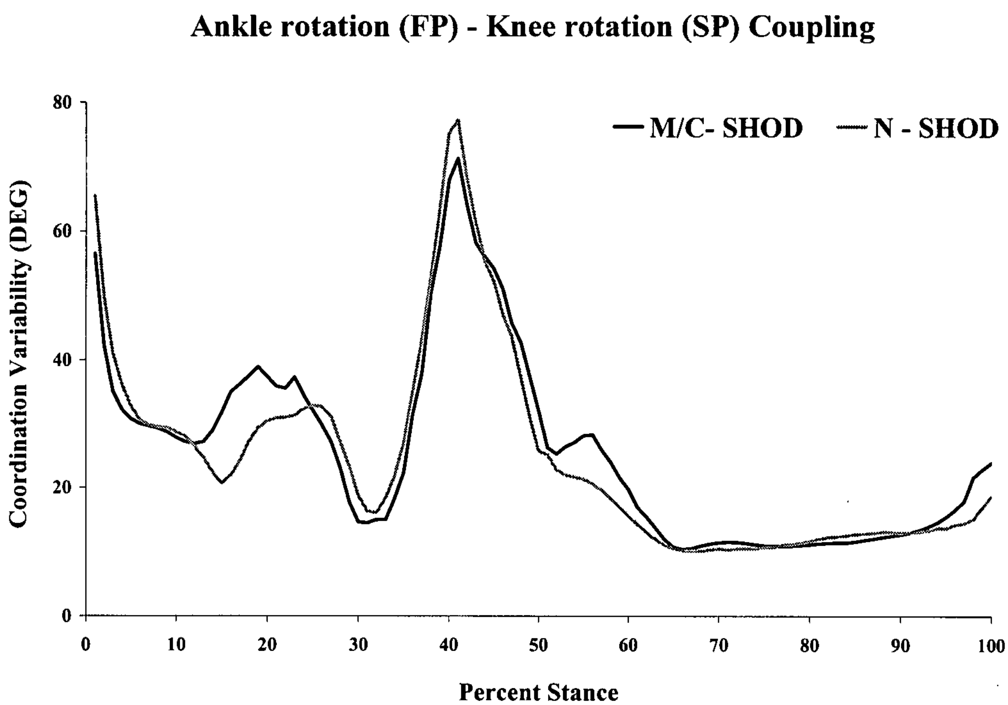

Figure 4.4. Mean coordination variability across the stance phase of a treadmill run in the motion control shoe and neutral shoe without orthoses (SHOD).

The shod condition produced very different results. Over the entire stance phase, the two shoes produced statistically different results with the motion control shoe exhibiting 0.92° increased variability ($p = 0.033$; ES = 0.22). Their was no difference in coordination variability between the two shoes from 0 – 40% of stance ($p = 0.953$; ES = 0.009), however, in the region between 10 and 30% of stance, there was a significant difference with the motion control shoe exhibiting 3.17° more variability ($p = 0.023$; ES = 0.54). Finally, during the midstance region, 31 – 60% of stance, there was no significant difference between the two shoes ($p = 0.376$; ES = 0.16).

Table 4.2 Average variability (°) of intralimb joint coordination across the entire stance phase as well as regions of interest within the stance phase of running on a treadmill in a motion control shoe and neutral shoe. (*) denotes a significant difference.

Coupling	Stance	0 - 40%	10 - 30%	31 - 60%
M/C – shod	25.77 (14.72)	32.16 (11.46)	30.58 (6.57)	36.50 (17.03)
N – shod	24.85 (15.60)	32.21 (12.99)	27.41 (4.23)	35.73 (19.31)
Difference	0.92	-0.05	3.17	0.7663
ES	0.22	0.009	0.54	0.16
P-value	0.033*	0.953	0.023*	0.376

4.1.2 Pressure Analysis

Figures 4.5 and 4.6 are graphs showing the different loading patterns for the two shoes, with and without orthoses. Loading was measured as a percentage of the total F*t integral over the entire stance phase, over the entire insole. 'Mask number' refers to the areas of interest on the insole as illustrated by the diagram in the top left hand corner of the graph. 'M/C_ORTH' refers to the condition where subjects ran in the motion control shoe with orthoses, 'N_Orth' refers to the condition where subjects ran in the neutral shoe with orthoses.

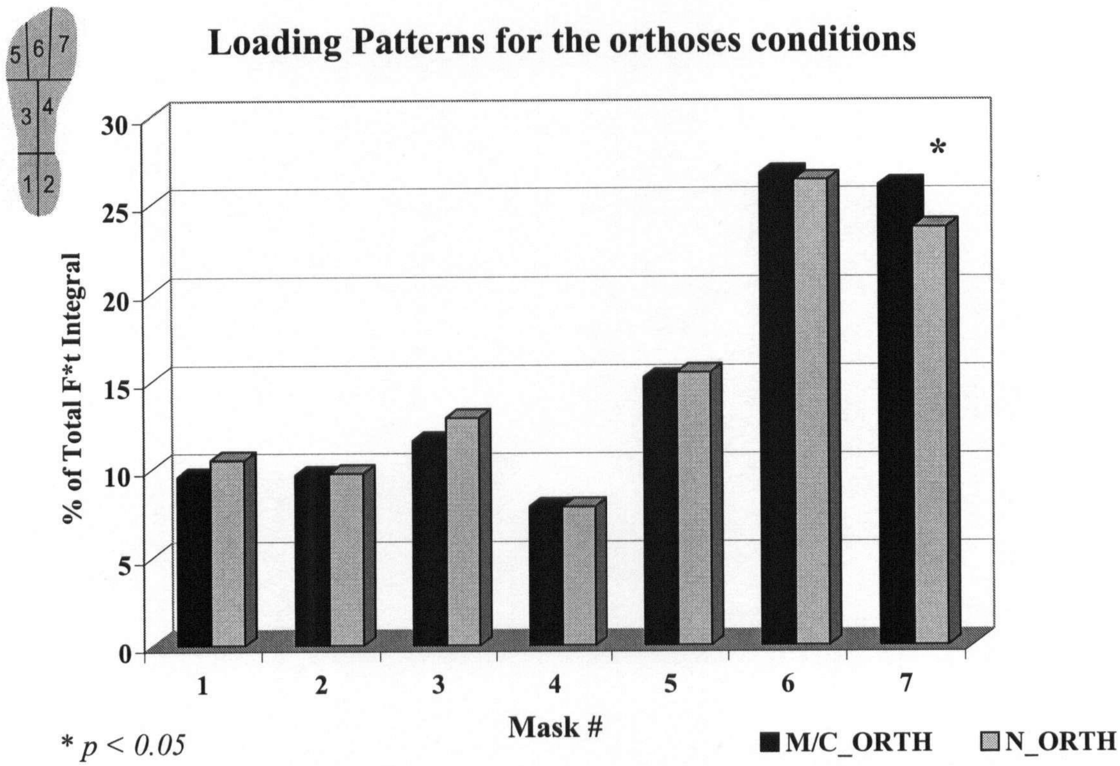

Figure 4.5 Loading patterns for orthoses conditions

A paired samples t-test was used to assess the differences between the two shoes for the orthoses conditions. The bars indicate the amount, in terms of percent, that each mask contributed to the total force*time integral. To clarify, we calculated the F*t integral over the entire stance phase for each mask we then calculated the percentage that impulse made up of the total F*t integral, over the entire stance phase, over the entire surface of the foot. It was hypothesized that there would be a significant difference in the plantar surface loading patterns between the two shoes however as was the case with the coordination variability hypothesis, there is a significant lack of research in this area to warrant a directional hypothesis. The asterisk (*) above "Mask 7" in Figure 1.0 indicates that the two shoe conditions exhibited statistically different loading patterns with the motion control shoe averaging 2.39% greater F*t impulse than the neutral shoe ($p = 0.01$; ES = 1.19). Overall, the two shoes, when used with a foot orthoses, did not show statistically significant differences, in loading patterns ($p = 0.494$; ES = 0.69).

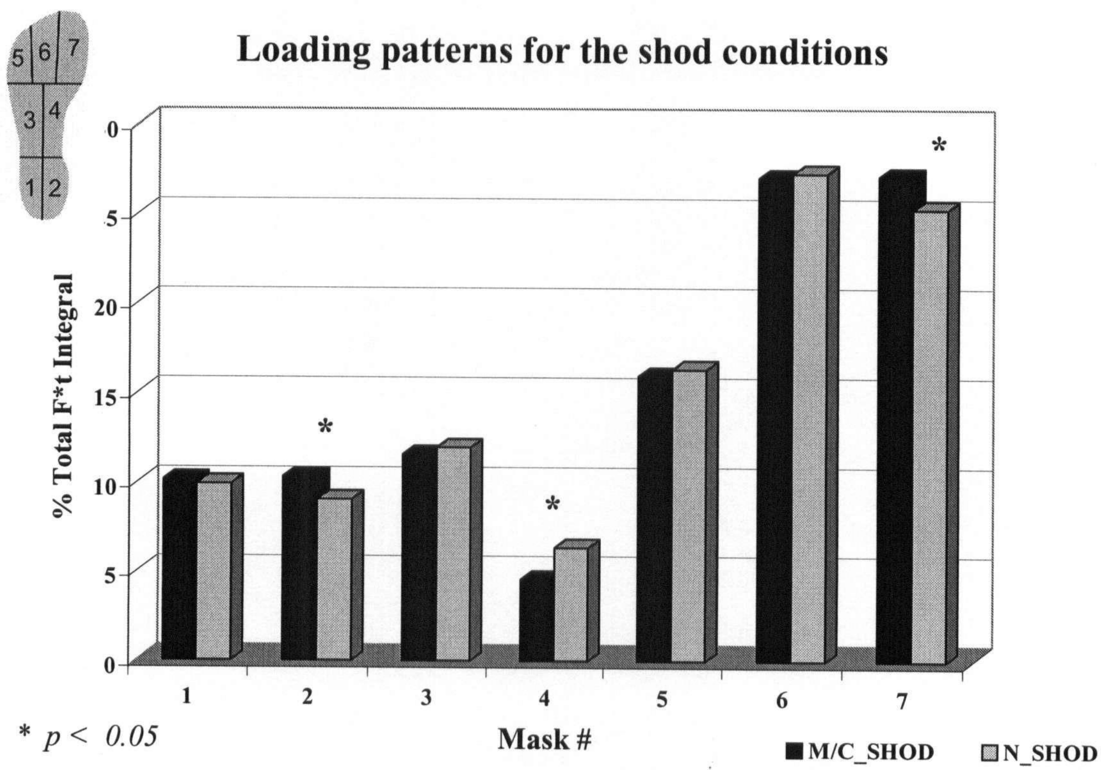

Figure 4.6: Loading Patterns for shod conditions

The conventions for *Figure 4.6* are the same as those for *Figure 4.5*, however *Figure 4.5* represents the data for the shod condition. Overall, there was no difference between the shoe conditions for loading patterns ($p = 0.718$; ES = 0.04), however, when each mask was compared individually, using paired-samples t-tests, significance was achieved in Mask two, four and seven, the three medial masks. The motion control shoe had a

1.27% greater impulse than the neutral shoe for Mask 2 ($p = 0.007$; ES = 0.85). The neutral shoe had a 1.8% greater impulse for Mask 4 ($p = 0.001$; ES = 1.15) and finally, the motion control shoe produced a 1.83% greater F*t impulse than the neutral shoe ($p = .009$; ES = 0.82).

4.1.3 Questionnaire Results

A within-subjects, repeated measures ANOVA (condition x question) was conducted for the results of the questionnaire. Figure 4.7 shows the average response for each question, for each shoe condition. The areas of assessment were:

1. Overall Comfort (O. Comfort)
2. Heel Cushioning (H. Cush)
3. Forefoot Cushioning (FF Cush)
4. Medio-lateral Control (M-L Cont.)
5. Arch Height (A. Height)
6. Heel Counter Fit (HC Fit)
7. Heel Width (H Width)
8. Forefoot Width (FF Width)

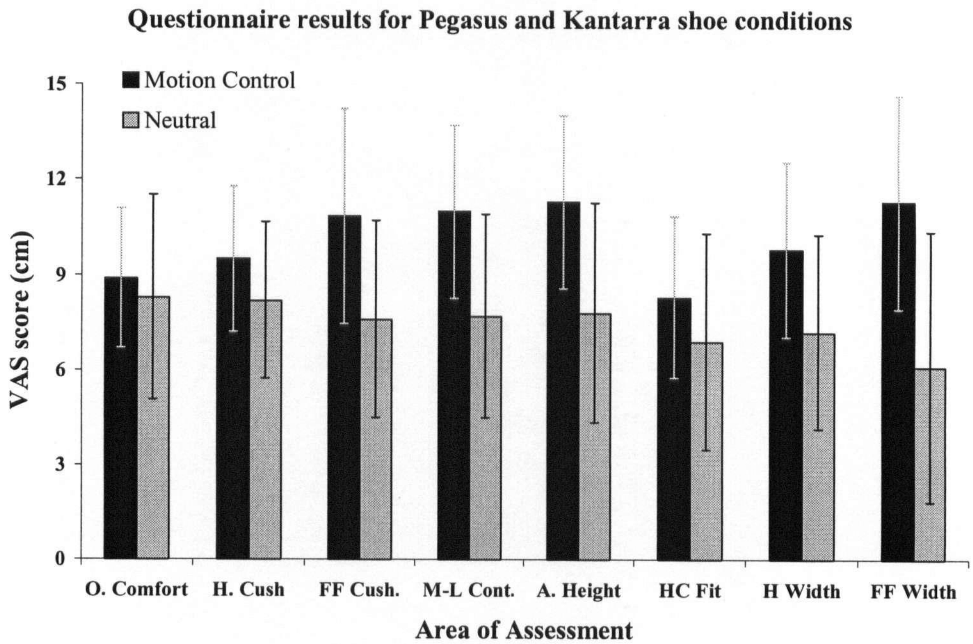

Figure 4.7. Shoe Comfort Questionnaire Results. Results are the average VAS score for the motion control shoe and the neutral shoe. Error bars equal one standard deviation.

Though there was a consistent trend of results for the Motion Control shoe exhibiting higher comfort ratings than results from the Neutral shoe, the results from our paired samples t-tests indicate that the differences are not statistically significant.

The results of the correlation between navicular drop and shoe comfort are presented in Table 4.3. Differences between the answers for each question were calculated by subtracting the response from the motion control condition with the response given in the neutral condition. Positive differences indicate the motion control shoe received a higher comfort score and negative results indicate that the Pegasus shoe received a higher score. The differences were then correlated with the navicular drop from the right foot (the foot used for testing). A positive correlation indicates the relationship that as navicular drop increased, the greater the difference in average VAS response for that condition indicating that the motion control shoe was more likely to be preferred. A negative correlation describes a situation where as navicular drop increased the smaller the average difference between the two shoes and the likelihood of the motion control being perceived to be the most comfortable shoe being reduced. We hypothesized based on previous research that correlated arch height to comfort ratings, that there would be a significant relationship between the amount of navicular drop and the design of the shoe that is found to be most comfortable. There were no statistically significant relationships found between navicular drop and perceived comfort in the areas of assessment.

Table 4.3. Correlation matrix for Navicular drop and Questionnaire results; n = 14

	O. Comf	**H. Cush**	**FF Cush.**	**M-L Cont.**	**A. Height**	**HC Fit**	**H Width**	**FF Width**
Navic. Drop	0.373	-0.258	0.254	-0.090	-0.481	-0.421	-0.296	0.474
Sig.	*0.210*	*0.395*	*0.402*	*0.770*	*0.096*	*0.152*	*0.327*	*0.102*

4. 2 Secondary Outcome Measures

Table 4.4. Comparisons (mean (SD)) of secondary outcome measures for the ORTH condition. (*) denotes a significant difference.

Variable of Interest	Neutral Shoe	Motion Control Shoe	ES	*p* value
Maximum Rearfoot Angle (deg)	8.57 (7.09)	8.39 (7.74)	0.27	0.918
Total Rearfoot Excursion (deg)	19.48 (4.66)	17.00 (4.85)	0.59	0.038*
Time to Maximum Eversion (% of Stance)	40.10 (11.58)	45.11 (14.77)	0.41	0.132
Time to Maximum Eversion Velocity (% of Stance)	14.36 (7.84)	12.80 (3.89)	0.29	0.28

Table 4.4. Comparisons (mean (SD)) of secondary outcome measures for the SHOD condition. (*) denotes a significant difference.

Variable of Interest	Neutral Shoe	Motion Control Shoe	ES	*p* value
Maximum Rearfoot Angle (deg)	8.47 (6.25)	7.11 (5.12)	0.55	0.05*
Total Rearfoot Excursion (deg)	20.44 (4.24)	16.21 (4.03)	1.47	0.00*
Time to Maximum Eversion (% of Stance)	41.82 (11.82)	45.42 (13.03)	0.25	0.35
Time to Maximum Eversion Velocity (% of Stance)	14.51 (4.30)	12.78 (4.06)	1.26	0.00*

Tables 4.3 and 4.4 summarize the mean values (±SD) for each of the secondary outcome measures. The secondary outcome measures in this study are examples of traditional analyses tools for assessing different interventions in running. The secondary outcome measures were included to augment the more novel primary outcome measures. Because there have been limited published research using vector coding techniques to measure joint coordination, it is difficult to compare the findings of this study with other gait analysis type research. The use of more traditional analysis methods allows us to compare our findings with other similar research. It also allows us to make inferences about the areas of increased coordination variability.

It was hypothesized that the motion control shoe would produce a significantly smaller rear foot angle than the neutral shoe. The maximum rear foot angle is a measure of the maximum degree to which the rear foot is everted. For the orthoses condition, the motion control shoe did produce a smaller maximum rear foot angle, however the difference was not significant ($p = 0.918$; ES = 0.27). When the runners removed their orthoses, the maximum rearfoot angle was significantly smaller than that of the neutral shoe ($p = 0.05$; ES = 0.55).

It was hypothesized that the motion control shoe would produce significantly smaller total rear foot excursion values. Total rear foot excursion is a measurement of the range of motion the rear foot moves through from the position of maximum inversion to the position of maximum eversion. When running with orthoses, the motion control shoe did produce significantly smaller rear foot excursion values ($p = 0.038$; ES = 0.59); similarly, when the orthoses were removed, the motion control shoe produced significantly smaller rearfoot excursion values ($p = 0.00$; ES = 1.47).

"Time to maximum eversion" allows the reader to identify at what point during the stance phase the rear foot was in a position of maximum eversion, a position that has been identified in pervious research to be a position at which the sub-talar joint begins to invert, the tibia begins to externally rotate and the knee begins to extend; this period of the stance phase has been theorized to be a position of instability which may be related to increased coordination variability when discussing joint coupling between the knee and rear foot. It was hypothesized that time to maximum eversion would occur significantly later than when running with the motion control shoe versus the netural shoe. When running with orthoses, the time to maximum eversion occurred at 45.11% of stance phase compared with the neutral shoe, where it occurred at 40.10% of stance, the difference was not significant ($p = 0.132$; ES = 0.41). When the orthoses were removed, time to maximum eversion did not change considerably. The values were 45.42% and 41.82% of stance for the motion control and neutral shoe respectively; the difference was not significant (*p* = *0.35*; ES = 0.25).

The final secondary outcome measure is one which measures the time at which the maximum eversion velocity occurs. This variable has been implicated in previous research as a possible mechanism for running injury. It has been suggested by previous research that increased rates of rear foot eversion during the loading phase of stance reduces the amount of time that impact forces are able to be attenuated. It was hypothesized that the 'time to maximum eversion velocity' would occur significantly earlier in the motion control shoe, than in the neutral shoe. When wearing the motion control shoe with orthoses while running, maximum eversion velocity occurred 1.56% earlier in stance than when running with the neutral shoe; this difference was not statistically significant (*p* = *0.28;* ES = 0.29). When the orthoses were removed, maximum eversion velocity occurred 1.73% earlier in stance than when running with the neutral shoe without orthoses, a difference that was statistically significant (*p* = *0.00;* ES = 1.26).

Chapter 5.0 Discussion

The purpose of this investigation was to conduct a comprehensive investigation including the analysis of the coordination variability of the knee/rearfoot joint coupling, rearfoot kinematics, an analysis of plantar surface loading patterns and a shoe-comfort questionnaire to provide a better understanding of the role two commonly prescribed shoe-orthoses combinations plays in running gait. Each component of the investigation is not without its limitations; however, in many cases where one component is limited, another component may be able to augment the analysis. The true value of this investigation is its use of vector coding to quantify coordination variability and how it differs when two functionally different shoes are worn with the same foot orthoses.

Vector coding and its measure of coordination variability is an analysis tool that is relatively new to those who are interested in the pathomechanics of running injury. There has been a plethora of research that uses rearfoot and knee kinematics, either in isolation, or how they compare in a discrete analysis (i.e. specific points in time as opposed to a continuous analysis) to study running pathomechanics, however there are only a handful that use coordination variability as a clinical measure.

In order to better understand the results of this analysis and the interpretation, it is important to have a clear understanding of variability and it's relation to joint coupling coordination. Traditionally, when discussing variability and locomotion, variability was thought to be a limitation to successful locomotion and has been associated with pathology (Heiderscheit, 2000). For instance, in human locomotion, within-subject variability of stride characteristics such as stride length, stride frequency, stride duration etc has been associated with reduced gait stability and have been regarded as good predictors of falling (Heiderscheit, 2000). However, when joint coordination patterns are the outcome measure, variability is said to be an essential component. Variability of joint coordination is said to provide the necessary flexibility for adaptation to uncertain terrain, external perturbations and is necessary for shock attenuation, while running (Heiderscheit, 2000). It is thought that reduced coordination variability could lead to an overuse situation whereby the constant stress repeatedly being placed on soft tissues (i.e. cartilage, tendons and ligaments) could lead to degenerative changes (Hamill *et al.* 1999). One might argue that running on a treadmill eliminates the uncertainty that exists when running overground or on varying terrain, however, there is likely a degree of uncertainty prior to heel contact on the moving treadmill (Heiderscheit *et al.*, 2002).

It was hypothesized that there would be a significant difference in coordination variability of the knee/rearfoot rotation couple while running in the motion control shoe and the neutral shoe, with and without orthoses. The results of this investigation reveal that differences between the two shoes exist when comparing joint coordination variability with and without orthoses. When averaging the variability across the stance phase in the orthoses condition, there were no statistically significant differences in average joint coordination. Previous research has found this same trend, and thus on the suggestion by Heiderscheit (2002), a more sensitive analysis was conducted whereby regions of the stance phase were isolated for comparison. However, it was interesting to find that the average variability across the stance phase of the SHOD condition was statistically different with the motion control shoe allowing more variability in joint coordination. A more specific regional analysis revealed that when wearing orthoses, there was a trend toward the neutral shoe allowing greater joint coordination variability with the neutral shoe statistically different than the motion control shoe in the 0 – 40% and 31 – 60% of stance regions. When the orthoses were removed, there was a trend toward the motion control shoe providing more coordination variability with the entire stance region, and the region of stance between 10 and 30% being statistically different.

The opposing trends between the orthoses and SHOD conditions with respect to coordination variability would indicate that too much control was being provided by the motion control shoe when orthoses were worn thus preventing the foot from adapting to the terrain effectively, and when the orthoses were not worn, the motion control was a more appropriate shoe to wear. As the subjects were prescribed their orthoses as a result of running-related injury (i.e ITBFS and PFPS), wore their orthoses daily and ran with them habitually, it would make sense that when the orthoses were removed, the stability features of the motion control shoe provided the needed control during stance. Comparing the coordination variability while running with the neutral shoe and orthoses and running with the motion control shoe and no orthoses produces a statistically different result with the neutral shoe and orthoses combination allowing greater coordination variability ($p = 0.038$; ES = 0.21).

The pattern of coordination variability exhibited by the knee rotation/ankle rotation joint couple is consistent with previous work by Heiderscheit, Hamill and Van Emmerik (2002). Areas of increased coordination variability are anticipated prior to and at heel contact, and during changes in joint coordination when there is a reversal of direction of one joint or both. This was exhibited for both the SHOD and orthoses conditions. At heel contact, there is increased coordination variability. As the rearfoot begins to evert and the knee flexes coordination variability decreases. Maximum rearfoot eversion velocity occurs 12.5% and 14.5% of stance for both the orthoses and SHOD conditions, during this time, there is a marked increase in coordination variability. This allows the body to be flexible and adaptable to the terrain and external perturbations while the rearfoot is everting the quickest in an effort to attenuate shock. The next large increase in coordination variability is around 40-45% of stance, the point at which our subjects reached maximum rearfoot eversion, this corresponds to the literature that indicates that increased joint coordination occurs around joint reversal. Following maximum rearfoot eversion, the rearfoot begins to invert and the knee begins to extend, both are examples of joint reversal. Finally, the coordination variability is reduced and remains stable for the final 40% of stance.

Our second primary outcome measure was plantar surface loading patterns. It was hypothesized that there would be a difference in loading patterns between the two shoes. Our hypothesis was based on both previous literature, and the fact that the two shoes have been designed with functionally different features. Specific features such as the shape of the last, the midsole construction and durometer as well as heel counter all contribute to the controlling nature of the shoe. The only difference between the two shoes for the orthsoes condition was in mask seven, which encompasses the hallux region of the foot. A larger F*t impulse was exhibited in this region of the foot while running in the motion control shoe which indicates that the runner likely produced more force and or time in this region of the foot, while wearing orthoses. If there were statistically larger F*t integrals in masks two and four, the two medial masks in the heel and midfoot respectively, for the motion control shoe, one could infer that force transfer tracked medially and did not exhibit the typical "s" shape of force progression. However, this was not the case; the F*t impulses were identical for the motion control and neutral shoes, in both mask two and four. Finally, there is no specific design feature of either shoe in the toe region that would produce a greater F*t impulse there, this in combination with the lack of significant difference in masks two and four for F*t impulse make it difficult to explain the significant difference in mask seven.

The SHOD condition exhibited different loading patterns with more masks exhibiting statistically different F*t impulses. Because the subjects were running without orthoses, their plantar pressure loading patterns were influenced directly by the shoe construction and not by the shoe-orthoses combination. There was the same significant difference in mask seven, with the motion control shoe exhibiting a larger F*t impulse than the neutral shoe, however, there were also significant differences in masks two and four. Mask two is on the medial side in the heel region; this is the area of the foot that is in direct contact with the dual density midsole of the motion control shoe. Based on previous literature that suggests too much rearfoot control can often block motion causing the foot to act more like a rigid structure with reduced ability to attenuate shock, the finding that the larger F*t impulse in mask two while running in the motion control shoe is likely the result of large impact forces rather than extended periods of time spent in this region. The argument may be raised then "why didn't the orthoses and motion control shoe combination produce an even greater F*t impulse." A possible answer to this question is that, an orthoses is designed to control motion and enhance foot function while a motion control shoe acts to block rearfoot movement, rather than control it. The difference being, that the act of "controlling" motion allows the foot to evert and invert within healthy limits, without limiting necessary rotations. The act of "blocking" motion on the other hand often prevents the foot from functioning in a healthy manner by reducing eversion beyond healthy limits. The neutral shoe produced a significantly larger F*t impulse than the motion control shoe in mask four. The subjects in this study habitually wore their foot orthoses while running, in order to control and enhance their rearfoot motion. The design of their orthoses was such that it prevented prolonged and excessive eversion. When the runners ran without their orthoses, in a shoe that has no compensatory controlling features, their foot reverted back to its uninhibited, anatomical function. Although maximum eversion occurred earlier in the neutral shoe it was likely prolonged, as was the case in Hamill *et al.*'s research (1992), thus contributing to the increased F*t impulse in mask four.

The pressure analysis portion of the investigation was the least reliable of the tools used thus introducing limitations to the conclusions that can be drawn from this aspect of the research. At the time of inclusion, the Pedar pressure analysis system was the best system available to us and was thought to be an excellent addition to the kinematic and joint coupling data. And in fact, the Pedar pressure analysis system is the best system available to those interested in studying plantar pressures. However, its strengths and reliability depend on proper calibration of the insoles and all sensing units working properly. If the insoles have been calibrated properly, and all sensors are reading pressure changes properly, the data can provide excellent insight as to what is happening at the shoe-foot interface. The insoles are approximately $700 USD/pair (Nike price) therefore regular replacement is not a financially viable option. Nike lent us the system as well as their best insoles. The insoles were calibrated accurately using a pressure chamber that calibrated the insoles every 0.5 bar from 0 to 7.5 bar, however at the time of data collection, there was still noise being measured when the insole was not loaded. Prior to reducing the data with the program provided by the University of Tübingen, the noise was filtered from the data, however, it is likely some pressure data was lost during filtering. The data reduction program used to analyze the plantar surface loading patterns has been tested for its reliability in shoe research and has been found to be reliable; however, it is limited by the accuracy of the data collected by the insoles. In an ideal world, the insoles would be brand new and perfectly calibrated however this study was limited by the availability of reliable insoles. Nonetheless, the results found are of interest however must be interpreted with caution.

The final primary outcome measure was the linear relationship between navicular drop and shoe comfort. Navicular drop was calculated by subtracting the navicular height at 'resting calcaneal stance position' from the height of the navicular at a position of 'subtalar joint neutral.' This was deemed to be the most appropriate arch measure as it had the fewest limitations and provided insight into foot morphology with respect to foot flexibility. Shoe comfort was assessed using a questionnaire designed by Mündermann and colleagues (2002).

The questionnaire was tested for validity and reliability and was found to be both valid and reliable when used with a control shoe condition and measured over five to seven trials. It was not possible to measure each of our subjects between five and seven times for each shoe condition, however, this questionnaire is still the most reliable shoe comfort questionnaire published so was used regardless of the inability to test each subject multiple times. The control shoe is to be presented prior to the test shoe condition so that all test shoe conditions are compared to the same footwear, rather than to each other. This investigation used a control shoe, the Nike Free shoe. This shoe is thought to be the ideal control shoe as it has no features built into it that are characteristic of any supporting shoe as it provides minimal support and minimal cushioning. The Nike Free has been designed to function similar to the bare foot. The control shoe was run in for two minutes with the foot orthoses prior to running in each of the motion control and neutral shoes. The questionnaire was administered following each shoe-orthoses combination. The questionnaire was specifically not issued following the SHOD conditions as currently; the subjects are running with their orthoses in their day-to-day activity. This investigation aimed to find the most ideal shoe-orthoses combination so it was thought that the results of the questionnaire would provide insight as to what shoe is most comfortable when orthoses are worn.

It was hypothesized that there would be a significant linear relationship between navicular drop and the shoe-orthoses combination that was most comfortable. This hypopthesis was based on the assumption that different shoe designs are designed for different structures of feet. There were no statistically significant correlations between navicular drop and shoe comfort. The strongest correlation was arch height ($r = -0.481$; $p = 0.096$) and the weakest was medio-lateral control ($r = -0.090$; $p = 0.770$). These results indicate that as navicular drop increased, the smaller the average difference in VAS score thus reduces the likelihood that the motion control shoe would be perceived as more comfortable than the neutral shoe. Overall there were no differences between two shoes for the average VAS response. The motion control shoe did have higher perceived comfort ratings in each of the areas of evaluation, however there were no statistically significant differences.

It was interesting that the motion control shoe with the orthoses was perceived to be more comfortable than the neutral shoe with the orthoses, though not significantly different, there was a trend. The joint coordination data would indicate that this is the least appropriate shoe-orthoses combination. This analysis was limited to responses following a seven minute run in each test shoe condition. Perhaps after the shoes had been run in regularly for three months, the repercussions of reduced joint coordination would present.

The questionnaire was designed so that all the instructions were written at the top such that the person administrating the questionnaire did not provide information that would bias the results, or give inconsistent instructions between subjects (Mündermann *et al.*, 2002). Unfortunately, the instructions were not clear enough for the subjects in this study and in most cases, clarification was required. The question that provided the most difficulty was that which assessed 'medio-lateral' control. 'Heel counter' was not well understood either. The investigator tried to give consistent information to clarify the questionnaire; however, the subject's uncertainty was a limitation of this method of analysis.

This investigation included four secondary outcome measures that were concerned with rearfoot kinematics. The variables measured were maximum rearfoot angle, total rearfoot excursion, time to maximum eversion (as a % of stance) and time to maximum eversion (as a percentage of stance). The majority of shoe, orthoses and biomechanics literature that has investigated lower extremity actions have focused on the aforementioned rearfoot kinematics rather than addressing the interaction between the joints (Hamill *et al.,* 1999). Because so little research has been conducted using vector coding and joint coordination, the inclusion of the secondary outcome measures allowed for comparison with previously conducted research. Joint coordination measures are based on rearfoot kinematic and knee kinematic data, however, joint coordination is thought to be a more valuable tool for studying running pathomechanics and the trend in research likely will move this way.

It was hypothesized that the motion control shoe would produce a smaller maximum rearfoot eversion angle than the neutral shoe. This hypothesis was based on research conducted by Hamill *et al.* (1992) and Clarke *et al* (1983) that found that pronation can be decreased by shoes that have a stiffer midsole, a stiffer heel counter and a wider heel base, all of which are components that distinguish the motion control shoe from the neutral shoe. The motion control shoe did exhibit a smaller rearfoot angle than the neutral shoe however the difference was not significant; however there was a significant difference for the SHOD condition with the motion control shoe exhibiting a smaller rearfoot eversion angle. These results may suggest that when an orthoses is combined with a shoe, regardless of the controlling features of the shoe, it is able to enhance motion rather than block rearfoot eversion entirely. The rearfoot angles were larger for the orthoses condition than the SHOD condition; this finding does not make sense as we would expect an increased inversion moment due to the addition of a post (intrinsically built into the orthoses) to the medial aspect of the calcaneus. These data are further evidence to suggest the SHOD conditions were limiting healthy range of motion while the orthoses acted to facilitate healthy range of motion. Features of the orthoses that would contribute to this facilitation would be the EVA top cover in combination with the material used to post the orthoses. Several layers of different EVA foam are used to prevent this idea of "blocking" motion and instead, acts to "facilitate" the motion.

It was hypothesized that the motion control shoe would produce less total rearfoot excursion than the neutral shoe as the control features of a motion control shoe are designed to block rearfoot eversion. The motion control shoe produced significantly less rearfoot excursion than the neutral shoe in both the orthoses and SHOD conditions. The neutral shoe in the shod condition produced greater rearfoot excursion than the orthoses condition, however the maximum rearfoot angle was smaller, this would suggest that the heel contacted the ground in a relatively higher degree of inversion than was the case in the orthoses condition. The amount of rearfoot excursion is an important variable because injury is often sustained if the controlled eversion does not occur. The greater the range of motion the rearfoot must evert through, the larger the demand placed on the supporting structures, whether it be muscles, ligaments, tendons, boney structures, shoes and orthoses, to control the eversion. The values obtained in this investigation are generally greater than those obtained in the study by Clarke *et al.* (1983) where three midsoles, three heel flares and three heel heights were compared.

It was hypothesized that the time to maximum eversion would occur significantly later in the motion control shoe than the neutral shoe. This hypothesis was based on the work by Hamill, Bates and Holt from 1992. They found the shoe with the hardest midsole to exhibit increased time to maximum rearfoot angle when compared with two softer midsoles. The trend was that maximum rearfoot eversion occurred later in the motion control shoe than the neutral shoe, however this trend wasn't significant. This variable was valuable when interpreting the joint coordination values as it is a measure of a point in time when the joints are theorized to begin reversal, a point where increased joint coordination is said to occur. The findings in the study by Hamill *et al.*(1992) were similar to ours with respect to time at maximum rearfoot velocity, the data for their hard midsole corresponds to the data for the motion control shoe in this study, and the medium durometer in Hamill's study corresponds to the data from the neutral shoe in this study.

The final outcome measure was time to maximum eversion velocity (as a percentage of stance). Increased eversion velocity has been linked to overuse injuries as it reduces the amount of time available for shock attenuation. Based on the research by Hamill et al (1992) we hypothesized the time to maximum eversion velocity to be increased in the neutral shoe. For both the orthoses and SHOD conditions, this was evident however the differences were only significant in the SHOD condition. The mean values obtained were similar between orthoses and SHOD conditions which is a possible indicator that the shoe is more closely related to eversion velocity than foot orthoses. The conclusions were consistent with those from Hamill's study and Clarke's study.

Rearfoot motion in the frontal plane is said to be one of the most variable rearfoot kinematic variables, with variability being an undesired outcome in this instance. Reliability measures were conducted on the rearfoot and knee kinematic data and were found to be reliable between testing sessions. The rearfoot and knee kinematic data was further analysed to produce joint coordination information. If the kinematic data is reliable, then the joint coordination data can be regarded as reliable.

5.1 Conclusion

The current investigation sought to answer the question proposed by Hintermann and Nigg back in 1998: "to what extent may shoe modifications and supports reduce [unhealthy] foot pronation and thus, help to prevent injuries?" Through the use of vector coding techniques, pressure analysis, analysis of rearfoot dynamics and a shoe comfort questionnaire this study produced information relating to joint coordination variability, plantar pressure loading profiles and information regarding rearfoot dynamics for a motion control shoe and a neutral shoe when run in by female runners with and without foot orthoses. The analysis of coordination variability provided information that can be used to better understand running pathomechanics as well as suggest one shoe over another. The rearfoot dynamic results augmented the coordination variability results allowing better understanding of the joint coordination that was taking place. The pressure analysis data, though limited in its level of reliability shed some light on the difference in loading profiles between SHOD and orthoses-shoe combinations. The questionnaire unfortunately did not produce any conclusive evidence regarding shoe comfort ratings as they relate to foot structure.

Based on the research presented in this investigation, it would appear as if a neutral shoe be worn with moderately posted rearfoot orthoses rather than a motion control shoe. The motion control shoe in combination with the orthoses appeared to block rearfoot motion rather than control it, and thus failed to enhance foot function. This resulted in reduced coordination variability which has been found to be an inflexible locomotion pattern and could be linked to, or lead to running injury such as cartilage damage or tendon tears. When running without moderately posted orthoses, the motion control shoe provided the necessary rearfoot support allowing the foot to function within variable ranges, thus being more adaptable to unsuspected external perturbations. The ideal combination of shoe and insert is the neutral shoe with the moderately posted orthoses. This combination enhances foot function and guides it through the stance phase. Though the motion control shoe without orthoses was preferred to the neutral shoe without orthoses, it did not produce greater coordination variability than the neutral shoe with foot orthoses, further lending to the belief that the motion control shoe is blocking proper rearfoot function. Based on this research, a shoe with controlling features is needed however one that does not block healthy mechanics such as rearfoot eversion. The shoe needs to be designed to facilitate and enhance healthy foot function.

The mask analysis indicated that the orthoses overcame the design features of either shoe to produce similar plantar pressure loading profiles. The mask analysis from the SHOD condition was much more indicative of the effects of design features of each shoe, on the plantar surface of the foot. There was a greater F*t impulse on the medial heel region for the motion control shoe, that could have implications for increased force at touch down. The greater F*t impulse in the medial mid-foot region for the neutral shoe compared with the motion control shoe suggests that the rearfoot was delayed in inverting following max eversion. Though max eversion occurred earlier in the neutral shoe, it may have been delayed in re-supinating. This could have implications for the timing of events, whereby the knee extension could have begun to extend prior to the rearfoot supinating.

The trend from the questionnaire indicating the motion control shoe and orthoses combination was perceived to have higher comfort ratings could indicate potential for overuse injury, if runners perceive this shoe to be the most comfortable upon initial wear.

Based on the coordination variability data, this could lead to injury as training volume increases. Though discomfort wasn't present in the short-term, pain and discomfort could present after several miles run.

This research is the first of its kind to assess joint coordination variability and how it differs between two different shoes, with and without orthoses. The analysis however could be improved with a six, high-speed video analysis. This would allow movement to be quantified in more than one plane. The knee angle data is likely quite accurate as flexion/extension occurs in the sagittal plane only, however measurement of rearfoot motion in the frontal plane underestimates the amount of movement that is taking place at the rearfoot as other planes of motion are involved. Finally, brand new insoles would add to the reliability of the plantar pressure data.

5.2 Future Direction

Future research should focus on the amount of coordination variability that is desired while running. Is the coordination variability that is required different for running on consistent terrain such as on a sidewalk vs. inconsistent terrain such as a trail. What are the ranges of coordination variability expected for an uninjured population of runners versus an injured population? More studies need to be conducted with injured and uninjured populations to establish tolerance ranges. In addition, can this tool be used for injury prevention? Future prospective studies could be conducted on high-risk runners to determine whether we can use this tool as a predictor for injury and early intervention to prevent running injury. Measuring joint coordination variability is an excellent step forward in the analysis of running pathomechanics, and future research should focus on expanding this research rather than continuing to address lower limb kinematics and its measures in discrete moments in time.

Finally, with respect to shoe design, a shoe needs to be designed that has greater controlling features than a neutral shoe, but is less restrictive than a motion control shoe for those runners who require moderately posted rearfoot orthoses. This new shoe would ideally have a midsole which acts to support and guide the foot through the stance phase rather than restrict motion and would allow the foot to function in a healthier manner, enhancing and facilitating its biomechanical function.

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Appendix A

ALLAN McGAVIN SPORTS MEDICINE CENTRE

Running History and Anthropometric Information

Name:

Address: **E-mail:**

Training Pace: **Competition Pace:**

# of Years Running: **Average Weekly Mileage:**

Type of Training Shoe:

Height: ________ **Weight:** ________ **Age:** ________

Absolute Leg Length (ASIS – Medial Malleolus): L ________ R ________

Relative Leg Length (Umbilicus – Medial Malleolus): L ________ R ________

Pelvic Obliquity: Y N

Tibial Varum: Y N

Navicular Drop: STJ N ________ RCSP ________

Forefoot Varus: L ________ R ________

Heel Varus: (NWB) L ________ R ________ (WB) L ________ R ________

Static Weight Bearing
Rearfoot Angle (γ): L ________ R ________

Appendix B

Nike Air Kantara II – Motion Control Shoe

Width: Medium

Sole Construction:
Blow molded air bag in the rearfoot. Dual chamber, dual pressure air bag in the forefoot region. The lateral forefoot outsole is blown rubber. There is a higher density foam piece along the rear midsole covered with a plastic foot bridge to provide additional motion control.

Nike Air Pegasus – Neutral Shoe

Width: Medium

Sole Construction:
Full length, constant pressure air cushion encapsulated in the foam midsole from rearfoot to forefoot.

Appendix C

Variable	Variable symbol	Definition
Rearfoot		
Maximum rearfoot angle	β_{max}	The greatest amount of rearfoot eversion achieved during foot contact
% Time to max RF angle	β_T	Time between foot strike and when max RF angle occurs
Total rearfoot excursion	β	Total amount of rearfoot eversion exhibited during stance phase = $\beta_{max} + \beta_0$
% Time to maximum eversion velocity	$\beta_{Tmaxvel}$	Time between foot strike and when maximum velocity of eversion occurs

Appendix D

Shoe Comfort Questionnaire

Please answer the following questions to the best of your ability. The questions are designed to establish shoe preference based on your personal experience running in the shoe. Please place a mark along each line, indicating your opinion of each category. The further to the right a mark is placed, the more comfortable the shoe. Below are some definitions to help you.

Overall Comfort:	Overall impression of the shoe
Forefoot Cushioning:	Softness/hardness of the shoe in the forefoot region
Heel Cushioning:	Softness/hardness of the shoe in the heel region
Medio-lateral control:	Position of the foot controlled by the shoe
Arch height:	Medial arch height of the shoe
Heel cup fit:	Fit of the shoe in the heel region
Shoe heel width:	Width of the shoe in the heel region.
Shoe forefoot width:	Width of the shoe in the forefoot region

Questionnaire adapted from: Mundermann A *et al.* Development of a reliable method to assess footwear comfort during running. *Gait and Posture* 16: 38 – 45, 2002.

Condition: **Name:**

Overall Comfort:

Not comfortable at all — Most comfortable shoe

Heel Cushioning:

Not comfortable at all — Most comfortable shoe

Forefoot Cushioning:

Not comfortable at all — Most comfortable shoe

Medio-lateral control:

Not comfortable at all — Most comfortable shoe

Arch height:

Not comfortable at all — Most comfortable shoe

Heel Cup Fit:

Not comfortable at all — Most comfortable shoe

Shoe Heel Width

Not comfortable at all — Most comfortable shoe

Shoe Forefoot Width

Not comfortable at all — Most comfortable shoe