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Kerry RAMBARRAN
AUTEUR DE LA THÈSE - AUTHOR OF THESIS

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Effectiveness of the Kinetic Wedge Foot Orthosis Modification to Improve Gait Posture

E. Lemaire
DIRECTEUR DE LA THÈSE - THESIS SUPERVISOR

G. Robertson
CO-DIRECTEUR DE LA THÈSE - THESIS CO-SUPERVISOR

EXAMINATEURS DE LA THÈSE - THESIS EXAMINERS

J. Kofman

H. Sveistrup

J.-M. De Koninck, Ph.D.

LE DOYEN DE LA FACULTÉ DES ÉTUDES SUPÉRIEURES ET POSTDOCTORALES

SIGNATURE

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Effectiveness of the Kinetic Wedge Foot Orthosis Modification to Improve Gait Posture

Kerry K. Rambarran
B.Sc., University of Ottawa

Thesis submitted to the
Faculty of Graduate and Postdoctoral Studies
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Département of Human Kinetics
Faculty of Health Sciences
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Dedications

To: Taryne Smith,
her love is my strength.

My parents: Bhagwatee (Golly)
and Chandrika (David)
Their hard work and sacrifices for
my education will never be forgotten.

ABSTRACT

Clinically, Functional Hallux Limitus (FHL) has been identified as a contributor to the disruption of proper progression of the inverted pendulum through the sagittal plane. The Kinetic Wedge Custom Foot Orthotic (CFO) modification has been used by the podiatric community to facilitate proper first metatarsal-phalangeal (MTP) joint function and improve gait posture. The aim of this study was to determine the effectiveness of the Kinetic Wedge custom foot orthosis (CFO) modification to improve gait, posture, centre of pressure velocity, plantar pressures of the foot, and perceived pain. Fifteen healthy subjects (9 women, 6 men) 22 to 53 years of age diagnosed with moderate to severe FHL volunteered for the study. Kinematic and plantar pressure data were collected without Kinetic Wedge (NKW) and with Kinetic Wedge(KW). There was a significant reduction of plantar pressure under the first MTP, however, there were no significant changes in plantar pressures under the hallux and fifth metatarsal. There was no significant difference in centre of pressure velocities. In addition, there were no significant increases in trunk, hip, knee, and ankle range of motion. Furthermore, self perceived pain did not significantly reduce 2 months after testing.

TABLE OF CONTENTS

List of Abbreviations

Centre of Gravity - CoG

Centre of Pressure - CoP

Custom Foot Orthotics - CFO

First metatarsophalangeal joint - first MTP joint

Functional Hallux Limitation - FHL

Head-arms-trunk segment - HAT

M = mean difference

Range of Motion - RoM

Subtalar neutral custom foot orthotics without kinetic wedge modifications - NKW

Subtalar neutral custom foot orthotics with kinetic wedge modifications - KW

List of Tables

List of Figures

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1.0 INTRODUCTION

Humans are the only species to successfully walk using an "erect" bipedal posture (Dananberg *et al.* 1996). However, modern humans were not the first bipedal species. Humans evolved from now extinct hominids that used less-erect gait postures.

An early hominid that resembled modern humans was *Australopithecus afarensis*. Fossil records indicate that *A. afarensis* populated eastern Africa three and-a-half to four million years ago (Figure 1). The *A. afarensis* species was unique, because unlike their hominids predecessors, *A. afarensis'* were bipedal.

Homo Habilis or "handy man" is believed to be a direct ancestor to modern human beings. *Homo Habilis* was the first hominid to use tools and adopt a gait posture that resembles modern humans (Lindsey, 2002). According to fossil records, the first metatarsophalangeal (MTP) joint of *Homo Habilis* differed greatly compared to its predecessors. The first MTP joint of *Homo Habilis* permitted an unprecedented range of dorsiflexion between the late stance phase and toe-off during gait (Dananberg *et al.* 1996). The early human was now able to more efficiently produce motion in the sagittal plane; the first MTP joint was used as a pivot for the inverted pendulum (Winter, 1993; Winter, 1995; Dananberg *et al.* 1996).

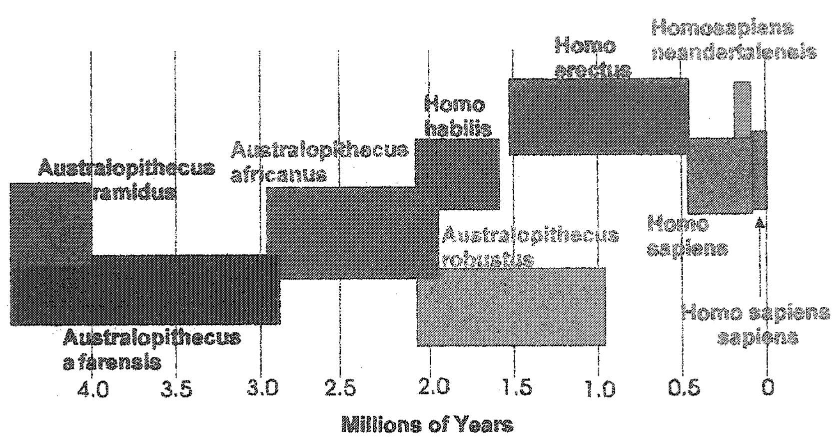

Figure 1: Re-created chronological chart of the main Hominid species (Lindsey, 2002).

The unique design of human feet allows us to walk in a relatively upright position. This relatively upright position facilitates the efficiency of the inverted pendulum during gait (Winter, 1995). An obstruction, inability or delay of the inverted pendulum to produce motion in the sagittal plane is referred to as sagittal plane blockade. A major source of sagittal plane blockage is the inability or delay of the first MTP joint to adequately dorsiflex from late stance phase to toe-off during the gait cycle (Root, 1977; Hicks, 1953; 1954; Dananberg, 1993; 1986; Dananberg *et al.*, 1996; Drago *et al.*, 1984; Roukis *et al.*, 1996). This condition is referred to as **Functional Hallux Limitus (FHL)**.

The effects of FHL can be divided into the two areas of focal and global. Focal effects can be viewed as symptoms or conditions at the first MTP joint itself. Global effects of FHL are believed to be expressed elsewhere (proximally) in the body such as the ankle, knee, hip and lower back.

1.1 Focal effects of FHL

The focal effects of FHL involve the inability or delay of the first MTP joint to achieve adequate dorsiflexion during the single limb support phase of gait (late stance). Some individuals with FHL may experience pain at the first MTP joint during the late stance phase of gait (Dananberg, 1993, 1986; 1986; Dananberg *et al.*, 1996). Although the first MTP joint lacks the ability to perform normally during gait, it may achieve full range of motion during conventional non-weight-bearing range of motion tests. If neglected, or improperly treated, many clinicians believe FHL may lead to more serious conditions; such as hallux rigidus, osteoarthritis of the first MTP joint, and hallux abducto valgus, or bunions.(Root, 1977; Hicks, 1953; 1954; Dananberg, 1993, 1986; 1996; Drago *et al.* 1984).

1.2 Global effects of FHL

Clinicians in the podiatric community suggest that the global effects of FHL can result in interruptions of a body's centre of gravity (CoG) progression through the sagittal plane (Dananberg, 1993; 1986). This can be better described as a slight disruption of the inverted pendulum as it moves through the sagittal plane (Winter, 1993;1995). According to Dananberg (1993; 1996), the inability of the first MTP joint to dorsiflex during stance phase leads to compensatory postural changes to force an individual's CoG anteriorly. These compensations can occur in other articulations in the foot as well as articulations proximal to the foot (ankle, knee, hip).

Previous clinical papers have also shown that FHL may result in a lateral compensatory strategy at the foot. Clinicians believe that this compensatory strategy uses MTP joints lateral to the first MTP joint to allow the CoG to progress anteriorly. This strategy manifests itself as relatively elevated plantar pressures under the fourth and fifth rays of the foot (Stokes et al., 1979; Lord and Hosien, 1994; Lord et al., 1986; Dananberg, 1995; 1993; 1986; Payne and Dananberg, 1997; Dananberg et al., 1996).

Related to body posture, FHL may result in proximal (ankle, knee, hip) kinematic changes during the stance phase of gait. Global effects stemming from such compensatory changes are suggested as significant contributors to various postural complaints.

Increased trunk flexion, or slouch has been described as a compensatory strategy to facilitate the progression of their CoG anteriorly in the sagittal plane during late stance (Dananberg, 1995; 1993; 1986; Dananberg & Payne, 1997; Dananberg et al., 1996). Dananberg suggested that this compensatory action may be a contributor to low back pain experienced by those diagnosed with FHL (Figure 2).

According to Dananberg (1993), the improper function of the first MTP joint during gait may be a cause of secondary conditions such as:

- Over-pronation of the Subtalar joint
- Achilles Tendinitis
- Patellofemoral Pain Syndrome
- Hip dysfunctions

Figure 2: Compensatory trunk flexion due to FHL.

1.3 Statement of the Problem

Recently, an unconventional method to reduce compensatory gait strategies emerged. The podiatric community referred to this new method as sagittal plane facilitation (Payne and Dananberg, 1997). Unlike the traditional Root paradigm which focussed on rear-foot biomechanics, sagittal plane facilitation emphasized the importance of proper forefoot biomechanics (Payne and Dananberg, 1997). Modified custom foot orthoses (CFOs) were used to induce proper forefoot biomechanics. One such modification was the kinetic wedge modification.

The podiatric community, including pedorthists and chiropodists, endorse the use of custom foot orthoses with the Kinetic Wedge modification (Langer Biomechanics Group Inc. - Figure 3) to promote proper MTP joint function in individuals diagnosed with FHL (Anthony, 1991;

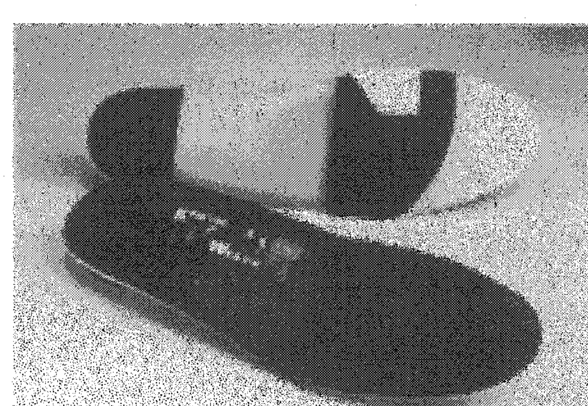

Figure 3: Subtalar neutral orthoses with Kinetic Wedge modification (Langer Biomechanics Group).

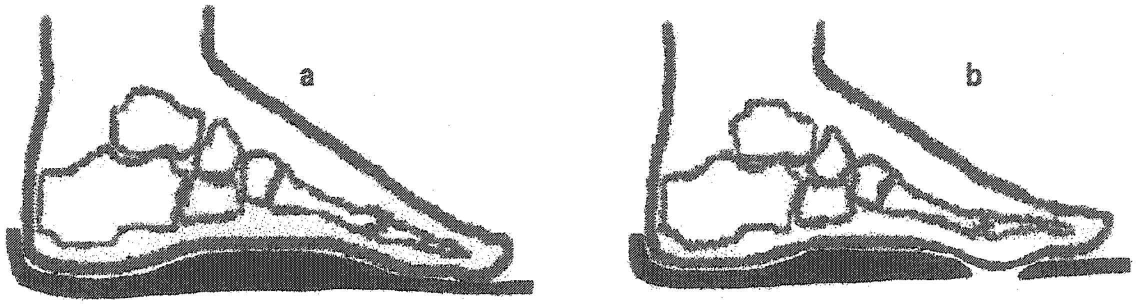

Figure 4: a) CFO without Kinetic Wedge modification. b) CFO with Kinetic Wedge modification inducing a relatively greater dorsiflexed first MTP joint position (sagittal view). Adapted from Dananberg (1995: 397).

Dananberg, 1995; 1993; 1986; Payne and Dananberg, 1997; Dananberg et al., 1996). The kinetic wedge was designed to promote "plantarflexion and eversion of the first metatarsal during the peak stress point during the step" (Dananberg, 1993).

The purpose of the Kinetic Wedge is to place the first ray in a greater plantar flexed position and the proximal phalanx more dorsiflexed relative to the first metatarsal. Clinicians believed that the limitation of the joint is bypassed with the joint at this initial position, thereby increasing the ability of the first MTP joint to dorsiflex (Figures 4 & 5).

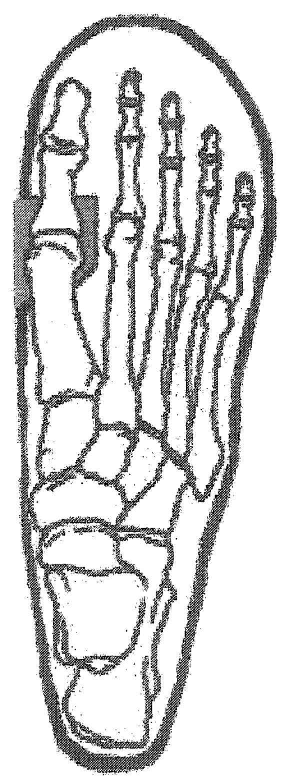

Figure 5: Transverse view of first MTP joint position relative to the Kinetic Wedge modification. Adapted from Sadurski (2001).

The Kinetic Wedge is most crucial at heel-off, when initial first MTP joint dorsiflexion occurs (Dananberg, 1995; 1993; 1986; Payne and Dananberg, 1997; Dananberg et al., 1996). Podiatric clinicians also believed that proper first MTP joint function could prevent FHL compensatory strategies during gait, thereby facilitating progression of the body's CoG through the sagittal plane. Although the use of the Kinetic Wedge is common in the podiatry community, no empirical studies were conducted to evaluate associated kinematic and kinetic changes.

1.4 Purpose of The Study

The purpose of this study was to determine if a CFO with a kinetic wedge modification would improve the posture of an individual with FHL during the late stance phase of gait, more specifically, whether or not the CFO would improve trunk, hip, knee, and ankle ranges of motion during stance. Secondly, the study evaluated the ability of the CFO with such a modification to reduce plantar pressures on the hallux, first MTP joint and lateral aspect (fifth metatarsal) of feet that exhibited FHL. Differences in plantar centre of pressure velocities were also investigated. The study determined whether or not application of the kinetic wedge modification would increase anterior/posterior plantar centre of pressure velocities.

1.5 Rationale

According to clinical findings, FHL can result in first MTP joint compression as the body's centre of gravity passes over the joint. Repetitive first MTP joint compression can eventually manifest itself as joint pain. Data supporting reduced and/or eliminated first MTP joint pain with modified CFO use may encourage more widespread use with patients who exhibit focal effects of FHL (Dananberg, 1986, 1993, 1995).

Data supporting clinical findings that unnecessary trunk flexion is reduced with modified CFO use may prove to be valuable to the medical community. Since low back pain has been linked to poor gait posture, verification of prior clinical success of Kinetic Wedge applications may support CFO use as an unconventional and indirect intervention for lower back pain (Dananberg, 1986, 1993, 1995). Data from this study may promote investigations to verify CFO use to reduce and/or eliminate other postural complaints such as: knee, hip, and lower back complications.

Repetitive peak plantar pressures of FHL combined with conditions that reduce normal tissue regeneration, such as diabetes, can result in tissue deterioration (Lord et al., 1994). The verification of improved plantar pressures can enable the podiatric community to manage diabetic feet, thereby avoiding amputations of digits and/or foot segments, or worse - entire feet.

1.6 Statement of Hypothesis

According to clinicians, Kinetic Wedge use instantly improved the first MTP joint's ability to dorsiflex during stance. Improved first MTP joint function resulted in a reduction of the focal effects of FHL. Adequate first MTP joint function reduced elevated plantar pressures under the joint (Dananberg, 1995; 1993; 1986; Payne and Dananberg, 1997; Dananberg et al., 1996). Assuming these clinical findings are true, the first hypothesis was that the use of a CFO modified with the Kinetic Wedge (KW) would reduce plantar pressures at the first MTP joint compared to a CFO without the Kinetic Wedge modification (NKW).

The second hypothesis also pertained to the focal effects of FHL. This hypothesis sought to determine whether or not a CFO modified with the Kinetic Wedge would reduce subjects' perceived first MTP joint pain after at least two months of use.

The third hypothesis was that reduced focal effects would reduce global affects of FHL. Plantar pressures under the hallux segment and fifth metatarsal would also decrease.

Clinically, CFOs modified with the Kinetic Wedge would instantly improve gait posture (Dananberg, 1995). Assuming these findings are true, the fourth hypothesis was that forward trunk lean during stance would instantly decrease during KW testing. In addition, hip extension, knee extension and ankle plantar flexion would increase during late stance.

Improved MTP joint function should facilitate sagittal plane motion, thereby increasing foot (centre of pressure) CoP velocities (Payne and Dananberg, 1997). Therefore, the fifth hypothesis was that anterior plantar CoP velocities would increase during KW testing.

1.7 Indicators

Plantar pressure during stance phase was measured using the F-Scan (Tekscan Inc.) system. Initial heel contact pressure and termination of forefoot contact pressure was used to denote stance. Reductions in plantar pressures at the first MTP joint, hallux, and fifth metatarsal of the forefoot were used to indicate efficacy of the Kinetic Wedge modification.

Kinematic data, specifically trunk, hip, knee and ankle angles were collected to determine the effects of the custom foot orthoses. Videographic data (sagittal plane) were collected and compared with and without the use of CFOs with the Kinetic Wedge modification.

1.8 Scope of the Study

While clinicians have maintained that kinematic changes may occur at each of the articulations proximal to the first MTP joint (ankle, knee hip), this study was mainly concerned with kinematic changes of the trunk segment. Two dimensional (sagittal plane) kinematic data of the trunk was collected and analysed.

Previous clinical papers have shown that FHL can cause changes to the plantar presure pattern at several areas under the foot. (Dananberg, 1995; 1993; 1986; Dananberg & Payne, 1997; Dananberg et al., 1996). Dananberg has shown a timing or sequencing relationship of the plantar pressures of heel off and hallux loading. While this relationship may exist, this study was concerned with the relative plantar pressures of the first MTP joint, hallux segment, and 5th metatarsal segment.

1.9 Assumptions and Limitations

The investigator assumed that the initiation of MTP joint dorsiflexion coincided with heel-off during late stance. A rapid reduction in plantar pressure at the heel coupled with a rapid increase of hallux plantar pressure was used to identify heel-off and forefoot loading. The investigator also assumed that the termination of plantar pressure at the toes signified toe-off.

Practice time between testing conditions were also related to time and cost restraints. Subjects were tested using the modified CFO after 30 minutes of practise time. The testing protocol reduced the subject time requirements. Reducing the number of visits to the testing facility also reduced parking costs. Parking and transportation costs were paid by the investigator. Reducing the time between testing conditions was also an attempt to reduce subject attrition.

Since first MTP joint range of motion occurs mainly in the sagittal plane, the study limited kinematic data collection and analysis in the sagittal plane. Compensatory movements in the frontal and transverse plane were not considered.

Like previous 2D kinematic studies, seven markers were used to identify major body joints and segments. The most proximal marker was the shoulder joint, rather than cervical spine or head markers. This choice was related to the purpose of the study, as more proximal markers will offer neck and head kinetics in the sagittal plane. It was assumed that hip and shoulder joint markers were sufficient to depict trunk kinematics during stance.

The F-Scan system (Tekscan Inc.) was susceptible to weaknesses. One weakness was the durability of insoles. Compared to other in-shoe plantar pressure systems, the F-scan insoles were less elastic. This characteristic decreased insole durability, and consequently longevity. In the event of insole damage, additional insoles were used. The need to replace damaged insoles occurred during data collection for two of the subjects. The need to use additional insoles was related to another weakness of the F-Scan insole sensitivity of plantar pressure readings as a result of insole temperature change. In the event of replacing insoles, subjects were required to wear the new insole(s) for at least 10 minutes. It was assumed that wearing the insoles for this period of time allows sensors to attain "in-shoe" temperature and thereby reduce variability.

One pair of CFO was constructed for each subject. Subjects were tested without the Kinetic Wedge modification first. The inability to provide subjects with two pairs of CFOs reduced the ability to randomize subject testing.

2.0 THEORETICAL BACKGROUND

2.1 Anatomy and Range of Motion of The First Ray and First MTP Joint

The first ray is comprised of a combination of articulations. These articulations include: the medial cuneonavicular joint (between the navicular medial cuneiform), the intercuneiform (between the medial cuneiform and intermediate cuneiform), and the medial tarsometatarsal joint (between the medial cuneiform and base of the first metatarsal). The first ray produces motion at the tarsometatarsal joint, the intercuneiform joint, and the medial cuneonavicular joint. Each of these joints move in unison through the sagittal plane to "function as a single combined unit or ray" (Roukis *et al.*, 1996). The axis of rotation of the first ray lies obliquely to the sagittal plane. This axis ranges from the proximal-plantar-medial margin of the navicular tuberosity to the distal-dorsal-lateral aspect of the base of the third cuneiform. (Figure 6).

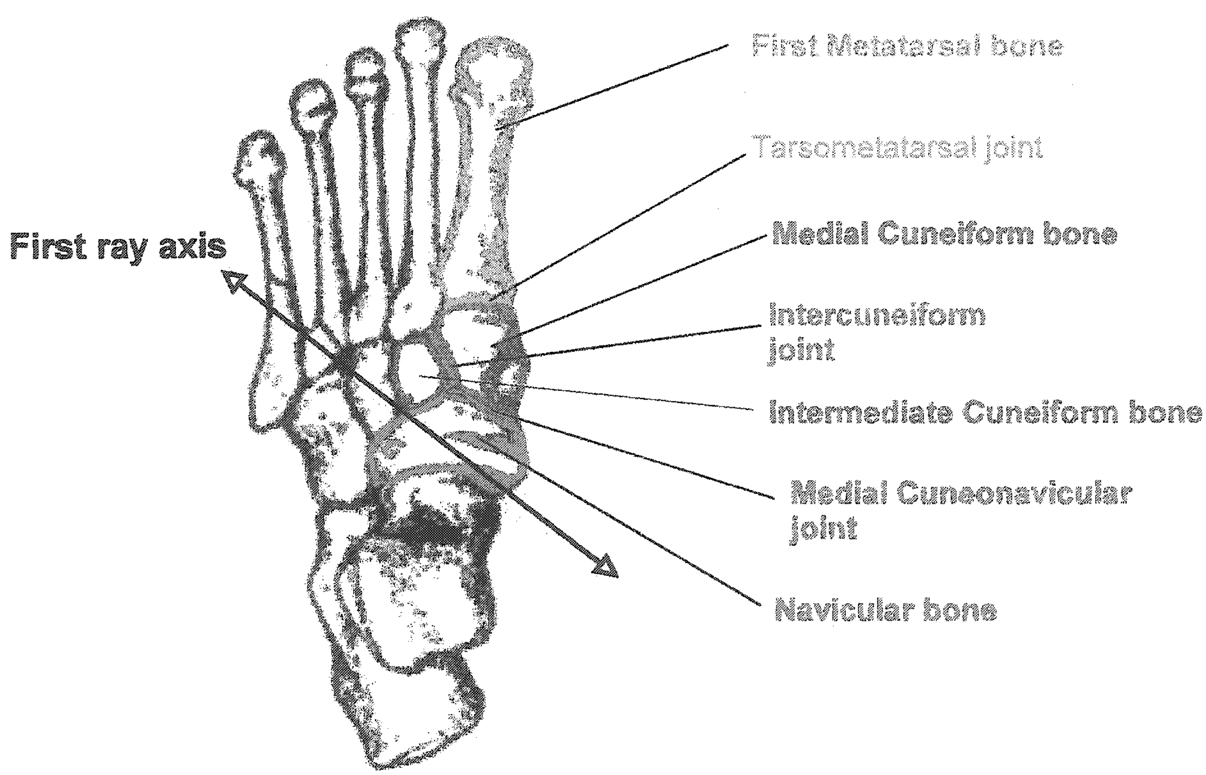

Figure 6: The oblique axis in which sagittal plane motion occurs. Adapted from Roukis *et al.*, (1996).

The orientation of the first ray prior to heel-lift during late stance can influence the ability of the first MTP joint to dorsiflex (Roukis, 1996). From midstance to heel-lift, the foot is in a state of pronation. Subtalar pronation involves slight dorsiflexion (lowered medial longitudinal arch) and first ray eversion. At heel-off, the first MTP joint begins to dorsiflex. Normal dorsiflexion of the first MTP continues until toe-off. Normal dorsiflexion of first MTP during late stance produces a "Windlass effect" which results in a return to the normal medial longitudinal arch height, and consequently foot supination. Pronation and subsequent foot resupination from late stance to toe-off results in a cushioning effect. This lowering (pronation) and rising of the arch (supination) is a spring-like action that can be partially attributed to shock attenuation of ground reaction forces experienced during each step (Root, 1977; Hicks, 1953; 1954).

If the foot is in a state of over-pronation during late midstance to heel-off, first MTP dorsiflexion is diminished (Root, 1977; Hicks, 1953; 1954; Dananberg, 1993; 1986; Dananberg *et al.*, 1996; Drago *et al.*, 1984; Roukis *et al.*, 1996). The normal range of motion of the first MTP joint (independent of the first ray) during stance is 65 to 75 degrees (Figure 7).

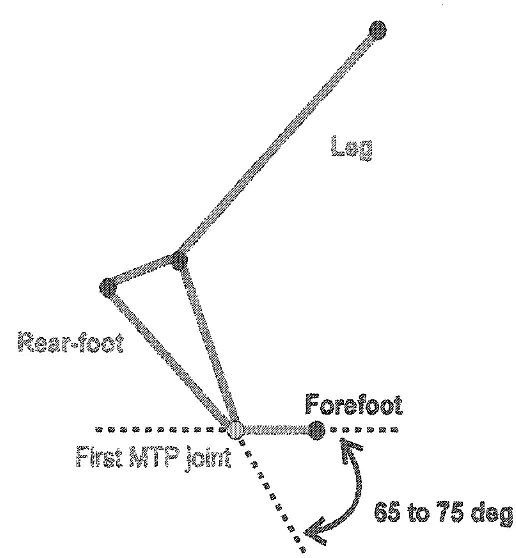

Figure 7: Normal range of first MTP joint dorsiflexion during late stance

2.2 Definition of FHL

Uninterrupted first MTP joint motion during late stance is conducive to "a proper environment to maintain adequate joint nutrition" (Dananberg *et al.*, 1996: p. 71). FHL is the inability of the first

MTP joint to dorsiflex normally during the stance phase of gait. FHL can be either a lack or delay of first MTP joint dorsiflexion from late stance to toe-off (Root, 1977; Hicks, 1953; 1954; Dananberg, 1993; 1986; Dananberg *et al.*, 1996; Drago *et al.*, 1984; Roukis *et al.*, 1996).

2.3 Etiology of FHL

Hallux Limitus can be structural or functional. Structural limitations result in reduced hallux dorsiflexion during both weight-bearing and non-weight-bearing conditions (Drago *et al.* 1984). FHL is apparent during weight-bearing. An individual who has a functionally limited hallux is able to achieve full dorsiflexion range of motion during non-weight-bearing.

Disagreement exists amongst the podiatric community regarding the principal cause(s) of FHL. Nilsonne (1930, as cited in Drago *et al.* 1984) attributed FHL to an elongated first metatarsal as well as flatfeet (pes planus). Lambrinudi (1938), and Kessel, and Bonney (1958, as cited in Drago *et al.* 1984) associated FHL with metatarsus primus elevatus. Jack (1940) and Bingold and Collins (1950, as cited in Drago *et al.* 1984) proposed that recurring minor injuries due to hallux hypermobility was the major causative factor of FHL. Goodfellow (1965; and McMaster, 1978, as cited in Drago *et al.* 1984) claimed that osteochondritis dissecans was the major causative factor of FHL. Lapidus (1940, as cited in Drago *et al.* 1984) related Iatrogenic and neuromuscular disorders resulting in tibialis anterior hyperactivity or peroneus longus weakness which in turn resulted in metatarsus primus elevatus. In 1977, Root *et al.* identified the causes of FHL as: 1) hypermobility, 2) immobilization, 3) elongated first metatarsal, 4) metatarsus primus elevatus, 5) primary osteoarthritis, 6) acute tauma, 7) osteochondritis dissecans, 8) gout, 9) rheumatoid arthritis. Drago *et al.*, (1984) attributed FHL development to a number of conditions such as: metatarsus primus elevatus, hypermobility of the first ray, long first metatarsal, trauma, osteochondritis dissecans, immobilization, gouty arthritis, osteoarthritis, rheumatoid arthritis, neuromuscular disorders, and iatrogenic disorders.

More recently, the following conditions have been suggested as etiologic factors directly responsible for decreasing motion at the first MTP joint: "1) heredity, 2) primary or secondary osteoarthritis, 3) osteochondritis dissecans, 4) avascular necrosis, 5) various arthritides (eg, rheumatoid, psoriatic, gouty and septic), 6) neuromuscular disorders resulting in either weakness of peroneus longus or spasticity of tibialis anterior, 7) acute trauma, 8) iatrogenic causes related to previous surgery of the first metatarsal or first MTP joint, 9) congenital fragmentation of the epiphysis in the base of the proximal phalanx of the hallux" (Roukis et al, 1996). Persons experiencing decreased motion of the first MTP joint may possess one or more of these deficiencies.

2.4 Modern Theory of FHL Etiology

Dananberg *et al.* (1996) stated that the cause of FHL can be attributed to physiological changes of the first MTP joint and biomechanical deficiencies of the foot. Physiological changes include neurogenic inflamation of the joint. Biomechanical deficiencies of the foot involve improper first MTP joint motion resulting in compensatory actions through the oblique axis of the midtarsal (sub-talar) joints.

2.4.1 Physiological Changes

The classic view of FHL has been described simply as unavoidable wear and tear of the first MTP joint. Reduced joint function is a natural progression of age and lifestyle (Dananberg *et al.*, 1996). Dananberg attributed this wear and tear to articular (hyaline) cartilage degeneration at the distal end of the first metatarsal bone and the proximal end of the first proximal phalange. The major error associated with this view is that hyaline cartilage does not have pain receptors (nociceptors). However, there are nociceptors (pain receptors) in the joint capsule and surrounding ligaments. The role of these receptors is to provide feedback when a joint moves beyond its normal range of motion.

The joint tolerance can be reduced over time; thereby reducing the range of motion that facilitates a pain response. Three known situations can reduce this threshold: joint immobility, nerve compression, and muscle hypertonus (Dananberg *et al.*, 1996).

2.4.2 Biomechanics of the foot

Human gait has been described as an inverted pendulum moving in the sagittal plane (Winter, 1995). The mass of an individual (lever) is moved through the sagittal plane at the first MTP (Figure 8). Motion of the swinging limb provides the energy necessary to move the body mass over the pivot point (first MTP joint). The first MTP joint on the supporting foot is relatively motionless in the antero-posterior direction. From heel-lift to toe-off, the first MTP joint dorsiflexes while loaded (i.e. the joint accepts the body's weight). The first MTP joint continues to dorsiflex at contralateral heel-strike, as the ipsolateral knee begins to flex. At this point the body's centre of mass passes over the weightbearing foot. Load on the weightbearing foot peaks between 1.4 and 1.7 times body weight (Dananberg, 1995). During mass acceptance, the first metatarsal is relatively perpendicular in relation to the ground. At toe-off, 65 degrees of dorsiflexion is achieved by the first MTP joint (Root, 1977; Hicks, 1953; 1954; Dananberg, 1993; 1986; Dananberg *et al.*, 1996; Drago *et al.*, 1984; Roukis *et al.*, 1996). Contralateral heel-strike occurs at this maximum range of first MTP joint dorsiflexion. Contralateral heel-strike rapidly reduces the total weight experienced by the first MTP joint.

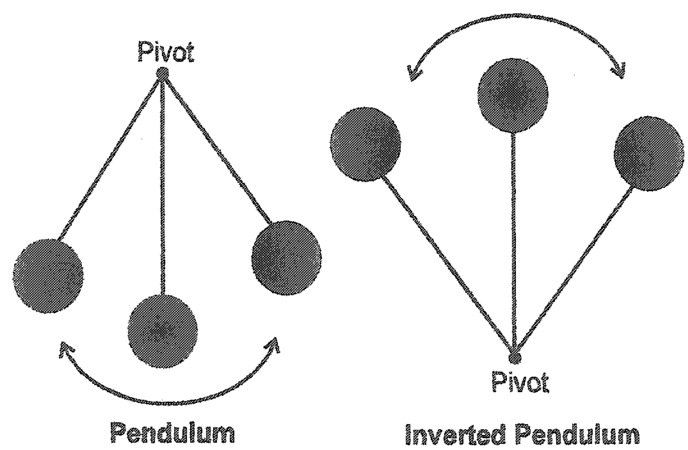

Figure 8: The inverted pendulum model.

The first MTP joint inability to dorsiflex at heel-lift can cause a compensatory action through the oblique axis of the midtarsal (subtalar) joints. The compensatory motion occurs between the talus and the navicular, and the calcaneus and the cuboid. The compensation expresses itself as a lengthening of the foot, and consequential drop of the longitudinal arch (Figure 9). This motion is otherwise known as overpronation of the foot during midstance of single support (Root, 1977; Hicks, 1953; 1954; Dananberg, 1993; 1986; Dananberg *et al.*, 1996; Drago *et al.*, 1984; Roukis *et al.*, 1996). In turn, compression of the first MTP joint capsule occurs (Dananberg, 1993; Dananberg *et al.*, 1996; Roukis, 1996; Drago *et al.* 1984). The first MTP joint capsule is designed to achieve compression at the range of motion end. Joint capsule compression in turn inhibits the ability of the first MTP joint to achieve further dorsiflexion.

Figure 9: Skeletal feet during stance: a) Normally pronated foot, b) overpronated foot. Adapted from Dananberg (1995).

The first MTP joint is subjected to unusual stress if proper range of motion is not achieved as the centre of mass advances during stance. Stress during late stance can result in joint capsule compression. With repetitive compression up to 2000 to 3000 times a day, capsule damage can occur resulting in reduced joint function (Dananberg, 1993; Dananberg *et al.*, 1996,). Increased first MTP joint capsule pressure during gait may express itself as pain. In reaction to this pain, surrounding musculature splints the joint, causing joint stiffness (Dananberg, 1993; Dananberg *et al.*, 1996). This progression shows that the development of FHL is a neurologic reaction to biomechanical deficiencies of the foot.

3.0 Review of Literature

The literature reviewed for this study can be categorized into the general areas of: 1) plantar pressure studies of the foot, 2) kinematic analysis of the trunk during normal human gait, 3) kinetic and kinematic analyses of footwear, foot orthoses and other devices, and 4) kinetic and kinematic analyses of the first ray and metatarsals.

3.1 Plantar pressure studies of the foot

Past plantar pressure studies have been conducted to investigate both normal and pathological gait patterns. Researchers described normal-footed subjects as those who do not experience pain during gait, have not experienced recent injury to the lower extremity, have not experienced surgery to either limb, have no obvious foot deformities (Akhagli et al., 1994: 64), have no hereditary or congenital foot defects requiring professional treatments, nor other problems resulting in pathological gait patterns (Elftman, 1934; Bennett and Duplock, 1993; McPoil *et al.*, 1995; Hughes et al., 1991; Atkinson - Smith and Betts, 1992; and Morag and Cavanagh, 1997).

The first study to describe normal plantar pressures was conducted by Beeley in 1882 (Elftman, 1934). This rudimentary study collected data as subjects stepped into a thin sack filled with plaster of paris. Beeley suggested that the deepest indentations indicated the areas of greatest weight concentration. Future studies using similar methods were criticized, because these methods measured mainly the shape of the foot rather than pressures (Elftman, 1934). In later years, some researchers adopted methods that resembled large rubber ink stamp pads onto which subjects stepped (Elftman, 1934).

In 1935, Elftman evaluated a pressure plate system previously designed by Balser in 1927. The apparatus incorporated a pyramidal studded rubber mat placed on thick glass plate. The studded surface contacted the glass plate with the smooth side facing up. The glass plate was mounted on a suspended frame under which was placed a 16 mm cinecamera. The camera was run at 72 frames per second. When stepped on, the studs in contact with the glass would deform. The film was played back at 16 frames per second for analysis.

Elftman found that pressure at the heel decreased after heel-strike while increasing at the ball of the foot. Pressure at the heel was non-existent at heel-lift as pressure at the ball of the foot reached maximum. Pressure at the toes increased while decreasing at the ball of the foot. Finally pressure at the toes terminated as the contralateral heel struck the ground.

In 1978, Nocol and Hennig tested an apparatus designed to record plantar pressure. The purpose of the study was to design a relatively large pressure sensing mat that subjects could hit with greater ease. The larger pressure sensing mat was produced by assembling "six elementary mats" (Nicol and Hennig, 1978: p. 375). Data from the enlarged mat were recorded and stored using a video tape recorder.

The researchers chose the broad jump, hopping on ground and bouncing on a trampoline to be analysed using the new system. The system was able to show "a good impression of the force distribution and its change during take-off and landing" (Nocol and Hennig, 1978: p. 378).

In 1991 Hughes et al. investigated the reliability of a pressure platform (EMED F system). This version of the EMED F system used 1344 transducers (2 per cm^2) within a 34 x 20 cm sheet. The sensor sheet was placed level to the surrounding walkway used for data collection. The purpose of this study was to collect peak pressure information while 10 normal subjects walking at three defined velocities. The study examined peak pressure at the heel, each metatarsal head, each toe, and the base of the 5^{th} metatarsal.

The study concluded that peak pressure increased linearly as walking velocity increased. However, increases in peak pressure were not uniform for the entire plantar surface of the foot. While the EMED F system reliability increased as the number of trials increased, valid single recordings were infrequent.

The major limitation of the methodology used by Hughes et al. (1991) was the inability of the system to be placed within the shoe. The use of non-in-shoe pressure transducers seemed inappropriate for the current study. Such a experimental design may measure interaction between footwear and the ground rather than foot and CFO. Since CFOs were worn within a shoe, it would be more appropriate to place a plantar pressure transducer between the foot and the CFO. Non-in-shoe systems may not possessed the sensitivity to detect differences between the conditions of NKW and KW.

Atkinson - Smith and Betts (1992) attempted to draw relationships between footprints, plantar pressure distributions, rearfoot motion and foot function during running. Nine subjects with normal to high arches (pes cavus) were chosen for the study. The area of each midfoot was evaluated using a simple footprint (Atkinson - Smith and Betts, 1992: p. 149). The areas of each foot were compared to plantar pressure collected using a pressure platform (pedobarograph). Pedobarograph data were collected during barefoot walking trials. The centre of pressure was expressed relative to the centre line of the foot.

Subjects were filmed (50 fps) in the frontal plane while running on a treadmill. Images from cinefilm were used for kinematic data analysis. Tibiocalcaneal angles were measured in each frame during running stance.

The researchers found that low midfoot areas (consistent with a high arched subjects) created high pressures on the lateral aspect of the foot. The pressure peaks consistently continued on the lateral side of the foot from midfoot to the fifth metatarsal. These two factors were also consistent with a relatively low (below the normal range) tibiocalcaneal range of motion during stance.

A limitation of this study was that walking data were used to make inferences regarding subjects' running gait. While the study did fulfill its purpose to draw correlations measuring the three indicators, it still lacked the use of an in-shoe pressure sensing system.

Bennett and Duplock (1993) conducted a descriptive study of plantar pressures. The purpose of the investigation was to collect plantar pressure characteristics of normal, healthy adults. The investigators used a pressure sensing system (Musgrave Footprint) to collect the data. Data collected for each subject were represented as the mean of three good trials.

The investigators found that "it was reasonable to presume that the first ray relative to the lesser metatarsals play a more functional role in the distribution of pressure beneath the foot" (Bennett and Duplock, 1993: p. 676). The researchers speculated that plantar pressure distribution is facilitated by the available first ray range of motion. The study was able to provide useful information regarding plantar pressure distributions.

In 1997, Morag and Cavanagh conducted a study to identify structural and functional factors to predict plantar pressures during walking gait. The purpose of the study was to develop a model to predict variances in peak plantar pressure based on those structural and functional factors. The study used 55 healthy subjects free of any known disorders of the lower extremity.

The researchers collected electromyographic data from four muscles in the lower extremity. Subjects were filmed while walking. Plantar pressure data were recorded using the EMED - SF2 system (Novel Inc. Minneapolis MN). "[S]ignificant predictors of regional peak plantar pressures were identified by multiple regression" (Morag and Cavanagh, 1997: p. 362). The authors separately discussed the results for each region of the foot (heel, midfoot, MTH1 [metatarsal head 1], and hallux). In addition, the authors divide stance phase into four quarters for discussion.

According to their model, the authors found that an increase in gastrocnemius activity during the third quarter was consistent with an increase in plantar pressure under MTH1. The study was able to show a relationship between dynamic talocrural joint range of motion and plantar pressure at the MTH1. This "demonstrate that subjects who tend to walk with a larger dynamic ROM at the talocrural joint from plantar - to dorsiflexion experienced higher pressures on MTH1" (Morag and Cavanagh, 1997: p. 367). The study supported the notion that FHL causes compensatory motion (overpronation) at the talocrural joint during late midstance (Dananberg, 1993; Dananberg *et al.*, 1996; Drago *et al.* 1984). Another significant finding from the study was that limited first MTP joint dorsiflexion resulted in excessive loading under the joint.

For each of these aforementioned studies, the investigators used pressure platform systems. Such non-in-shoe systems can require targeting by subjects (McPoil *et al.*, 1995). Rather than walking in a genuine manner, gait style could have altered in attempted to consistently step on the platforms. Variability in step length and cadence can result as subjects attempt to consistently step on platforms.

As technology improved, so did the ability to measure plantar pressure. Manufacturers developed instruments to measure plantar pressures between the foot-shoe interface. One of the first studies to produce and test an in-shoe plantar pressure sensing system was conducted by Hennig *et al.* in 1980. The system was composed of 512 piezoelectric transducers within an insole. The authors chose to use piezoceramics because of "their high charge output, low noise and disturbance characteristics" (Hennig *et al.*, 1980: p. 120).

The investigators found the apparatus showed "good linearity (< 2%) and hysteresis (< 1%)" (Hennig *et al.*, 1980: p. 120). The researchers also found a high signal-to-noise ratio. With the use of piezoelectric technology, relatively high spacial resolutions were possible (the system was able to use transducers as small as 1 mm x 1 mm). While the system possessed desirable characteristics, the authors stressed the need to produce a more suitable imaging program. The program used at the time was limited to reproducing and displaying single images. The authors required software capable of displaying plantar pressure data histories at a rate of 60 images per second.

In 1994, Akhagli *et al.* conducted a study to determine peak pressure patterns during normal walking using an in-shoe pressure sensing system (Gaitscan). Normal subjects were described as those who had no obvious foot deformities.

The Gaitscan system was comprised of eight separate piezoelectric pressure sensing transducers on an insole. Transducers were placed at "clinically relevant loadbearing areas of the foot" (Akhagli *et al.*, 1994: p. 63). The authors placed the transducers at: "the five metatarsal heads, the center of the hallux, the base of the fifth, metatarsal, and the center of the heel" (Akhagli *et al.*, 1994: p. 64). Each site was identified using palpation techniques. Transducers were fixed to insoles constructed of leather and rubberized cork. The authors found that as "pressure under the first metatarsal head increased, that under the fifth decreased and visa versa" (Akhagli *et al.*, 1994: p. 65). Using this relationship, the authors suggest that plantar pressure varied from medial to lateral. However, the authors did not offer insight as to when this trend occurred.

The authors state that insole firmness "caused neither discomfort nor alteration of gait" (Akhagli *et al.*, 1994: p. 64). However, the rubberized cork insole could have acted as a shock absorber, thereby altering plantar pressure data.

Systems that use individual sensors at predetermined locations are referred to as "discrete systems" (McPoil *et al.*, 1995: 95). The major limitation of discrete systems is the possibility of sensors moving as a result of shear forces between foot and insole. Secondary to this reason is the possibility that displaced sensors "act as a foreign body in the shoe," thereby attenuating plantar pressure measurements (McPoil *et al.*, 1995: 95).

In 1995 McPoil *et al.* conducted a study to compare performance between the EMED Pedar (Novel) and F-Scan version 3.622 (Tekscan Inc.) systems. The calibration protocol for the F-Scan required subjects to stand on the insoles. However, both systems were calibrated using the EMED Pedar calibration tool (a pressurized chamber).

Data were collected from both systems while subjects walked on a treadmill. The study used four healthy male subjects with no history of foot or ankle problems that resulted in pathological gait patterns. Each subject used the same model of shoe during data collection. The researchers found that the EMED system was capable of validly and reliably measuring pressure and normal force. However, the researchers critique the ability of the F-Scan to measure the same parameters.

Woodburn and Helliwell (1996) conducted a study to evaluate the F-Scan (Tekscan Inc.) system. The F-Scan was evaluated based on physical characteristics, accuracy and repeatability. The authors evaluated the calibration technique as described by the Tekscan company. This method involves static loading the insole with the subject's body mass. The authors state that the "calibration procedure consistently overestimated body mass across a normal range by 4%, with individual error as high as 14%, the greatest error being in the 75 - 90 kg range" (Woodburn, and Helliwell, 1996: p. 302). This is inaccuracy may be due to the fact that the F-Scan measures normal forces at each sensor. The overestimating error may be due to non-linearity of the sensors at higher forces. The overestimating error encountered by Woodburn and Helliwell (1996) should have no significant effects on the current study since subjects were compared to their own scores in each testing condition..

Like previous studies, the authors praised the spatial resolution of the F-Scan system (four pressure sensing cells per cm^2). The authors stated that such a high resolution enables the detection of plantar pressure differences between individual metatarsal heads. The authors stated that the "[s]ampling rate (1 - 100 Hz) was adequate for most clinical applications" (Halliday *et al.*, 1993: p. 302). Furthermore, the F-Scan insole was thin enough to be "unobtrusive in the shoe" (Halliday *et al.*, 1993: p. 302). However, this thinness reduced durability. The researchers recommend the use of an EVA (ethylene vinyl acetate) material to increase durability. However, this may have altered plantar pressure measurements because EVA is used as a cushioning component in athletic footwear.

In 1997, Rash *et al.* compared the PEDAR (Novel Inc.) and F-Scan (Tekscan Inc.) in-shoe pressure sensor systems. Previous investigations found the F-Scan system to be inferior. However, this study used an up-to-date F-Scan system with newer insoles and software that allowed its calibration with an air pressure bladder. The investigators addressed four questions: "(a) Which system measures known pressures more accurately? (b) Which system is more repeatable? (c) Which system has lower measurement variance across a uniform distribution? (d) Is F-Scan repeatability improved with re-calibration?" (Rash *et al.*, 1997: p. 1).

Each day both insoles were calibrated using a Novel air bladder. The researchers stated that this was an appropriate tool because the Novel bladder can generate and apply predetermined pressures. In addition, the Novel air bladder could contain both a Novel and a F-scan insole at the same time.

The Novel air bladder was also used to test both systems. Data collection was conducted over three separate days. The Novel air bladder applied pressures up to 500 kPa and down to 100 kPa at 100 kPa intervals to both insoles. Data from each insole were collected at each 100 kPa interval.

The investigators found that there "were no significant differences in the average absolute pressure difference between the two systems" (Rash *et al.*, 1997: p. 2). The investigators note that use of the F-Scan may be beneficial to ascertain trends across days "if insoles and calibrations initially used by the patient are used again on the second or third visit" (Rash *et al.*, 1997: p. 2).

Despite the relatively higher variance of the F-Scan plantar pressure system, past authors stated a few advantages. The first is its substantially lower price compared to other in-shoe plantar pressure systems. The second is its thickness; the F-Scan is considerably thinner (0.15 mm) than other in-shoe plantar pressure systems. Thin sensor insoles reduce altered footwear fit. (Rose *et al.*, 1992; Woodburn, and Helliwell, 1996). More importantly, each of the authors recognized the greater spacial resolution of the F-Scan plantar pressure system compared to the PEDAR.

3.2 Kinematic analysis of the trunk during "normal" human gait

Several studies have been conducted to quantify the kinematic characteristics of human gait. Some studies were devoted to the lower extremity, others to total body, and others to the upper extremity and trunk during gait (Carlson et al., 1988; Cromwell et al., 2001; Gill et al., 2001; Gilchrist and Winter, 1997; Grasso et al., 2000; Prince et al., 1994; Winter, 1991; 1995; 1995; Winter et al., 1993; Öberg et al., 1993; 1994; and Krebs et al., 1992; Kubo et al., 1997).

Few studies have offered descriptions of the kinematic characteristics of the head-arms-trunk (HAT) segment during walking and running. A kinematic study of normal subjects by Carlson et al.(1988) found that angular displacement of the trunk segment in the sagittal plane was 4 degrees from the vertical (y) axis. In 1993, Winter et al. developed an integrated

EMG/biomechanical model of the upper body during gait. The study was consistent with previous findings regarding angular changes of the trunk in the anterior posterior direction (sagittal plane). Winter et al. (1993) found that trunk angle varied 2 degrees from the vertical axis during the gait cycle.

Prince et al., (1994) investigated both the kinematic and electromyography characteristics of healthy male subjects while walking with and without arm swinging. While this study described the linear accelerations of the head and hip, they did not offer insight into the angular position of the trunk segment during walking.

Similar to the current study, the previous studies utilized video data collection techniques to calculate trunk kinematics. These studies were of value to the current study since they provided normative parameters for trunk kinematics; namely trunk angles during the stance phase of walking gait. While these studies are valuable resources, they do not offer insight to the kinematic patterns of individuals diagnosed with FHL.

Case studies by podiatrists using similar testing designs have noted changes in gait kinematics. According to Dananberg (1995), individuals with CFO intervention experienced significant reductions in low back pain. Dananberg also found improvements in hip extension for subjects using CFOs modified to manage FHL. One subject was tested using a quasi-experimental (A-B-A-B treatment) format to determine whether or not CFO intervention would reduce low back pain. The subject was asked to perform 30 trials with and without modified CFOs (shod). Trials were then repeated using the same format. Data presented in his 1995 paper were of selected trials of one subject. Dananberg claims that a "consistency of hip extension with orthotics is clearly stated" (1995: p. 404).

Generalizations of the study conducted by Dananberg (1995) was limited in that the case study data were from a single subject. Secondly, the author could not use any statistical tools to discern significant differences between the two conditions. In addition, the author chose to represent the data for selected trials separately (i.e. kinematic data were not ensemble averaged).

3.3 Kinetic and Kinematic analyses of footwear, foot orthoses and other devices

As biomechanical analysis techniques became more available, affordable and reliable, the ability to evaluate footwear, foot orthoses and other in-shoe devices improved. Researchers used videographic, goniometric, force plates and in-shoe plantar pressure systems to conduct such studies.

In 1996, Bennett *et al.* investigated the effects of custom-fit foot orthoses on temporal and plantar pressure parameters. The objective of the investigation was to determine if the timing of maximal plantar pressure changed with the use of custom-fit foot orthoses. The investigators collected plantar pressure data at each of the metatarsal heads. The researchers compared plantar pressure between: 1) bare foot, 2) running shoe, and 3) running shoe with custom-fit orthoses (replacing original insole). The custom-fit foot orthoses were constructed with subortholene (4mm) and modified with an EVA heel post. The Musgrave Footprint (W.M. Automation) was used to collect plantar pressure data during barefoot trials. Plantar pressure data for the latter two conditions were collected using the Electrodynogram (Tekscan Inc.).

The foot was divided into the areas of hallux, metatarsal heads, medial midfoot, lateral midfoot and the heel. The time for each of these areas to achieve maximum plantar pressure was calculated and compared between each of the conditions.

The investigators found that the orthosis plus shoe condition reduced maximal plantar pressures on the medial aspect of the foot and increased plantar pressure at the lateral aspect of the midfoot. This finding was questionable because the plantar pressure data were collected using two different systems. The Electrodynogram uses pressure sensing transducers placed at discrete locations on the plantar aspect of the foot. This system is susceptible to error because plantar tissues may move as a result of foot elongation during stance. In contrast, the Musgrave Footprint reported all pressures of the foot, thereby indicating each maximal pressure.

Rose *et al.* (1992) evaluated the F-Scan (Tekscan Inc.) and investigated the effects of shoe modifications on plantar pressure distribution. More precisely, the investigators addressed the reproducibility, durability, variability, and validity of measuring plantar pressure. In addition, the investigators used the F-Scan system to evaluate the effects of heel wedges on center of pressure paths, and plantar pressure distribution.

Eleven normal subjects were used for the study. Subjects were required to perform walking trials without shoe modifications and with four heel wedges. Subjects shoes were equipped with $^1/_4$ inch medial heel wedges, $^1/_4$ inch lateral heel wedges, ½ inch medial heel wedges, and ½ inch lateral heel wedges. Separate new sensors were used for each subject.

The researchers found that insole sensitivity declined over time. After approximately 30 gait cycles, plantar pressure readings decreased by 3.50%. After approximately 80 gait cycles readings decreased by 20.5%. The study also found that removing and replacing the insole affected measurements. The researchers did not offer any insight as to why insole removal and replacement affected measurements.

The study also found that lateral heel wedge modifications displaced plantar centre of pressure medially. This medial displacement was greatest at the midfoot and metatarsal areas. One quarter inch lateral heel wedge, and ½ inch lateral heel wedge displaced plantar centre of pressure 7mm and 10 mm, respectively (compared to the no wedge condition). In addition, medial heel wedge modifications displaced plantar centre of pressure laterally. Like the lateral wedges, medial wedge displacement was greatest at the midfoot and metatarsal areas. Lateral centre of force displacement increased with wedge height. One quarter inch medial heel wedges and ½ inch medial heel wedges displaced plantar centre of pressures 2.50 mm and 5.00 mm respectively (compared to the no wedge condition).

The study by Rose *et al.* provided information regarding the effects of various heel posts (devices added externally to the CFO used to increase medio-lateral stability) on a normal, non-pathological foot. However, the authors did not provide any direct information with regard to foot pathology. The study did not provide information regarding the effects of such shoe modification techniques on patients with foot pain or deformities.

In 1992, Cornwall and McPoil conducted a study to evaluate the ability of custom-fit foot orthoses to reduce forefoot forces. More precisely, the purpose of the study was to "assess the effectiveness of a semirigid foot orthoses with a varus wedge on forefoot vertical forces in a 24-year-old female with a compensated rearfoot varus deformity" (Cornwall and McPoil, 1992: p. 371). The study used a single-subject alternating treatment design. The different treatments were: 1) shoe only, 2) shoe with a semirigid orthoses, 3) shoe only, and 4) shoe with semirigid orthoses modified with a heel post (varus wedge).The EMED (Novel Inc.) insole system was used to collect vertical force data for each of the three conditions while the subject walked on a treadmill. (Cornwall and McPoil, 1992). This particular version of the EMED insole was composed of 85 pressure transducers. Data were sampled at 70 Hz. The insole was placed on the orthoses, under the foot. The researchers evaluated the average impulse at the forefoot for multiple trials (Cornwall and McPoil, 1992).

The researchers found significant differences between the non-orthoses conditions and the orthoses conditions. However, no significant differences were found between the two orthoses conditions. This suggested that the "rearfoot varus wedge did not effectively decrease the forces acting on the forefoot" (Cornwall and McPoil, 1992: p. 374). Therefore, the ability of an orthosis to reduce forefoot forces was not dependent on the rearfoot varus wedge modification. The authors also found that the orthoses decreased impulse at the forefoot.

The authors stated that the orthoses significantly increased the total contact area with the EMED insole. Some custom fit insole are designed to maintain total contact with the entire plantar surface of the foot.

Like previous studies (McPoil *et al.*, 1995) the authors used plantar pressure data to represent vertical force in Newtons. The authors reported this measurement in relation to time (impulse). This is questionable, because plantar pressure systems are not designed to measure ground reaction forces. As stated earlier, "pressure is a scalar, having no directional quantities" (Halliday *et al.*, 1993). Ground reaction force and impulse are vector quantities.

Sanderson and Taunton (1992), investigated the ability of EVA and Thermoplastic custom-fit foot orthoses to control rearfoot motion during running. The study was "an attempt to identify if there are differences that would make one type more suitable than the other" (Sanderson and Taunton, 1992: p. 2).

Thermoplastic foot orthoses were fitted using a non-weight bearing plaster cast method. Positive foot molds were made from the plaster impressions. Orthoses were made from the molds using a vacuum press. Orthoses were then modified with rearposts. The orthoses were designed to fit subjects' footwear. EVA orthoses were fitted with subjects semi-weight bearing, using the Amfit apparatus (Amfit Inc.) EVA orthoses were also designed to fit subjects' footwear.

Ten subjects with custom-fit foot orthoses were requested to run on a treadmill. Each subject needed forefoot and rearfoot realignment. Subjects ran under three conditions 1) shoe without orthoses (control), 2) shoe with EVA custom-fit orthoses, and 3) shoe with thermoplastic custom-fit orthoses. Two reflective markers were placed on the posterior aspect of the heel counter of each subject's right shoe. An additional two markers were placed along the posterior of each subject's right calf. Subjects were filmed at 200 Hz while running on a treadmill. Subjects were filmed for each of the three conditions. Markers were digitized, and tibiocalcaneal angles were calculated. Tibiocalcaneal angles were used to calculate: touchdown angle, heel off angle, maximum rearfoot angle (pronation), peak rearfoot angular velocity, time to maximum rearfoot angle, and time to peak angular velocity.

The researchers found no significant differences between the two orthoses in controlling rearfoot motion. Both orthoses reduced the maximum angle of pronation and the maximum rearfoot angular velocity.

It is the opinion of this researcher that EVA orthoses will lose their ability to control rearfoot motion over longer periods (months). The durometer (resistance of a material to indentation) of EVA is substantially less than that of thermoplastics. This study was valuable as it demonstrates the ability of EVA orthoses versus conventional thermoplastic orthoses to control rearfoot motion during short-term use. The current study will not attempt to investigate differences after prolonged use.

Smart and Robertson (1985) investigated the effects of corrective running orthotic devices (CRODs). The authors investigated kinematic changes at the ankle and knee with and without the use of CRODs. To measure such kinematic changes, the investigators used triplanar electrogoniometers.

The group of subjects was comprised of competitive distance runners. Each subject wore the same model of running shoe. A foot switch was secured to the lateral plantar side of the heel of each runner's shoe. Foot switch signals were used to indicate heel strike during running. Goniometric data at the knee and ankle were collected as subjects ran on a treadmill at 80% of the subjects' average racing speed. A 3 X 2 (repeated measures) analysis of variance was used to determine differences between the CROD and non-CROD conditions.

The authors found that CRODs significantly reduced the amount of subtalar pronation during stance. In addition, CROD use resulted in increased valgus knee angles (genu valgum). The authors described this position as a transfer of frontal plane motion proximally (knee), as motion was reduced distally (ankle). The investigators demonstrated a relatively faster and reliable method to obtain kinematic parameters of running gait. The expediency of this method is relative to the time consuming task of digitizing cinefilm.

A delimitation of this study was the use of goniometers at just the knee and ankle. Because of the size of such devices, measurements of angles achieved at the first MTP joint was not possible.

While Smart and Robertson (1985) found significant kinematic changes at the knee, Nester et al., (2000) did not. Nester *et al.* (2000) investigated possible kinetic and kinematic changes at the knee and hip with and without the use of medially and laterally wedged foot orthoses. The investigators used videographic (infrared) data to obtain kinematic parameters. In conjunction with force plate data, inverse dynamics calculations were conducted to obtain moment force data at the ankle, knee, and hip. Nester *et al.* (2000), found mediolateral changes in ground reaction forces between the two wedges. The medial heel wedge created more lateral ground reaction forces, while the lateral wedge created more medial ground reaction forces. The investigators found "some effects on the moments at the knee" (Nester *et al.*, 2000: p. 2). However, "kinematics of the knee, hip, and pelvis, were minimally affected" (Nester *et al.*, 2000: p. 2 p. 2). Perhaps the inability to find differences between the shod and shod plus orthoses lay within digitizing errors.

The aforementioned studies were conducted to identify kinetic and kinematic differences with and without the use of custom built foot orthoses and/or orthosis modifications. Most of those studies were mainly concerned with kinematic differences at the ankle between conditions. Some studies continued further to investigate hip and knee joint parameters. However, none of these studies investigated kinematic parameters at the forefoot. These studies were not concerned with such data at the metatarsals, phalanges or metatarsophalangeal joints.

3.4 Kinetic and kinematic analyses of the first ray and metatarsals

Biomechanics of the forefoot and metatarsals are not fully understood. Kinematic and kinetic studies of the forefoot and metatarsals have been conducted to better understand forces and ranges of motion during normal gait. These studies proved to be valuable, since pathological gait data can be compared and used to formulate strategies to restore proper foot function.

In 1979, Stokes *et al.* investigated the distribution of vertical forces in the foot. The researchers examined loads on the metatarsophalangeal joints, the metatarsal bones, and selected muscles and ligaments. Six subjects were studied while they walked across a strain-gauged force platform. The force platform was divided into 12 sections, each 144 mm long and 12 mm wide. Multiple walking trials were collected. For each trial an ink impression of the foot was made as subjects stepped on the force platform (longitudinally). The force plate was rotated 90° and trials were repeated (transversely). Longitudinal and transverse ink impressions for both sets of trials were matched based on force platform data. X-rays of each subject's foot were superimposed on the force plate and ink impressions. Markers were placed on the heads and bases of the first and fifth metatarsals. The angles of the forefoot were calculated using cinefilm recorded in the sagittal plane. Angles and force plate data were used to calculate moment force data about the metatarsophalangeal joints.

The investigators estimated the peak force experienced by the first toe (hallux) to be 30% of body weight. They also estimated that the second toe experienced 10% of body weight. They claim that combined with forces experienced by the musculature and plantar aponeurosis, the metatarsophalangeal joints experienced forces equal to body weight. The greatest force was experienced by the first metatarsophalangeal joint. The authors state that the first metatarsal is mechanically strongest. During late stance the toes experience 40% of body weight, mostly at the first toe.

Stokes *et al.* (1979) study was relevant to the current study. The authors speculated that foot disorders that reduced the toe's load-bearing ability resulted in greater bending of the metatarsal bones (compensatory movement). In addition, such compensatory movements "produce greater stresses at the attachments of the metatarsals to the tarsus" (Stokes *et al.*, 1979: p. 590). Such stresses may be responsible for pain in the metatarsals and consequent "progressive changes and deformities in the midfoot joints" (Stokes *et al.*, 1979: p. 590). One such disorder of the foot may be FHL.

In 1988 Stuck *et al.* investigated the forces under the foot of patients diagnosed with hallux rigidus. The purpose of the study was to evaluate the ability of a custom-fit orthoses (modified with a heel post) to alter the weightbearing function of the great toe after a surgical procedure. Each of the subjects underwent interpositional arthroplasty of the first metatarsophalangeal joint. The surgical procedure attempted to increase joint space and reduce bone-on-bone friction with the use of bone-end implants.

Five subjects were chosen for the study. Each subject experienced pain and limited range of motion (less than 10° of dorsiflexion) at the first metatarsophalangeal joint. Each of the patients were unresponsive to shoe modifications and medications.

The Electrodynogram (Langer Biomechanics Group Inc.) was used to collect plantar pressure data for a number of conditions. The tested conditions were: barefoot, wearing foam healing shoes, and wearing foam healing shoes plus custom-fit orthoses. The data collection protocol was repeated at 3, 6, and 9 months after surgery.

Average first MTP joint dorsiflexion prior to surgery was 6.83°. Average first MTP joint dorsiflexion post-surgery was 26.17°. In addition the authors found a noticeable reduction "in the duration of stance of the hallux postoperatively with the use of the functionally posted foot orthoses" (Stuck *et al.*,1988: p. 467). The authors assumed that there was a significant reduction in subtalar pronation. The authors made this claim despite not collecting kinematic data of the tibiocalcaneal angles.

Although the authors did show an improvement in most of the cases, two patients developed metatarsalgia (pain in the forefoot) at the second metatarsal head postoperatively. To address this problem, a sulcus modification to the custom-fit orthoses was used for one subject. This is puzzling, since a sulcus modification is used to decrease the range of motion of the metatarsophalangeal joints. The use of the sulcus modification is contrary to the goals of interpositional arthroplasty of the first MTP joint. The sulcus modification reduces MTP joint dorsiflexion as the rigid portion of the orthoses is extended to the proximal ends of the proximal phalanges of the toes. The rigid portion of most orthoses would extend just proximally to the distal end of the metatarsals.

The authors stated that there were significant increases in first MTP joint dorsiflexion. However, the authors did not state the nature of such range of motion (i.e. passive, active, non-weightbearing, or weightbearing). This question is critical as proper first MTP joint function facilitates the efficiency of the inverted pendulum.

In 1987, Creighton, and Olson investigated the passive and active ranges of first MTP joint ranges of motion for a group of runners. A control group of six subjects who were not experiencing any foot pain was compared to the experimental group of six runners who were diagnosed with chronic plantar fasciitis. Each subject was to lie (supine) with their ankle immobilized at 90°. A modified ankle-foot- orthosis was used to keep the foot at 90°. A "standard international goniometer was used to measure the range of motion at the first MTP joint" (p. 359). Multiple trials of maximal active and passive dorsiflexion and plantar flexion were performed. The mean of three trials per condition were used for data analysis. To reduce variability, the ranges of motion were recorded by one individual.

The authors found significant differences in the ranges of motion. T-statistics showed significant differences between the two groups for passive plantar flexion, passive dorsiflexion, and active dorsiflexion. The group of runners diagnosed with plantar fasciitis exhibited lower ranges for each of the three motions. Variation may have been further reduced by using an automated device (electrogoniometer). Although the investigators tested the active and passive ranges of motion, a dynamic test was not conducted.

In their discussion, the authors raised an important question: "if they did have a predisposed decreased ROM, would ROM exercises or mobilization treatment of the first MTP have prevented disabling plantar fasciitis?" (p. 360). This question is important as the current study aims to evaluate the use of the Kinetic Wedge in restoring first MTP mobility.

In 1988 Buell *et al.* compared a number of static techniques to measure first MTP joint range-of-motion including radiographic measurements. In 1995, Hopson *et al.* compared dynamic measurement techniques to the static techniques investigated by Buell *et al.* (1988) (excluding radiographic measurements).

The two static range of motion techniques examined by Buell *et al.* (1988) included surface landmark and radiographic methods. To measure the range of motion with the landmark technique, the center of rotation of the first MTP was identified and marked at the medial aspect. A line representing the bisection of the first metatarsal was drawn (on the medial aspect) which ended at the centre of rotation mark. A second line was drawn representing the bisection of the proximal phalanx. This second line also connected to the joint center mark. Using the drawn lines, the investigators measured the angle of the first MTP joint for the static conditions of: 1) relaxed hanging (neutral), 2) straight line position (0°), 3) active maximal dorsiflexion, 4) passive maximal dorsiflexion, 5) active plantar flexion, and 6) passive plantar flexion. The straight line position was used as a baseline to measure angles achieved during the other conditions. For each of the measurements the foot was maintained in the subtalar neutral position.

Each condition was repeated and measured using radiographs. Bisections were again drawn between the first proximal phalanx and the first metatarsal on the radiographic images. The bisection lines were used to measure the angles achieved at the first MTP joint for each of the conditions. During passive dorsiflexion and plantar flexion, an examiner wore a lead glove.

Fifty subjects were used (100 feet) to determine normal values for first MTP joint range of motion,. This control group was comprised of individuals who had "no specific foot complaints or clinically evident pathology of the first ray" (p. 444). The second (experimental) group was comprised of pre-operative patients "with specific first ray pathology" (p. 444). 52 patients that represented 69 feet were used for this group.

The researchers found differences in the measured ranges of motion between the radiographic and clinical technique. The researchers found that the pathological group had a significantly lower range of motion than the control group. This trend was apparent for all of the measured first MTP joint positions. Although the measured angles differed between the two techniques, the authors state that either method is reliable and repeatable.

Buell *et al.*'s (1988) study is valuable because it provided information regarding normal range of motion. The authors state that the maximum active dorsiflexion range of motion of the first MTP was 89° according to the radiographs. The authors found the maximum passive dorsiflexion range of motion at the joint was 100°. The ranges of motion for the active and passive dorsiflexion ranges of motion were 90° and 105°, respectively, using the clinical (surface landmark) method.

These measurements represented static measurements rather than dynamic measurements. While these findings are valuable, they fail to represent the ranges of motion experienced by the first MTP joint during gait.

In 1995 Hopson *et al.* conducted a study to measure first MTP joint RoM during gait. The researchers also re-evaluated the clinical method used by Buell *et al.* (1988), as well as three additional static measurements.

The authors evaluated 2 non-weightbearing, 1 partial weight bearing, and 1 weightbearing technique. One of the non-weightbearing techniques was the same technique used by Buell et al. (1988); i.e similar bisection lines were drawn on each subject's foot. The other non-weightbearing technique was a modification of one used by Buell et al. (1988) The technique placed the axis of the goniometer on the dorsal aspect of the first MTP joint rather than on the medial aspect of the foot. For both techniques the foot was placed in the subtalar neutral position.

For the partial weightbearing technique, subjects were seated on a chair with the knee and ankle at 90° angles. Subjects were asked to raise their heel off the floor (plantarflex the ankle) maximally, thereby maximally dorsiflexing the first MTP. The angle achieved was measured using the bisection lines.

The static weightbearing technique was a modified dynamic technique. This method involved having subjects walk barefooted across a 24' long walkway covered with kraft paper "after dipping their feet in mineral oil" (Hopson *et al.*, 1995: p. 200). The footprints in the middle 8' of the paper were used to measure step length. Each subject's foot was marked using the same method by Buell *et al.* (1988). Subjects were asked to reposition their feet on two successive footprints on the paper. They were "then asked to raise the heel of the left foot extending the first metatarsophalangeal joint as far as possible while maintaining step length and hallux contact with the floor" (p. 201). Using a goniometer, the angle of the first MTP joint was measured.

The authors evaluated one dynamic range of motion technique. This technique was designed to measure the dynamic range of motion of the first MTP joint during walking gait. This method involved filming the foot at 60 frames per second. Markers were placed at the most distal point on the hallux, the joint center, and the most proximal point on the first metatarsal. Markers were digitized using the Peak Performance software (Peak Performance Inc.) for kinematic analysis.

The authors found significant differences between the all five measurement techniques. However, the authors did not specify which method was the most accurate. The authors do state that "once a specific measurement technique is selected to determine first metatarsophalangeal joint extension, the same technique must be used for reassessment" (p. 203). Their reasoning for their position was because different measures were obtained using different techniques.

The other major finding was that the amount of dorsiflexion achieved by the first MTP joint during gait was less than that achieved during any of the static measurements. Therefore, the authors suggest the dynamic range of motion required by the first MTP joint during gait is significantly less than static measurements. The authors identify this maximal range as 65° of first MTP joint dorsiflexion. This finding is congruent with findings of others authors (Dananberg *et al.*, 1996; Drago *et al.* 1984).

This study is of value because it provided information regarding the range of motion required by the foot during walking gait. However, those experimental protocols required subjects to walk while barefooted. The aim of the present study is to evaluate the effectiveness of custom foot orthoses with a Kinetic Wedge modification. The efficacy of foot orthoses to improve foot function requires it to be placed within the patients shoe. The methods used by Buell *et al.* (1988), and Hopson *et al.* (1995) cannot be used since bisection lines cannot be viewed while inside a shoe.

The methodologies used by Buell *et al.* (1988),and Hopson *et al.* (1995) specified the importance of placing the foot in the sub-talar neutral position. The authors did not justify this method. However, past authors indicate the ability of the first MTP joint to dorsiflex in inhibited when the foot is in an overpronated position. This position involves dorsiflexion of the first ray and eversion at the talocrural joint (Root 1977; 1954; Dananberg, 1993; 1986; Drago *et al.* 1984; Roukis *et al.*, 1996). While studies by Root 1977; 1954; Dananberg, 1993; 1986; Drago *et al.* 1984 and Roukis *et al.*, 1996 endorsed a relationship between first ray position and first MTP joint function, their studies failed to provide suitable empirical data to support their position.

In 1996, Roukis *et al.* conducted a study to investigate the ability of the first metatarsophalangeal joint to dorsiflex, when the first ray was dorsiflexed. The authors justified their study because

> "although the medical literature supports the theory that the position of the first ray has an effect on motion of the first metatarsophalangeal joint, there exists no known quantitative documentation of this effect" (Roukis *et al.*, 1996: p. 539).

Ten subjects (20 feet) were chosen for the study. Normal subjects were required to meet a number of conditions similar to those discussed previously. To measure first MTP joint dorsiflexion, a weightbearing goniometer was constructed. The goniometer was composed of two pieces of wood articulated at a brass hinge. The head of the first metatarsal was placed on the proximal end of the goniometer. The hallux was placed on the distal end. The first MTP joint was placed on the brass hinge. The goniometer allowed the first MTP joint to be passively dorsiflexed. The hallux was dorsiflexed until resistance was felt by the experimenter. The experimenters measured the magnitude of first MTP joint dorsiflexion with the first ray placed in: weightbearing resting position, dorsiflexed 4 mm, and 8 mm from the weightbearing resting position. To create the dorsiflexed first ray positions, 4 mm acrylic platforms were placed under the first ray.

The researchers found a "proportionate decrease of the first metatarsophalageal joint dorsiflexion that occurs with dorsiflexion of the first ray" (Roukis *et al.*, 1996: p. 541). With 4 mm first ray dorsiflexion, MTP joint dorsiflexion decreased by 19.0% compared to the resting position. With 8 mm first ray dorsiflexion, MTP joint dorsiflexion decreased by 34.7% compared to the resting position.

The authors suggest that the reduced ability of the first MTP joint to dorsiflex (as a result of dorsiflexion of the first ray) is a predominant factor resulting in the development of hallux rigidus and hallux abducto valgus deformities. The authors suggested that methods to influence the position of the first ray should be considered when addressing pathological first MTP joint conditions. The authors suggest the need to prescribe "functional foot orthoses to control abnormal biomechanical foot function...These considerations warrant further investigation" (Roukis *et al.*, 1996: p. 544).

Numerous studies were conducted to investigate kinetic and kinematic foot parameters. Many of these studies identified plantar pressure patterns during gait. With the availability of new technologies, researchers were able to measure plantar pressures during shod conditions rather than barefooted walking. Some of these studies were also concerned with kinetic and kinematic differences with and without the use of custom foot orthoses. These studies were limited to the investigation of the hip, knee and ankle. Researchers were unable to investigate the kinematic characteristics of the distal joints of the foot during shod conditions. Studies designed to measure the range of motion of the first MTP joint were limited to barefooted conditions.

A study by Roukis *et al.* (1996) was conducted to support hypothesis that first MTP joint RoM is dependent on the position of the first ray. This study measured dorsiflexion potential while subjects were static. Since data were collected while subjects were barefooted, there was still a need to investigate the dynamic range and pattern (during walking) of the first MTP joint motion during shod conditions. The purpose of the current study was to assess the effectiveness of a custom foot orthoses modified with a Kinetic Wedge. Custom orthoses are designed to place the foot in the sub-talar neutral position, thereby inverting and plantarflexing the first ray. The Kinetic Wedge works in conjunction with custom orthoses. By further plantarflexing the metatarsal of the first ray, the kinetic wedge theoretically restores normal first MTP joint function.

4.0 Methodology

4.1 Subjects

The subjects for this study were of individuals who were diagnosed with moderate to severe FHL. Fifteen subjects (9 females, 6 males) were recruited from the Total Foot Care Clinic (The Rehabilitation Centre, Ottawa Hospital), and the University of Ottawa. The average age, height, and weight of the female subjects were 29 yrs, 162.4 cm, 63.3 kg. The average age of the height, and weight of the male subjects were 40 yrs, 180.4 cm, 87.9 kg. Five of the female subjects were diagnosed with moderate FHL, and 4 of the female subjects were diagnosed with severe FHL. Three of the male subjects were diagnosed with moderate FHL, and 3 of the male subjects were diagnosed with severe FHL.

4.1.1 Selection Criteria

Potential subjects were examined by a chiropodist at the Total Foot Care Clinic of The Rehabilitation Centre. Potential subjects were excluded if they had any current injuries to the lower extremity joints, or experiencing intolerable pain during gait because walking was a requirement by each subject for data collection.

In addition, subjects must not have had chronic neurological or musculoskeletal conditions associated with the foot and ankle, thereby altering walking gait. All subjects must have exhibited normal ranges of motion for each of the foot joints. Furthermore, subjects must not have had any recent history of injury and/or surgery to the lower extremity.

4.2 Clinical Test for FHL

An initial non-weight bearing test of the first MTP joint range of motion was necessary. This test involved holding the foot and first ray in the "neutral" position (the calcaneus was placed perpendicular to the floor). Next, the hallux was dorsiflexed about the first metatarsal. A joint without limitation should achieve dorsiflexion without crepitus or pain. With the foot in the neutral position, the hallux should be able to dorsiflex "straight back without any varus or valgus rotation"..."Any amount of motion less than 65 degrees [may] indicate the presence of hallux limitus or hallux rigidus..." (Dananberg, 1996:55).

Although normal range of motion may have been observed after the initial non-weight bearing test, a secondary test of the first MTP joint was also conducted. The metatarsal head was palpated on its plantar surface. Once the metatarsal head was established, the head was dorsiflexed to its end range and the hallux was dorsiflexed with the other hand (Dananberg, 1996; Chapman, 1999). "If the first 15 -20 degrees of motion is difficult to achieve, or no range of motion is available, then the presence of functional hallux limitus can be anticipated" (Dananberg, 1996: p. 55).

Each foot of each subject was examined by the chiropodist. For subjects who presented FHL in both feet, the more severely effected foot was chosen for the investigation (for NKW and KW).

4.3 Consent and Ethical Considerations

Upon expressing an interest in participation, each volunteer was required to sign a consent form and information letter. These documents conveyed all the requirements of the testing protocol that are necessary for the collection of valid data (Appendix A and B).

Additional subject information collected for this study included: modifications to each subject's orthoses, shoe models used by each subject, and plantar foot impression data obtained using the Footfax (Amfit Inc.) (see next section).

4.4 Custom Foot Orthoses Construction

The Amfit (Amfit Inc.) CAD/CAM system was used to manufacture all custom foot orthoses. The plantar surface of a patient's foot was digitized using the Footfax device (Amfit Inc.). The digitizing protocol required recipients to be semi-weight-bearing with the foot placed in the subtalar neutral position. Multiple pins rise from the surface of the Footfax. The pins are pushed up under the plantar surface of the foot by a pressurized bladder. Once the pins reach a maximum height they are locked in position. The Amfit software used the pin height information to create a model of the plantar surface of the foot (Figure 10).

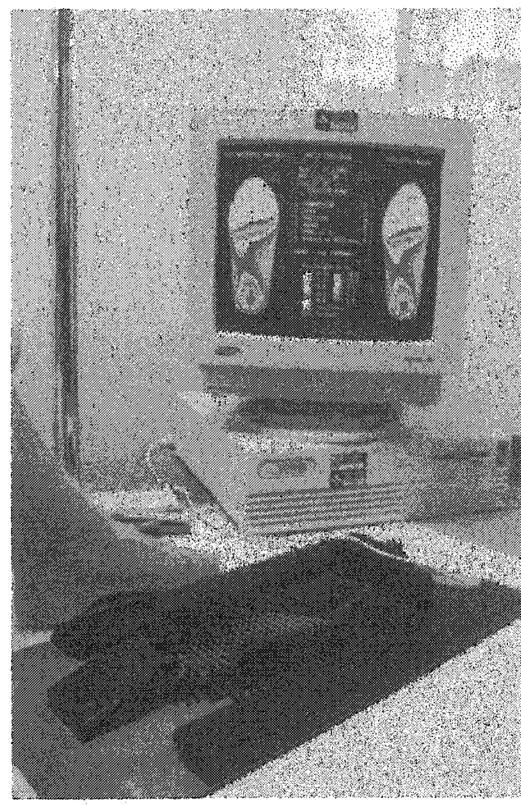

Figure 10: Footfax system. Impression of a left foot

Each subject diagnosed with FHL received a pair of custom foot orthoses. All orthosis modifications except the Kinetic Wedge were completed using the Amfit software. Each AMFIT CFO was milled using 65 durometer EVA. Each pair of orthoses had a 3 degree lateral forefoot wedge, 2 mm forefoot thickness, and 3 mm heel lifts. The arch of each pair of orthosis was reduced by 30%. During manufacturing, Kinetic Wedge modifications were not applied to orthoses. This modification was added just before KW testing.

4.5 Experimental Design

All subjects were tested under the following 2 conditions: with subtalar neutral foot orthoses without the Kinetic Wedge modification (NKW) and subtalar neutral foot orthoses with a Kinetic Wedge modification (KW).

4.6 Testing Protocol / Data Collection

4.6.1 Qualitative Data Collection

After gaining informed consent each subject was asked to complete a numeric pain questionnaire at the end of their initial visit with the chiropodist. The questionnaire was used to indicate subjects' initial state of discomfort (without custom orthoses intervention). The questionnaire used ordinally scaled scores (0-10) to express their level of pain (Appendix C).

Subjects were asked to complete the questionnaire again two months after KW data collection. Data from this survey were used to qualitatively determine the efficacy of the CFOs modified with the Kinetic Wedge lateral forefoot post to reduce pain at the first MTP joint after two months of use.

If subjects answered no to the first question - Do you experience pain in your first (big) toe joint? - the subsequent three questions did not apply (Appendix C). For those subjects who did answer yes, the last question was used to quantify the severity of perceived pain before and after modified CFO use. If subjects answered no to the first question, the investigator rated their perception of pain as zero. This was relevant during post test inquiries; if they had experienced pain prior to testing, and none after testing, their self selected score was compared to a zero score.

4.6.2 Quantitative Data Collection

Data collection was conducted in the Gait and Motion Analysis Lab (GAMA) at The Rehabilitation Centre. The lab was equipped with a 7 metre walkway. Subjects were required to complete at least 10 walking trials along the walkway for each of the two experimental conditions (NKW and KW). Plantar pressure (F-Scan, Tekscan) as well as joint and segment kinematic data were collected.

Before data collection, subjects had the opportunity to warm-up to prevent injury. This time was also used by subjects to familiarize themselves with the lab surroundings. After the warm-up, subjects were asked to walk along the lab walkway at a comfortable, self-selected pace.

Subjects were asked to provide their own appropriate footwear for testing. Desirable characteristics of such shoes included forefoot flexibility to at least 80 degrees with relative ease. Most walking and running shoes exhibited such traits. Subjects wore their own footwear with CFOs. The CFOs replaced the original shoe insoles. Removing the original insole and replacing it with the CFO reduced the likelihood of a poorly fitted shoe.

4.6.2.1 Kinetic Data Collection

Subjects were required to walk along the walkway wearing F-scan insoles for plantar pressure data collection (Figure 11). Each insole was trimmed to fit the subject's shoe with respect to the centre of the foot. The origin of the insole corresponded to the centre of the foot (Rose *et al.*, 1992). The F-Scan insole was placed between the subject's foot

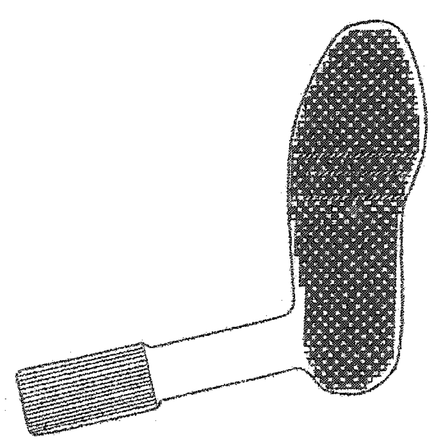

Figure 11: Schemata of an F-Scan insole. The origin is marked red at its center

and the insole of the shoe. In the event of F-Scan insole damage, a new insole was cut using the aforementioned method. F-Scan insoles were also used as footswitches to determine heel-strike and toe-off during walking. To signify the beginning of each trial, an audible tone was generated by the F-Scan software. This short tone prompted subjects to begin walking. At the initiation of each trial, the F-Scan software simultaneously generated a pulse signal. This signal was sent to a light emitting diode (LED) via an RS-232 (serial) port. The LED was used to identify the start of data collection on videotape and to synchronize kinematic data to the kinetic data. F-Scan data were collected at a sampling rate of 120 Hz.

4.6.2.2 Kinematic Data Collection

Kinematic data were collected in the sagittal plane using JVC (DVL - G800U) digital video cameras. Subjects were videotaped at 60 frames per second. Reflective markers were placed at the shoulder, hip, knee, ankle, heel, ball of the foot, and (fifth) toe (from end of shoe). Kinematic data were collected from heel-strike to toe-off (stance-phase) during the gait cycle.

A reflective marker was placed in the foreground of the field to act as a fixed point. To reduce the likelihood of creating false data, the background of the lab was painted black. In addition, subjects were provided a black sleeveless t-shirt. The sleeveless shirt allowed the shoulder marker to be placed on the skin, thereby reducing the likelihood of marker translation. Subjects were responsible for supplying their own black athletic shorts and footwear. Ideal lighting conditions were achieved with the use of a single spotlight mounted beside the video camera. This minimized unwanted shadows and artificial data points.

After all the trials were recorded control points were filmed. The control points and fixed point were used by the *Ariel Performance Analysis System* (*APAS*) and *Biomech for Windows '95* system as a reference. The known coordinates of the control points were digitized and were used to determine the scaling from screen units to actual coordinates. Control points were used as a scaling factor to maintain consistency between frames.

4.6.2.3 Subject testing

Subjects were asked to perform 10 walking trials along the walkway at a comfortable/self-selected cadence for NKW and KW. One step taken by the subject along the middle (3rd and 4th metre) of the walkway was chosen for data analysis. Trials in which the hip marker was visible for the greatest number of frames were chosen for data analysis. Fives trials for each condition, for each subject were selected for data analysis. Thus seventy-five trials were collected for each condition (15 subjects total). A grand total of 150 trials were collected. After completing trials for NKW, subjects' CFO were modified with the kinetic wedge bilaterally, and were allotted 30 minutes practice/accommodation time before KW testing.

4.7 Data Reduction

4.7.1 Kinematic Data

Digital video data for each of the trials were captured using the *Ulead* video capture tool (*Ulead Systems, Inc.*). The video data were then digitized using the *APAS* Digitize tool. *APAS* system then used the previously mentioned reflective markers to create a digital animation of the subjects actions.

The data were filtered using the APAS *Filter* module. The *Filter* module was used to smooth data, thereby reducing digitizing artifacts. The *Filter* was essential for smoothing the estimated hip data.

Data for each of the trials were normalized by time to provide a common number of data points for ensemble averaging. Kinematic data during the stance phase of each trial were normalized to represent 100 percent of stance. Once data were normalized, ensemble averages were calculated within and between subjects. These averages were used to represent the typical lower extremity joint and segment kinematic parameters. To determine validity, trunk, hip, knee and ankle angle data collected for this study were compared with the published results of Winter (1991).

4.7.2 Kinetic Data

Once video data for selected steps were collected and digitized, it was then possible to isolate the corresponding plantar pressure data. For example: the third right heel strike after the LED signal was matched to the third heel strike from the corresponding F-Scan trial.

The step taken at the middle of the walkway was isolated using the "keep frames" option in the F-Scan Edit menu.

Using the "Add Box" tool, selected areas of the plantar surface of the foot were isolated. Peak plantar pressure for the first MTP joint (Figure 12a), hallux segment(Figure 12b) and 5^{th} metatarsal (figure 12c) were isolated and saved in an ASCII format. These files were exported to the *BioProc* (Robertson, 2002) program for normalizing and ensemble averaging. Each trial was normalized to 100 percent of stance.

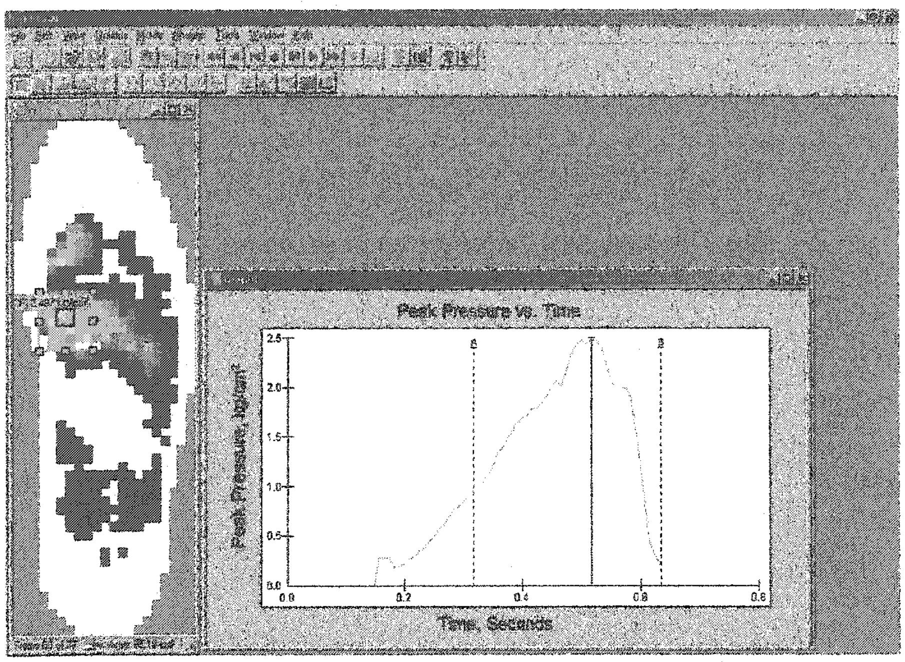

Figure 12a: Screen image of F-Scan (Tekscan Inc.) software with 1st MTP joint mask, and relative plantar pressure during stance

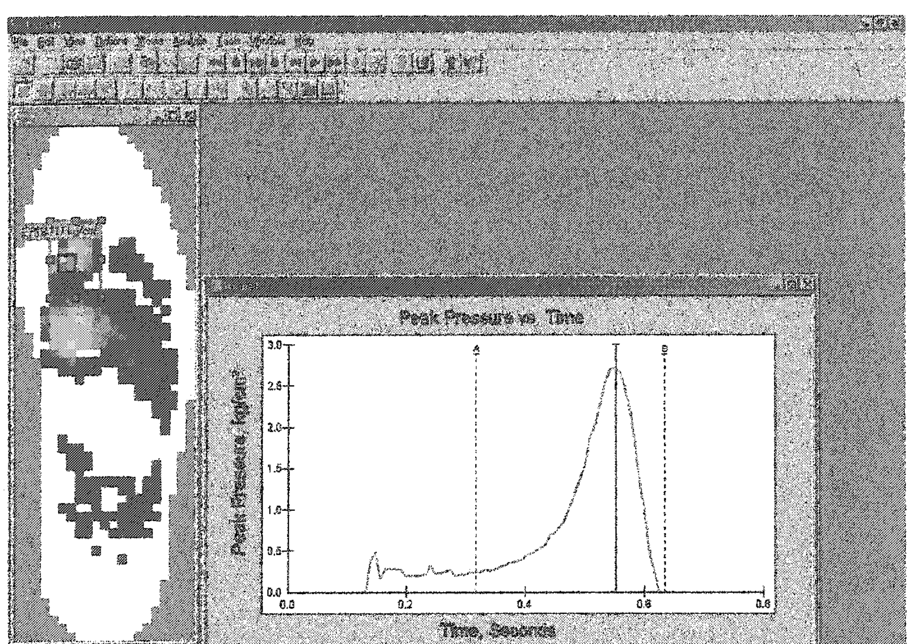

Figure 12b: Screen image of F-Scan (Tekscan Inc.) software with hallux segment mask, and relative plantar pressure during stance

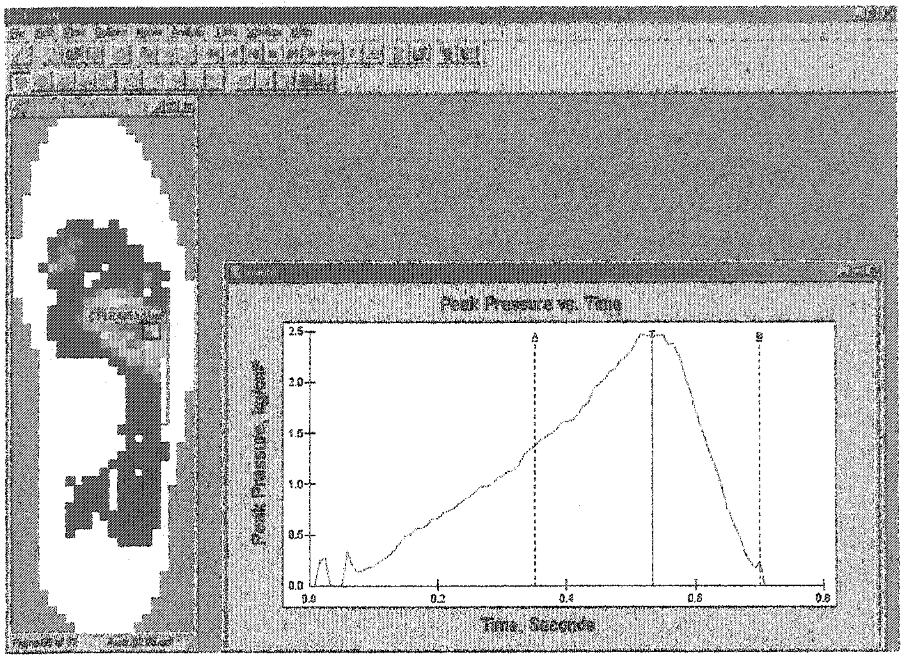

Figure 12c: Screen image of F-Scan (Tekscan Inc.) software with 5th metatarsal segment mask, and relative plantar pressure during stance

In addition to the plantar pressure data, differences in CoP velocities during stance were also investigated. Since the study was limited to planar motion in the sagittal plane, CoP velocities in the anterior posterior direction were examined.

The F-Scan system provided centre of normal force position for the duration of each stance phase. This data were used to calculate CoP displacement, and inturn, CoP velocity. The F-Scan system provided CoP locations for each frame of stance. Raw location data were in the form of insole coordinates. Row and column data for each frame was saved generated in an ASCII format. ASCII coordinates were imported to a spreadsheet program to determine average CoP velocities in the antero-posterior direction. To calculate distances, a scaling factor was used. The distance between the centre of each sensor was 0.508 cm. Since the sampling rate was known stance duration times were calculated.

For example: if the most anterior sensor that recorded pressure data was column 44, and the most posterior sensor that recorded pressure data was 16, then the distances from the origins would be 22.35 cm (0.508 cm x 22) and 8.13 (0.508 cm x 16) cm respectively. Therefore, $\Delta d = 22.35 - 8.13 = 14.22$ cm. If such CoP displacement occurred over 88 samples at a rate of 120 Hz or 0.008333 seconds per sample, them $\Delta t = 88 \times 0.008333 = 0.7333$ s. Since $v_{avg} = \Delta d / \Delta t$ then $v_{avgCoF} = 14.22 / 0.7333 = 19.39$ cm/s. This calculation was repeated for each of the five trials for each condition. Means were calculated for both conditions, and then compared.

4.8 Statistical Analysis of Data

Paired T-tests were used to determine differences between the two conditions. The tests were conducted to discern differences in kinetic and kinematic data collected during stance for each of the conditions.

The *Wilcoxon matched-pairs signed-rank test* was used to determine significant differences in pain experienced pre- and post-custom orthoses intervention. This test was valid as it was intended to discern significant differences between paired nonparametric data. More specifically, this test was used to compare paired, ordinally-scaled data.

5.0 Results

5.1 Qualitative Data

The qualitative data presented in this chapter represented ordinal data. Ordinal data were collected during initial subject examining and selection. Data were again collected after subjects were allotted two months of CFO (modified with the kinetic wedge) use.

Two months later one subject was not available for questioning, therefore only data for the remaining 14 subjects were analysed. When asked to self-select the severity of their pain, six subjects stated no change in perceived pain, thus the data were discarded prior to analysis; retaining difference scores of zero increases the chances of failing to reject the null hypothesis. The remaining eight pain scores were rank ordered by the magnitude of the change in perceived pain, and a *Wilcoxon T* was used to evaluate the data. Two subjects had an improvement of 1 point, one subject had an improvement of 2 points, 1 subject had an improvement of 3 points, and 4 subjects had an improvement of 4 points after using the modified CFOs for two months. The results showed no significant reduction in perceived pain after 2 months, $T = 32.0$, $p > 0.05$, with the ranks of the decreases totalling 32.0 (Appendix D).

5.2 Quantitative Data

The quantitative data presented in this chapter represent ensemble averaged trials for each of the two treatment conditions of:

- NKW - walking without Kinetic Wedge modifications to custom foot orthoses

  75 trials total - 5 trials per subject, 15 subjects total

- KW - walking with Kinetic Wedge modifications to custom foot orthoses

  75 trials total - 5 trials per subject, 15 subjects total

5.2.1 Kinematic Data

Average maximal ranges of motion (in tables and text) for the trunk, hip, knee, and ankle are the averages of the peak values of each subjects' kinematics, rather than the peak of the ensemble average. Ensemble averaging requires that each ensemble have the same number of data points. Thus, the peak of an ensemble average is typically less than the average maximum of the original data since the peak values of different curves do not occur at the same instant. In this study the concern was with maximal ranges of motion; thus, it was more appropriate to use the average of the peaks of all trials. However, data presented in each of the kinematic graphs are that of the ensemble averaged data of 75 normalized trials for each condition.

5.2.1.1 Trunk Posture During Stance

The trunk position during stance is represented with respect to the positive x-axis. Trunk flexion (less than 90 degrees) at the end of stance was used to represent forward trunk lean. Carlson et al.(1988) stated that normal trunk posture during stance varied by 4 degrees with respect to the positive y-axis. A 4 degree change would represent a normal relative trunk flexion of 86 degrees. Nine of the 15 subjects exhibited trunk flexion angles greater than 86 degrees during NKW.

The average maximum trunk flexion during NKW was 85.2 degrees ($\pm$ 2.09). For condition NKW, subject 5 exhibited the lowest average trunk flexion with 88 degrees while, subject 14 exhibited the greatest average trunk flexion at 81.3 degrees (Table 1).

Average trunk flexion during KW was 85.5 degrees ($\pm$ 2.76). For condition KW, subject 12 exhibited the lowest average trunk flexion at 89.2 degrees, while subject 14 exhibited the highest average trunk flexion at 79.4 degrees (Table 1).

Nine subjects showed an improvement in trunk posture (decreased trunk flexion). Subject 2 showed the greatest improvement in posture with an average decrease of 1.600 degrees. Five subjects showed an increase in trunk flexion. Subject 14 showed an increase of 1.900 degrees.

Overall, application of the Kinetic Wedge modification to custom foot orthoses resulted in a decrease (M = 0.212, ± 0.9879, where M = mean difference) in trunk flexion before toe-off. The decrease in trunk flexion was not statistically significant, $t(14) = 0.831$, $p > 0.05$, two tailed (Table 1). Ensemble averaged trunk angles during stance for all subjects during both conditions are represented in Figure 13.

	Condition	
	NKW	KW
Average trunk flexion (deg)	85.2	85.5
p	0.420	
Mean difference	0.212	
Std. dev of mean difference	± 0.9879	
Lowest trunk flexion (deg)	88.0 (subject 5)	89.2 (subject 12)
Greatest trunk flexion (deg)	81.3 (subject 14)	79.4 (subject 14)

Table 1: Trunk segment flexion angles before toe-off for both conditions

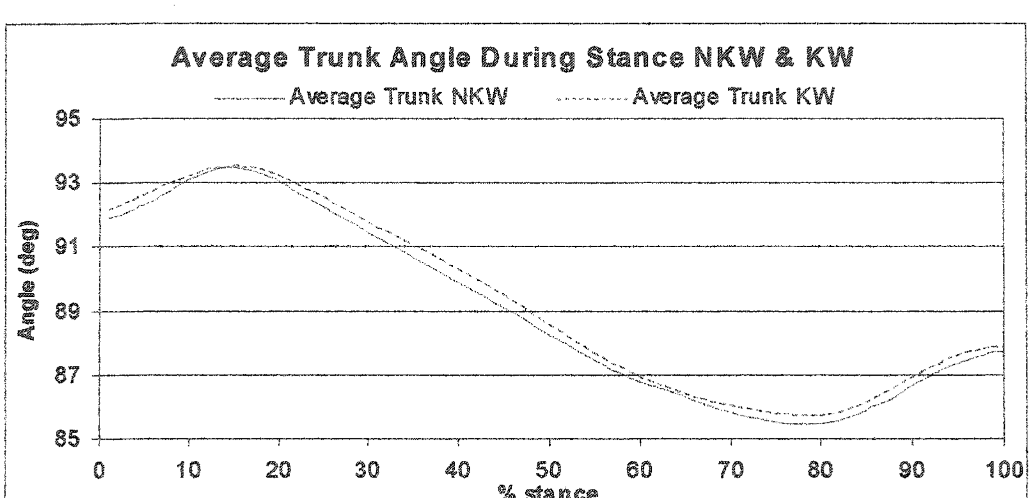

Figure 13: Ensemble averaged kinematic data of trunk segment during stance for both conditions

5.2.1.2 Hip Joint Kinematics During Stance

Hip joint kinematics were derived using the Winter method (Winter 1991) that defines the hip angle as the difference between thigh segment angle and trunk segment angle ($\theta_{hip} = \theta_{thigh} - \theta_{trunk}$). Thus, positive angles represent flexion and negative angles represent hyperextension. Thigh and trunk segment angles were represented with respect to the positive x-axis. The lowest angle achieved by the hip at the end of stance was used to represent maximum hip hyperextension.

The average maximum hip hyperextension during NKW was -8.26 degrees. For condition NKW, subject 12 exhibited the greatest average hip hyperextension at -16.22 degrees, and subject 13 exhibited the lowest average hip hyperextension at -4.83 degrees (Table 2).

The average maximum hip hyperextension KW was -8.28 degrees. For condition KW, subject 12 exhibited the greatest average hip hyperextension at -14.32 degrees, and subject 9 exhibited the lowest average hip hyperextension at -2.90 degrees (Table 2).

Eight subjects showed an increase in maximum hip joint hyperextension (between NKW and KW). Subject 2 showed the greatest increase of hip hyperextension during condition KW with 2.80 degrees. Seven subjects showed a decrease in hip hyperextension during KW. Subject 9 showed greatest decrease in hip hyperextension with 2.09 degrees.

Overall, application of the Kinetic Wedge modification to custom foot orthoses resulted in a increase ($M = -0.022, \pm 1.301$) in maximal hip hyperextension at the end of stance phase. This increase in maximal hip hyperextension at the end of stance was not statistically significant, $t(14) = -0.252$, $p > 0.05$, two tailed (Table 2). Ensemble averaged hip angles during stance for all subjects during both conditions are represented in Figure 14.

	Condition	
	NKW	KW
Average hip hyperextension (deg)	-8.26	-8.28
p	0.949	
Mean difference	-0.022	
Std. dev. of mean difference	1.301	
Lowest average hip hyperextension	-4.83 (subject 13)	-2.90 (subject 9)
Greatest average hip hyperextension	-16.22 (subject 12)	-14.32 (subject 12)

Table 2: Peak hip joint hyperextension angles at toe-off for both conditions

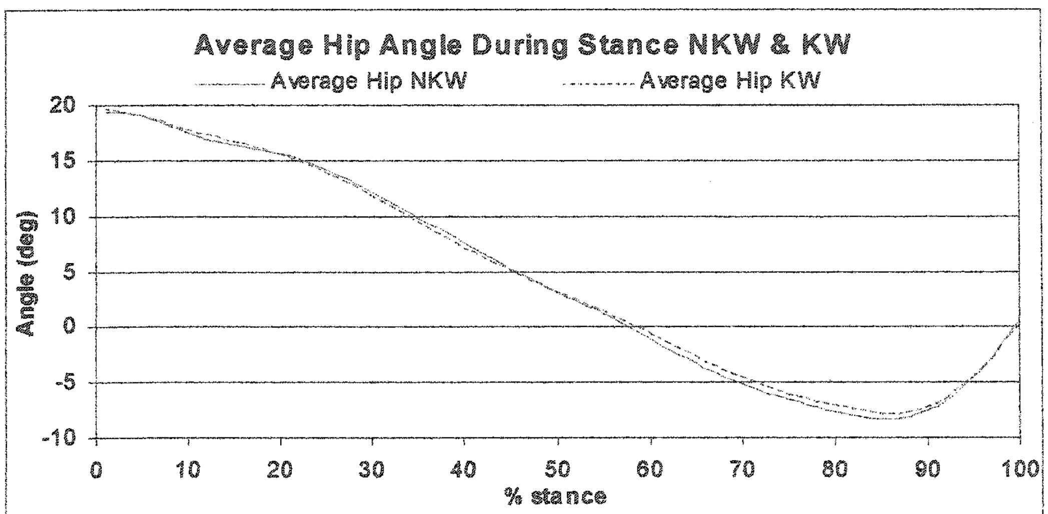

Figure 14: Ensemble averaged kinematic data of hip joint during stance for both conditions

5.2.1.3 *Knee Joint Kinematics During Stance*

Knee joint kinematics were also derived using the method described by Winter (1991). Knee joint angles were calculated by subtracting leg segment angles from the thigh segment angles ($\theta_{knee} = \theta_{thigh} - \theta_{leg}$, where positive angles represented flexion, and negative angles represented hyperextension). The lowest values during the latter 50% of stance was used to represent maximum knee extension.

The average maximum knee extension during NKW was 14.2 degrees. For condition NKW, subject 3 exhibited the greatest average knee extension with 20.2 degrees, and subject 7 exhibited the lowest average knee extension with 7.89 degrees (Table 3).

The average maximum knee extension during KW was 14.45 degrees. For condition KW, subject 3 exhibited the greatest average knee extension at 21.61 degrees, and subject 4 exhibited the lowest average knee extension at 8.17 degrees (Table 3).

Nine subjects showed an increase in maximum knee joint extension between NKW and KW). Subject 5 showed the greatest improvement of knee joint extension; during condition KW with 2.39 degrees. Six subjects showed a decrease in knee extension (knee extension decreased during condition KW). Subject 11 showed greatest decrease in knee extension, with 2.39 degrees.

Overall, application of the Kinetic Wedge modification to custom foot orthoses resulted in an increase ($M = 0.256, \pm 1.417$) in maximal knee extension at the end of stance phase. The increase in knee extension at the end of stance was not statistically significant, $t(14) = 2.71$, $p > 0.05$, two tailed (Table 3). Ensemble averaged knee joint angles during stance for all subjects during both conditions are represented in Figure 15.

	Condition	
	NKW	**KW**
Average knee extension (deg)	14.20	14.45
p	0.496	
Mean difference	0.256	
Std. dev. of mean difference	1.417	
Lowest average knee extension	7.89 (subject 7)	8.17 (subject 4)
Greatest average knee extension	20.2 (subject 3)	21.61 (subject 3)

Table 3: Knee joint extension angles before toe-off for both conditions

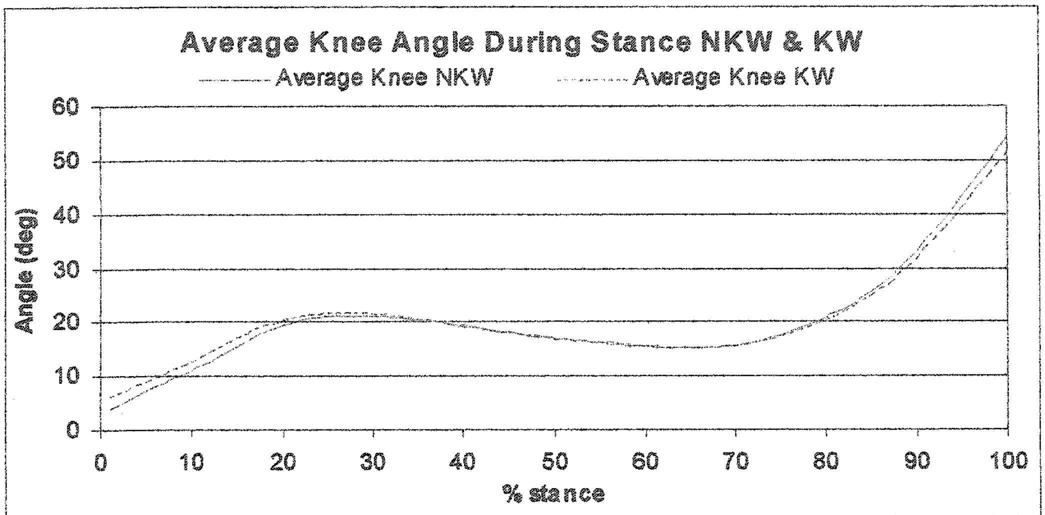

Figure 15: Ensemble averaged kinematic data of knee joint during stance for both conditions

5.2.1.4 Ankle Joint Kinematics During Stance

Ankle kinematics were also derived using the method described by Winter (1991). Ankle angle was calculated by subtracting foot segment angle from the leg segment angle with respect to the negative x-axis ($\theta_{ankle} = \theta_{foot} - \theta_{leg} - 90°$), where positive angles represented dorsiflexion, and negative angles represented plantar flexion. Maximum negative angles at toe-off was used to represent maximum ankle plantar flexion.

The average maximum ankle plantar flexion during NKW was -9.87 degrees. For condition NKW, subject 12 exhibited the greatest average ankle plantar flexion with -32.4 degrees, and subject 4 exhibited the lowest average ankle extension with -1.839 degrees (Table 4).

The average maximum ankle plantar flexion during KW was -9.80 degrees. For condition KW, subject 7 exhibited the greatest average ankle plantar flexion at -18.12 degrees, and subject 4 exhibited the lowest average ankle extension with 0.3973 degrees (Table 4).

Eleven subjects showed an increase in maximum ankle joint plantar flexion between NKW and KW. Subject 7 showed the greatest increase of ankle joint plantar flexion with 7.02 degrees. Four subjects showed a decrease in ankle plantar flexion between NKW and KW. Subject 12 showed greatest decline in ankle plantar flexion with 19.90 degrees.

Overall, application of the Kinetic Wedge modification to custom foot orthoses resulted in a decrease ($M = 0.071, \pm 6.00$) in maximal ankle plantar flexion at toe-off. The decrease in ankle plantar flexion at the end of stance was not statistically significant, $t(14) = 0.1772$, $p > 0.05$, two tailed (Table 4). Ensemble averaged ankle joint angles during stance for all subjects during both conditions are represented in Figure 16.

	Condition	
	NKW	**KW**
Average peak plantar flexion (deg)	-9.87	-9.80
p	0.9642	
Mean difference	0.071	
Std. dev. of mean difference	6.00	
Lowest average ankle plantar flexion	-1.839 (subject 4)	0.3973 (subject 4)
Greatest average ankle plantar flexion	-32.4 (subject 12)	-18.12 (subject 7)

Table 4: Peak ankle joint plantar flexion angles at toe-off for both conditions

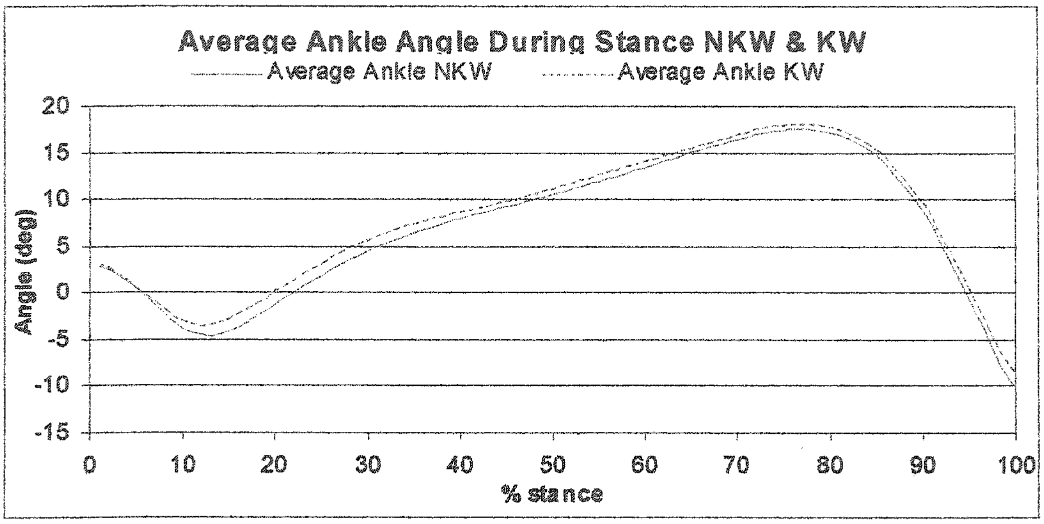

Figure 16: Ensemble averaged kinematic data of ankle joint during stance for both conditions

5.2.2 Kinetic Data

Kinetic data are presented as the maximum relative plantar pressures for three (plantar) foot areas: first MTP joint, hallux segment, 5th metatarsal segment. As in the previous section average peak plantar pressures (in tables and text) for the first MTP joint, hallux segment, and 5th metatarsal segment are the averages of the peak values of each subjects' kinetics, rather than the peak of the ensemble average. However, data presented in each of the kinetic graphs are that of the ensemble averaged data of 75 normalized trials for each condition.

5.2.2.1 First MTP Plantar Pressure During Stance

Maximum plantar pressure under the first MTP joint was used to represent maximal joint loads during the stance phase of gait. The average maximum plantar pressure under the first MTP joint during NKW was 1.871 kg/cm^2 ($\pm$ 0.459). For condition NKW, subject 15 exhibited the greatest average plantar pressure with 2.73 kg/cm^2, and subject 1 exhibited the lowest average plantar pressure with 1.094 kg/cm^2 (Table 5).

The average maximum plantar pressure under the first MTP joint during KW was 1.554 kg/cm^2 ($\pm$ 0.409). For condition KW, subject 8 exhibited the greatest average plantar pressure with 2.46 kg/cm^2, and subject 7 exhibited the lowest average plantar pressure with 0.987 kg/cm^2. (Table 5).

Thirteen subjects showed an decrease in maximum first MTP joint plantar pressure between NKW and KW. Subject 13 showed the greatest decrease in first MTP joint plantar pressure with 0.978 kg/cm^2. Two subjects showed an increase in first MTP joint plantar pressure between NKW and KW. Subject 4 showed greatest increase with 0.319 kg/cm^2.

Overall, application of the Kinetic Wedge modification to custom foot orthoses resulted in a decrease ($M = -0.317, \pm 0.326$) in maximum plantar pressure under the first MTP joint during stance. The decrease in first MTP joint plantar pressure during stance was statistically significant, $t(14) = -14.58$, $p < 0.05$, two tailed (Table 5). Ensemble averaged first MTP joint plantar pressures during stance for all subjects during both conditions are represented in Figure 17.

	Condition	
	NKW	**KW**
Average peak plantar pressure (kg/cm^2)	1.871	1.554
p	0.0021	
Mean difference	-0.3168	
Std. dev. of mean difference	0.3260	
Lowest average peak plantar pressure	1.094 (subject 1)	0.9872 (subject 7)
Greatest average peak plantar pressure	2.73 (subject 15)	2.46 (subject 8)

Table 5: Peak first MTP joint plantar pressures during stance for both conditions

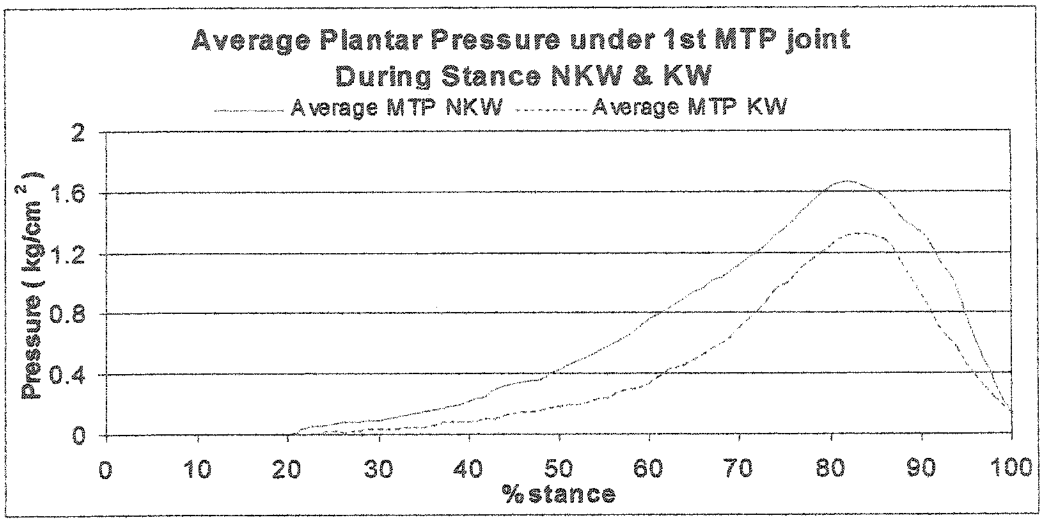

Figure 17: Ensemble averaged plantar pressures under the first MTP joint during stance for both conditions.

5.2.2.2 Hallux segment Plantar Pressure During Stance

Maximum plantar pressure was used to represent maximal loads experienced by the hallux segment during the stance phase of gait. The average maximum plantar pressure under the hallux segment during NKW was 2.66 kg/cm^2 ($\pm$ 0.832). For condition NKW, subject 2 exhibited the greatest average plantar pressure with 4.10 kg/cm^2, and subject 5 exhibited the lowest average plantar pressure with 1.163 kg/cm^2. (Table 6).

The average maximum plantar pressure experienced under the hallux segment during KW was 2.28 kg/cm^2 (Std. dev = 0.7813). For condition KW, subject 4 experienced the greatest average plantar pressure with 4.26 kg/cm^2, and subject 5 experienced the lowest average plantar pressure with 1.267 kg/cm^2. (Table 6).

Eleven subjects showed a decrease in maximal hallux segment plantar pressure between NKW and KW. Subject 15 showed the greatest improvement in hallux segment plantar pressure with 1.469 kg/cm^2. Four subjects showed an increase in hallux segment plantar pressure between

NKW and KW. Subject 14 showed greatest increase with 0.8320 kg/cm^2.

Overall, application of the Kinetic Wedge modification to custom foot orthoses resulted in a decrease (M = -0.3733, $\pm$ 0.7426) in maximum plantar pressure under the hallux segment during stance. The decrease in hallux segment plantar pressure during stance was not statistically significant, t(14) = -7.54, p > 0.05, two tailed (Table 6). Ensemble averaged hallux segment plantar pressures during stance for all subjects during both conditions are represented in Figure 18.

	Condition	
	NKW	KW
Average peak plantar pressure (kg/cm^2)	2.66	2.28
p	0.0719	
Mean difference	-0.3733	
Std. dev. of mean difference	0.7426	
Lowest average peak plantar pressure	1.163 (subject 5)	1.267 (subject 5)
Greatest average peak plantar pressure	4.10 (subject 2)	4.26 (subject 4)

Table 6: Peak hallux segment plantar pressures during stance for both conditions

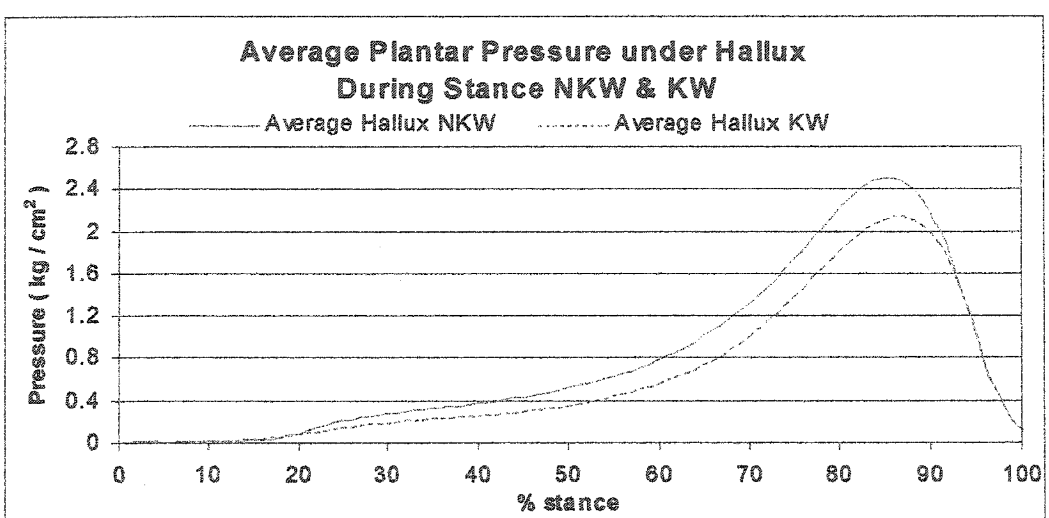

Figure 18: Ensemble averaged plantar pressure data under the hallux segment during stance for both conditions.

5.2.2.3 *Fifth Metatarsal segment Plantar Pressure During Stance*

The average maximum plantar pressure experienced under the 5th metatarsal segment during NKW was 1.749 kg/cm^2 ($\pm$ 0.6083). For condition NKW, subject 8 experienced the greatest average plantar pressure with 3.24 kg/cm^{2,} and subject 14 exhibited the lowest average plantar pressure with 0.7846 kg/cm^2. (Table 7).

The average maximum plantar pressure experienced under the fifth metatarsal segment during KW was 1.748 kg/cm^2 (Std. dev = 0.7890). For condition KW, subject 8 experienced the greatest average plantar pressure with 4.33 kg/cm^2, and subject 15 experienced the lowest average plantar pressure with 1.009 kg/cm^2. (Table 7).

Eight subjects showed a decrease in maximal fifth metatarsal segment plantar pressure between NKW and KW. Subject 9 showed the greatest decrease in fifth metatarsal segment plantar pressure with 0.4400 kg/cm^2. Seven subjects showed an increase in fifth metatarsal segment plantar pressure between NKW and KW. Subject 8 showed greatest increase with 1.099 kg/cm^2.

Overall, application of the Kinetic Wedge modification to custom foot orthoses resulted in a decrease ($M = -0.0005$, $\pm$ 0.363) in maximum plantar pressure under the fifth metatarsal segment during stance. The decrease in fifth metatarsal segment plantar pressure during stance was not statistically significant, $t(14) = -0.0198$, $p > 0.05$, two tailed (Table 7). Ensemble averaged fifth metatarsal segment plantar pressures during stance for all subjects during both conditions are represented in Figure 19.

	Condition	
	NKW	KW
Average peak plantar pressure (kg/cm^2)	1.749	1.748
p	0.9960	
average difference	-0.0005	
Std. dev. of mean difference	0.3629	
Lowest average peak plantar pressure	0.7846 (subject 14)	1.009 (subject 15)
Greatest average peak plantar pressure	3.24 (subject 8)	4.33 (subject 8)

Table 7: Peak fifth metatarsal segment plantar pressures during stance for both conditions

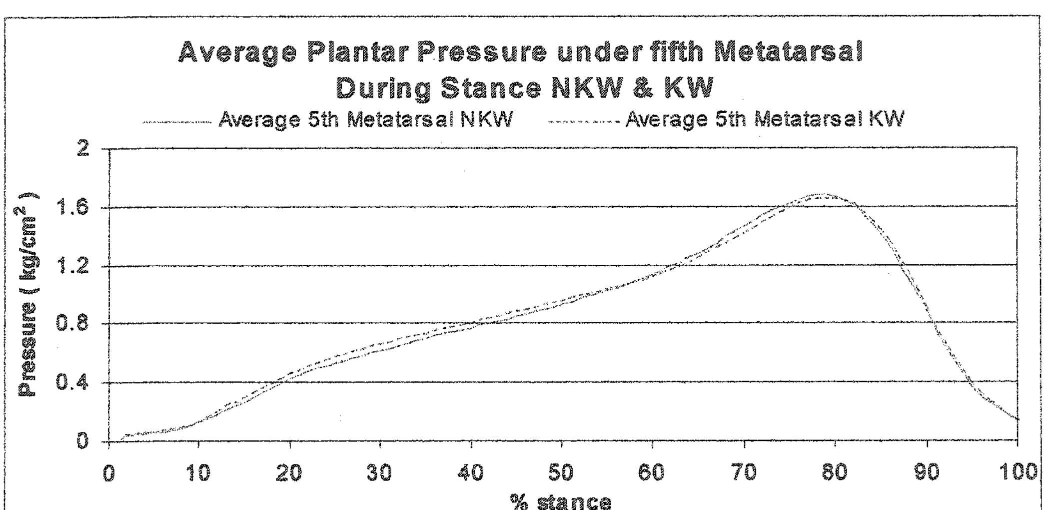

Figure 19: Ensemble averaged plantar pressures under the fifth metatarsal during stance for both conditions.

5.2.2.4 Centre of Pressure Velocities During Stance

Average CoP velocity during NKW was 22.49 cm/s. (± 2.38). For condition NKW, subject 1 had the greatest average CoP velocity with 26.4 cm/s, and subject 5 exhibited the lowest average CoP velocity with 18.56 cm/s (Table 8).

Average CoP velocity during KW was 22.91 cm/s. (± 2.28). For condition KW, again subject 1 had the greatest average CoP velocity with 26.9 cm/s, and again subject 5 exhibited the lowest average CoP velocity with 19.84 cm/s (Table 8).

Eleven subjects showed an increase of CoP velocity between NKW and KW. Subject 14 showed the greatest increase in CoP velocity, with 2.90 cm/s. Four subjects showed a decrease in CoP velocity. Subject 8 showed the greatest decrease with 3.79 cm/s.

Overall, application of the Kinetic Wedge modification to custom foot orthoses resulted in increased ($M = 0.415$, ± 1.418) CoP velocity during stance. The increase in CoP velocity during stance was not statistically significant, $t(14) = 4.39$, $p > 0.05$, two tailed (Table 8). Ensemble averaged CoP velocities for all subjects during both conditions and difference scores during stance are represented in Figure 20.

	Condition	
	NKW	**KW**
Average CoP velocity (cm/s)	22.49	22.91
p =	0.2762	
mean difference	0.415	
Std. dev. of mean difference	1.418	
Lowest CoP velocity	18.56 (subject 5)	19.84 (subject 5)
Greatest CoP velocity	26.40 (subject 1)	26.92 (subject 1)

Table 8: Average centre of pressure velocities during stance for both conditions

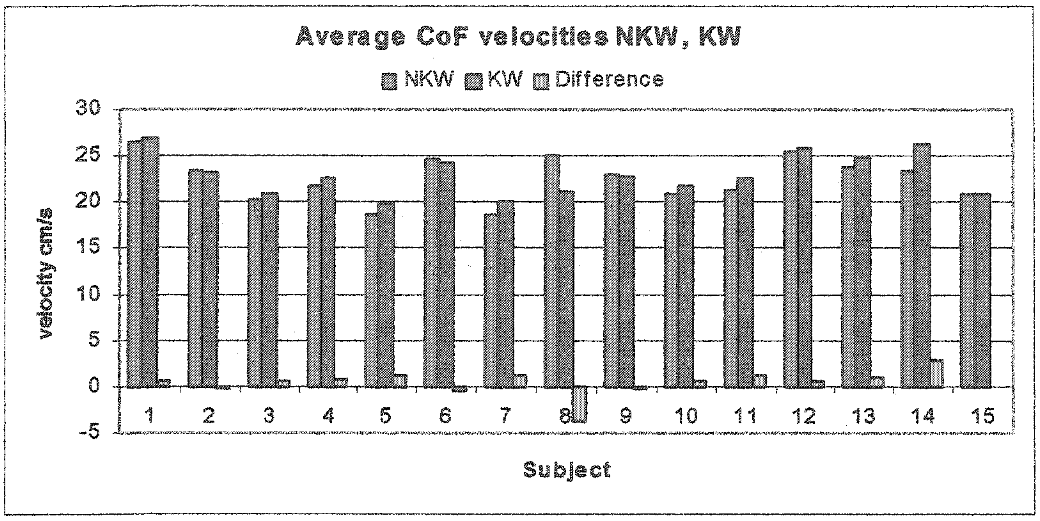

Figure 20: Average and difference centre of force velocities for both conditions

5.3 Statistical Power Analysis

Using a statistical power analysis, it was possible to determine the sample size necessary to find a significant difference (Dupont and Plummer, 1990). For the variable of trunk posture, given the established values of $\alpha = 0.05$, M = 0.212, SD = 0.988 and n = 15, the statistical power was 0.11005 To establish a higher power, a greater sample size would be necessary. For example, with each of these statistical characteristics, an *n* of 171 subjects would be necessary to generate a power of 0.8, and an *n* of 399 would be necessary to achieve a power of 0.99.

For the variable of hallux plantar pressure, given the established values of $\alpha = 0.05$, M = -0.373, SD = 0.743 and *n* = 15, the statistical power was 0.423. To establish a higher power, a greater sample size would be necessary. For example, with each of these statistical characteristics, an *n* of 33 subjects would be necessary to generate a power of 0.80, and an *n* of 76 would be necessary to achieve a power of 0.99.

For the variable of 5^{th} metatarsal pressure, given the established values of $\alpha = 0.05$, $M = -0.0005$, $SD = 0.363$ and $n = 15$, the statistical power was 0.050. To establish a higher power, a greater sample size would be necessary. Unfortunately it was not possible to calculate sample sizes as it resulted in failure of the program.

6.0 DISCUSSION

The purpose of this study was to determine whether a CFO modified with a Kinetic Wedge had positive effects on the body both focally and globally. More specifically, the purpose was to determine if such a change would reduce plantar pressures at the first MTP joint, hallux segment, and fifth metatarsal segment, and/or improve trunk posture, and/or increase CoP velocity during stance.

After statistical tests, there was no significant reduction of perceived pain in the first MTP joint by subjects after at least 2 months of use. Subjects were only asked the four survey questions. Some subjects offered rationale for their answers. Without being asked additional questions, few subjects did admit to not wearing the CFO everyday. There were times when subjects forgot to include the CFOs after changing footwear. Subjects also stated that the CFO did not fit each and every pair of their shoes. In addition, testing sessions occurred during summer months. Three subjects stated a preference for sandals during the warmer season. Subjects stated that it was not possible to wear their CFOs in their sandals.

Subjects may have not worn CFOs often while at home. Preference to be barefooted while at home may have been a factor for a lack of inconsistent CFO use. As stated earlier, the efficacy of the CFO is greatest when used with properly fitted footwear.

Clinically, patients would be advised to return, in the event that no appreciable change in conditions were achieved. During return visits, necessary changes or modifications would be made to CFOs to accommodate the needs of the client. After the second survey, subjects were encouraged to return to the clinic to address any existing discomforts.

There was a significant reduction of plantar pressure under the first MTP joint with the use of the modified CFOs. There was an average of 16.03% reduction of plantar pressure under the first MTP joint. Placing the first MTP joint in a relatively greater dorsiflexed position allowed the joint to avoid functional limitations (Figure 4). In addition to avoiding functional limitations, dorsiflexion of the first MTP joint may have established an anatomical forefoot rocker. Such a shaped surface would create a path of lesser resistance as the body centre of mass progresses forward over the first MTP joint.

Functional locking of the first MTP joint would result in a longer lever as the first ray and hallux acts as a single unit. Moment of force is a product of applied force and lever (moment) length. Moments created and experienced by the longer lever as a result of FHL would result in relatively high plantar pressures at the hallux.

Considering a significant reduction of first MTP joint plantar pressure, a significant reduction of hallux plantar pressure was expected. There was an average reduction of 0.038 kg/cm^2 (14.05%) with modified CFO use. However, this difference was not statistically significant. Apparently, there was no significant reduction of the moment arm length with modified CFO use.

Previous research stated that 1.4 to 1.7 times that of body weight is experienced by the foot during stance (Dananberg, 1995). Studies also suggested that the first ray relative to the lesser metatarsals played a greater role to distribute plantar pressure (Bennett and Duplock, 1993; Morag and Cavanagh, 1997). According to the podiatric community, first MTP joint limitation, resulted in compensatory movements to establish a path of lesser resistance to move the body's centre of mass through the sagittal plane. One method to lessen this resistance would be to use centre of mass through the sagittal plane. One method to lessen this resistance would be to use the lesser (4 lateral) MTP joints to move the body through the sagittal plane. Displacing the body's centre of mass more laterally would result in relatively higher planter pressures under the lesser metatarsals. Dananberg refers to this strategy as an avoidance of the first MTP joint. Ten of the fifteen subjects displayed this strategy during NKW testing. Average plantar pressure under first MTP and fifth metatarsal segment for both conditions by subjects were similar.

Since the data showed an improvement in first MTP joint plantar pressure, a reduction of plantar pressures experienced by the lesser metatarsal heads pressure was also expected (Dananberg, 1993 and Dananberg *et al.*, 1996). With the use of modified CFOs, subjects experienced a reduction of 0.001 kg/cm^2 under the fifth metatarsal segment. However, this reduction of plantar pressure was not statistically significant. According to the results, modified CFOs did not reduce the tendency of subjects to use the compensatory strategy of first MTP avoidance. The data did not indicate where pressure under the first MTP joint was redistributed since additional plantar pressure areas were not included in the study design.

One of the hypotheses of this study was that an improvement in first MTP joint function would increase plantar centre of pressure velocities (CoP). Eleven of the fifteen subjects' CoP velocities increased by 1.844 %. This increase of CoP velocity was not statistically significant. According to the data, CFOs did not increase CoP velocity.

Unlike studies performed by the podiatric community (Dananberg, 1995), changes in gait posture were not noted by this study. According to statistical (*t*) tests, there was no significant improvement in trunk posture with the application of the Kinetic Wedge to CFOs. Considering there was no significant effect at the trunk, the investigators decided to consider whether or not there were indeed significant changes lower in the kinetic chain. However, it was found that there were no significant changes during stance at the hip, knee and ankle joints. There were no significant effects of the CFOs with the Kinetic Wedge globally.

No significant change between treatments may have been due to little or no chance for learning effects; since little time was allowed between treatments. Each subject was asked to perform gait trials using the modified CFOs after only 30 minutes of practice time. Perhaps months, weeks, or even days of modified CFO use may have caused significant kinematic changes along the kinetic chain.

However, using a similar experimental design, Dananberg claimed that a "consistency of hip extension with orthotics is clearly stated" (1995: p. 404). Dananberg's study may have used semi-rigid CFOs modified with the Kinetic Wedge. Semi-rigid CFOs were produced by Langer Biomechanics (Langer Biomechanics Inc.). It is possible that the use of a more rigid device may have reduced the magnitude of sub-talar overpronation. A reduction of sub-talar overpronation, results in a less everted and dorsiflexed position of the first ray during late stance. Since the most distal bone of the first ray is the first metatarsal, the first MTP joint itself would be subject to greater plantar flexed position, thereby establishing an anatomical forefoot rocker. (figure 4).

Additionally, CFOs provided to each of the subjects may have not offered adequate support under the midfoot, because the arches of each pair of CFO was reduced by 30% using the Amfit software. Lord and Hosien (1994) stated that custom orthoses can provide "a redistribution of the load away from the metatarsal region, primarily due to increased support under the midfoot" (p. 214).

Although trunk, hip, knee ankle kinematics, hallux and fifth metatarsal plantar pressures revealed no significant statistical differences with KW, there may be some relevant clinical significance. The findings of plantar pressure reductions under each area of the foot was that of five stance phases per subject. Perhaps such reductions may be clinically beneficial in the long term. The clinical significance of such plantar pressure reductions may depend on the effect of thousands of steps. Reduced plantar pressures is important to the podiatric community to decrease occurrences of plantar foot ulcerations of diabetic feet. Clinically, common areas for neuropathic plantar diabetic ulcers include the metatarsal heads (Stokes et al., 1979; Lord and Hosien, 1994; and Lord et al., 1986). Avoiding ulcers, and in turn dangerous infections are important to avoid the need for lower-extremity amputations.

7.0 Conclusions

The present study demonstrated significant differences focally with the use of modified CFOs. Plantar pressure under the first MTP were significantly reduced using the Kinetic Wedge modification. There were, however, no significant differences with the use of modified CFOs globally. Hallux and fifth metatarsal plantar pressures were not significantly reduced with modified CFO use. In addition, neither ankle, knee, hip, nor trunk sagittal plane kinematics were significantly changed by the Kinetic Wedge modification.

Furthermore, after two months of CFO use, there was no significant reduction in pain as perceived by the subjects. Inconsistency of use by subjects may have been a contributing factor. Better dialogue between clinicians and patients may be needed to maximize patient care.

7.1 Recommendations

For future study into this area, the following recommendations are suggested:

Future study should include more subjects and collect more trials for data analysis. A lack of statistically significant global effects may be related to sample size. Statistical power analysis indicated that a larger sample size was necessary to find a significant statistical differences (Dupont and Plummer, 1990).

Subjects should be allotted more practice time between testing conditions. To better understand the manner in which CFOs modified with the Kinetic Wedge improve foot leg, and thigh function, greater periods of time be allotted between testing sessions. Clinically, foot care specialists usually recommend a 2 week trial or break-in period. During this time patients are advised to wear CFOs for approximately 1 hour during the first day. As each day passes, patients are advised to gradually increase CFO use until they are able to wear CFOs for a full day.

Increased CFO use may create a learning effect and in turn improved first MTP joint function.

To verify the first MTP joint dorsiflexion, it would be advantageous to use an electrogoniometric device directly on the first ray and hallux. Direct joint angle data could offer greater insight to the kinematic characteristics of the first MTP joint such as maximum joint dorsiflexion, and changes in dorsiflexion velocity during gait. The ability to use this device with its axis on the centre of the first MTP joint may prove to be important.

If through using the previously mentioned recommendations, differences between conditions are found, it would be a point of interest to make contrasts and comparisons to Winter's previously published data (Winter, 1991). If indeed statistically significant improvements in gait posture are found, it would be worthwhile to consider whether or not the improved gait posture parameters are similar or dissimilar to "normal" walking. Establishing a more "normal" walking patterns with the use of CFO modified with a kinetic wedge modification would justify CFO use as an unconventional intervention for postural complains.

The plantar pressure data collected for this study cannot suggest where pressure from the first MTP joint was redistributed. To determine plantar pressure redistribution, it may be necessary to investigate additional plantar foot areas such as: second, third, fourth MTP joints, phalanges, first ray, and heel.

Due to restrictions set by the research ethics board, kinematic, and plantar pressure data of subjects without any CFO intervention was not possible. Including this testing condition would be of value, as it would present kinematic and plantar pressure data of subjects with FHL prior to treatment. Such an inclusion may allow greater distinctions between treatment conditions.

For future study, it is also recommended that further comparisons be made between KW and NKW. To determine the effect of a CFO without the kinetic wedge modification, it is recommended that future research use an "A-B-A" design. Data would begin with subjects tested during NKW (treatment A). Secondly, data would be collected for subjects during KW (treatment B). Finally, a third collection of data would be conducted for subjects during NKW (repeat of treatment A). To conduct an A-B-A design each subject would have to be fabricated two pairs of CFOs: one without (NKW), and one with the kinetic wedge (KW) modification.

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Amfit Inc.
2850 Bowers Avenue
Santa Clara, CA 95051
http://www.amfit.com

Langer Biomechanics Group Ltd.,
Deer Park, NY
http://www.langerbiomechanics.com/

Tekscan, Inc.
307 West First Street
South Boston, MA 02127-1309
http://www.tekscan.com

APPENDIX A

INFORMATION LETTER

"THE EFFECTIVENESS OF THE KINETIC WEDGE FOOT ORTHOSES MODIFICATION TO IMPROVE POSTURE DURING WALKING"

Investigator: Kerry K. Rambarran B.Sc.(Hons)
Masters Candidate (Biomechanics),University of Ottawa,

Advisors:
D. Gordon E. Robertson PhD, University of Ottawa
Edward Lemaire PhD, **The Rehabilitation Centre**,

The Research Ethics Board Chair:
Dr. Shawn Marshall, **The Rehabilitation Center**

Study Description
The purpose of this study is to investigate how people walk with Custom Foot Orthoses (custom insoles). More precisely, the study is mainly concerned with changes in posture during walking, with the use of a pair of custom orthoses **(with a "kinetic wedge"modification)**.

Research Study Summary:
You are being **asked to participate** in this study as you were diagnosed with Functional Hallux Limitus (FHL) by a Chiropodist during a visit to the Total Foot Care Clinic at **The Rehabilitation Centre**. To help with this condition you have been prescribed a pair of custom foot orthoses.

As procedure of Total Foot Care, you will be asked to return to **The Rehabilitation Centre** three weeks after receiving your custom insoles. At this time you will also be invited into the Gait and Motion Analysis Lab at **The Rehabilitation Centre**. This will be the site for conducting the study.

As a participant you will wear a special insole (in your shoe) to measure pressures under your foot. The insole will be used to identify various events during each step.

Participants will also be asked to wear reflective markers on their body. Markers will be placed on: 1) the shoulder, 2) hip, 3) knee, 4) ankle, 5) heel, 6) ball of the foot and 7) toe. Markers 5, 6, and 7, will be placed directly on your shoe. Back tape may be placed on your shoes. This ensures better images of your foot while videotaping.

For proper placement of markers you will be asked to wear black shorts and a black sleeveless shirt. Clean shorts and shirts (laundered after each use) are available in the event you do not have either.

While wearing the markers and pressure insole, you will be filmed using a video camera. You will be given time to become accustomed to wearing your custom orthoses with the pressure sensing insole in your shoe.

Participants will be asked to walk trials along a 7-metre walkway. You will be asked to walk with your soles (in your shoe) while being video taped. Your insole will then be adjusted. You will then be given time to become accustomed to the adjustment. You will then be asked to walk while being taped again.

The total time for this procedure should last approximately 1 hour. **Your travel and/or parking fees will be reimbursed to you for this time.**

Risks / Discomforts and Benefits
There are no anticipated risks in this research project.

Client Information
To better understand the individual needs of each client, it may be necessary to consult your medical chart. Only information pertinent to the study will be used, such as: height, weight, shoe size, and features of your custom orthoses.

Right of Confidentiality
All information will be collected and used for research purposes only. All information will be treated as confidential. Complete anonymity, and the confidentiality of the data will be maintained by the researcher at all times. All client data will be reported in numerical form. Each participant will be designated a unique numerical code. All data will be presented in a pooled form. No clients' identities will be disclosed outside of the laboratory/clinic. All raw data including video tapes will be secured in locked file cabinets.

Rights as a Volunteer

As a volunteer in this study you are under no obligation to participate, and may withdraw at any time. Refusal to participate or withdrawing from the study will in no way affect present and/or future treatment at **The Rehabilitation Centre**.

Contact Information

Participants have the right to speak to the principal investigator, Gordon Robertson, Edward Lemaire, and Shawn Marshall **Chair of the Research Ethics Board** concerning participation in this study, and can withhold participation until any questions or concerns have been addressed. All contact information for each of these individuals can be found on page 1.

APPENDIX B

CONSENT FORM

"THE EFFECTIVENESS OF THE KINETIC WEDGE FOOT ORTHOSES MODIFICATION TO IMPROVE POSTURE DURING WALKING"

Investigator: Kerry K. Rambarran B.Sc.(Hons)
Masters Candidate (Biomechanics),University of Ottawa,

Advisors:
D. Gordon E. Robertson PhD, University of Ottawa,
Edward Lemaire PhD, The Rehabilitation Centre,

Research Ethics Board Chair:
Dr. Shawn Marshall, **The Rehabilitation Center**

I ______________________________ (print name) understand that this study part of a Masters of Arts thesis at the University of Ottawa. The purpose of this study is to find out how a "kinetic wedge" modification to your orthosis changes your walking style.

During testing, a special insole will be put in my shoe and reflective markers will be taped on my clothing or on my skin. I will be asked to walk along a walkway while we collect pressure data between your foot / shoe and video tape to see my walking style. The testing session lasts approximately 60 minutes. More detailed explanations are on the information sheet.

I understand that if I need any further explanation of these risks I can seek the advice of a qualified medical practitioner before participating in the study. The researchers will advise me if any new information has become available that may effect your decision to continue in the study.

The data collected during this project will be stored at **The Rehabilitation Centre**. All identities will be kept confidential and only used by the project team during data analysis. The results from this project will be presented at scientific conferences and published in scientific journals and as a thesis. I understand that all raw data will be destroyed 10 years after completion of the study.

I have received a copy of this consent form as well as a copy of the background information concerning this study.

"I agree to participate in this study with the understanding that information will be collected and used for research purposes only and will be treated as confidential. I have been informed about the purpose of this study and realize that I am under no obligation to participate and may withdraw at any time. Refusal to participate or withdrawing from the study will in no way affect my present and/or future treatment at The Rehabilitation Centre."

Volunteering Subject:______________________ **Date:**____________

Print name: ______________________

Signature of Witness:______________________ **Date:**____________

Print name: ______________________

Signature of Investigator:__________________ **Date:**____________

COPY

The Rehabilitation Centre

Le Centre de réadaptation

April 4, 2002

Re: Research Ethics Review of project entitled "**The Effectiveness of the Kinetic Wedge Foot Orthoses Modification to Improve Posture During Walking** "

Dear Mr. Rambarran:

The REB approved the proposed protocol providing the following conditions are met before proceeding:

- A budget needs to be submitted for review.
- Under ethical considerations 6.1 B - the wording should be changed to "the ability to provide better treatment for foot pathology" instead of "ability to better diagnose the foot pathology". This similar change should be made in the literature review on page 4.
- The summary page 12 line 6 should read "three weeks <u>after</u>".
- On page 3 under statement of the problem - line 4, the word "those" should be changed to "these".
- On page 1 of methodology, under subjects - line 2 "for the" needs to be removed and the phrase " and volunteer students of the University of Ottawa" needs to be removed.
- Information letter - all references to the Research Ethics Committee should be changed to Research Ethics Board.
- All "Rehabilitation Centre" should read "The Rehabilitation Centre" with capital letters on each word.

.../2

April 4, 2002

- On the information letter under the heading "Research Study Summary" it should read "you are being asked to participate in this study" versus "you were chosen for this study.

- Also within the same paragraph, it needs to be clarified that the visit could have been at the Total Foot Care Clinic or Curryer Specialists.

- Under the subheading Client Information the word "clients" should be changed to "client". Similarly , under Study Description "individuals" should be changed to "individual's". Also in the Information Letter, travel reimbursement needs to be clarified.

- For the Information Letter, it should be clarified that the shoes may be taped with hockey tape or electrical tape. Further, black shorts and sleeveless T-shirt may be supplied if the patient does not bring their own and these would need to be laundered between usages.

- Consent Form. Once again, Research Ethics Committee referral should be changed to Research Ethics Board throughout the letter. Also, the phrasing should be done in the first person consistently versus using second person references.

- Also, the purpose of the study varies between the Consent Form and the Information Letter, this should be consistent and made consistent with the protocol.

Revisions are to be submitted to the Chair, of the Research Ethics Board.

Sincerely,

Shawn Marshall MD MSc FRCPC
Chair, Research Ethics Board
The Rehabilitation Centre

c.c.: Jeff Blackmer, Vice-Chair, Research Ethics Board
J.C. MacDougall, Director of Research

The Rehabilitation Centre Le Centre de réadaptation

APPENDIX C

THANK-YOU FOR VOLUNTEERING YOUR TIME TO FURTHER THE BODY OF KNOWLEDGE ON FOOT CARE.

PLEASE TAKE A MOMENT TO COMPLETE THIS BRIEF QUESTIONNAIRE.

WE APPRECIATE YOU TAKING THE TIME TO DO SO, AS YOUR COOPERATION WILL AID US IN PROVIDING YOU WITH THE BEST CARE POSSIBLE.

NAME:____

PLEASE CIRCLE YOUR ANSWERS.

1. DO YOU EXPERIENCE PAIN IN YOUR FIRST (BIG) TOE JOINT?

   YES NO

2. DO YOU EXPERIENCE YOUR PAIN AT ALL TIMES?

   YES NO

IF YOU ANSWERED NO TO QUESTION 2:

3. DO YOU EXPERIENCE PAIN MAINLY DURING ACTIVITY (I.E., WALKING OR RUNNING)?

   YES NO

4. COULD YOU EXPRESS THE SEVERITY OF YOUR PAIN ON A SCALE OF 0 TO 10?

   0 = NO PAIN **10 = PAIN AS BAD IT COULD BE**

   0 1 2 3 4 5 6 7 8 9 10

APPENDIX D

Survey Results

	Perceived Pain			
Subject	Before	After	Difference	Rank
1	0	0	0	-
2	0	0	0	-
3	0	0	0	-
4	0	0	0	-
5	0	0	0	-
6	0	0	0	-
7	2	1	-1	1.5
8	2	1	-1	1.5
9	4	2	-2	3
10	4	1	-3	4
11	4	0	-4	5.5
12	6	2	-4	5.5
13	7	3	-4	5.5
14	7	3	-4	5.5
			$\sum R_-$	32
			n	14
			T	32
			p	0.05
			T crit	21

Perceived pain scores between initial assessment and after two months of modified CFO use.

The results showed no significant reduction in perceived pain after 2 months, $T = 32.0$, $p > 0.05$, with the ranks of the decreases totalling 32.0

Appendix E

Ensemble Averaged Plantar Pressure Data
of all trials (75) for both conditions during stance

	MTP Joint		Hallux		5th Metatarsal	
	kg/cm^2	kg/cm^2	kg/cm^2	kg/cm^2	kg/cm^2	kg/cm^2
% Stance	NKW	KW	NKW	KW	NKW	KW
1	0	0	0.01	0	0.037	0.048
2	0	0	0.009	0	0.043	0.052
3	0	0	0.009	0	0.05	0.057
4	0	0	0.008	0	0.056	0.064
5	0	0	0.008	0.004	0.062	0.07
6	0	0	0.009	0.005	0.072	0.08
7	0	0	0.011	0.006	0.084	0.091
8	0	0	0.01	0.008	0.106	0.105
9	0	0	0.013	0.012	0.125	0.126
10	0	0	0.014	0.016	0.152	0.162
11	0	0	0.015	0.018	0.18	0.203
12	0	0	0.019	0.024	0.208	0.243
13	0	0	0.024	0.027	0.237	0.266
14	0	0	0.026	0.029	0.261	0.297
15	0	0	0.03	0.034	0.295	0.336
16	0	0	0.039	0.045	0.328	0.368
17	0	0	0.057	0.053	0.36	0.396
18	0	0	0.076	0.068	0.392	0.423
19	0	0	0.088	0.08	0.421	0.456
20	0.016	0	0.119	0.093	0.446	0.484
21	0.048	0	0.149	0.104	0.47	0.508
22	0.059	0	0.171	0.117	0.492	0.532
23	0.06	0.006	0.194	0.135	0.513	0.557
24	0.071	0.023	0.209	0.144	0.531	0.577
25	0.079	0.017	0.221	0.153	0.547	0.596
26	0.076	0.018	0.239	0.169	0.563	0.614
27	0.078	0.028	0.253	0.177	0.582	0.631
28	0.088	0.037	0.263	0.183	0.598	0.649
29	0.095	0.035	0.277	0.189	0.614	0.663
30	0.104	0.037	0.287	0.196	0.631	0.679
31	0.115	0.04	0.299	0.206	0.65	0.693
32	0.126	0.041	0.309	0.214	0.668	0.709
33	0.136	0.042	0.317	0.22	0.685	0.722
34	0.146	0.049	0.327	0.226	0.701	0.736
35	0.16	0.061	0.335	0.232	0.717	0.755
36	0.172	0.076	0.347	0.236	0.731	0.769
37	0.184	0.081	0.354	0.239	0.742	0.781
38	0.198	0.086	0.366	0.244	0.756	0.795

39	0.224	0.085	0.378	0.249	0.769	0.81
40	0.243	0.093	0.387	0.259	0.787	0.823
41	0.258	0.108	0.396	0.266	0.803	0.837
42	0.294	0.108	0.41	0.275	0.817	0.853
43	0.326	0.124	0.423	0.284	0.831	0.871
44	0.332	0.139	0.436	0.294	0.849	0.886
45	0.342	0.15	0.449	0.306	0.867	0.9
46	0.354	0.148	0.462	0.317	0.883	0.917
47	0.369	0.163	0.477	0.327	0.9	0.93
48	0.399	0.176	0.499	0.336	0.917	0.944
49	0.431	0.181	0.519	0.351	0.933	0.961
50	0.457	0.193	0.541	0.369	0.952	0.977
51	0.478	0.198	0.556	0.385	0.971	0.991
52	0.508	0.21	0.583	0.401	0.989	1.005
53	0.541	0.228	0.607	0.419	1.008	1.018
54	0.57	0.24	0.634	0.443	1.027	1.037
55	0.6	0.249	0.655	0.461	1.047	1.054
56	0.635	0.282	0.683	0.484	1.071	1.068
57	0.671	0.302	0.712	0.505	1.092	1.086
58	0.712	0.315	0.747	0.532	1.114	1.105
59	0.756	0.335	0.785	0.56	1.139	1.125
60	0.796	0.377	0.824	0.593	1.164	1.147
61	0.823	0.427	0.862	0.622	1.192	1.171
62	0.859	0.44	0.909	0.655	1.221	1.197
63	0.901	0.465	0.963	0.695	1.252	1.221
64	0.937	0.492	1.016	0.737	1.283	1.25
65	0.969	0.527	1.068	0.781	1.317	1.285
66	1.01	0.563	1.125	0.831	1.352	1.32
67	1.04	0.597	1.178	0.883	1.392	1.351
68	1.076	0.646	1.246	0.941	1.427	1.387
69	1.12	0.7	1.316	1.003	1.463	1.421
70	1.16	0.758	1.391	1.075	1.498	1.455
71	1.205	0.816	1.469	1.151	1.537	1.489
72	1.257	0.882	1.552	1.225	1.574	1.527
73	1.308	0.949	1.645	1.306	1.601	1.561
74	1.36	0.992	1.738	1.385	1.627	1.594
75	1.416	1.044	1.835	1.469	1.647	1.623
76	1.477	1.1	1.935	1.559	1.663	1.644
77	1.533	1.147	2.031	1.645	1.675	1.658
78	1.582	1.194	2.125	1.733	1.676	1.66
79	1.633	1.24	2.219	1.815	1.666	1.653
80	1.66	1.286	2.308	1.89	1.645	1.64
81	1.666	1.315	2.382	1.96	1.61	1.617
82	1.653	1.327	2.44	2.022	1.558	1.576
83	1.628	1.326	2.481	2.073	1.493	1.514
84	1.607	1.31	2.508	2.114	1.414	1.444

85	1.565	1.282	2.507	2.135	1.329	1.359
86	1.498	1.219	2.471	2.135	1.23	1.262
87	1.429	1.131	2.41	2.113	1.118	1.158
88	1.378	1.026	2.31	2.066	1.001	1.04
89	1.331	0.924	2.178	1.99	0.889	0.916
90	1.25	0.82	2.007	1.879	0.778	0.797
91	1.161	0.715	1.799	1.72	0.663	0.681
92	1.067	0.637	1.556	1.532	0.56	0.576
93	0.937	0.563	1.283	1.297	0.464	0.487
94	0.788	0.47	1.003	1.028	0.377	0.401
95	0.632	0.386	0.748	0.76	0.305	0.331
96	0.483	0.312	0.522	0.528	0.26	0.275
97	0.348	0.247	0.342	0.345	0.216	0.228
98	0.238	0.201	0.211	0.212	0.178	0.182
99	0.155	0.138	0.122	0.128	0.141	0.14
100	0.096	0.066	0.04	0.048	0.107	0.126
max	1.666	1.327	2.508	2.135	1.676	1.66
min	0	0	0.008	0	0.037	0.048

Appendix F

Ensemble Averaged Segment and Joint angle Data
of all trials (75) for both conditions during stance

	Ankle		Knee		Hip		Trunk	
	deg	deg	deg	deg	deg	deg	deg	deg
% Stance	NKW	KW	NKW	KW	NKW	KW	NKW	KW
1	2.856379	3.059685	3.816647	6.017602	19.34804	19.66695	91.91401	92.18183
2	2.458654	2.607761	4.668146	6.725625	19.35135	19.57572	91.98793	92.27533
3	1.899552	2.018224	5.568588	7.484451	19.30371	19.44688	92.07831	92.38163
4	1.188197	1.300455	6.500557	8.280289	19.20273	19.27823	92.18479	92.50065
5	0.37198	0.511521	7.365831	9.042471	19.02911	19.06625	92.30996	92.62741
6	-0.550605	-0.331034	8.154553	9.773178	18.78088	18.81808	92.45474	92.75899
7	-1.480536	-1.153519	8.904953	10.49518	18.48984	18.54961	92.60937	92.89158
8	-2.376787	-1.907901	9.609674	11.21331	18.15692	18.272	92.77361	93.01963
9	-3.182646	-2.548175	10.31981	11.96146	17.81521	18.00276	92.93654	93.13877
10	-3.821171	-3.024947	11.07454	12.75566	17.48726	17.74978	93.08965	93.24597
11	-4.280298	-3.325019	11.87713	13.59518	17.18183	17.51542	93.22644	93.33847
12	-4.524811	-3.431276	12.75108	14.48232	16.9187	17.30419	93.33686	93.41273
13	-4.556981	-3.356408	13.68229	15.39193	16.69488	17.10692	93.41749	93.46803
14	-4.400562	-3.119861	14.64014	16.29544	16.50825	16.91462	93.46159	93.50303
15	-4.075704	-2.746619	15.60588	17.16973	16.35187	16.72411	93.46995	93.51451
16	-3.623724	-2.270009	16.53446	17.98228	16.21034	16.52171	93.44207	93.50358
17	-3.079259	-1.718793	17.40065	18.71435	16.07595	16.30307	93.37835	93.46925
18	-2.472366	-1.11641	18.18308	19.35861	15.93337	16.0683	93.2861	93.41048
19	-1.832995	-0.484989	18.86355	19.90904	15.77192	15.81333	93.16868	93.329
20	-1.178516	0.161464	19.44126	20.37359	15.58734	15.54044	93.03022	93.2272
21	-0.523288	0.81261	19.91694	20.75773	15.36794	15.25192	92.88094	93.10592
22	0.123079	1.457161	20.2993	21.07293	15.11555	14.94768	92.72184	92.97036
23	0.756763	2.091303	20.59792	21.32496	14.82772	14.62661	92.55881	92.82438
24	1.370716	2.706038	20.82347	21.51752	14.50561	14.28808	92.39566	92.6712
25	1.964601	3.295919	20.98713	21.65568	14.1543	13.93004	92.23261	92.51553
26	2.534413	3.85898	21.09541	21.73831	13.77931	13.5485	92.07002	92.36099
27	3.076255	4.388147	21.15485	21.76689	13.38227	13.14568	91.91062	92.20762
28	3.591163	4.882173	21.17195	21.74482	12.97185	12.72203	91.75087	92.05769
29	4.075564	5.341566	21.14813	21.67301	12.54869	12.27796	91.59291	91.91121
30	4.529293	5.764844	21.08752	21.5599	12.11516	11.82269	91.43657	91.7652
31	4.956012	6.153019	20.99503	21.40809	11.67525	11.35596	91.28047	91.62137
32	5.355889	6.510235	20.87304	21.22532	11.22747	10.88245	91.12644	91.47831
33	5.732869	6.839211	20.72683	21.02083	10.77515	10.40984	90.97274	91.33404
34	6.091898	7.141633	20.56042	20.79526	10.31803	9.935278	90.81965	91.1904
35	6.433956	7.425381	20.37754	20.55955	9.856755	9.463868	90.66741	91.0467
36	6.763093	7.693138	20.18105	20.31638	9.393723	8.99891	90.51411	90.9009
37	7.080287	7.946654	19.97132	20.06672	8.927357	8.537613	90.36041	90.7546
38	7.386223	8.193184	19.7528	19.81609	8.460881	8.082838	90.20546	90.60656

39	7.682126	8.431955	19.52418	19.56348	7.995902	7.635148	90.04773	90.45543
40	7.967484	8.665403	19.28721	19.30853	7.531601	7.19302	89.88852	90.30131
41	8.244018	8.897339	19.04662	19.05302	7.073597	6.756197	89.72646	90.14474
42	8.512477	9.128282	18.80043	18.79795	6.619144	6.326266	89.56303	89.984
43	8.772249	9.360223	18.55165	18.54371	6.169998	5.903673	89.39839	89.819
44	9.027992	9.597167	18.30467	18.29506	5.727242	5.489056	89.23362	89.65086
45	9.278747	9.839423	18.05623	18.05382	5.286179	5.084373	89.07058	89.47834
46	9.526543	10.08894	17.8116	17.82268	4.85037	4.691247	88.90786	89.30182
47	9.775621	10.3482	17.57268	17.60436	4.417238	4.309213	88.7473	89.12235
48	10.02553	10.61445	17.33982	17.39756	3.987132	3.936895	88.58743	88.94037
49	10.27832	10.88999	17.11865	17.20337	3.565468	3.573879	88.42566	88.7568
50	10.53713	11.17346	16.9107	17.02009	3.149887	3.217509	88.26324	88.57256
51	10.80106	11.46288	16.71766	16.84532	2.743491	2.863441	88.09799	88.38979
52	11.07039	11.75858	16.54042	16.6786	2.346768	2.510515	87.92994	88.20898
53	11.34824	12.05888	16.37934	16.51894	1.954322	2.156661	87.76276	88.0309
54	11.63113	12.36066	16.23042	16.36322	1.563081	1.796826	87.59752	87.85798
55	11.92161	12.66461	16.09244	16.21377	1.16806	1.432477	87.43768	87.68996
56	12.21921	12.96669	15.96268	16.06813	0.761577	1.060404	87.28732	87.52801
57	12.52349	13.26582	15.83871	15.92632	0.340646	0.678056	87.14817	87.37422
58	12.8336	13.56099	15.71877	15.78999	-0.096753	0.286839	87.02095	87.228
59	13.1499	13.85051	15.60531	15.65886	-0.549277	-0.115121	86.90538	87.09065
60	13.46939	14.13629	15.49851	15.53715	-1.012952	-0.525409	86.79928	86.96155
61	13.78876	14.4174	15.40065	15.42635	-1.480682	-0.94304	86.69864	86.8406
62	14.10762	14.69746	15.31745	15.33159	-1.94574	-1.364927	86.60126	86.72698
63	14.42101	14.9778	15.25062	15.25824	-2.402951	-1.787546	86.50472	86.61976
64	14.72719	15.25865	15.20578	15.20878	-2.844127	-2.208856	86.40553	86.51886
65	15.02672	15.5441	15.1884	15.19132	-3.269513	-2.624476	86.30593	86.42372
66	15.31763	15.83058	15.20312	15.20741	-3.676361	-3.032845	86.20558	86.33426
67	15.60117	16.11903	15.25602	15.26253	-4.063519	-3.430872	86.10534	86.25094
68	15.87861	16.40712	15.35264	15.36056	-4.433439	-3.816524	86.00851	86.17414
69	16.14952	16.68937	15.49977	15.50256	-4.785144	-4.188034	85.91595	86.10348
70	16.41225	16.96309	15.70089	15.69334	-5.117264	-4.542185	85.82788	86.03901
71	16.66448	17.2221	15.95898	15.93273	-5.43295	-4.878682	85.74725	85.98066
72	16.90029	17.46012	16.27724	16.22371	-5.729789	-5.192908	85.67356	85.92643
73	17.11382	17.67118	16.65443	16.56724	-6.008524	-5.485476	85.60774	85.87701
74	17.29494	17.84954	17.08737	16.96356	-6.274027	-5.75672	85.55248	85.83263
75	17.43731	17.98695	17.57806	17.41337	-6.523713	-6.005293	85.50718	85.79246
76	17.52919	18.07749	18.11963	17.91629	-6.763429	-6.236573	85.47501	85.7599
77	17.56161	18.11391	18.71195	18.47121	-6.994014	-6.4526	85.45636	85.73612
78	17.52831	18.08582	19.35743	19.07717	-7.213743	-6.656019	85.45166	85.72344
79	17.41694	17.98625	20.05358	19.7347	-7.42604	-6.851349	85.46368	85.72566
80	17.22144	17.8051	20.80736	20.44304	-7.624379	-7.039731	85.4903	85.74508
81	16.93321	17.52975	21.62327	21.20707	-7.808277	-7.219036	85.53462	85.78397
82	16.54009	17.15454	22.50796	22.03091	-7.971717	-7.388641	85.59629	85.84479
83	16.03595	16.66618	23.4706	22.92433	-8.107681	-7.539355	85.67546	85.92747
84	15.41052	16.05801	24.52034	23.89511	-8.210325	-7.664848	85.77372	86.03217

	85	14.65405	15.32301	25.66717	24.95795	-8.268518	-7.74989	85.88952	86.1553
	86	13.76338	14.45421	26.92156	26.12007	-8.27275	-7.785672	86.02187	86.29529
	87	12.73249	13.44896	28.29124	27.39239	-8.211728	-7.75696	86.16832	86.4471
	88	11.55721	12.30424	29.78372	28.7804	-8.073973	-7.652677	86.32523	86.60601
	89	10.23816	11.01881	31.40113	30.28368	-7.849646	-7.465831	86.48833	86.76883
	90	8.772376	9.59287	33.14048	31.8993	-7.532529	-7.189773	86.65366	86.93126
	91	7.156336	8.024129	34.99179	33.61831	-7.118126	-6.820024	86.81622	87.08858
	92	5.390085	6.316335	36.94696	35.42896	-6.605591	-6.35943	86.97326	87.23939
	93	3.489892	4.480054	38.98246	37.31497	-5.998671	-5.805764	87.12067	87.37803
	94	1.453483	2.518969	41.0843	39.2631	-5.299662	-5.163384	87.2559	87.50291
	95	-0.670006	0.481992	43.23592	41.25657	-4.514598	-4.435362	87.37781	87.61085
	96	-2.805653	-1.578399	45.40896	43.27485	-3.64984	-3.628141	87.48253	87.69985
	97	-4.93941	-3.629632	47.60317	45.31321	-2.707564	-2.743685	87.57128	87.76958
	98	-6.867279	-5.479748	49.77801	47.3347	-1.70114	-1.799547	87.64283	87.82226
	99	-8.543629	-7.093932	51.92654	49.33308	-0.634166	-0.799904	87.69805	87.85965
	100	-10.00391	-8.50421	54.05557	51.31458	0.495849	0.25838	87.73694	87.88117
max		17.56161	18.11391	54.05557	51.31458	19.35135	19.66695	93.46995	93.51451
min		-10.00391	-8.50421	3.816647	6.017602	-8.27275	-7.785672	85.45166	85.72344