EFFECT OF FOOT SUPPORTS ON KNEE JOINT LOADING

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EFFECT OF FOOT SUPPORTS ON KNEE JOINT LOADING

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A thesis submitted in partial fulfilment of the requirements for

the Degree of Master of Philosophy

March 2011

CERTIFICATE OF ORIGINALITY

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Abstract of thesis entitled

Effect of foot supports on knee joint loading

Submitted by

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Knee pain is one of the most prevalent problems that impairs knee functions. Mechanical loading is one of the most important determinants of knee pain. Changing the alignment of lower limb may vary knee loading. Clinicians prescribe wedged insoles primarily for pain-relieving. However, how wedged insoles influence the load transfer through foot-ankle-knee have not been fully elucidated. The purpose of this study is to systematically understand the effectiveness of wedged insoles on lower-limb alignment and knee loading. Experimental gait analysis and multi-body model of wedged lower-limbs were used. Six different wedges were analyzed for their effects on the center of pressure, joint angles and the joint moments.

Ten healthy female subjects participated in this experiment. Their gait patterns were assessed with six different wedged conditions: forefoot lateral wedge (LF), forefoot medial wedge (MF), rearfoot lateral wedge (LR), rearfoot medial wedge (MR), full-length lateral wedge (LW) and full-length medial wedge (MW) in addition to a control flat insole (FF). They walked with a controlled cadence of 120 step/min and were assigned with random insole each time. Gait was monitored using Motion Analysis System with eight infrared cameras and

two force platforms. Forty-four reflective markers were attached on the pelvis, thigh, shank, forefoot and rearfoot segment, to monitor lower extremity motion. The multi-body musculoskeletal model was developed using OpenSim (SimTK.org) to estimate the joint and major muscle forces at the knee. The gait data were input into this model to obtain the joint and muscle forces during walking. The kinematic and kinetic data were analyzed by one-way repeated measures analysis of variance.

The results showed that LW and MF shifted the center of pressure laterally up to 0.18% foot width. The first peak knee adduction moment was reduced by an average of 8.75% in all wedged insole conditions. For the second peak, the knee adduction moment was reduced by 34.00% in the LW condition and by 9.80% in the MW condition, while the moment was increased in the MR group (1.96%). It was suggested that certain medial wedge could be used in the treatment of knee osteoarthritis. All conditions had an apparent relationship between the peak angles of forefoot dorsiflexion and the ankle eversion moment, between the ankle eversion moment and the ankle inversion angle and between the ankle inversion and the peak knee adduction moment. The data showed that the changes in knee adduction moment occurred due to the combined effects of increased forefoot dorsiflexion angle and decreased ankle inversion angle. These findings suggested that knee joint kinematics and kinetics were altered not only by the knee locomotion during walking, but also by responses of other parts of the lower limb, such as the ankle and forefoot, that responded to the influence of the wedged insoles. Adjusting the forefoot angle and the ankle moment were more efficient for reducing knee adduction moment compared to adjusting the center of pressure. These adjustments could be made by custom orthoses. Ankle joint and foot compensation should be considered when using wedged insoles.

Current findings indicated that proper-wedged support decreased peak knee adduction moment and might alleviate knee pain around the medial compartment where OA often occurs. These results provide information for designing foot supports for footwear and knee orthoses as well as providing information for the input and validations for computational models.

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CHAPTER I **INTRODUCTION**

1.1 Knee Osteoarthritis – Prevalence

Many people suffer from knee-related problems that may produce pain and influence normal activities (Woo et al., 2004, Hoaglund et al., 1973). Approximately 25% people over the age of 55 reported a significant episode of knee pain. The proportion of painful knee osteoarthritis (OA) in adults is high(Guccione et al., 1994).

In 2005, the World Health Organization (WHO) reviewed existing prevalence studies of OA. In China, more than 50% of the population over 65 years of age had OA, while Americans had an OA prevalence of 60% (male) and 70% (female) (Table 1-1). Although OA can occur in any joint, knee joint OA was the most commonly reported type (Dickson and Kim, 2003). The prevalence rates for OA knee are summarized in Figure 1-1. Knee OA significantly impairs mobility particularly for females (Symmons et al., 2003).

Table 1-1. Prevalence of OA in different countries/regions(Symmons et al., 2003)

Region	**Age (years. old)**	**Prevalence**
China	>65	>50% ~ 13% among total population
Taiwan	>50 >70	>50% >80%
USA	>65	Male: ~ 60% Female: ~ 70%
UK	45-65	~ 17%
World	>60	~ 10%
~, Approximately		

According to data collected in Hong Kong in 1995, 90% of the patients suffering OA knee pain were either middle aged (40-60 years) or older (over 60 years) (Leung et al., 1996). In the mid-2000s, the Census and Statistics Department of Hong Kong reported that middle-aged (40-60 years) and older (over 60 years) people represented

27.9% and 15% of the whole population, respectively. The middle-aged and older age groups were expected to increase to 30.8% and 18.8% of the total population by 2014, as projected by Hong Kong's population growth (Leung et al., 1996).

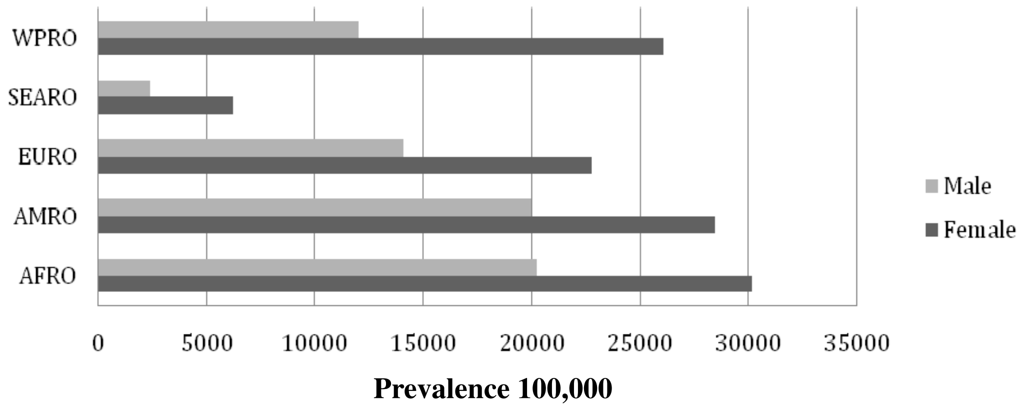

Figure 1-1. Prevalence of knee osteoarthritis in different WHO-defined regions: Africa (AFRO), the Americas/Pan American Health Organizations (AMRO), Europe (EURO), South-East Asia (SEARO) and the Western Pacific (WPRO) (Symmons et al., 2003)

The incidence and prevalence of knee OA increases with age. An increasing life expectancy without concurrent joint replacement support has become a problem in some developing countries. Gender also contributes to the risk of knee OA. Knee OA was more common in females than in males (Anthony et al., 2003) (Fig.1-2). One study estimated that approximately 10% of the population will have symptomatic OA by age 60. The studies were not consistent in comparing the prevalence rate of OA in developing versus developed countries. However, gender, ethnic and geographical differences were shown. Family history increases the risk of OA, while other factors, such as diabetes mellitus, osteoporosis, obesity, history of trauma, smoking, and habits such as wearing high-heeled shoes, are associated with developing OA. Workers who perform repeated knee bending for specific occupations, including miners(Kellgren and Lawrence, 1952), dock workers (Partridge and Duthie, 1968), and farmers (Croft et al., 1992), all presented a higher risk of OA.

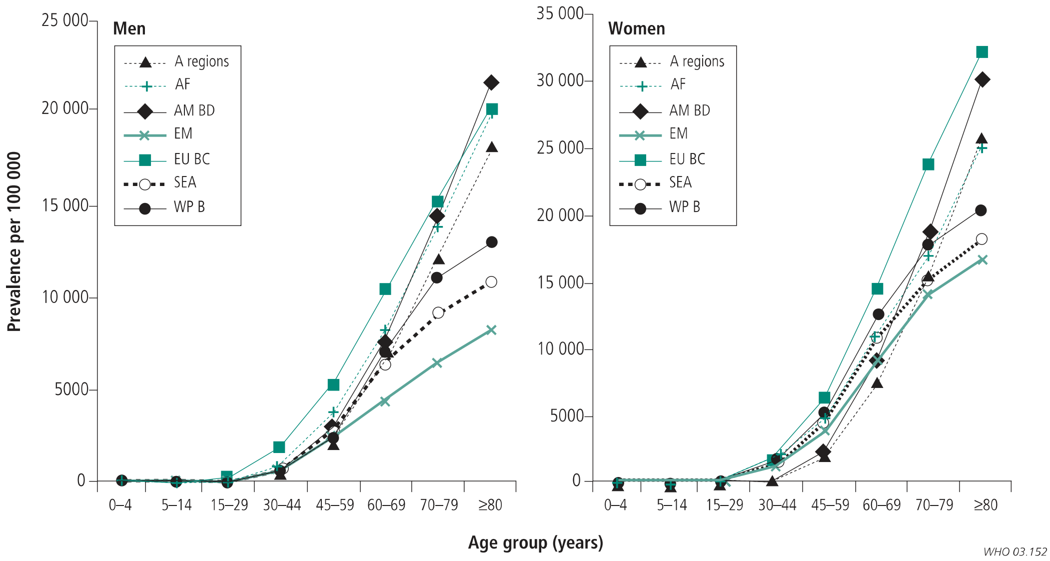

Figure 1-2. The prevalence of knee osteoarthritis by age group, sex, and region: A regions = developed countries in North America, Western Europe, Japan, Australia, New Zealand. AF = countries in sub-Saharan Africa. AM BE = developing countries in the Americas, E<= countries in the Eastern Mediterranean and North African regions. EU BC = developing countries in Europe. SEA = countries in Southeast Asia. WP B = countries in the Western Pacific region (Anthony et al., 2003).

1.2 Knee Osteoarthritis – Treatment

The treatment goals of OA knee must be individualized for each patient based on their pain and discomfort. Medication and physical therapy can help reduce pain and restore movement for stiff joints. It is difficult to cure osteoarthritis, and the degenerative process may often proceed. However, a well-prescribed treatment program can reduce pain, improve joint function and allow the patient to live more actively and comfortably.

Medical treatments for OA knee include patient education, exercise, weight reduction, quadriceps strengthening, and analgesics, such as paracetamol, non-steroidal anti-inflammatory drugs (NSAIDs), and nutriceuticals. However, when prescribing pain relief medication for people with OA knee, doctors should be aware that the administration of analgesia is associated with an increase in the magnitude of the knee adduction moment during gait, which may cause progression of medial tibiofemoral disease (Sharma et al. 1998).

Nonsurgical interventions, such as foot orthoses, have been developed for OA patients to alleviate painful symptoms while reducing knee joint load. Studies have reported that lateral wedge orthoses can reduce knee adduction moments, which can alleviate medial knee pain in OA patients (Kerrigan, 2002). Interventions that reduce the loading on the medial tibiofemoral compartment may have a disease-modifying effect in OA knee. Lateral foot wedges effectively reduced knee adduction moments (Hinman et al., 2009, Kuroyanagi et al., 2007). Modifying the toe-out angle also effectively changed knee adduction moments (Hinman et al., 2009, Kuroyanagi et al., 2007). However, it is currently unknown whether these interventions are clinically significant for people with established OA or individuals at risk of disease. More clinical trials and systematic biomechanical studies are needed to ascertain the consequences of such interventions and to identify the perpetrating biomechanics that lead to disease onset and/or progression before beginning therapy.

The purpose of this study is to establish biomechanical evidence of the effects and load transfer mechanisms of different foot supports during walking and standing using controlled experiments and musculoskeletal models. The quantitative information obtained from this study will provide guidelines for treatments and footwear design.

1.3 Objectives of this Study

The overall objective of this study is to establish biomechanical evidence for the effectiveness of the foot supports in reducing knee joint loading. The objective will be met by:

Investigating the load transfer mechanism and the biomechanical effects of different foot supports on walking via experiments and predicting knee joint loading via musculoskeletal model simulation.

1.4 Outline of the Dissertation

Chapter II starts with the functional anatomy of knee joint and then provides a review of the existing advances in foot orthotic and biomechanical research for knee joint loading, followed by a discussion of existing gait analysis and musculoskeletal models and a summary of the innovation of this study.

Chapter III presents the study methodology, including the subject information, the equipment and materials used in the experimental protocol, the data processing technique, and the statistical methods.

The results of the study are presented in Chapter IV. The first section reports on the kinetics and kinematic analyses results from one typical subject. The second section discusses the statistical analysis of all subjects. The final section presents the relationships between the kinetic and kinematic variables.

Chapter V discusses the results and findings of this study. The effect of lateral and medial wedges is discussed, and a comparison of current results with existing gait analyses is presented. The correlation between different biomechanical variables is analyzed to understand the transfer mechanism. Limitations of this study are also discussed.

Chapter VI provides conclusions from this project and offers suggestions for future developments.

CHAPTER II **LITERATURE REVIEW**

2.1 Functional Anatomy of Lower Extremities

The lower extremities are responsible for supporting the weight of the whole body. This part introduces the anatomy of the foot, ankle and knee, which are relevant to current research.

2.1.1 Functional Foot Anatomy

The foot is an incredibly complex mechanism. Each foot contains 26 bones, 33 joints, 107 ligaments and 19 muscles, and a network of blood vessels, nerves, skin, and other surrounding soft tissues (Fig. 2-1). These components work together to provide the body with support, balance and mobility. Figure 2-1 presents the top and side views of the bones in the human foot.

The foot is important for stability, which supports crucial functions such as weight bearing and propulsion. Additionally, the foot must be flexible to adapt to uneven surfaces. The multiple bones and joints of the foot provide the necessary flexibility and form an arch to support the body's weight.

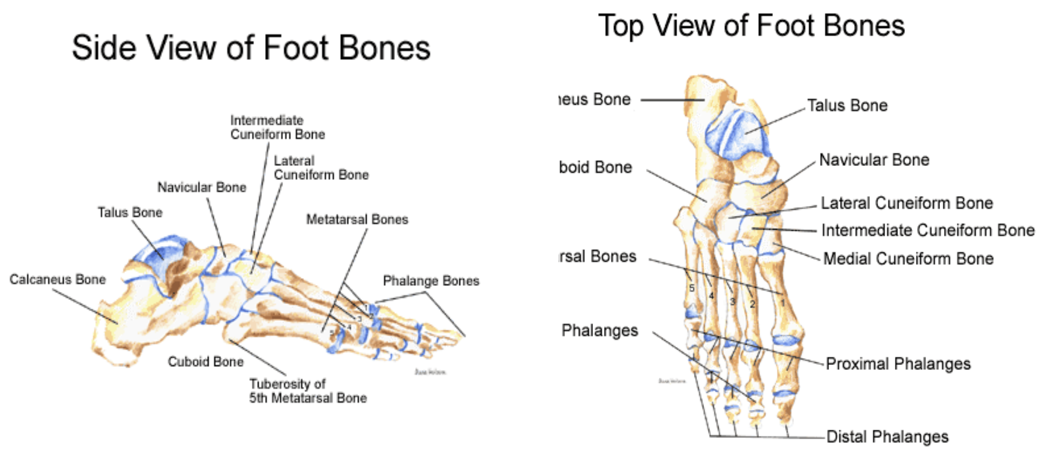

Figure 2-1. Anatomy of the foot bones Left: side view; Right: top view (http://www.podiatrychannel.com)

There are generally three arches in the foot: the lateral longitudinal arch, the medial longitudinal arch and the transverse arch (Fig. 2-2).

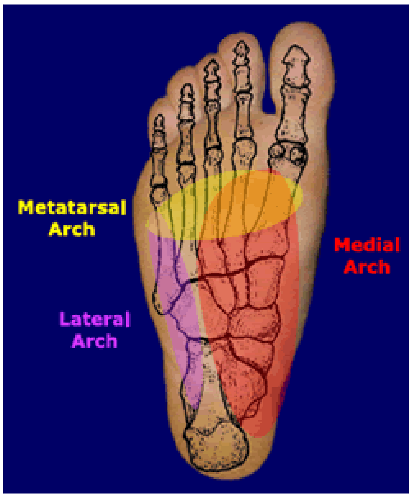

Figure 2-2. The lateral longitudinal arch, the medial longitudinal arch and the transverse arch (http://www.feetrelief.com)

The arch structure of the foot is not present at birth. Instead, it develops as the person grows. Arches usually develop by approximately 5 years of age (Woollacott and Assaiante, 2002). The medial longitudinal arch is the most apparent and important of the three arches. It is composed of the calcaneus, talus, navicular, cuneiforms and the first three metatarsals. The lateral longitudinal arch is lower and flatter than the medial arch. It is composed of the calcaneus, cuboid and the fourth and fifth metatarsals. The transverse arch runs across the midfoot, which is composed of the cuneiforms, the cuboid and the five metatarsal bases. Not only had the shape of the bones maintained the arches but also by the foot's soft connective tissues, such as the ligaments, muscles and tendons.

In the foot/ankle region, ankle joint, tarsal joints and metatarsal joints are easily identifiable, but the foot also has some smaller joints (Fig. 2-3). The movement of the joints is defined by their location within the foot structure. In addition, foot movements ordinarily are combined with ankle movements.

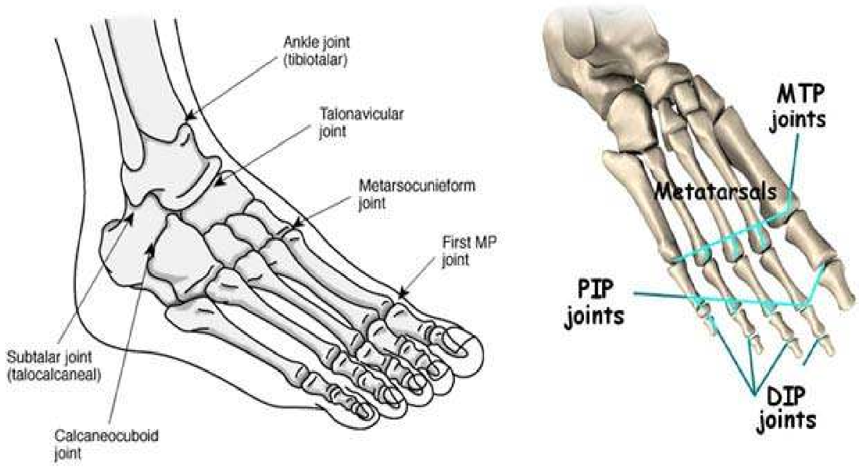

Figure 2-3. The location of different joints in the human foot (http://www.podiatrychannel.com)

Toe movements occur at the joints. These joints are capable of motion in two directions, plantarflexion or dorsiflexion. Additionally, the joints permit toe-in and toe-out. The foot as a whole has two movements: supination (or inversion) and pronation (or eversion) (Fig. 2-4).

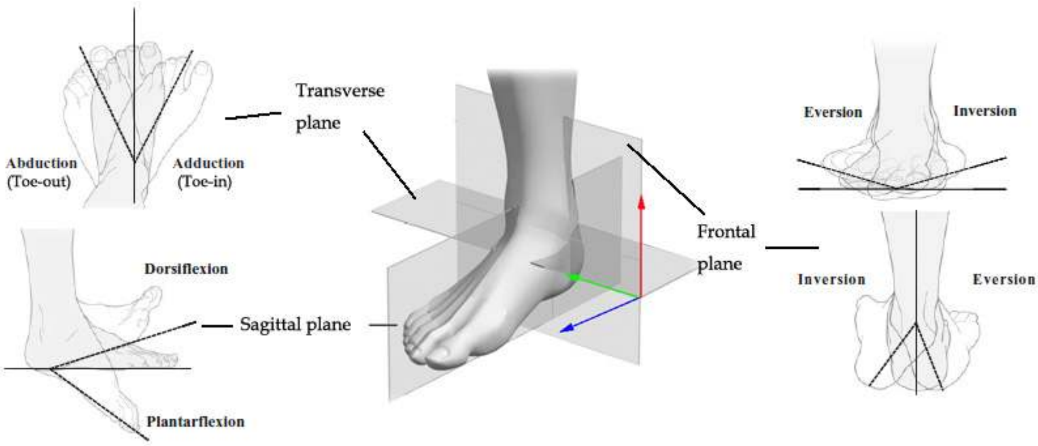

Figure 2-4. The movements of the foot (Doya et al., 2010)

All the joints of the hindfoot and midfoot from the subtalar contribute to these complex movements and multi-faceted movements. The location, movement and movement range of the main foot-ankle joints are shown in Table 2-1 (Muscolino, 2010).

Table 2-1. Main joints locations and movements (Muscolino, 2010)

Main Joint	**Location**	**Movement**
Ankle joint	Between distal tibia/fibula and talus	Plantarflexion 50°/ Dorsiflexion 20°
Subtalar joint (talocalcaneal joint)	Between talus and calcaneus (Major tarsal joint)	Eversion10°/Inversion 20°; Dorsiflexion2.5°/ Plantarflexion 5°; Abduction10°/Adduction 20°
Transverse tarsal joint	Between talus/calcaneus and navicular/cuboid (Important tarsal joint)	
Distal intertarsal joint	Other joints formed between the tarsal bones	
Tarsometatarsal (TMT) joints	Between the tarsal bones located proximally and the metatarsal bones located distally	Eversion/Inversion; Dorsiflexion/Plantarflexion
Intermetatarsal joints	Between metatarsal bones (Functionally related to MTM joints)	Plane synovial articulations
Metatarsophalangeal(MTP) joint	Between the metatarsal bones located proximally and the phalanges bones located distally	Toes 2-5：Extension60°/Flexion40°; Big Toe（Toe 1）：Extension80°/Flexion40°
Interphalangeal (IP) joints	Between phalanges	Allow flexion and extension within the sagittal plane around a mediolateral axis.
Proximal Interphalangeal (PIP) joints	Between the proximal and middle phalanges of toes#2-5	
Distal Interphalangeal (DIP) joints	Between the distal and middle phalanges of toes2-5	

2.1.2 Functional Knee Anatomy

The knee joint is a synovial joint, comprised of the distal femur, the proximal tibia and the patella. The interface between the patella and the femoral trochlea forms the patello-femoral joint. The medial and lateral femoral condyles, together with the respective articular surfaces of the tibial plateau, are termed as the medial and lateral tibiofemoral joints (Fig. 2-5).

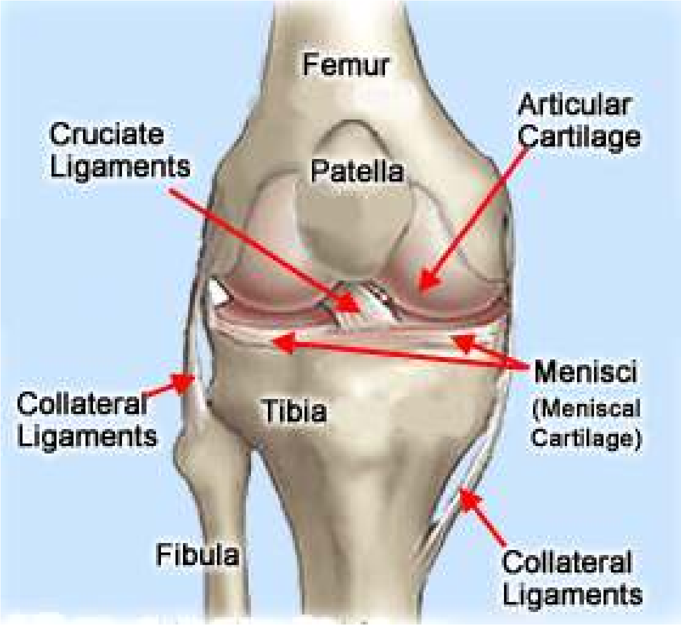

Figure 2 -5. Anatomical structure of knee joint (www. factotem.org)

The patella is the largest sesamoid bone in human body. It forms the extensor apparatus with the tendon of the quadriceps femoris muscles and the patellar tendon. One-quarter of its area comprises the insertion point of patellar tendon, beyond the joint's articular surface.

The stability of the knee joint depends on the special geometry of the bony structure, the constraint and balance of the cruciate ligaments and the strength of extensor apparatus, quadriceps and hamstring. In the knee joint, four main ligaments link the femur to the tibia and help stabilize the knee as it moves. These ligaments include the collateral ligaments along the medial and lateral sides of the knee, the anterior cruciate ligament (ACL) and the posterior cruciate ligament (PCL), which cross each other as they stretch diagonally from the bottom of femur to the top of tibia. A complete understanding of each individual ligament's role in restraining the knee's motion is essential for adequately diagnosing and assessing treatment procedures for a patient (Blankevoort and Huiskes, 1991).

The mechanical axis of the lower extremity is created by the center of the femoral head, the center of the knee joint and the center of the ankle joint in anatomical posture, with the transverse axis parallel to the ground. In fact, the axis of femur is approximately 6degrees varus with respect to the mechanical axis of the knee. The angle differs among individuals and depends on the femoral anteversion angle and the femur neck length. The joint line from the tibial plateau is approximately 3degrees valgus in balanced standing. Eventually, the asymmetry of the knee kinematics results in common medial compartment knee OA (Hurwitz et al. 2002).

In fact, the force vector of quadriceps and the tibial axis form a varus angle, Q-angle (Hungerford and Barry, 1979) (Fig. 2-6), which tends to dislocate the patella laterally. One study measured the alignment of the knee in a static state (Schollin-Borg et al., 2003). In OA patient also with degenerative patella, the lateral ligament is often stressed and the patella tends to be subluxed laterally.

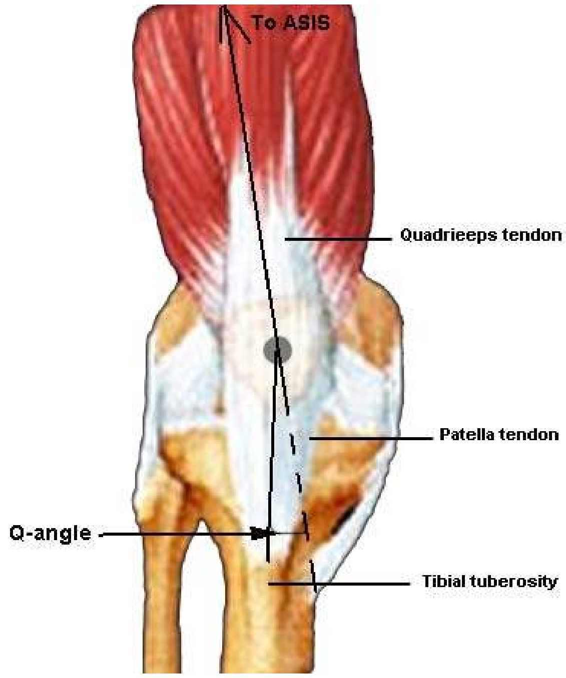

Figure 2-6. Q-Angle and the patellar tendon (www. factotem.org)

The maximum active flexion and maximum passive flexion of the knee is 135degrees and 160degrees, respectively, with a normal range of 145degrees flexion. When knee flexion is initialized, the femoral condyles begin a primary roll on the tibial plateau that occurs between 0 and 25degrees (Owolabi and Alawale, 1996). When knee flexion approaches 25degrees, the further rolling of the condyles tightens the cruciate ligaments and creates an anterior translational force. The displacement force results in a combined rolling and gliding action that further aided by the special geometry of the meniscus. The rolling and gliding action also prevents the condyles from rolling past the knee posteriorly. The knee extension action is the reversal of this mechanism.

The joint rotation is facilitated by the medial pivot geometry of the condyles because the medial condyle has a larger arc motion than its lateral counterparts based on its geometry. When the tibia rotates laterally, the lateral tibial condyle move a greater distance than the medial condyle, while during medial tibial rotation, the movement is the exact opposite. In a knee flexion of 90degrees, the tibia approaches its maximum rotation angle. The maximum medial angle is 15degrees, while the maximum lateral rotation is 20degrees. The flexion-rotation is also coupled with slight abduction or adduction. The maximum abduction angle is 13degrees, which occurs at 20degrees of flexion, while the maximum adduction (8degrees) occurs during full knee extension (Escamilla, 2001). These maximums are caused by the joint geometry that creates an oblique axis compared with the tibial plateau level.

2.1.3 Biomechanics of Lower Extremities

The biomechanics of the lower extremities can be modeled as joints connecting different rigid segments. The cup-like acetabulum of the pelvis and the rounded head of the femur form the hip joint mechanically as a ball-and-socket joint. All three rotary motions can occur at the joint- the flexion and extension move about a transverse axis, the adduction and abduction move about an anterior-posterior axis, and the medial and lateral motion about a vertical axis or the longitudinal axis of the femur (Rydell, 1973). Unlike the hip joint, the knee joint is more complex. It is formed by the shank and thigh segment and allows motion in three planes. Flexion and extension of the knee joint also include slight rotational and translational actions during movement, creating a complex motion at the center of the joint that is not easily identifiable. The ankle joint is formed by the medial and lateral malleoli (at the distal ends of the tibia and the fibula, respectively) and the talus bone of the foot. This joint normally allows plantar flexion and dorsiflexion about transverse axis, and inversion and eversion about an oblique axis (Doya et al., 2010). However, the foot pronation and supination movements are coordinated in three directions, with the relationship to the subtalar joint often described but hard to find. The metatarsal joint only allows one type of motion – rotation about a transverse axis – between the toes and the metatarsal bones.

Mathematical models could provide clinicians with detailed, patient-specific information about the internal forces in patient's' joints, as well as their spatial and temporal data. However, applying these models will first require a comprehensive validation. Muscles experience substantial mechanical loads due to both the tibiofemoral and patello-femoral joints, which has been modeled and validated with patient data (Besier et al. 2005). The degree of joint misalignment in specific patient may cause unpredictable change. Therefore, a sufficiently accurate model of knee joint kinematics is required to detail the joint's force distributions. By combining MR-based in vivo imaging techniques and data on the individual characteristics of knee kinematics, new mathematical models can help describe the effect of muscle activity on joint kinematics (Delp et al. 2007).

The magnitude of knee loading depends on individuals' daily activities, such as level walking, running and stair climbing. However, personal characteristics, such as walking habits, pace, body weight, and joint alignment (valgus/varus and tibial bowing) also contribute to the alteration of knee loading (Pandy and Andriacchi, 2010). Normally, knee joint loading is approximately 0.43times body weight (BW) in balanced standing and increases to 3.22times upon walking. Stair climbing increases the load up to 4.25times BW. In gait analysis, the knee loading starts to increase with the initial contact phase and reaches a maximum at the loading response phase (Kathleen et al., 2010). However, large shear and torque are experienced at the terminal stance. The contact pressure is reduced by the material properties of the meniscus and articular cartilage, which maximize the contact area between the joint. The uneven load distribution on the tibia shifts the center of mass medially to initiate single limb support. The posture is maintained by the tensor fascia latae and the gluteus maximus.

2.2 Review of Existing Research

2.2.1 Introduction

Scientists strive for better treatment guidelines for evaluating and managing OA because of its high prevalence in the general population. A meta-analysis conducted by (Zhang et al., 2007) indicated that the evidence for using insole modulation was applicable for treating knee OA. The same research team established an evidence-based and expert consensus guideline for managing knee OA (Zhang et al., 2008). According to the consensus, appropriate footwear and OA insoles should be one of the non-pharmacological modalities for treating OA to help relieve pain and improve ambulation. This recommendation was supported by the observational study (Zhang et al., 2008). Although patients showed no symptomatic improvements, they did reduce their NSAID dosage and showed better compliance. Zhang et al. (2008) also observed that existing studies lack controlled trials of footwear.

Most of the clinical research on knee OA used pain and functional assessments, such as the WOMAC (Western Ontario and McMaster Universities) Index, Lequesne Index, etc. Recently, quantitative assessment has gained popularity, with scientists looking for biomechanical evidence in presentation and evaluation of the disease. Gait analyses have been used to screen for disease.

The following sections will introduce existing research on knee OA and insoles, categorised in terms of pain assessment, radiographic assessment, gait analyses and musculoskeletal models.

2.2.2 Pain and Functional Assessment

The most commonly used pain assessment protocols are the WOMAC Index, the Lequesne Index and the Visual Analog Scale (VAS).

The WOMAC index evaluates patients' pain, functional ability and joint stiffness. The index uses 24 parameters, including social functions and emotional considerations. Table 2-2 presents an excerpt from the WOMAC index form.

Table 2-2. Excerpt form the WOMAC index (WOMAC, http://www.womac.org)

Functions/Scoring	**None**	**Slight**	**Moderate**	**Severe**
Pain Assessment				
Walking				
Stair Climbing				
………				
Stiffness				
Morning Stiffness				
Stiffness in the day				
Physical Functions				
Descending Stairs				
Ascending Stairs				
Rising from Sitting				
……..				

The lequesne Index (Lequesne,1997) is another index designed for evaluating osteoarthritis, particular for evaluating the effectiveness of interventions. It includes three sections: pain or discomfort, maximum distance walking and daily activities. The maximum index score is 24, with scores greater than 13 indicating an extremely severe handicap. Table 2-3 presents an excerpt from the Lequesne Index.

Table 2-3. Excerpt from the Lequesne index (Lequesne, 1997)

Parameters	Finding	Points
Pain or Discomfort		
Pain or discomfort during nocturnal bedrest	None	0
	Only on movement or in certain	1
	Without movement	2
……		
Maximum Distance Walked		
Maximum distance walked	Unlimited	0
	> 1 km but limited	1
	Approximately 1 km (about 15	2
	Approximately 500-900 m	3
	From 300-500 m	4
	From 100-300 m	5
	< 100 m	6
……		
Activities of Daily Living		
Can you put on socks by bending forward	Easily	0
	With mild difficulty	0.5
	With moderate difficulty	1.0
	With marked difficulty	1.5
	Impossible	2.0
……		

The unidimensional VAS is one of the most frequently used pain rating tests, with the advantage that no verbal or reading skills are required. It is especially useful in evaluating paediatric patients. However, there is large variation within the VAS, and most of the literature does not specify the type or characteristics of the chosen VAS. Figure 2-7 presents two of the common VAS in clinical research.

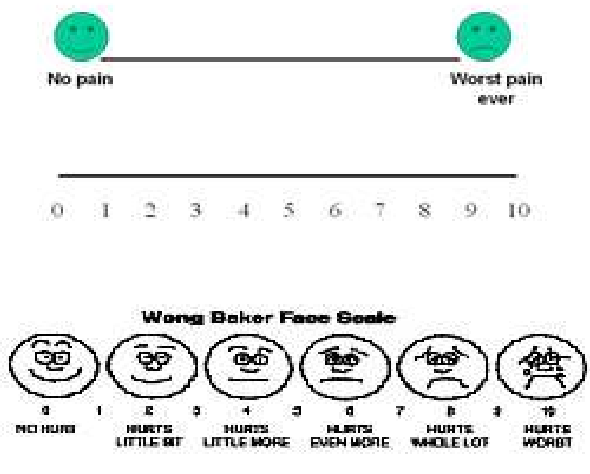

Figure 2-7. Examples of VAS (Department of Health, Australia)

Pain assessment is often used in evaluation of OA treatment. Toda et al. (2002) evaluated the treatment efficacy of wedged insoles combined with elevation and strapping with the patients' subjective information and discovered that the efficacy of symptom relief was related to a patients' age and Body Mass Index (BMI). An experiment looking at the relationship between knee adduction moment and a patient's pain and functional evaluation score showed that reducing the adduction moment had no effect on pain or functional improvement (Hinman et al., 2009). However, other studies presented a correlation index of 0.66 on WOMAC function score and 0.63 on WOMAC pain score with the knee adduction moment (Kim and Eng, 2004).

The results of existing pain and functional assessments showed large variation that may due to the differences in population, experimental settings, protocols and the duration of the experiments. The efficacy of wedged insoles was controversial. In the pain assessment by Roentgenogram classification, wedged insoles improved pain by 50%, with patients with milder OA receiving greater relief (Keating et al., 1993,Tohyama et al., 1991). Tohyama et al. (1991) compared the wedge insoles of an angle between 5 and 10degrees and compared them to treatment with analgesia alone. The results showed that wedged insoles used in combination with analgesia had significant differences in pain and functional improvement compared to analgesia alone. Full length lateral wedges (4 degrees) decreased the pain, stiffness and functional impairment during short-term use (Fang et al., 2006). Custom-made insoles demonstrated similar results (Sasaki and Yasuda, 1987), with one-third of the patients reporting pain relief and half reporting improved walking ability. However, other studies suggested that wedged insoles might not be effective. Baker et al. (2007)

compared the effect of 5-degree wedge insoles and flat insoles and showed no significant differences in pain or functional improvement. Maillefert et al. (2001) and Pham et al. (2004) failed to demonstrate symptomatic relief both in short term and long term studies, respectively. Nevertheless both reported good compliance.

Besides wedged insole, footwear and accessories were also evaluated. Toda and Tsukimura (2008) compared the efficacy of lateral wedges with subtalar strapping with ankle supporters accompanied with a wedged heel insert. The group with subtalar strapping had reduced symptoms significantly compared with the baseline. Different types of comparisons between elastic strapping and traditional insert wedges were also conducted (Toda et al., 2001,Toda and Tsukimura, 2006,Toda et al., 2004). Elastic strapping reduced the tibiofemoral angle and the talar tilt, while both the VAS and Lequesne assessments showed significant differences compared to baseline after 6 months and 2 years. Toda and Tsukimura (2008) further confirmed that the type of footwear, socks and accessories could affect the efficacy of the insole by conducting a comprehensive study comparing neutral insoles, wedge insoles, sock-type ankle supporters with wedges and socks of flat footwear, subtalar strapping insoles, and strapped insoles with socks of flat footwear. Specially designed shoes were of particular interest in this study. The MBT (Masai barefoot Technology) shoe was suggested to reduce general and walking pain (Nigg et al., 2006).

2.2.3 Radiographic Assessment

Radiographic assessment was used to diagnose OA, although many previous studies graded the severity of OA by X-ray (Andriacchi et al., 2004, Da Costa et al., 2005, Kaufman et al., 2001, Andriacchi and Mundermann, 2006). The severity of OA and disease progression was determined by looking at joint space narrowing in X-rays (Figure 2-8).

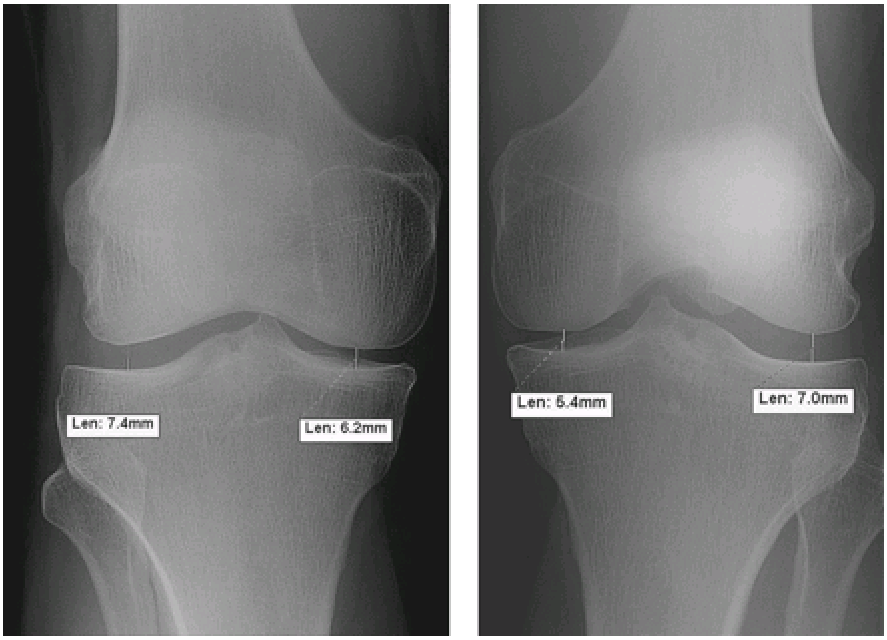

Figure 2-8. X-ray film showing the narrowing of the knee joint space as an indicator of OA and disease progression (Shelbourne and Dickens, 2007).

Table 2-4 Grading of OA knee (Kellgren and Leitner, 1953)

Grade 0	Normal
Grade 1	Unlikely narrowing of the joint space, possible osteophytes
Grade 2	Identified small osteophytes, possible narrowing of the joint
Grade 3	Multiple, moderately sized osteophytes, definite joint space narrowing, some sclerotic areas, possible deformation of bone ends
Grade 4	Multiple large osteophytes, severe joint space narrowing, marked sclerosis and definite bony end deformity.

Sharma et al. (1999b) characterised the knee alignment in both men and women by radiographic assessment. Women had a greater varus-valgus laxity than men, while the laxity also correlated modestly with age. Age-based laxity could be the reason that female OA was more prevalent. The uninvolved knees of OA patients had a greater varus-valgus laxity compared with control knees (Sharma et al., 1999a). Varus-valgus status was assumed to increase the risk of knee OA and the rate of deterioration (Sharma et al., 1999b).

Radiographic assessment has been compared with gait analyses. Knee adduction moment significantly correlated with the mechanical axis. The correlation between knee adduction moment and joint space and pain scores were negative (Miyazaki et al., 2002). A further estimation revealed that a 1% increase in adduction moment increased disease progression by 6.46 times. Barrios et al., (2009) found similar results by considering both static and dynamic measures. The tibial mechanical axis and the medial knee joint lines and rearfoot eversion angle were correlated with the knee adduction moment. Upon orthotic evaluation, a decrease in tibiofemoral angle was also reported in patients with lateral wedges and subtalar strapping (Toda et al., 2002).

2.2.4 Gait Analysis

2.2.4.1 Introduction

Gait analysis, or motion analysis, is an important investigation and diagnostic tool used currently in sports medicine. The analysis system can be categorised as motion recognition, motion tracking and analysis of segment movement. The motion recognition and tracking aspect consists of a video surveillance system in three dimensions, while the analysis of segment movement and preceding kinematics analysis are performed via a force platform and musculoskeletal models.

Vicon Motion System (Oxford, UK) is a motion capturing system on the market. The system uses optical tracking to acquire coordinate data using reflective markers and then calculates the kinematic solution with pre-set or manual parameters. Figure 2-9 demonstrates the application of the Vicon system.

Figure 2-9. Demonstration of the Vicon Motion analysis System

The force platform is another important device that can obtain kinetic data, such as ground reaction force and centre of pressure. The information can then be input into the musculoskeletal model to calculate the joint and muscle forces on any part of the body. Figure 2-10 shows one of the force platforms from AMTI (Advanced Mechanical Technology Inc., Newton, Mass, USA).

Figure 2-10. Force platform, BP400600-1000, AMTI Technology

Knee adduction moment was the key indicator in studying knee OA, reflecting the severity of the disease. This conclusion is supported by Hurwitz et al., (2002) who reported the correlation between radiographic measures of OA severity and peak adduction moment.

2.2.4.2 Characterization of OA knee

One of the main concerns in this study was the effect of walking speed on knee loading. Gok et al. (2002) characterised the gait characteristics of knee OA and discovered that OA patients had a reduced walking velocity and stride length, as well as a prolonged stance period. The extensor moment in loading response increased, along with a reduced flexor moment at the late stance. OA patients also showed greater valgus and internal rotation moment in stance phase. Gok et al. (2002) and Mundermann et al. (2004) discovered that walking style and speed were related to the severity of OA and hypothesised that slower speed could be associated with reduced knee adduction moment to lessen the pain.

The association between knee adduction moment and the mechanical axis was suggested by Hurwitz et al. (2002) and Miyazaki et al. (2002). Hurwitz et al. (2002) suggested that the toe-out angle could also act as a predictive indicator of OA. Mundermann et al. (2005b) found that patients with more severe OA had a greater knee adduction moment compared to normal subjects and less severe OA patients. OA patients had a more extended knee position at initial contact, while more severe OA had a lower hip adduction moment (Mundermann et al., 2005a). However, some research reported fewer kinematic differences between OA patients and normal subjects. Kaufman et al. (2001) showed no significant difference between the knee motion of patients and control patients and suggested that female OA was more prevalent due to greater knee flexion and extension moment. Messier et al. (2005) agreed with the previous studies and demonstrated no significant differences in joint force and moments. However, a general increase in compressive force (7%) and shear (8%) were discovered (Messier et al., 2005).

2.2.4.3 Evaluation of Wedged Insole

Wedged insole have been proven effective in reducing the load of OA in biomechanical analyses. Kerrigan et al. (2002) conducted an experiment with flat, 5-degree and 10-degree insoles. The 5-degree and 10-degree insoles reduced the knee varus torque compared with flat insoles, with the 10-degree insoles reducing the torque slightly more than the 5-degree insoles, though the 5-degree insoles were reportedly more comfortable. Comparing flat insoles and 6-degree insoles presented similar results (Kakihana et al., 2005). The wedged insole reduced the knee joint varus moment and increased the subtalar joint valgus moment significantly (Kakihana et al., 2005). Kakihana et al. also discovered that OA patients had a greater knee joint varus moment than normal subjects. Although there was reduction in kinematic parameters, no alteration of static alignment was reported. The knee adduction moment decreased with the use of 5-degree full length insoles, but alignment was not affected (Hinman et al., 2008). The results were consistent with the investigation of Kakihana et al. (2007), who reported no significant differences in radiological assessment, despite a 17.6% increase in knee varus moment and a medial shift in the centre of pressure.

The efficacy of lateral wedges on the severity of OA was assessed (Kuroyanagi et al., 2007, Shimada et al., 2006b). The Kellgren & Lawrence (1953) grading scale was used, whereas different levels indicate severity of symptoms. Kuroyanagi et al. (2007) demonstrated barefoot and subtalar strapping led to reductions in peak varus moment in both insoles (8% and 13%, respectively) compared to conventional insoles. Subtalar strapping presented a lower moment compared with conventional insoles in grade 2 and 3 OA, while no significant differences were reported in grade 4 OA.

Lateral wedge insoles significantly reduced the peak adduction moment in grade 1 and 2 OA, as well as the first acceleration peak (Shimada et al., 2006a). Custom-made insoles significantly reduced the knee adduction excursion in the mild to moderate OA group, while the resultant muscular force differences could only have effected small changes (Hamill et al., 2009).

2.2.4.4 Effect of Footwear

It was agreed that barefoot has less moment load on the knee joint. Kemp et al. (2008) suggested that walking in shoes significantly increased the peak knee adduction moment compared with barefoot walking, while Shakoor and Block (2006) demonstrated that barefoot walking could reduce knee adduction moment by 11.9%. Dress-shoes, sneakers, high-heeled shoes, clog and stability shoes all caused higher knee adduction moments (Kerrigan et al., 2005, Kerrigan et al., 2003, Shakoor et al., 2008). Yet, it was also suggested that the stiffness of the sole material could affect the knee adduction moment and could act as a good substitute to wedges (Erhart et al., 2008, Fisher et al., 2007). Specially designed shoes were suggested to be beneficial, with an 8% reduction in knee adduction moment compared with a self-chosen shoe (Shakoor et al., 2008).

2.2.4.5 Other Research

Other OA researches related to biomechanics are Bone Mineral Density (BMD) evaluations and Electromyogram (EMG) studies. A patient's BMD was significantly correlated with peak knee adduction moment and the mechanical axis (Wada et al., 2001), though it was independent of height, weight and BMI (Thorp et al., 2006). OA patients showed significantly greater knee instability and medial joint laxity in EMG studies, which suggested that gait alteration was one of the factors (Lewek et al., 2004). Specially designed shoes, MBT, may increase muscle intensity and help the locomotor system (Nigg et al., 2006).

2.2.4.6 Summary

Table 2-4 shows a summary of the existing literature, reporting the methodology and the parameters of interest. Gait analysis was used to evaluate the characteristics of knee OA and the biomechanical consequences of the orthotic intervention. Pain assessment and radiographic evaluation were performed before orthotic intervention. Different factors that may affect the performance of the orthoses include the insole/orthotic design, the accessories on the orthoses and the footwear. The evaluation of orthotic design is not comprehensive enough to suggest an optimal solution, and most of the research compared one particular design with the control condition (flat insoles). Moreover, the evaluation often focused on knee loading. A systematic investigation of the entire lower limb system – from the ankle to the knee – could better elucidate the load transfer mechanism.

The innovations of this study are:

- A comprehensive study and comparison of different insole designs, including, rearfoot, forefoot, medial, lateral and full-length wedges; and
- Evaluation of the kinematics of both the ankle and knee joints.

Table 2-5. Summary of existing literatures

Author and Year	**Parameter(s) of Interest**	**Pain Assess -ment**	**Radiogr -aphic Assess- ment**	**Gait Analysis**	**Insole/ Orthotic**	**Footwear/ Accessory**
(Yasuda and Sasaki, 1987)	5-degree wedge block	X	X		X	X

(Sasaki and Yasuda, 1987)	3 kinds of 5-degree customized insoles	X	X		X	
(Tohyama et al., 1991)	Lateral heel wedge	X	X		X	
(Wolfe and Brueckmann , 1991)	Wedge on Genu Varum and Valgus patients	X			X	
(Keating et al., 1993)	Lateral Wedge	X			X	
(Sharma et al., 1998)	Joint laxity with gender and age		X			
(Toda et al., 2001)	Elastic strapping and traditional insert wedge	X	X		X	X
(Maillefert et al., 2001)	Lateral and Neutral Wedge	X			X	
(Kaufman et al., 2001)	Level walking, stair ascend and descend	X		X		
(Wada et al., 2001)	Correlation of BMD and kinematics parameters	X	X	X		
(Kerrigan, 2002)	5 and 10-degree wedge			X	X	
(Toda et al., 2002)	Lateral wedge with subtalar strapping and ankle supporters with wedged heel insert	X	X		X	X
(Miyazaki et al., 2002)	Disease Progression	X	X	X		
(Ogata et al., 1997)	Insoles on acceleration	X			X	
(Toda and Segal, 2002	Wedge insole with elevation and strapping	X	X		X	X
(Gok et al., 2002)	Gait Characteristic s of OA	X		X		
(Hurwitz et al., 2002)	OA Indicator		X	X		

(Kerrigan et al., 2003)	Dress shoe, sneakers, barefoot			X		X
(Toda et al., 2004)	Subtalar strapping and traditional wedge	X	X		X	X
(Pham et al., 2004)	Lateral and neutral wedge	X			X	
(Mundermann et al., 2004)	Walking speed of different OA severity			X		
(Kim and Eng, 2004)	Disease Severity on kinematics parameters			X		
(Lewek et al., 2004)	EMG, joint laxity			X		
(Kakihana et al., 2005)	0-degree and 6-degree wedge			X	X	
(Kerrigan et al., 2005)	High-heeled shoes			X		X
(Messier et al., 2005)	Kinematic parameter of disease		X	X		
(Mundermann et al., 2005a)	Disease Severity and gait parameters			X		
(Teichtahl et al., 2006)	Relationship between kinematic parameters			X		
(Shimada et al., 2006a)	Lateral wedge on disease severity			X	X	
(Fang et al., 2006)	4-degree lateral wedge	X			X	
(Toda and Tsukimura, 2006)Toda et al	Subtalar strapping and traditional wedge	X	X		X	X
(Schmalz et al., 2006)	Wedge with rigid AFO and semi-rigid ankle support			X	X	X
(Shakoor and Block, 2006)	Barefoot, usual shoe			X		X
(Nigg et al.,	Special	X		X		X

2006)	design shoe and control, EMG					
(Thorp et al., 2006)	BMD			X		
(Butler et al., 2007)	Custom lateral wedge			X	X	
(Kuroyanagi et al., 2007)	Lateral wedge, severity of OA, barefoot, conventional insole and subtalar strapping			X	X	
(Kakihana et al., 2007)	0-degree and 6-degree lateral wedge		X	X	X	
(Baker et al., 2007)	5-degree and neutral wedge	X			X	
(Fisher et al., 2007)	Sole material			X		X
(Franz et al., 2008)	With and without arch support in shoe			X	X	
(Rutherford et al., 2008)	Relationship between kinematic parameters		X	X		
(Hinman et al., 2008)	5-degree full length and rearfoot wedge			X	X	
(Toda and Tsukimura, 2008)	Neutral, wedge insole, sock-type ankle supporter with wedge and socks of flat footwear, subtalar strapping insole, strapped insole with socks of flat footwear	X			X	X
(Rodrigues et al., 2008)	Medial and neutral wedge	X	X		X	
(Kemp et al., 2008)	Barefoot, usual shoe,			X		X

	cane					
(Shakoor et al., 2008)	Self-chosen shoe and mobility shoe, control			X		X
(Erhart et al., 2008)	Shoe stiffness			X		X
(Hinman et al., 2009)	Neutral, valgus, varus wedge			X	X	
(Butler et al., 2009, Barrios et al., 2009)	Relationship between kinematics parameters		X			

2.3 Review of Musculoskeletal Models

There are two fundamental limitations in understanding the dynamics of human movement in an experimental study. The first limitation is the feasibility of conducting invasive research to examine the internal physical structure, such as the muscle. The second limitation is the cause-and-effect relationships in complex dynamic systems. Therefore, musculoskeletal modelling helps researchers to solve these two problems.

The musculoskeletal modeling environment was introduced by (Delp et al., 1990) in the early 1990s and called SIMM (Delp and Loan, 1995). It could create, alter and evaluate models of many different musculoskeletal structures (Hutchinson et al., 2005). For biomechanical studies, the musculoskeletal models were created to simulate human movements such as walking (Jonkers et al., 2003, Neptune et al., 2001), cycling(Smak et al., 1999, Raasch and Zajac, 1999), running (Sasaki and Neptune, 2006), and stair-climbing (Peterman et al., 2001). It was also developed to examine the biomechanical consequences of surgical procedures including tendon surgeries(Saul et al., 2003), osteotomies (Delp and Maloney, 1993)and total joint replacements (Delp et al., 1994).

To generate and analyze three-dimensional (3D) simulations of large population, it was possible to establish the muscle functions for subjects with a range of sizes, strengths, and movement patterns using new software (OpenSim, SimTK.org). It was also practical to perform sensitivity studies to determine whether the conclusions drawn from a simulation are sensitive to variations in model parameters.

The OpenSim was used to generate forward simulations using models (Delp et al., 2007). The first step was to scale models to subject- specific geometry based on marker positions under static trial. Inverse kinematics and residual reduction algorithms were applied to calculate the joint kinematics during walking gait. Computed muscle control (CMC) was then used to find the optimal muscle excitation patterns that would drive the models along the desired trajectory (Thelen and Anderson, 2006). Forward dynamic simulation offers a potentially powerful methodology for characterizing the causal relationship between muscle excitations and multi-joint movement during gait. For example, recent studies have used simulations of normal walking to quantify the contributions of lower extremity muscles to vertical support, forward progression, and swing leg kinematics (Anderson et al., 2004, Anderson and Pandy, 2001, Neptune et al., 2001).

The CMC method is now available for use by researchers around the world. Thelen et al. 2003 provided a computationally efficient means to generate these simulations. The coupling of muscle excitation (u) to activation (a) was modeled as a first-order process with rise and decay time constants of 10 and 40 ms, respectively (Zajac, 1989).

Muscle-driven dynamic simulations complement experimental approaches by providing estimations for important variables, such as muscle and joint forces, which are difficult to measure experimentally. The theoretical framework reveals the cause-effect relationships between neuromuscular excitation patterns, muscle forces, and motions of the body. Many elements of the neuromusculoskeletal system interact to enable coordinated movement. Determining how individual muscles contribute to observed motions is difficult because a muscle can accelerate joints that it does not span and body segments to which it does not attach (Zajac and Gordon, 1989). The data of mechanics of muscle, the geometric relationships between muscles and bones, and the motions of joints were necessary.

Describing the anatomy and physiology of the elements of the neuromusculoskeletal system as a framework, the musculoskeletal modeling provided the mechanics of multi-joint movement by dynamic simulation of movement. Delp et al. (2007) studies had demonstrated the utility of musculoskeletal models and dynamic simulations for analyzing the causes of gait abnormalities and the effects of various treatments. Value of dynamic simulations of movement was broadly recognized (Zajac et al., 2002, Zajac et al., 2003). Individual investigators have made elegant contributions to simulation technology, including the development of novel methods to muscle model (Bhargava et al., 2004), simulate contact (Fregly and Sawyer, 2003), and sent musculoskeletal geometry (Menegaldo et al., 2004).

Dynamic optimization typically requires thousands of complete integrations of the model state equations to converge to a solution (Anderson & Pandy, 2001). It was translated into days, weeks or even months of computer time depending on the complexity of the model. Even then, to dynamic optimization of complex nonlinear problems were endemic numerical difficulties, which lead to sub-optimal solutions (Neptune and Hull, 1999).Other researchers have used this observation to refine anthropometric parameter estimates (Vaughan et al., 1982)and to improve the accuracy of joint moments computed using inverse dynamics (Cahouet et al., 2002). To compute pelvis translations and low back angles so as to enforce dynamic consistency between kinematics and external forces was approached. Justification for this approach is three-fold. First, the estimation of pelvis orientation and lower extremity joint angles during gait using motion analysis systems is fairly well established and widely used (Ounpuu et al., 1991).Dynamic optimization was used to generate the walking simulations at the five different speeds by fine-tuning the excitation onset, duration and magnitude for each muscle group using a simulated annealing algorithm until the difference between the experimental and simulated kinematic and ground reaction force data was minimized (Davy and Audu, 1987, Neptune and Hull, 1999, Neptune et al., 2001, Neptune et al., 2004). The specific quantities evaluated in the objective function included the time history of the right and left hip, knee and ankle joint angles, horizontal and vertical ground reaction forces, and the two components (x, y) of the trunk translation resulting in a total tracking variables.

Using of the dynamic optimization, the human body musculoskeletal model was developed including eight segments, twenty-one of freedom and ninety-two Hill type muscle tendon units (Delp et al., 1990). A lower-extremity model was used to estimate muscle-tendon lengths, velocities, moment arms, and induced accelerations during normal and pathologic gait (Arnold and Delp, 2001). It have been used to reproduce the features of normal gait in 3D (sagittal, transverse, and frontal planes)(Anderson and Pandy, 2001). The contributions of rectus femoris, vastus, and other muscles to knee flexion have been evaluated by altering muscle excitations in the simulation and computing the resulting changes in peak knee flexion. A musculoskeletal model relies on the use of advanced biomechanical analysis of muscle function during gait. In clinical practice, personalization of the model is usually limited to rescaling a generic model to approximate the patient's anthropometry, even in the presence of bony deformities, as in the case of cerebral palsy (CP). Generic models defining the musculoskeletal geometry of a healthy, average-sized adult male is constructed by motion capture experiment on large population.

Gait analysis together with information during the clinical examination, it supports the selection of patient-tailored treatment procedures (Sheldon et al., 2005). Related to musculoskeletal geometry, gait analysis techniques relied on musculoskeletal models for the calculation of gait kinematics and parameters as well as muscle-tendon dynamics (e.g. force generating capacity of muscles) (Arnold and Delp, 2001). Human walking requires the coordination of the neuromuscular system to support body weight. However, muscle forces were not directly measurable and each muscle can accelerate multiple segments and joints (Zajac et al., 2002, Zajac et al., 2003).

To simulate whole-body movements such as walking or running, motion-capture protocols that accurately describe patients 'joint axes, trunk motions, and foot motions are needed, along with ground reaction force data from consecutive strides.

Conventional protocols for clinical motion analysis were not designed with the intent of creating simulations, and they could be improved. Developing simulations of movement highlights the limitations of current motion capture data and demonstrates the need for improved experimental protocols. Muscle-driven simulations generate a wealth of data. Using simulations to elucidate the principles that govern muscle coordination and to achieve improved clinical outcomes, therefore, requires tools that can help reveal insights from these data.

The 3D models were conducted to investigate the treatment of individuals with spinal cord injury (Wilkenfeld et al., 2006), to analyze joint mechanics in subjects with patellofemoral pain (Besier et al., 2005, Kautz and Neptune, 2002), to calculate forces at the knee during running (Besier et al., 2003) and cutting (McLean et al., 2004), to examine the influence of foot positioning and joint compliance on the occurrence of ankle sprains, and to investigate causes of abnormal gait (Higginson et al., 2006, Kerrigan et al., 1998, Piazza and Delp, 1996). Recent studies have used simulations of normal walking to quantify the contributions of lower extremity muscles to vertical support, forward progression, and swing leg kinematics(Anderson et al., 2004, Anderson and Pandy, 2001, Neptune et al., 2001). While, some other studies (Davy and Audu 1987, Higginson et al. 2006, Piazza and Delp 1996) used two-dimensional (2D) models to simulate the sagittal plane movement only. Both 2D and 3D simulations reasonably reproduced experimental data but there were some differences in the results of predicted muscle function. Liu et al. (2006) using a 3D model found that hamstrings did not substantially contribute to either progression or support, while(Neptune et al., 2004) with a 2D model, reported that hamstrings accelerated the body forward and provided some support during early stance.

Moreover, it was found that ankle plantar flexors provided center of mass (COM) support from early stance phase (30% gait cycle) (Neptuneetal 2004), while the ankle plantar flexors only provided COM support from late stance phase (50% gait cycle) using a 3D model(Anderson and Pandy, 2003). It was possible that the variation in muscle function reported in the literatures (Anderson and Pandy 2003, Liu et al., 2006, Neptune et al., 2001, Neptune et al., 2004) was due to different degrees of freedom (DOFs) included in their models. Chen et al., (2006) found that individual joint moments to ground reaction forces and segmental powers were different between models, even when net contributions were identical across different models. Moreover, greater power redistribution attributed to the joint moments by increasing the DOFs of the model. Patel et al. (2007) reported that modeling the pelvis and trunk as separate segments impacted the interpretation of the role of the joint moment during normal walking.

To analyze individual muscle function, vertical COM accelerations induced by each muscle was computed (Xiao and Higginson, 2008). The perturbation tool (Liu et al., 2006) adjusted one muscle's force and simulated forward over a short time interval to observe the resulting change in position of the model's COM.

Although human walking is a 3D activity, there have been quite a number of simulation studies in the literature using models limited to the sagittal plane (Neptune et al., 2001). From the standing neutral trial, a kinematic model comprised of five skeletal segments: foot, talus, shank and thigh of the support limb, and the pelvis. There was12 degrees-of-freedom (DOF) defined using Mocap Solver software which performs model based kinematic analysis through global least-squares optimization

(Lu and O'Connor, 1999). The pelvis was assigned 6 DOF relative to the laboratory coordinate system. The three rotational DOF somersault, tilt, twist were defined using the rotation sequence (Yeadon et al., 1990). The hip joint possessed 3 DOF, with rotations, flexion-extension, abduction-adduction and internal-external rotation defined about three orthogonal axes, passing through a fixed joint center defined according to Bell et al. (1990).

The ankle joint was modeled as a 2-DOF mechanism allowing rotation about talocrural and subtalar joint axes. The talocrural joint center was defined as the midpoint between the lateral and medial malleoli, with the plantar-dorsiflexion axis extending laterally from this point. This simplified orientation was justified based on known variations within a normal population (van den Bogert et al., 1994). The subtalar joint was located 10 mm directly below that of the talocrural joint (van den Bogert et al., 1994). The 3D marker trajectories recorded during the 10 side-stepping trials were processed by the Mocap Solver software to solve the 12 generalized coordinates for each frame, corresponding to the 12 DOF of the skeletal model.

That indicated that modelers needed experimentalists to acquire parameters used in simulations and to test the accuracy of results derived from simulations. Experimentalists needed modelers to provide a theoretical framework within which to interpret experimental observations, and to help gain perspective from the wealth of data derived from biomechanical experiments. With access to open source software for developing and analyzing muscle-driven simulations, biomechanics researchers are now in a position to establish quantitative, cause-effect relationships between the neuromuscular excitation patterns, muscle forces, external reaction forces, and motions of the body that are observed in the laboratory. Coupled with high-quality experimental measurements, simulations will help elucidate how elements of the neuromusculoskeletal system interact to produce movement. We hope, improve the outcomes could give the guild to the treatments for persons with movement disorders.

The development of a digital human (a computational model of the human neuromusculoskeletal system with complexity comparable to a human) was a grand challenge. If a general and comprehensive model were available, then users could choose how to simplify the model to address a particular scientific question. The Physiome Project (Hunter and Borg, 2003) outlines this challenge and some of the important benefits of its success.

Summary

The musculoskeletal modeling is widely used in the field of biomechanics, especially for the human motion analysis. Internal joint forces, including muscle forces, can be calculated through musculoskeletal model simulation rather than trying to overcome the difficulty of getting the measurements from experimental studies. The dynamic simulation is based on kinematics, kinetics and inverse dynamic mechanisms. However, there still are some gaps that a systematic investigation of the insole was not done and the transfer mechanism of the entire lower limb system – from the ankle to the knee –are not clear.

CHAPTER III **METHODOLOGY**

3.1 Participants

Ten healthy Chinese female volunteers participated in this study. The Human Subject Ethical Application Review System at The Hong Kong Polytechnic University approved the experimental protocol. The experimental information was explained to all participants and consent form was signed before the experiment.

The subject selection criteria were:

- Age 18 to 30 years
- BMI from 18.5 to 24.3
- Q-angle ranged from 15 to 20degrees
- No history of symptoms with their back and lower extremities

Exclusion criteria were:

- Any problems on the hip, foot or knee
- Leg Length Discrepancies
- Congenital anomalies or neuromuscular disorders of the lower extremities
- Previous lower limb surgeries in the past 2 years
- History of lower limb extremity injury

The physical examination consisted of items selected for their appropriateness depending on the clinical assessments. The subject information acquisition form and consent form are attached in the appendix. Table 3-1 lists the subject information of each participant.

Table 3-1. Subject information and basic measurement

	Height (cm)	Mass (kg)	BMI	Age (year)	Q-angle (Degree)	Inter-condylar (Inter-malleolar) gap (mm)	RSCA (Resting Stance Calcaneal Angle) (degree)
1	165	51.5	18.92	24	15	8.0	2
2	158	56.0	22.43	30	17	1.0	2
3	160	45.5	17.77	27	16	0.0	0
4	159	48.8	19.30	26	18	(1.0)	2
5	161	51.0	19.68	27	17	(5.0)	2
6	156	50.0	20.55	31	17	2.6	4
7	162	50.0	19.05	28	16	5.0	2
8	162	51.0	19.43	25	15	1.0	2
9	155	48.0	19.98	28	16	6.0	2
10	157	54.8	22.23	24	17	(3.0)	0

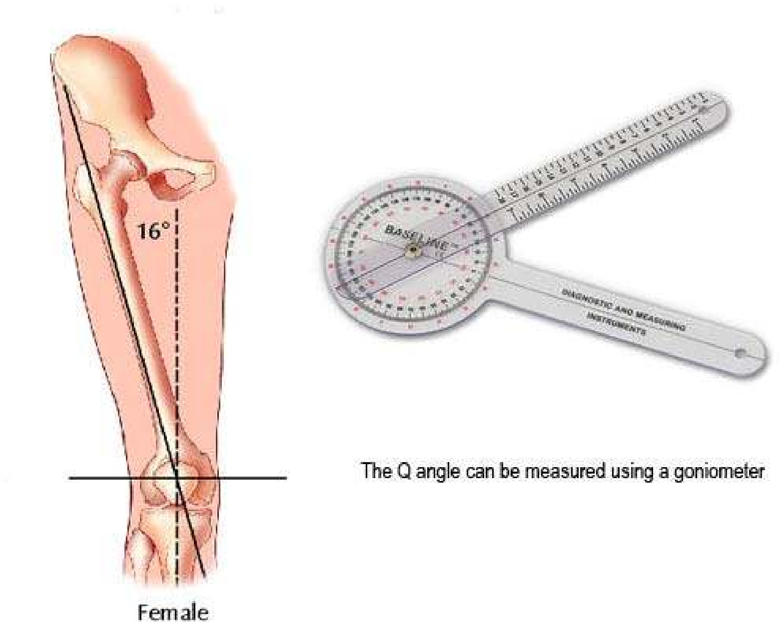

Figure 3-1.Measurement of the Q-angle (North Carolina School of Science and Mathematics http://www.ncssm.edu)

Three anatomical landmarks were used to determine the Q-angle. The ASIS, the centre of the patella (kneecap) and the tibial tubercle, which is the bump approximately 5cm below the kneecap on the front of the shin bone (tibia). The Q-angle is formed by a line drawn from the ASIS to the centre of the kneecap, and from the centre of the kneecap to the tibial tubercle. A normal Q-angle in women ranges from 15 to 20degrees.

The inter-malleolar/inter-condylar gap was used to determine the varus/valgus knee. Inter-condylar is a condition where the legs are bowed outwards in the standing position. When the knees are far apart during standing, it is called the inter-condylar gap; if the feet are far apart while the knees are closer together, the condition is termed an inter-malleolar gap. An inter-condylar gap greater than 3cm or an inter-malleolar gap greater than 5cm is considered abnormal (Michael et al., 2006) (Fig. 3-2).

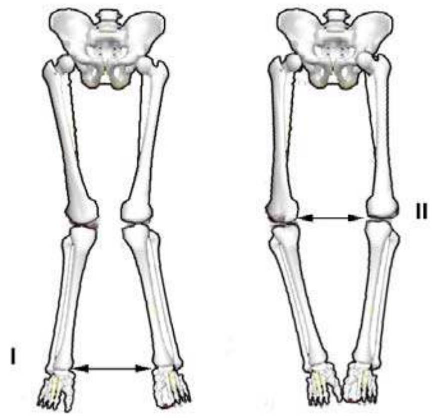

Figure 3-2. Inter-malleolar and inter-condylar gap: I: inter-malleolar distance II: inter-condylar distance

The Resting Stance Calcaneal Angle was measured as the angle between the bisection of the lower one third of the tibia and the bisection of the calcaneus and the tibia (Fig. 3-3).

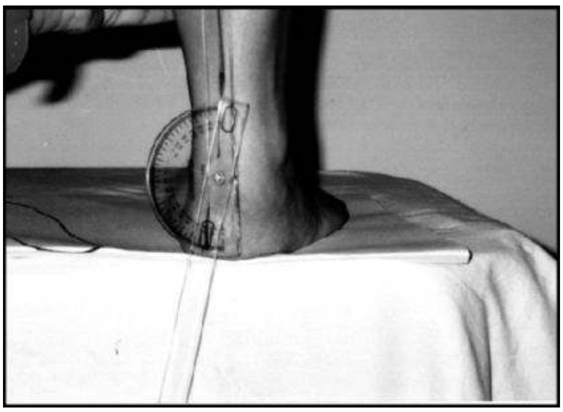

Figure 3-3. Positioning for the Resting Stance Calcaneal Angle measurement (Shuchi et al., 2008)

3.2 Shoe and Insoles

Seven different insoles were tested (Figure 3-4, Table 3-2): a control insole with no wedge (FF), a wedge with a 5-degree in the lateral rearfoot (LR), lateral forefoot (LF), full-length lateral (LW), medial rearfoot (MR), medial forefoot (MF) and full-length medial (MW). The wedges made of high-density (Shore A65) ethylene-vinyl acetate (EVA, Podotech®, UK) were placed under both feet (Fig. 3-5).

Table 3-2. Conditions for this study

Condition No.	**Symbols**	**Presentation**
1	FF	Flat insole (control group)
2	LR	Rearfoot lateral wedge
4	LF	Forefoot lateral wedge
6	LW	Full-length lateral wedge
3	MR	Rearfoot medial wedge
5	MF	Forefoot medial wedge
7	MW	Full-length medial wedge

Figure 3-4. Seven insoles conditions

Figure 3-5. Wedged insoles made of EVA

The insoles were trimmed to fit in the walking shoes (Superstar, Adidas Ltd., Germany). Five holes were punched in the shoe so that the markers on each foot could be observed (Fig. 3-6). The 5-degree wedge group was selected because it is commonly prescribed in clinical practice and has been reviewed in previous literature. A wedge greater than 5-degrees resulted in discomfort (Kerrigan, 2002). The sequence of different foot support conditions and controls was randomized during the experiment.

Figure 3-6.Footwear used in the experiment

3.3 Equipment

Experiments were conducted at the Human Locomotion Laboratory at the Hong Kong Polytechnic University. Kinematic information was collected using a motion analysis system (Vicon Nexus, Oxford) with 8 infrared cameras. The ground reaction forces were measured using two force platforms (Advanced Mechanical Technology Inc. Newton, Mass, USA) situated at the middle of the walkway (Fig. 3-7). The kinematic data were collected at 240 Hz and, the AMTI analogue force plated data were collected at 960 Hz.

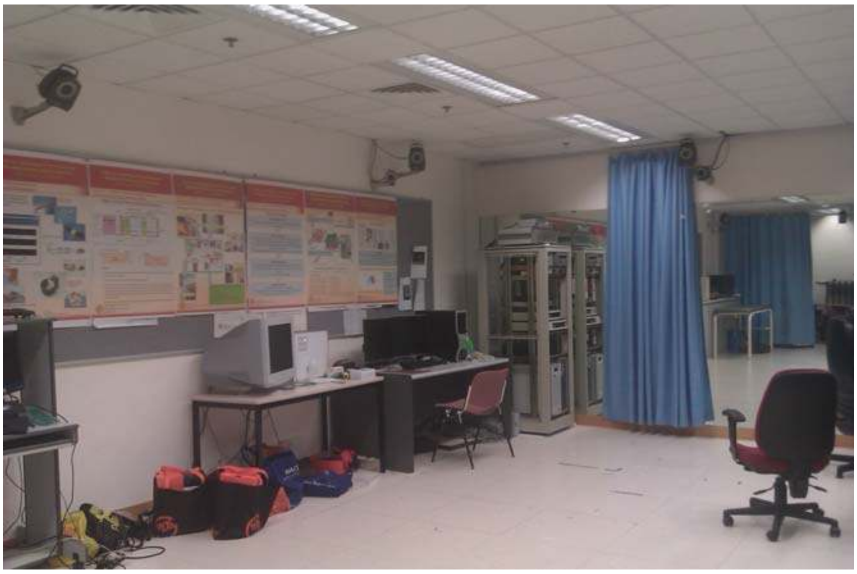

Figure 3-7. Experimental area including the motion captures system and AMTI force platform

3.4 Experimental Protocol

The subjects were asked to wear shorts and a tee shirt. Before the session started, the motion analysis system was calibrated using a rigid orthogonal jig. Forty-four reflective markers were then place over anatomical landmarks (Fig. 3-8, Table. 3-3) - bilaterally on the anterior and posterior superior iliac spines (ASIS, PSIS), the greater trochanter (GT), the lateral and medial femoral condyles (KL, KM), the medial and lateral malleoli (LMA, MMA), the middle calcaneus (Heel), the 1st and 5th metatarsal heads (M1, M5) and the top of the hallux, the rigid third marker to define the thigh, shank, forefoot (FF) and rearfoot (RF) segments.

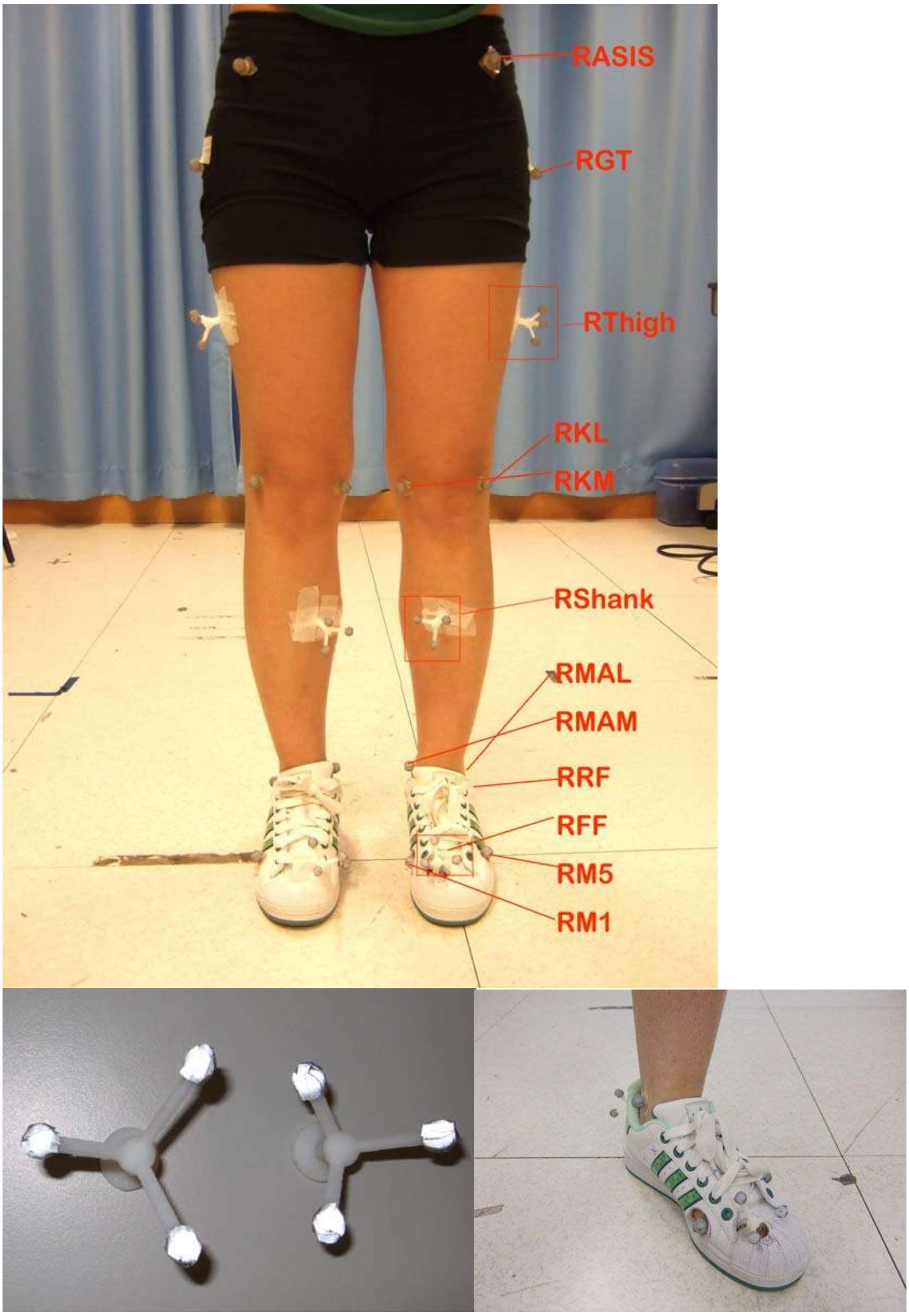

Figure 3-8. Demonstration of marker placement on different body segments

Table 3-3 Marker placement and body segments

Landmark	**Marker**		**Joint and Segment**
	Right	**Left**	
Right and left anterior and posterior superior iliac spines	RASIS, RPSIS,	LASIS, LPSIS	Pelvis
Greater trochanter	RGT, LGT		Hip Joint
Lateral side of thigh	RHUL, RHUM, RHDL	LHUL, LHUM, LHDL	Thigh
Lateral and medial femoral condyles	RKL, RKM	LKL, LKM	Knee Joint
Shaft of tibia	RTUL, RTUM, RTDL	LTUL, LTUM, LTDL	Shank
Lateral and medial malleoli	RLMA, LMMA	LLMA, LMMA	Ankle Joint
Top of hallux	RFX, RFY, RFZ	LFX, LFY, LFZ	Forefoot
1st and 5th metatarsal heads	RM1, RM5	LM1, LM5	
Heel center, Lateral calcaneus	RHEEL, RHX, RHY, RHZ	LHEEL, LHX, LHY, LHZ	Rearfoot

The lower body was divided into segments, including forefoot, rearfoot, shank, thigh, and pelvic segment. For each segment, a three-marker array was placed on the anatomical landmarks. Kinematic information for each segment could be deduced from this array. The components of the ground reaction force and moment in the vertical, antero-posterior (AP), and medio-lateral (ML) directions and their temporal variations were measured.

After the markers were attached, subjects were instructed to stand for five seconds to establish the relationship among the markers for the subject's initial anatomical position. Standing trials were then performed for participants by asking them to stand quietly on the force plate for thirty seconds in each insole condition. Using predefined lines and landmarks on the testing platform surface, the subjects' feet were carefully placed on the average preferred-stance position; they were instructed to place their arms over the chest and stare at a predetermined spot, 'X', while the data of marker trajectory and the ground reaction force were collected.

In the dynamic test, each subject was asked to walk following the metronome at a cadence of 120 steps per minute along a ten-meter walkway for every assigned condition (Table 3-2). Three minutes of walking accommodation were preceded for each condition before data capture. Five minutes of relaxation were allowed between trials. For each condition, there were ten walking trials.

3.5 Data Processing

The data optimization process was performed with a commercial data package because the data collected in the dynamic trials may process discrepancies, such as data missing or marker blockade (Fig. 3-9). Once the integrity of each trial data was confirmed, the track file was exported in c3d format, which could be inputted into post-processing program, Visual3D (C-motion Inc., USA). The knee joint motion data in the sagittal and coronal planes were analyzed bilaterally. Each subject's weight and height were inputted into the software. The linear and angular velocities and accelerations of each segment and joint centre position were calculated.

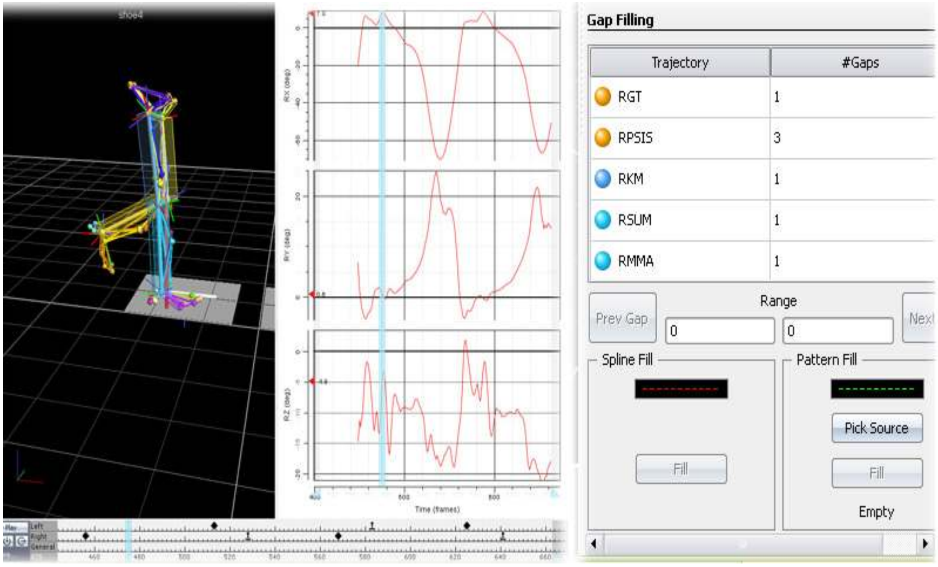

Figure 3-9. Data processing in the Vicon NUXE software, processing kinematics information

Static data collection was used to construct the joint and body segment of the Visual3D model (Fig. 3-10). The model could aid in the kinematic and kinetic analysis in the dynamic trials. The standing trials were performed for all conditions, as proposed by (Le Clair and Riach, 1996), who determined that these data were sufficient to ascertain differences in postural measures between conditions.

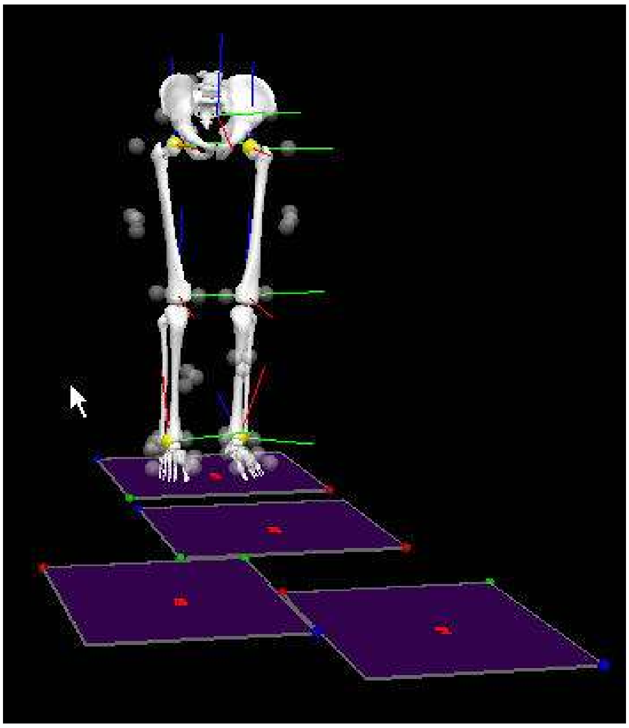

Figure 3-10. Processing software, Visual 3D, showing calculated joint and body segments

In the software environment of Visual3D, each segment is a rigid object, defined geometrically by a local coordinate system and length. The segment was transformed between sensor coordinates, segment coordinates, and joint constraints. In Visual3D, a joint used the term “sensor” very loosely because it assumed that the markers were attached rigidly to a segment and that the segment was connected to other segments. A local coordinate system is capable of moving in space markers, which are sensors that attach to the segment to track this movement. Visual3D does not allow a segment to be defined without a tracking sensor.

The CODA segment model was used for pelvis referred to Charnwood Dynamics (Bell et al., 1990). The pelvis segment was defined by the anatomical locations of the ASIS and the PSIS. The (x-y) plane of this segment coordinate system is defined as the plane passing through the right and left ASIS markers, and the mid-point of the right and left PSIS markers (Fig. 3-11).The x-axis was defined from the LASIS towards the RASIS. The z-axis was perpendicular to the (x-y) plane. The y-axis was then the cross product of the x-axis and z-axis. Estimates for the right and left hip joint centre are represented as landmarks that are created automatically when the CODA pelvis segment is created. The location of the landmark was defined as:

Right Hip Joint Center =(0.36*ASIS_Distance, -0.19*ASIS_Distance,
-0.3*ASIS_Distance)

Left Hip Joint Center =
(-0.36*ASIS_Distance,-0.19*ASIS_Distance,-0.3*ASIS_Distance)

AISIS_Distance was the distance between the right ASIS and left ASIS based on the marker RASIS and LASIS.

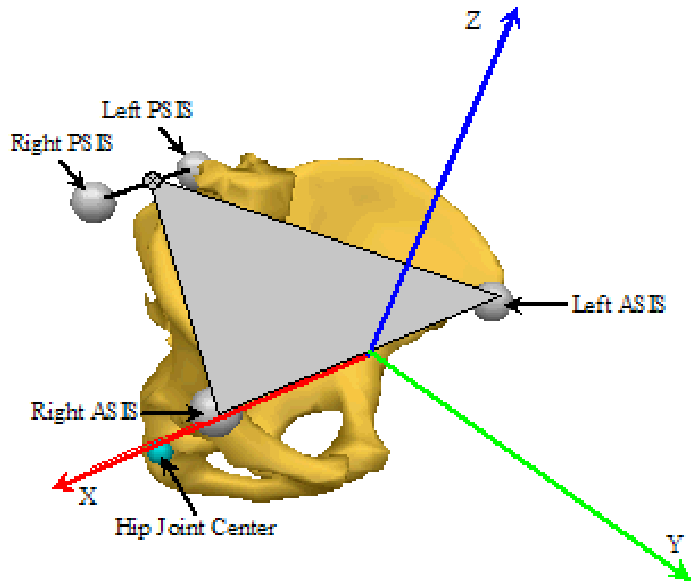

Figure 3-11. CODA Pelvis with local coordinate system (Visual3D)

Other segments and joints were defined by the conventions of the segment's own coordinate systems. The segment coordinate system's z axis was oriented along the length of the segment, the x axis was aligned medio-laterally, and the y axis was oriented at a right angle to both x and z. The tracking markers defined the segment body. Figure 3-12 showed that the shank and ankle joint were defined using the anatomical locations of KM, KL, LMA and MMA and the tracking markers SUM, SDL and SDM, which were the rigid third markers of the shank. The x-axis of the knee joint was defined from the KM to KL. According to the conventions, the forefoot and rearfoot segment forefoot/rearfoot joint were defined showed in Figure 3-13. The x-axis of the forefoot/rearfoot joint was same as the ankle joint.

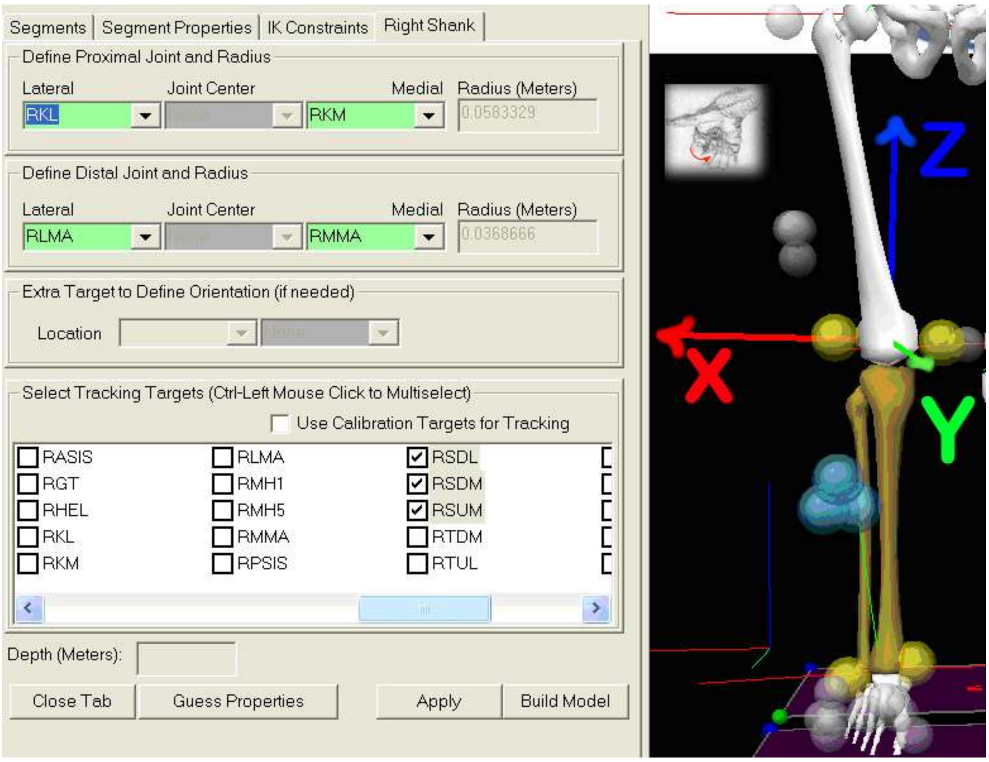

Figure 3-12. Right hand grip rule showing the positive direction of the coordinate system and the definition of segment of right shank and knee joint

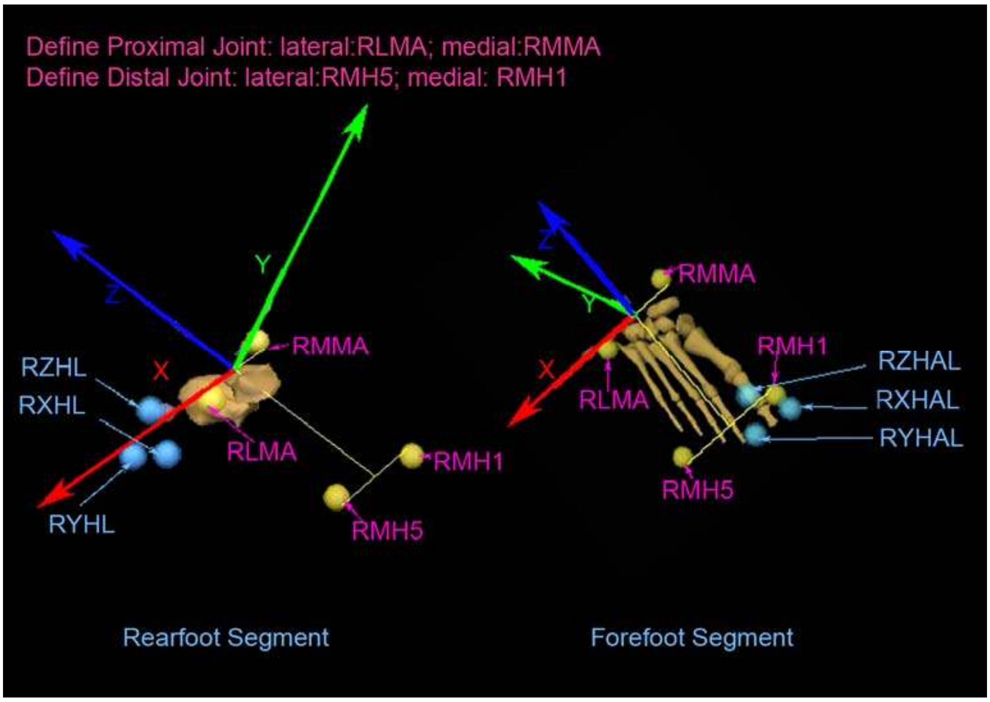

Figure 3-13.The coordinated system and definition of segments of the rearfoot and forefoot

Joint angles are defined as the orientation of one segment relative to another segment. Visual3D allows user to pick any 2 segments to define a joint angle. Joint angles are calculated as the transformation from one segment (A) to another segment (B). The local coordinate system was used as the frame of reference, which is the ordered sequence of rotations (x, y, z) that assumes that the Z axis is in the up/down/Axial direction and the y axis is anterior/posterior, following the direction it travels (Yeadon, 1990, Kadaba et al., 1990). An alternative approach is to create Virtual Segments that define the desired angle in the standing posture

The x-axis for both the left and right thigh segment coordinate systems points from the left to the right. The joint rotation followed the right-hand rule (Fig 3-12), where the thumb points in the direction of the x-axis. When the fingers are curled in the direction of shank extension relative to the thigh, the knee extension is a positive angle. Joint moments were normalized by dividing by subject's body mass. Temporal events defining the gait cycle were identified from the ground reaction force data.

3.6 Musculoskeletal Model Simulation

A musculoskeletal model was developed using the OpenSim simulating software (OpenSim 2.2.1 Released, US) to estimate the joint and major muscle forces in this study. Figure 3-14 shows the 3D musculoskeletal model developed in this study. The model consists of rigid bodies connected by joints. There were 91 muscles-tendon units, 12 joints and 20 DOFs (Delp et al., 2007) (Table 3-4). The joints were allowed to move in the sagittal, frontal and coronal planes. The head and torso were modeled as a rigid segment with locked back joints. The pelvis could rotate and translate in all three dimensions with respect to the ground. The hip joint was modeled as a ball-and-socket joint with three rotational DOFs. In the knee joint, the tibiofemoral translation and non-sagittal rotations were constrained as functions of the knee flexion angle. Two revolute joints aligned with anatomical axes represented the ankle-subtalar complex. The metatarsophalangeal joint was modeled as a hinge joint to allow toe flexion and extension. The 91 muscles in the model include the soleus (SOL), medial gastrocnemius (GAS), tibialis anterior (TA), tibialis posterior (TP) vastus intermedius (VASI), rectus femoris (RF), biceps femoris short head (BFSH), biceps femoris long head (BFLH), iliacus (IL), adductor magnus (ADD MAG), gluteus maximus (GMAX), gluteus medius (GMED) and other muscles.

Table 3-4. 3D Model configuration

No. of segments	No. of joints	No. of muscles	Total DOF	Torso DOF	Pelvis DOF	Hip DOF	Knee DOF	Ankle DOF	Toe DOF
11	12	91	20	0	6	3	1	2	1
Hip, knee, ankle and toe DOFs represent only one leg.									

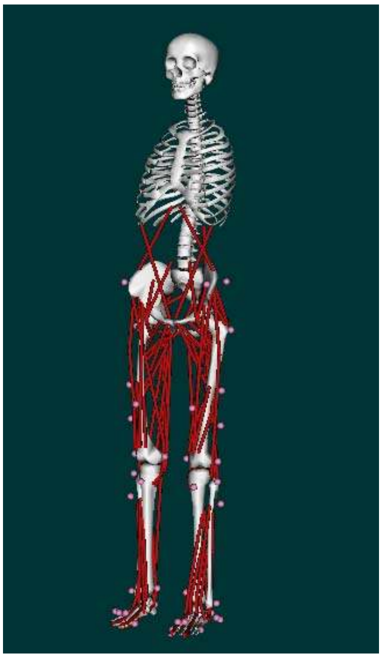

Figure 3-14.The musculoskeletal model of subject

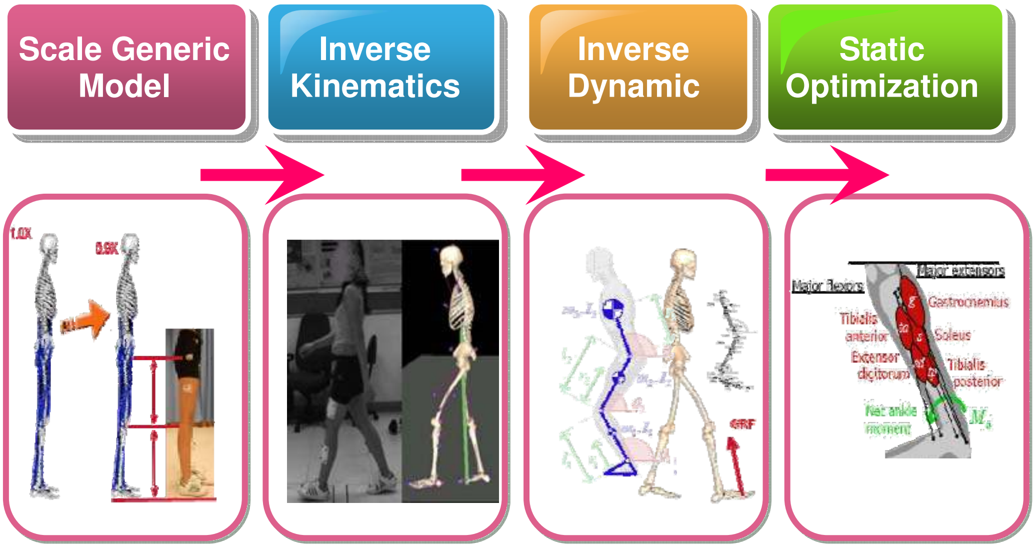

Figure 3-15. OpenSim workflow

The simulation workflow is shown in Figure 3-15. The marker set contained a set of virtual marker that was placed on the body segments of the model. The experimental position data of a static trial provided the position information of the subject for model setup through scaling and matching of the anthropometry of the subject. Scaling was based on the trajectories of the standing trial was and used a tool to move the virtual markers on the model to match those of the subject. The model was scaled to the dimensions of the subject.

Inverse kinematics was used to compute the joint angles for the musculoskeletal model and to reproduce the motion of the subject. The Weighted Least Squares Equation was used in inverse kinematics tool:

$$\min_{\mathbf{q}} \left[\sum_{i \in \text{markers}} w_i \left\| \mathbf{x}_i^{\exp} - \mathbf{x}_i(\mathbf{q}) \right\|^2 + \sum_{j \in \text{unprescribed coords}} \omega_j \left(q_j^{\exp} - q_j \right)^2 \right]$$

$$q_j = q_j^{\exp} \text{ for all prescribed coordinates } j$$

where $\mathbf{q}$ is the vector of generalized coordinates being solved for; $\mathbf{x_i}^{\exp}$ is the experimental position of marker *j*; $\mathbf{x_i}(\mathbf{q})$ is the position of the corresponding marker on the model which depends on the coordinate values; and q_j^{exp} is the experimental value for coordinate *j*. The weight sum of maker errors was minimized after this step.

The net reaction force and net moment at each of the joint can be calculated based on the joint angles, angular velocities and angular acceleration of the model, together with the GRF and moments during the inverse dynamic loading. The remaining term on the right-hand side of the equations of motion is unknown. The Inverse Dynamics Tool uses the known motion of the model to solve the equations of motion for the unknown generalized forces.

Computed muscle control (Thelen and Anderson, 2006) was then used to find the optimal muscles extension patterns that would drive the models along the desired trajectory, the experimental data. The muscle excitations were minimized by the weighted squared sum of muscle forces.

For the subject, we generated one walking trial with a model. Simulation results for this subject with lateral wedged insoles were normalized to the stance phase, starting from the heel strike to the toe off. Only muscles crossing the knee were used in the total muscle force calculation: medial GAS, VASI, RF, BFSH, tensor fasciae longus (TFL), sartorius (SAR), and gracilis (GRAC). The net knee joint force summed the muscle forces and joint reaction forces from the inverse dynamic.

3.7 Data Analysis

Variables describing forefoot joint kinematics, ankle and knee joint moments and center of pressure were quantified in this thesis. These variables are defined in Table 3-5.

Table 3-5. Kinematic and Kinetic variables included in analyses

Symbol	Definition
KADM1	First peak knee adduction moment in stance phase
KADM2	Second peak knee adduction moment in stance phase
KABM	Peak knee abduction moment in stance phase
KEM1	First peak knee extension moment in stance phase
KEM2	Second peak knee extension moment in stance phase
KFM	Peak knee flexion moment of the knee joint in stance phase
KIRM1	First peak knee internal rotation moment in stance phase
KIRM2	Second peak knee internal rotation moment in stance phase
KERM	Peak knee external rotation moment in stance phase
ADFM	Peak ankle dorsiflexion moment in stance phase
APFM	Peak ankle plantarflexion moment in stance phase
AEVM	Maximum ankle eversion moment in stance phase
AIVM	Maximum ankle inversion moment in stance phase
AABM	Peak ankle abduction moment in stance phase
AADM	Peak ankle adduction moment in stance phase
ADFA	Peak ankle dorsiflexion angle in stance phase
APFA	Peak ankle plantarflexion angle in stance phase
AIVA	Maximum ankle inversion angle in stance phase
AEVA	Maximum ankle eversion angle in stance phase
AADA	Maximum ankle adduction angle in stance phase
FDFA	Maximum forefoot dorsiflexion angle in stance phase
FPFA	Maximum forefoot plantar flexion angle in stance phase
FIVA	Maximum forefoot inversion angle in stance phase
FEVA	Maximum forefoot eversion angle in stance phase
FADA	Maximum forefoot adduction angle in stance phase
FABA	Maximum forefoot abduction angle in stance phase
COPL	Maximum shift center of pressure in lateral direction during stance phase
COPM	Maximum shift center of pressure in medial direction during stance phase
COPA	Maximum shift center of pressure in anterior direction during stance phase
COPP	Maximum shift center of pressure in posterior direction during stance phase
PKF	Maximum force of knee joint
PKMF	Maximum total muscle forces of knee joint
PKRF	Maximum reaction force of knee joint

A preliminary evaluation and comparison of a subject in stance phase under different wedge conditions was presented. Some of the key alterations by the various wedge conditions, including angles, moments, and the center of pressure (COP), were identified.

The following key parameters were extracted to be analyzed: the COP (ML and AP directions), the first peak knee adduction moment (KADM1), the second peak knee adduction moment (KADM2), the peak knee abduction moment (KABM), peak knee internal rotation moments (KIRM1, 2), the peak knee external rotation moment (KERM), the peak knee flexion moment (KFM), the peak knee extension moments (KERM1, 2) and the maximum ankle inversion moment (AIVM).

All the kinematic and kinetic dependent variables were assessed by a one-way repeated measures analysis of variance (ANOVA), including the within subject factor (insole). Multiple post-hoc comparisons were undertaken using Lest Significant Difference (LSD) to identify significant differences between conditions. Bivariate correlation analysis (Pearson correlation, r) was carried out to determine correlations of respective parameters. The level of significance was $P < 0.05$.

CHAPTER IV **RESULTS**

Sections 4.1-6 present the results of biomechanical variables during stance phase for one typical subject. The knee joint moments, ankle joint moments, forefoot angles, COPs, and the joint forces are shown. Section 4.7 presents the statistical analyses and the correlation analyses are presented at the end of this chapter.

4.1 Knee Joint Moment

Figure 4-1 shows the knee joint moments in flexion/extension rotation direction compared between the wedges and the FF condition.

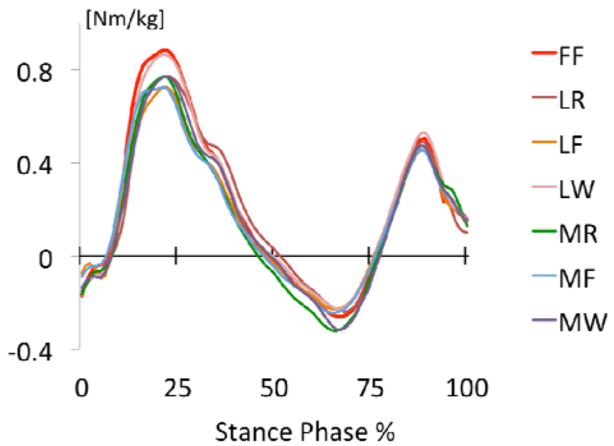

Figure 4-1. Knee joint extension (+) /flexion moments in different conditions during stance phase

There were two peaks of extension moment (KEM1, KEM2) at approximately 20% and 90% of the stance phase, while there was one peak flexion moment (KFM) at approximately 65-70% of the stance phase. At the first peak of extension moment, the higher extension moments at approximately 0.86Nm/kg, while the groups were apparently lower, 0.72Nm/kg. The second peak of extension moments was lower than the first approximately 0.51Nm/kg.

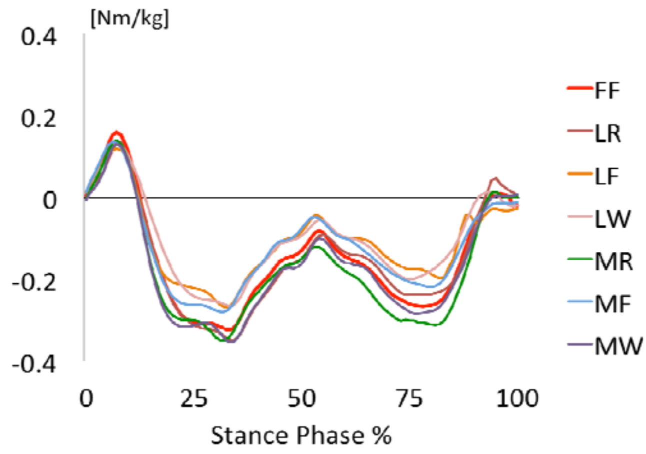

Figure 4-2. Knee joint abduction (+) /adduction moments in different conditions during stance phase

The knee adduction moment started neutral at the beginning of the stance phase (Fig. 4-2). A shoot-off of knee abduction moment appeared at approximately 5-10% of the stance phase, with a value of 0.16Nm/kg in the FF condition for the typical subject. The moment turned to adduction and reached a plateau at approximately 20-40% of the stance phase. The knee adduction moment was followed by a rebound at mid-stance, with similar ordering. The knee adduction moment reached another plateau at approximately 80% of the stance phase, at 0.27Nm/kg. No observable differences were observed at the terminal stance, where all conditions were approximately neutral.

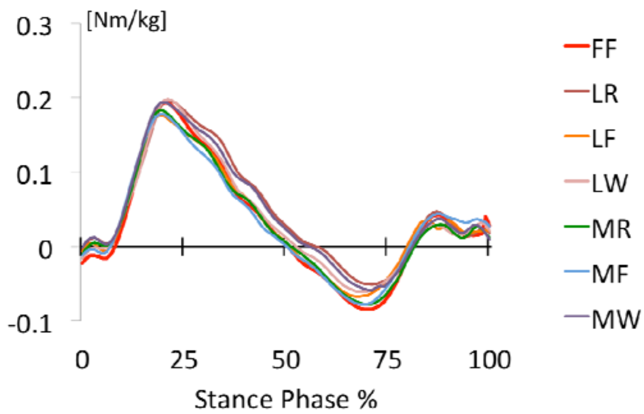

Figure 4-3. Knee joint internal (+) /external rotation moments in different conditions during stance phase

The internal rotation moment of the knee started out around neutral and reached a peak at approximately 20% of the stance phase (Fig. 4-3). The moment fell to its lowest value at approximately 70% of the stance phase, which is interpreted as a peak of moment of knee external rotation. The internal rotation moment headed back to neutral at terminal stance. The maximum knee external rotation moment was at 70% of the stance phase, with a value of 0.084Nm/kg of this typical subject.

4.3 Ankle Joint Moment

For the ankle dorsiflexion and plantarflexion moments, the peak values were almost the same at all conditions (Fig. 4-4). There was one peak on the ankle inversion moment at approximately 75% of the stance phase, while the peak of moment of ankle plantarflexion moment operated simultaneously (Fig. 4-5). The maximum value of the eversion moment was at the terminal stance phase. The peak ankle adduction moment had occurred around 20% stance phase, while the knee internal rotation moment reached the first peak (Fig. 4-6).

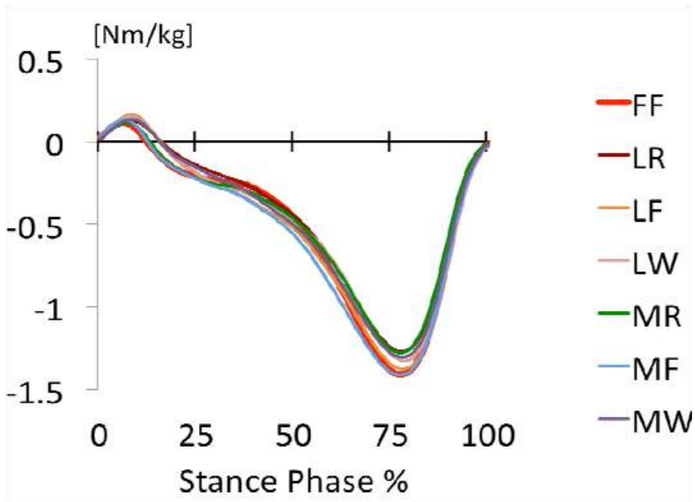

Figure 4-4. Ankle joint dorsiflexion (+) /plantarflexion moments in difference conditions during stance phase

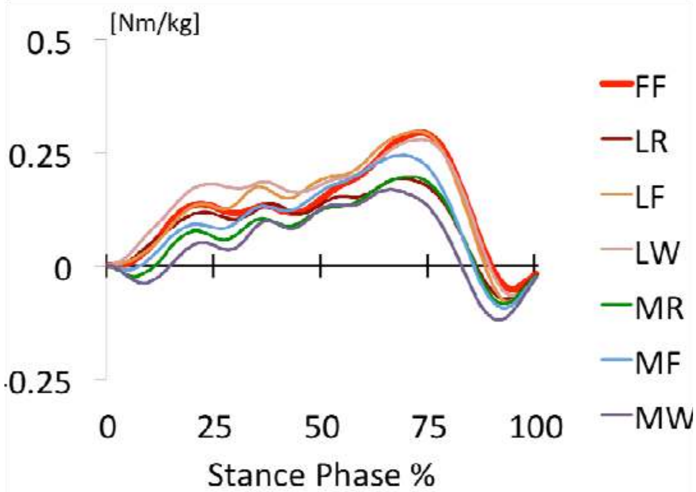

Figure 4-5. Ankle joint inversion (+) /eversion moments in difference conditions during stance phase

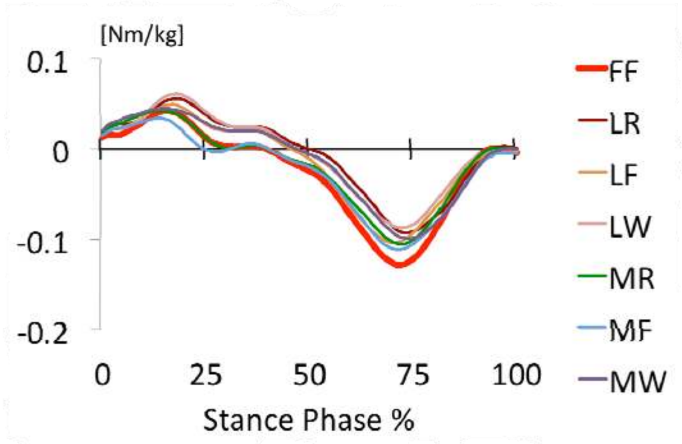

Figure 4-6 Ankle joint adduction (+) /abduction moments in difference conditions during stance phase

4.4 Forefoot Angle

The forefoot-rearfoot angle creates a motion that relates the forefoot to the rearfoot in three directions, forefoot dorsiflexion/plantarflexion was showing in Figure 4-7. Though all conditions had a small peak at the initial stance phase, the forefoot angle behaved differently. A peak dorsiflexion angle located at 90% of the stance phase along the toe-off phase was found for the conditions, approximately at 7.76degrees. The forefoot-rearfoot relationship was about neutral at the initial stance phase, while the FF condition apparently resulted in a plantarflexion of approximately 4.21degrees in this characterized subject.

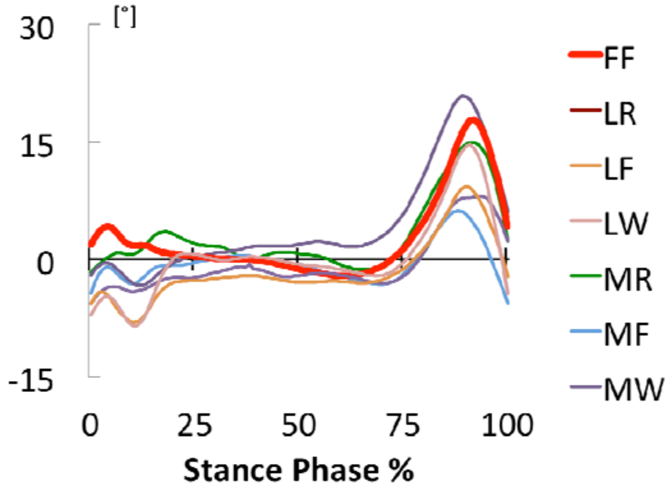

Figure 4-7. Forefoot dorsiflexion (+) /plantarflexion angles in different condition during stance phase

The peak inversion angle located at the terminal stance phase, at approximately 7.09degrees in FF condition, which was higher than the lateral wedges (Fig. 4-8). LW and LF conditions had lower eversion angles at the late stance phase. However, all three lateral wedge conditions had a higher inversion angle at the initial stance phase, with LW the highest, reaching 4.54degrees. LR led to inversion a majority of the time in initial stance and mid-stance phases. Medial wedges illustrated similar results as the lateral wedges in the investigation of eversion angle. Unlike the lateral wedge, all three medial wedge conditions had a lower eversion angle at the terminal stance phase compared with the FF condition. The MW condition had a peak inversion angle of 5.14degrees, while the MF reached an eversion angle of 4.54 degrees in the initial stance phase.

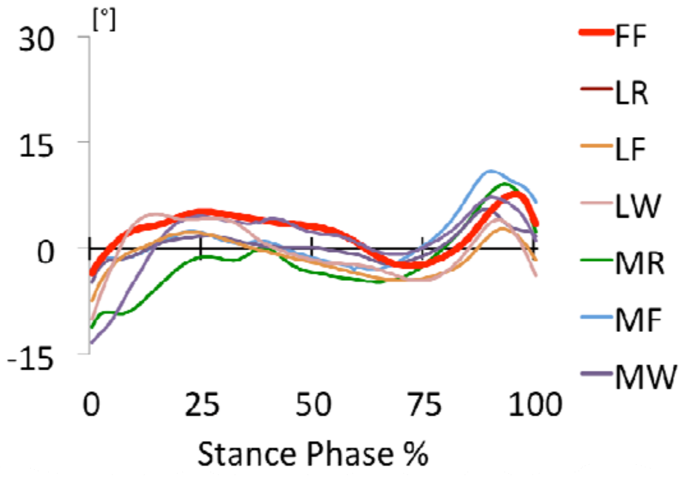

Figure 4-8. Forefoot inversion (+) /eversion angles in different condition during stance phase

There was a peak in the adduction angle that occurred at the terminal stance phases (Fig. 4-9). The adduction angle has a degressive tendency at the initial stance phase, reached the plateau of abduction angle after. In fact, the maximum abduction value of 6.23degrees was at approximately 10-15% stance phase. The FADA was at the initial stance phase. MF and MW conditions reached maximum adduction angles at approximately 60-70% of stance phase.

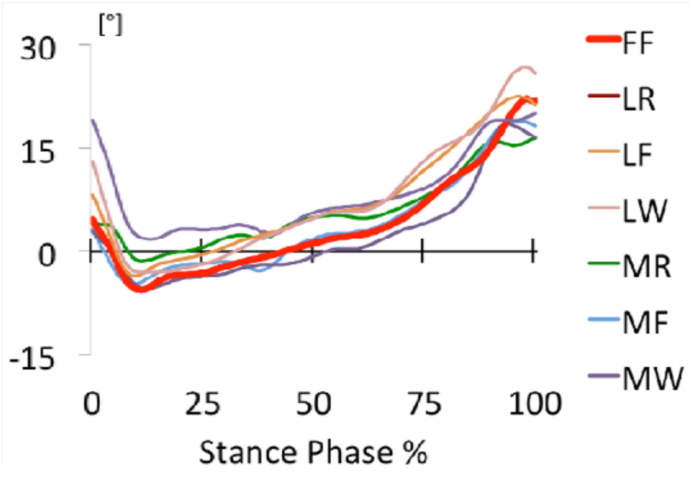

Figure 4-9. Forefoot adduction (+) /abduction angles in different condition during stance phase

4.5 Centre of Pressure

Figure 4-10 illustrates the center of pressure of one characterized subject and highlights differences between the insoles. The COP in AP direction normalized by the length of the foot segment (FtL) and the ML direction was normalized by the width of the foot segment (FtW). The LF and LW conditions laterally shifted the location of the COP during the whole stance phase compared with the FF condition. At the heel strike, the COP was at the lateral side in the LF, LW and controlled conditions. In the MF group, the COP was close to the center of the rearfoot. During the mid-stance phase, the COP for all conditions was at the lateral side of the foot. The LF, LW and MF conditions were more laterally than the FF condition, and the LF condition was the farthest (1.26FtL). The most medial one was 0.14FtW at the MR condition for this typical subject.

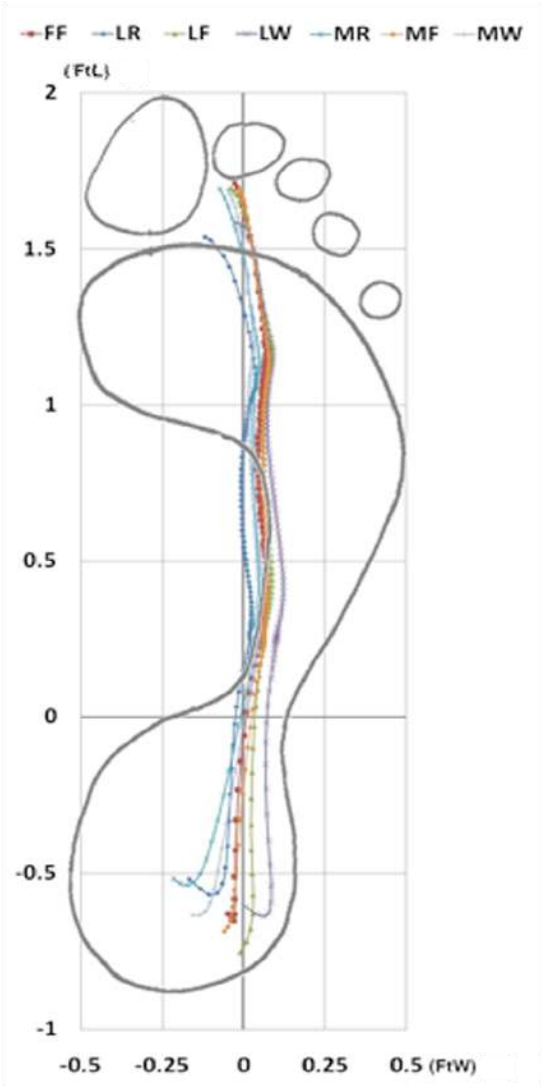

Figure 4-10. Sketch of right foot COP of individual subject during stance phase with different insoles

4.6 Knee Joint Force

The knee joint forces were calculated from the sum of the knee joint reaction forces and the total muscles forces. The overall peak knee joint resultant force (4.43 BW) occurred during the initial stance and PKF exhibited two distinct peaks (Fig. 4-11). The first PKF was reduced by 11.36% in the LW condition. There were small differences between LF, LR and LW through the stance phase. Figure 4-12 shows the muscles and knee reaction forces. The LW was 3.92BW and lower than other conditions for PKMF. The maximum value of PKRF among conditions was 0.41BW in FF condition. For the PKRF, LR was the lowest than others and the peak was 0.003BW.

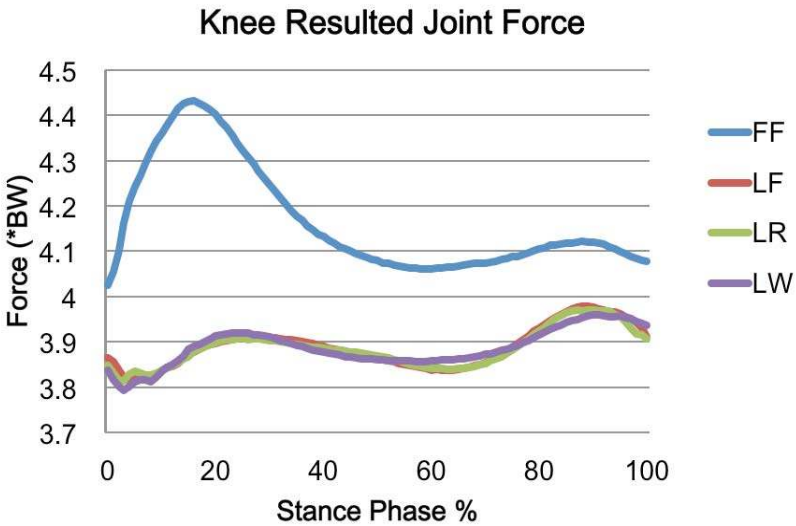

Figure 4-11. Knee joint resulted forces during stance phase for different conditions

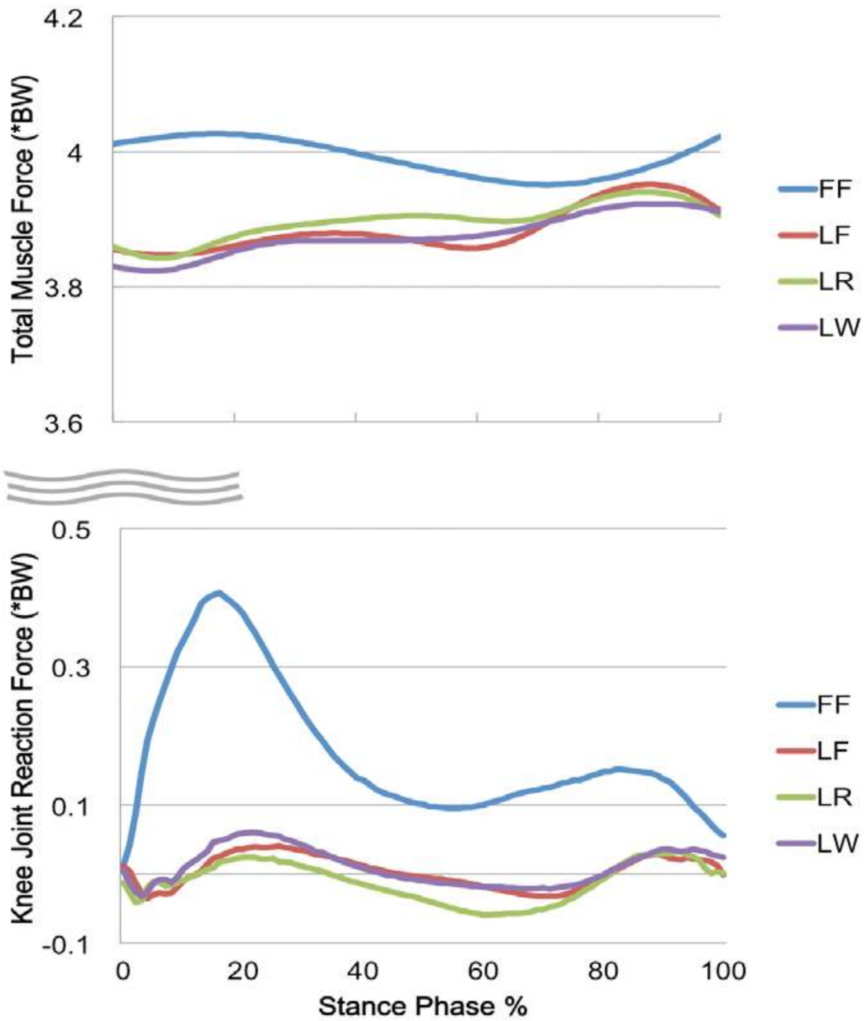

Figure 4-12. Total muscle forces and joint reaction forces

4.7 Statistical Analysis

4.7.1 Repeatability

The intraclass correlation coefficient (ICC) was used to measure the reliability. The data from the knee adduction moment were extracted for the repeatability analysis. The analysis involved the calculation of between-subject (n=10) and within-subject (n=7) reliability. The ICC computed by the SPSS showed that the within subject reliability was 0.946, and the between subject reliability was 0.898 (Table 4-19).

Table 4-1. Single measures ICC within subject and between subject

Intraclass Correlation Coefficient							
	Intraclass Correlation[a]	95% Confidence Interval		F Test with True Value 0			
		Lower Bound	Upper Bound	Value	df1	df2	Sig
Single Measures within-subject	0.946[b]	0.912	0.971	123.625	27	162	0.000
Single Measures between-subject	0.898[b]	0.841	0.944	89.263	27	243	0.000
Two-way mixed effects model where people effects are random and measures effects are fixed.							
a. Type C intraclass correlation coefficients using a consistency definition-the between-measure variance is excluded from the denominator variance.							
b. The estimator is the same, whether the interaction effect is present or not.							

4.7.2 Knee Adduction Moment

Figure 4-13 show the mean and standard deviation of knee adduction moments normalized by body weight for various conditions. LR, LF, MF, LW and MW conditions all had lower magnitude adduction moments, whereas MR condition had a higher moment compared to the FF condition in the KADM1. Both the LF and MW conditions had higher magnitude KADM1 than those of LR, LW, MF and LW. The

LW condition reached the lowest KADM1 around -0.26Nm/kg.

The knee adduction moment reached another plateau, KADM2, with MR condition having the lowest moment, at approximately -0.26Nm/kg. LW had the highest KADM2, valued at -0.17Nm/kg, while FF had a lower value at -0.26Nm/kg. The KADM2 was lower than the first peak by 26.17%. Compared the FF condition and wedge conditions, the changes of the second peak for was similar to that of the first peak, except that MR was increased in the second peak.

The peak knee abduction moment (KABM) was at the initial stance phase. The FF condition showed the largest maximum knee abduction moment at the first peak, while MF condition showed the smallest knee abduction moment at 0.09Nm/kg.

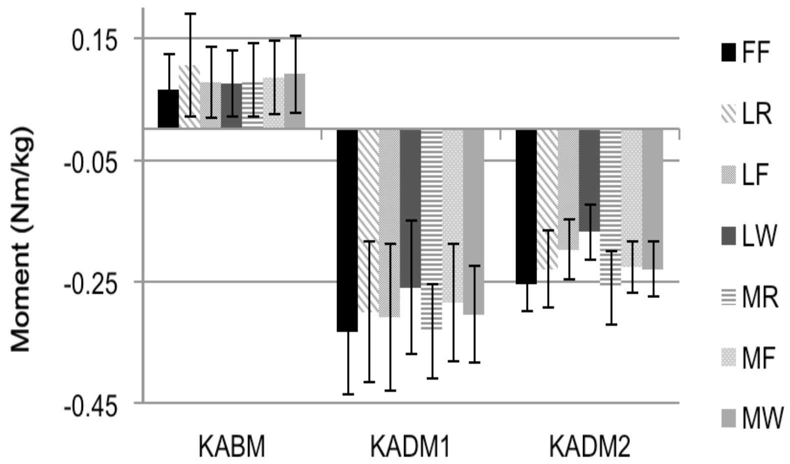

Figure 4-13.Peak knee adduction moments normalized by the body weight for different conditions

Table 4-2 shows the result of one-way repeated measures ANOVA for the peak knee adduction/abduction moment, with the independent variables as different conditions. There were no significant differences for the KABM (p=0.31). However, the differences respects to different conditions were significant at KADM1 (p=0.039) and KADM2 (p=0.000).

Table 4-2. One-way repeated measures ANOVA studying the effect of different conditions on the knee adduction and abduction moments

Tests of Within-Subjects Effects							
Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
KADM1	Huynh-Feldt	0.04	3.80	0.01	2.88	0.039	0.24
KADM2	Sphericity Assumed	0.06	6.00	0.01	8.73	0.00	0.49
KABM	Greenhouse-Geisser	0.01	2.65	0.00	1.26	0.31	0.12

Table 4-3 shows the mutual comparison of each condition with respect to the KADM1 and KADM2. There was no significant difference between any conditions at KABM. At the KADM1, there were generally no significant differences between conditions, despite the LW and FF (p=0.023), MR and LF (p=0.021), and MR and MW (p=0.003) exhibited significant differences. For the KADM2, there were significant differences among the conditions between LF, LW, MF and FF.

Table 4-3. Multiple post-hoc comparisons for KADM1 and KADM2 (Extract from Appendix A-E)

Pairwise Comparisons					
	Group (I)	Group (J)	Mean Difference (I-J)	Std. Error	Sig.a
KADM1	LW	LF	0.049*	0.018	0.023
		MR	0.072*	0.026	0.021
		MF	0.024*	0.007	0.008
		FF	0.072*	0.026	0.023
	MR	LF	-0.072*	0.026	0.021
		MW	-0.028*	0.007	0.003
KADM2	LR	LF	-0.032*	0.012	0.027
		LW	-0.061*	0.014	0.002
	LF	LW	-0.029*	0.010	0.019
		MF	0.029*	0.013	0.049
	LW	MF	0.058*	0.011	0.001
		MW	0.062*	0.018	0.008
	MR	LF	-0.063*	0.022	0.017
		LW	-0.092*	0.022	0.002
		MW	-0.030*	0.009	0.011
	FF	LF	-0.058*	0.012	0.001
		LW	-0.087*	0.013	0.000
		MF	-0.029*	.011	.026

Based on estimated marginal means
a. Adjustment for multiple comparisons: Lest Significant Difference (equivalent to no adjustments)
*. The mean difference is significant at the 0.05 level

4.7.3 Knee Internal Rotation/External Rotation Moment

The knee internal/external rotation moments in various wedge conditions shows in Figure 4-14. The peak of knee external rotation moments was occurred at the end of mid-stance phase. LR condition had a maximum magnitude of knee external rotation moment, with a value of -0.07Nm/kg, which was the lowest among all the conditions. However, LW had a higher KERM than FF condition, decreased approximately 20.97%.

Except for LR and MR conditions, all other wedge conditions showed a larger KIRM1 compared with the FF condition. The magnitudes of the KIRM2 were smaller than those of the first peak. The magnitude was reduced by 68.30% in FF condition. The medial wedge conditions were slightly smaller than the FF condition for KIRM2, while the lateral wedge conditions had a slightly larger.

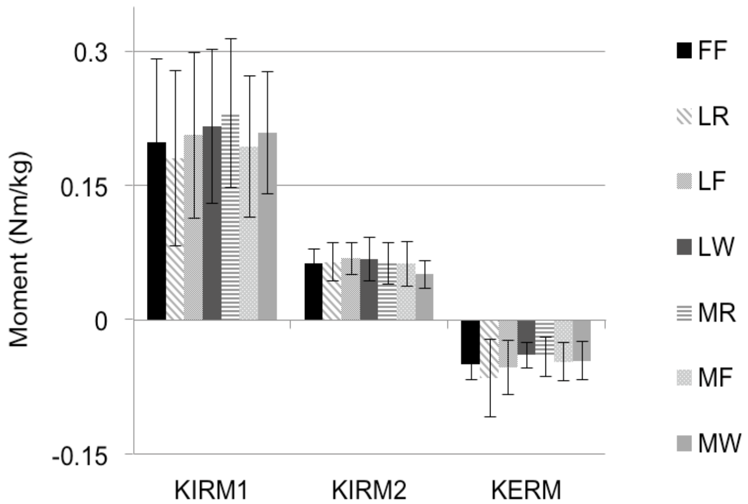

Figure 4-14. Peak knee internal and external rotation moments normalized by body weight for different conditions

Table 4-4 shows a statistical analysis on the effect of wedges on the first and second peak knee internal rotations moment. The effect of different wedge conditions had no significant differences on the first ($p=0.205$) and second ($p=0.409$) peak knee internal rotations moments.

Table 4-4. One-way repeated measures ANOVA studying the effect of different conditions on the peak knee internal/external rotation moments

Tests of Within-Subjects Effects							
Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
KIRM1	Greenhouse -Geisser	0.016	2.87	0.006	1.643	0.205	0.154
KIRM2	Sphericity Assumed	0.002	6.00	0.000	1.042	0.409	0.104
KERM	Greenhouse -Geisser	0.005	2.64	0.002	2.526	0.088	0.219

Table 4-5 shows the multiple comparisons between each condition. There were significant differences between the MF condition and LW condition (p=0.041) in the KIRM1 and LR and LW (p=0.049) in the KIRM2. From the table, there was no observable difference compared with the FF and all wedge conditions for KIRM1 and KERM.

Table 4-5. Multiple post-hoc comparisons for the peak knee internal/external rotation moments (Extract from Appendix A-E)

Pairwise Comparisons					
	Group (I)	Group (J)	Mean Difference (I-J)	Std. Error	Sig. a
KIRM1	MF	LW	-0.023*	0.010	0.041
KIRM2	MW	FF	-0.012*	0.005	0.029
		LF	-0.018*	0.006	0.021
KERM	LR	LW	-0.026*	0.011	0.049
Based on estimated marginal means					
a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).					
*. The mean difference is significant at the .05 level					

4.7.4 Knee Flexion/Extension Moment

Figure 4-15 shows the first and second peak knee extension (KEM1, KEM2) and the peak knee flexion (KFM) moments for different conditions. At the KEM1, the MR group had higher extension moments at approximately 0.66Nm/kg, while the LW and MW group were apparently lower, at 0.62Nm/kg and 0.59Nm/kg, respectively.

Compared to the FF condition, LR, LF and MF conditions had smaller KEM1. LR had a 6.79% lower magnitude than the FF condition. However, in examining the

KEM2, the deviation was smaller compared to the KEM1 and the minimum value was 0.43Nm/kg at MW condition. Upon knee flexion moment, the magnitude of the peak was similar except that LR was larger.

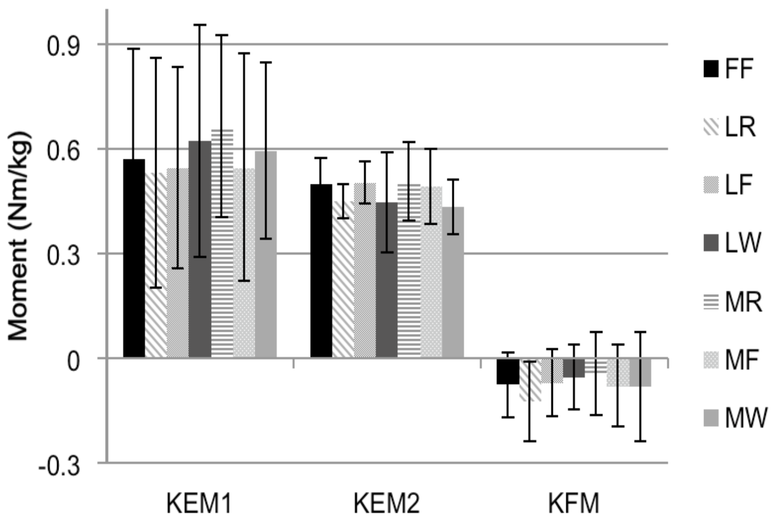

Figure 4-15. Peak knee flexion/extension moments normalized by body weight in different conditions

Table 4-6 shows one-way repeated measures ANOVA for studying the KEM1, KEM2, and KFM with respect to different conditions. There was no significant difference among different wedge conditions in studying the knee flexion and extension moments. However, the main differences between groups of FF and LR (p=0.030), FF and MW (p=0.024) for EKM2 and FF and LR (p=0.006) for KFM, showing a marginally significant difference (Table 4-7).

Table 4-6. One-way repeated measures ANOVA studying the effect of different conditions on the f peak knee flexion/extension moments

Tests of Within-Subjects Effects							
Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
KEM1	Sphericity Assumed	0.136	6.00	0.023	1.088	0.381	0.108
KEM2	Sphericity Assumed	0.058	6.00	0.010	1.526	0.187	0.145
KFM	Sphericity Assumed	0.037	6.00	0.006	1.780	0.121	0.165

Table 4-7. Multiple post-hoc comparisons for the peak knee flexion/extension moments (Extract from Appendices A-E)

Pairwise Comparisons					
	Group (I)	Group (J)	Mean Difference (I-J)	Std. Error	Sig.a
KEM 1	MF	LW	-0.076*	0.032	0.041
KEM 2	FF	LR	0.050*	0.019	0.030
		MW	0.068*	0.024	0.022
	LF	LR	0.052*	0.018	0.020
		MW	0.070*	0.028	0.032
KFM	LR	MR	-0.078*	0.028	0.020
		FF	-0.049*	0.014	0.006
		LW	-0.069*	0.023	0.016
Based on estimated marginal means a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments). *. The mean difference is significant at the .05 level					

4.7.5 Ankle Moment

Figure 4-16 shows the peak ankle inversion (AIVM) and eversion (AEVM) moments, peak ankle dorsiflexion and plantarflexion moments (ADFM, APFM), and peak ankle adduction and abduction moments (AADM, AABM) for different conditions. The LW condition had the highest magnitude for AIVM, but a lowest for AEVM among other conditions. There were the decreases for AIVM in LR, MF and MR conditions, the most reduction in MR was about 9.86%, compared with FF condition. For the AEVM, all wedge conditions had an apparently lower than with the FF condition.

In consideration both of AADM and AABM, the lateral wedge conditions were higher than FF condition, while all medial wedge conditions was lower for AABM. Though the MF condition had apparently higher peak than the FF condition, the other medial conditions reduced the AADM. The magnitudes for APFM were higher than other moments.

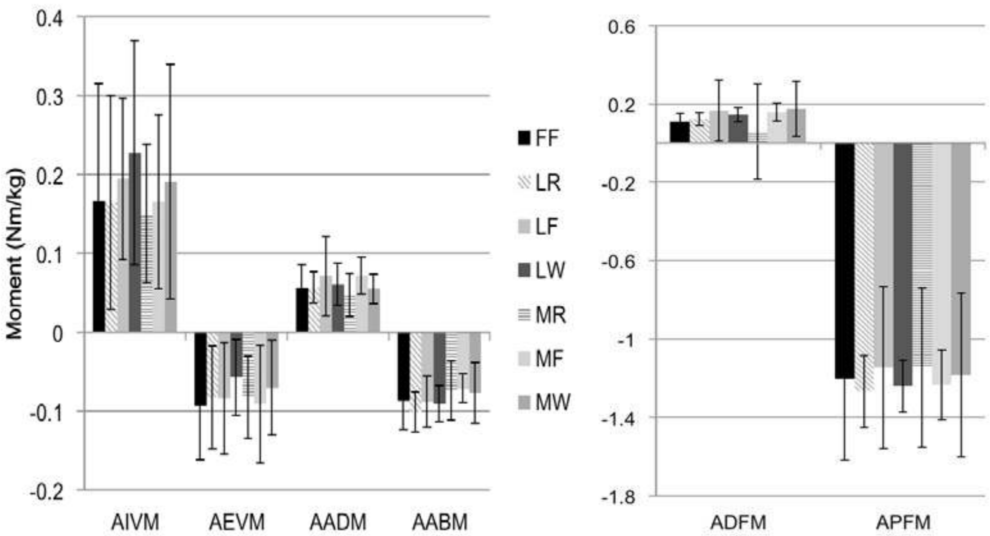

Figure 4-16. Ankle moments normalized by body weight in different conditions

Table 4-8 presents one-way repeated measures ANOVA for studying the ankle moment with respect to different conditions. There were no significant differences among different conditions for the ankle moments except the AIVM (p=0.008). There were significant differences between LW and FF in the AIVM (p=0.002), AEVM (p=0.039), ADFM (p=0.010), and AADM (P=0.033) (Table 4-9). The LW group presented significant differences between groups except for the LF and MW conditions for the AIVM. The conditions LR and MF had significant differences compared with LW for AEVM.

Table 4-8. One-way repeated measures ANOVA for studying the effect of different conditions on the peak ankle moments

Tests of Within-Subjects Effects							
Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
AIVM	Sphericity Assumed	0.041	6.00	0.007	3.292	0.008	0.268
AEVM	Sphericity Assumed	0.010	6.00	0.002	1.126	0.360	0.111
ADFM	Greenhouse-Geisser	0.099	2.47	0.040	1.134	0.349	0.112
APFM	Sphericity Assumed	0.138	6.00	0.023	0.246	0.959	0.027
AADM	Greenhouse-Geisser	0.005	2.34	0.002	1.463	0.254	0.140
AABM	Sphericity Assumed	0.007	6.00	0.001	1.494	0.198	0.142

Table 4-9. Multiple post-hoc comparisons for the ankle moments (Extract from Appendix A-E)

Pairwise Comparisons					
	Group (I)	Group (J)	Mean Difference (I-J)	Std. Error	Sig.a
AIVM	FF	LW	-0.061*	0.014	0.002
	LF	MR	0.045*	0.015	0.016
		MF	0.029*	0.008	0.007
	LW	LR	0.063*	0.016	0.003
		MR	0.078*	0.025	0.012
		MF	0.062*	0.015	0.003
AEVM	LW	FF	0.036*	0.015	0.039
		LR	0.026*	0.01	0.026
		MF	0.034*	0.011	0.01
ADFM	FF	LF	-0.036*	0.010	0.006
		LW	-0.049*	0.015	0.010
	LW	LR	0.037*	0.015	0.033
AADM	FF	LW	-0.015*	0.006	0.033
	LW	LR	0.014*	0.003	0.002
		LF	0.010*	0.004	0.032
		MW	0.016*	0.006	0.018
AABM	LR	LW	-0.029*	0.007	0.003
		MW	-0.024*	0.010	0.033
	LF	LW	-0.019*	0.007	0.030

Based on estimated marginal means
a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).
*. The mean difference is significant at the .05 level

4.7.6 Ankle Angle and Forefoot Angle

Table 4-10 presents one-way repeated measures ANOVA for studying the ankle angles with respect to different conditions. There were significant differences among different conditions for the AIVA (p=0.03) and AADA (p=0.01). Other peak angles were no significant differences. Though the peak forefoot angles were no significant differences among different conditions (Table 4-11), the differences between FF condition and LW condition were significant for FEVA (p=0.025) and FIVA (p=0.042) (Table 4-12).

Table 4-10. One-way repeated measures ANOVA for studying the effect of different conditions on the peak ankle angles (Extract from Appendix A-E)

Tests of Within-Subjects Effects							
Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
ADFA	Greenhouse-Geisser	1185.63	1.68	706.57	0.90	0.41	0.09
APFA	Greenhouse-Geisser	1281.42	1.56	819.12	1.17	0.33	0.11
AIVA	Greenhouse-Geisser	185.88	2.79	66.59	3.53	0.03	0.28
AEVA	Sphericity Assumed	99.45	6.00	16.58	0.78	0.59	0.08
AADA	Sphericity Assumed	158.19	6.00	26.36	3.11	0.01	0.26

Table 4-11. One-way repeated measures ANOVA for studying the effect of different conditions on the peak forefoot angles (Extract from Appendix A-E)

Tests of Within-Subjects Effects							
Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
FEVA	Sphericity Assumed	242.11	6.00	40.35	2.22	0.05	0.20
FIVA	Sphericity Assumed	138.76	6.00	23.13	1.89	0.10	0.17
FPFA	Greenhouse-Geisser	212.21	2.00	105.87	1.18	0.33	0.12
FDFA	Sphericity Assumed	311.00	6.00	51.83	1.27	0.29	0.12
FABA	Sphericity Assumed	96.67	6.00	16.11	0.64	0.69	0.07
FADA	Sphericity Assumed	239.17	6.00	39.86	1.66	0.15	0.16

Table 4-12. Multiple post-hoc comparisons for the ankle angles and forefoot angles (Extract from Appendix A-E)

Pairwise Comparisons					
	Group (I)	Group (J)	Mean Difference (I-J)	Std. Error	Sig.a
AIVA	LR	LF	4.064*	1.636	.035
		LW	3.007*	1.231	.037
	MR	LF	3.209*	0.931	.007
		MF	2.835*	1.207	.043
		LW	2.152*	0.791	.024
	LF	LR	-4.064*	1.636	.035
		MR	-3.209*	0.931	.007
		MW	-4.255*	1.118	.004
	MW	LF	4.255*	1.118	.004
		MF	3.881*	1.268	.014
AADA	LR	LF	4.104*	1.536	.026
		LW	3.234*	1.286	.033
	MR	LF	2.709*	1.020	.026
	MW	LF	3.559*	1.063	.009
		MF	3.220*	1.218	.027
FEVA	FF	MR	4.855*	1.909	.032
		LW	6.014*	2.242	.025

FIVA	FF	LW	2.991*	1.266	.042
	MR	MF	-3.614*	1.337	.024
FADA	MR	FF	-6.205*	2.030	.014
		LR	-3.253*	.927	.007
		LF	-4.512*	1.697	.026
		MF	-5.257*	2.251	.044
FPFA	LW	MW	-2.193*	.912	.040
Based on estimated marginal means					
a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).					
*. The mean difference is significant at the .05 level					

4.7.7 Centre of Pressure

An extract of COP is shown in Table 4-11. The maximum and minimum COP in the AP and ML directions were calculated by ruling out the data beyond the stance phase cycle. The information shows the overall shifting of the COP due to the insoles, with positive denoted medial and anterior directions. A natural lateral shift was shown for the FF group with flat insole, with a value of approximately 0.16FtW. All wedge conditions had greater lateral shifts in their COPs, while LW condition had the longest shifting distance (0.18FtW). MW resulted in a maximum medial shift of COP, and was medially shifted approximately 0.07FtW compared with FF. Upon AP direction, the COP was nearly neutral, with a value of 1.54FtL. MW and MF resulted in a further anterior shift of the COP to approximately 1.64FtL. The MR condition had the most posterior shift of COP (0.13FtL) compared with FF condition.

Table 4-13. The mean and standards deviation of center of pressure in both directions

	COPL		COPM		COPA		COPP	
Condition	Mean	Std. Deviation	Mean	Std. Deviation	Mean	Std. Deviation	Mean	Std. Deviation
FF	0.16	0.10	-0.14	0.21	1.54	0.36	-0.72	0.32
LR	0.10	0.06	-0.23	0.21	1.53	0.36	-0.70	0.41
MR	0.11	0.06	-0.23	0.22	1.36	0.56	-0.59	0.30
LF	0.16	0.08	-0.15	0.18	1.43	0.40	-0.62	0.26
MF	0.15	0.10	-0.16	0.18	1.64	0.24	-0.76	0.38
LW	0.18	0.10	-0.16	0.21	1.58	0.16	-0.65	0.37
MW	0.11	0.10	-0.24	0.22	1.64	0.16	-0.66	0.43

Table 4-13 shows a shift of the center of pressure for different conditions. There was significant difference at the COPL. However, when all subjects were considered, the conditions showed no significant differences at COPM, COPA, and COPP (Table 4-14).

Table 4-14. One-way repeated measures ANOVA on COP in both directions with different conditions

Tests of Within-Subjects Effects							
Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COPL	Greenhouse-Geisser	0.047	3.21	0.015	3.421	0.028	0.275
COPM	Sphericity Assumed	0.111	6.00	0.019	1.929	0.093	0.177
COPA	Huynh-Feldt	0.684	2.77	0.247	1.711	0.193	0.160
COPP	Greenhouse-Geisser	0.201	2.72	0.074	0.909	0.443	0.092

In comparing each pair of conditions (Table 4-13), there was significant difference in COPL. However, there were significant differences in the COPM for the pair of conditions MR and FF (p=0.035) and MW and FF (p=0.038). The AP direction showed no differences among all the groups.

Table 4-15. Multiple post-hoc comparisons for the COPM and COPL (Extract on appendix A-E)

Pairwise Comparisons					
	Group (I)	Group (J)	Mean Difference (I-J)	Std. Error	Sig.a
COPL	LF	LR	0.054*	0.016	0.009
		MR	0.044*	0.014	0.014
		MW	0.043*	0.018	0.040
	LW	LR	0.074*	0.026	0.018
		MR	0.063*	0.026	0.035
		MW	0.063*	0.024	0.028
COPM	MR	FF	-0.082*	0.033	0.035
		MF	-0.068*	0.027	0.034
	MW	FF	-0.096*	0.039	0.038
		LW	-0.075*	0.026	0.018
Based on estimated marginal means a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments). *. The mean difference is significant at the .05 level					

Summary

Table 4-16 shows the means and standard deviation for all kinematic and kinetic variables and figures the increased or decreased percentage for the wedge conditions compared with FF condition. There were significant differences for KADM2, AIVM, AIVA, AADA and COPL. In the multiple post-hoc comparisons, the significant differences were between LW condition and others for more variables (Table 4-17).

Table 4-17 Summary of the multiple post-hoc comparisons among conditions for 30 variables

	MR	MF	MW	LR	LF	LW
FF	COP MFA DA	KADM2	COPM, KAM2, KEM2	KEM2 KFM	ADFM, KADM2	FIVA, AIVM, ADFM, AADM, AEVM, KADM1, KADM2
MR		COPM, FIVA, FADA, AIVA	KADM1, KADM2	KFM, FADA	COPL, AIVA, AIVM, AADM, KADM1, KADM2	COPL, AIVA, AIVM, KEM1, KADM1, KADM2
MF			AADA		AIVM, KADM2	AIVM, AADM, AEVM, KIRM1, KADM1, KADM2
MW				AABM	COPL, AIVA, KEM2, AADA	COPL, COPM, FPFA KADM2,
LR					COPL, AIVA, AADA, KEM2, KADM2	COPL, AIVA, AADA, ADFM, AADM, AABM, AEVM, AIVM, KFM, KERM, KADM2
LF						AADM, AABM, KADM1, KADM2

Table 4-16. Mean and standard deviations values for all kinematic and kinetic variables (n=10 subjects)

		FF		LR			LF			LW			MR			MF			MW		
	Sig.	Mean	Std. D	Mean	Std. D	Inc. (%)	Mean	Std. D	Inc. (%)	Mean	Std. D	Inc. (%)	Mean	Std. D	Inc. (%)	Mean	Std. D	Inc. (%)	Mean	Std. D	Inc. (%)
Knee moment (Nm/kg)																					
KADM1	0.039	-0.33	0.10	-0.30	0.12	-9.6	-0.31	0.12	-7.0	-0.26	0.11	-21.7	-0.33	0.08	0.0	-0.28	0.10	-14.5	-0.30	0.08	-8.4
KADM2	0.000	-0.26	0.04	-0.23	0.06	-10.2	-0.20	0.05	-22.6	-0.17	0.05	-34.0	-0.26	0.06	2.0	-0.23	0.04	-11.4	-0.23	0.05	-9.8
KABM	N/A	0.07	0.06	0.11	0.08	60.3	0.08	0.06	17.1	0.08	0.05	15.0	0.08	0.06	24.1	0.09	0.06	30.1	0.09	0.06	38.6
KEM1	N/A	0.57	0.32	0.53	0.33	-6.8	0.55	0.29	-4.2	0.62	0.33	9.2	0.66	0.26	16.3	0.55	0.33	-4.2	0.59	0.25	4.3
KEM2	N/A	0.50	0.07	0.45	0.05	-9.9	0.50	0.06	0.5	0.45	0.14	-10.8	0.50	0.11	1.0	0.49	0.11	-1.7	0.43	0.08	-13.6
KFM	N/A	-0.08	0.09	-0.12	0.11	64.7	-0.07	0.10	-6.1	-0.05	0.09	-27.5	-0.05	0.12	-39.1	-0.08	0.12	5.3	-0.08	0.16	7.2
KIRM1	N/A	0.20	0.09	0.18	0.10	-8.9	0.21	0.09	4.1	0.22	0.09	8.9	0.23	0.08	16.6	0.19	0.08	-2.6	0.21	0.07	5.1
KIRM2	N/A	0.06	0.02	0.06	0.02	2.3	0.07	0.02	8.8	0.07	0.02	7.1	0.06	0.02	-0.1	0.06	0.03	-0.5	0.05	0.02	-19.5
KERM	N/A	-0.05	0.02	-0.07	0.04	30.8	-0.05	0.03	7.0	-0.04	0.01	-21.0	-0.04	0.02	-17.2	-0.05	0.02	-6.1	-0.05	0.02	-8.2
Ankle moment (Nm/kg)																					
ADFM	N/A	0.11	0.04	0.12	0.03	10.6	0.17	0.16	48.8	0.15	0.04	32.2	0.06	0.24	-47.6	0.16	0.05	44.1	0.17	0.14	57.7
APFM	N/A	-1.20	0.41	-1.27	0.18	5.4	-1.14	0.41	-4.8	-1.24	0.13	3.1	-1.14	0.41	-4.9	-1.23	0.18	2.7	-1.18	0.42	-1.6
AIVM	0.008	0.17	0.15	0.16	0.14	-0.9	0.19	0.10	17.0	0.23	0.14	36.9	0.15	0.09	-9.9	0.17	0.11	-0.4	0.19	0.15	14.5
AEVM	N/A	-0.09	0.07	-0.08	0.07	-11.0	-0.08	0.07	-10.4	-0.06	0.05	-39.0	-0.08	0.05	-11.6	-0.09	0.07	-2.3	-0.07	0.06	-24.6
AADM	N/A	0.06	0.03	0.06	0.02	1.2	0.07	0.05	26.4	0.06	0.03	8.1	0.05	0.03	-15.9	0.07	0.02	26.5	0.06	0.02	-1.9
AABM	N/A	-0.09	0.04	-0.10	0.03	15.6	-0.09	0.03	1.3	-0.09	0.02	4.0	-0.07	0.04	-14.9	-0.07	0.02	-18.0	-0.08	0.04	-12.0
Ankle angle (Degree)																					
ADFA	N/A	101.67	15.06	96.05	31.54	-5.5	104.06	7.37	2.3	105.20	10.22	3.5	108.25	11.39	6.5	106.28	7.97	4.5	109.12	4.87	7.3
APFA	N/A	80.04	12.13	72.86	30.60	-9.0	81.66	6.86	2.0	82.90	8.11	3.6	85.84	8.73	7.2	85.47	5.21	6.8	85.61	3.66	7.0
AIVA	0.032	2.07	3.56	4.66	4.49	125.0	.60	1.92	-71.3	1.65	2.33	-20.2	3.80	2.96	83.7	.97	3.19	-53.2	4.85	3.41	134.2
AEVA	N/A	-14.23	6.70	-14.87	6.37	4.5	-16.94	5.35	19.0	-15.35	10.42	7.8	-13.35	5.96	-6.2	-13.17	6.49	-7.5	-14.97	5.01	5.2
AADA	0.011	2.68	3.17	5.58	4.12	108.2	1.48	1.67	-44.9	2.35	2.73	-12.5	4.19	2.56	56.2	1.82	3.11	-32.3	5.04	3.18	87.9
Forefoot angle (Degree)																					
FDFA	N/A	24.08	5.39	22.42	8.73	-6.9	20.82	10.22	-13.5	22.90	7.33	-4.9	23.73	8.95	-1.4	22.97	9.12	-4.6	28.17	6.55	17.0
FPFA	N/A	-1.75	9.35	-0.72	5.45	-58.6	-2.52	5.02	44.0	-3.97	3.35	126.7	-0.10	7.47	-94.5	-5.54	5.06	216.7	-1.77	3.32	1.3
FEVA	N/A	11.06	5.35	6.64	4.21	-39.9	8.39	3.91	-24.2	5.04	4.26	-54.4	6.20	3.21	-43.9	8.85	6.71	-19.9	6.78	2.24	-38.7
FIVA	N/A	1.68	1.94	-0.46	3.57	-127.6	1.22	5.22	-27.1	-1.31	3.82	-178.0	-1.22	3.45	-172.5	2.40	4.95	42.5	1.61	2.08	-4.2
FADA	N/A	14.06	8.17	11.11	6.01	-21.0	12.37	6.27	-12.0	10.99	7.64	-21.8	7.86	5.09	-44.1	13.12	6.30	-6.7	10.94	7.97	-22.2
FABA	N/A	-4.54	3.98	-5.52	4.68	21.6	-4.58	3.26	0.9	-2.56	4.26	-43.7	-2.81	3.84	-38.0	-2.68	6.39	-40.9	-5.22	6.13	14.9
Center of pressure (FtL or FtW)																					
COPL	0.028	0.16	0.10	0.10	0.06	-33.8	0.11	0.06	-27.1	0.16	0.08	0.6	0.15	0.10	-4.0	0.18	0.10	13.1	0.11	0.10	-26.9
COPM	N/A	-0.14	0.21	-0.23	0.21	63.0	-0.23	0.22	56.7	-0.15	0.18	4.4	-0.16	0.18	9.4	-0.16	0.21	14.3	-0.24	0.22	66.7
COPA	N/A	1.54	0.36	1.53	0.36	-0.1	1.36	0.56	-11.7	1.43	0.40	-7.2	1.64	0.24	6.6	1.58	0.16	3.1	1.64	0.16	6.9
COPP	N/A	-0.72	0.32	-0.70	0.41	-2.7	-0.59	0.30	-17.4	-0.62	0.26	-13.4	-0.76	0.38	5.9	-0.65	0.37	-9.9	-0.66	0.43	-7.6

4. 8 Correlations between variables

This section illustrates the correlations between kinetic and kinematic variables across different body segments, including the knee and the ankle. Table 4-18 presents the correlation between knee adduction moments and the resultant ankle segment parameters.

Table 4-18. Correlation between knee adduction moments and other variables

Variables	**Parameters**	**Knee Adduction Moments**		
		KADM1	KADM2	KABM
Shift of COP	COPL		0.304*	
	COPA			0.428*
Angles	AIVA	0.325*		
	AEVA	0.323*		
	AADA		-0.253*	
	AABA			0.290*
	FEVA			-0.256*
	FADA			0.238*
Moments	KEM1		-0.313**	0.468**
	KEM2	-0.344**	0.322**	
	KIRM1	-0.353**	-0.357**	
	KIRM2		0.346**	
	KERM		0.250*	
	AIVM			
	APFM			-0.335**
	AABM	-0.253*	0.357**	0.359**
*, Correlation was considered significant at p<0.05 (2-tailed) **, Correlation was considered significant at p<0.001 (2-tailed)				

Knee adduction moments correlated with both ankle angles and moments. KADM1 was moderately correlated with ankle inversion angle (r=0.325), while less correlated with AABM (r=-0.253) negatively. Also, there were negative correlations between KADM1 and KEM2 (r=-0.344) and KIRM1 (r=-0.353), respectively.

KADM2 was significantly related to ankle abduction moment (r=0.357). The first peak knee extension and internal rotation moments and KADM2 presented negatively correlation coefficient of 0.313 and 0.346. However, the KAD2 moderately correlated with KEM2 and KIRM2 was positive (r=0.322, r=0.346). For the COPL, the correlation with KADM2 was moderately (r=0.304).

In terms of the KABM, there was a fair correlation with FEVA (r=-0.256) and little relationship with FADA (r=0.238). The KABM was correlated with COPA (r=0.428) and KEM1 (r=0.468) moderately.

Table 4-19 shows the correlation between a shift in COP and other variables. In the ML-direction, COP correlated with the forefoot inversion angle and the ankle inversion moments. KEM1 had negative COPL correlation factors of 0.297. The maximum medial shift COP and the KEM1 were also negatively correlated (r=-0.459). The adduction (toe-out) angle at the ankle joint correlated with the COP in the ML direction (r=-0.405; -0.278). The COPM was a negative, which was decreased present the COP shift more laterally. COPA correlated with AABA (r=0.311) and COPP positively correlated with FADA and AABM (r=0.335, 0.306). Interestingly, there was less of the correlation between the AABA and COPM than between the AABA and COPL moments.

Table 4-19. Correlation between a shift in COP and joint angles and moments

Variable	**Parameter**	**Shift in COP**			
		COPL	COPM	COPA	COPP
Angle	APFA		-0.237*		
	ADFA			0.278*	
	AEVA	0.313**			
	AIVA		0.517**		0.276*
	AABA	-0.405**	-0.278*	0.311**	
	FIVA	-0.260*			
	FADA	0.265*	0.256*		0.335**
Moments	AABM				0.306**
	AIVM	0.385**		0.300*	
	KEM2	0.410**	0.350**		
	KEM1	-0.297*	-0.459**		

*, Correlation was considered significant at p<0.05 (2-tailed)
**, Correlation was considered significant at p<0.01 (2-tailed)

The peaks of ankle moments correlated with the joint angles and moments are shown Table 4-20. There were high correlations between moments and angles of the ankle joint. The highest correlation was between AIVA and AEVM (r=1.00). The high negatively correlated with AIVM and AEVA (r=-0.746), while the AIVA positively correlated with AIVM, (r=0.730). ADDM correlated with the knee moments KFM, KIRM1 and KERM (r=0.445, 0.258, 0.273). The APFM correlated with the forefoot angles FPFA, FDFA and FABA (r=-0.358, -0.336, 0.237).

Table 4-20. Correlation between ankle moments and other variables

Variable	**Parameter**	**Ankle Moments**					
		AEVM	AIVM	ADFM	APFM	AADM	AABM
Moment	KEM2		0.240*				
	KFM		-0.304*			0.445**	
	KIRM1					0.258*	
	KERM					0.273*	
Angle	AIVA	1.000**	0.730**				0.283*
	AEVA	-0.592**	-0.746**				
	ADFA		0.278*		-0.300*		
	APFA		0.235*				
	AADA	-0.497**		0.259*			
	AABA	-0.258*	-0.309**				
	FDFA	-0.307**		0.363**	-0.358**		-0.396**
	FPFA				-0.336**		
	FABA				0.237*		0.263*
*, Correlation was considered significant at p<0.05 (2-tailed) **, Correlation was considered significant at p<0.01 (2-tailed)							

CHAPTER V DISCUSSIONS

The current studies investigated the biomechanical effects of different wedge locations, including forefoot, rearfoot, lateral and medial posting, aiming to provide comprehensive information for designing foot orthoses for osteoarthritis patients. The study investigated the effect of these interventions on kinetic and kinematic parameters measured from walking. The wedges were analyzed for their effect on knee moments, ankle moments, ankle angle, forefoot angle, and center of pressure. The relationships between the variables were also determined. A summary of change percentage for each of the respective variables listed in Table 4-16 and the effects of the wedge insole listed in Table 4-17.

Several aspects of the study are unique. First, the objectives were based on a proposed mechanism which hypothesized that pain relief reported by people with medial compartment knee OA using heel wedges was the result of shifting the center of pressure laterally, thus reducing the knee adduction moment. The mechanism did not hypothesize that pain relief associated with these interventions was related to forefoot-rearfoot coupling. Second, this study investigated both kinematic and kinetic variables in subject use of both lateral/medial wedges and forefoot/rearfoot/full-length wedge. The results demonstrated that the use of a full-length lateral wedge affected plurality of the parameters, such as the peak knee adduction moment (a predictor of medial joint loading) and peak displacement of center of pressure in the ML direction. The forefoot medial wedge insole also affected the peak knee adduction moment. Therefore, these results supported the theory that insole supports relieve pain associated with knee OA through influences on biomechanics or media1compartment joint loading.

All lateral wedge conditions (LR, LF, and LW) reduced the first peak knee adduction moment. The LW condition had the highest reduction (21.72%), while LR and LF had reductions of 9.64% and 7.10%, respectively. The significance was demonstrated between LW condition and FF condition (p=0.023). Because the reduction of the first peak knee adduction moment helps to reduce the symptoms of osteoarthritis, current studies demonstrated the serviceability of lateral full-length wedge. However, the magnitude of reduction in current studies was larger than the previous research. Kerrigan et al. (2002) made use of 5-degree and 10-degree full-length lateral wedges and demonstrated significant reductions of 5.4% and 9.0%, respectively, although the 5-degree insole group reported better comfort. Kakihana et al. (2005) found that 6-degree full-length lateral wedges reduced the mean knee adduction moment by 6.7% during stance phase compared with flat insoles. The use of 5-degree full length lateral wedges on subjects without osteoarthritis also demonstrated reduction in knee adduction moment (Crenshaw et al., 2000). Customized lateral wedges ranging from 5 to 15 degrees showed up to a 9% reduction in the first peak knee adduction moment (Butler et al., 2009). However, some research found no significant reductions when using lateral wedges (Nester et al., 2003, Maly et al., 2002).

Most of the research focused on full-length wedges. Investigations on different locations of the wedge, such as the forefoot and rearfoot, were seldom reported. Hinman et al. (2008) compared 5-degree full-length wedges with 5-degree rearfoot wedges. The results of this study were similar to that of Hinman et al. (2009), reporting a relative effective reduction with full-length wedges. Additionally, current studies revealed that the forefoot lateral wedge had a better effect than the rearfoot wedge, although it was less effective than a full-length wedge. Hinman et al. (2008)

reported a 12-14% reduction of the first peak knee adduction moment with a full-length lateral wedge. They postulated that the higher reduction, compared with previous literature, could be due to the difference in insole stiffness.

The second peak knee adduction moment showed a greater magnitude reduction compared with that of the first peak and it was significance (p=0.00). LW had a reduction of 34.00%, while LF had a reduction of 22.60% and the LR group showed a decrease of 10.20%. The significant differences could be found by comparing LW to the other conditions and the LF condition to the conditions except MW. The reduction was larger than that reported in current literature, possibly due to the difference in materials used for the insole. The overall reduction in the literature ranged from 4-9% (Hinman et al., 2008, Kerrigan, 2002), with a 14% reduction being reported by Hinman et al. (2009). Lateral wedged insoles might be more effective for reducing the knee adduction moment at terminal stance phase.

Current studies demonstrated medial-lateral and anterior-posterior shifts for wedges in terms of the shift of the COP. The COPL was shifted medially in lateral conditions LR and LF compared with FF condition. The study could not demonstrate significance between conditions for the peak center of pressure both in the COPA and COPP. But the effect of insoles was significance in the COPL (p=0.006). That means the wedges could shift the COP in lateral direction. It was found that the between LF and LR conditions was significant and the comparing with LR and LW was significant also. This could indicate the forefoot wedge in lateral played a main role for shift the COP.

The lateral shift caused by a lateral wedge was demonstrated by Hinman et al. (2008) and Maly et al. (2002) showed a significant reduction (p=0.036) in the shift of the center of pressure along the medial-lateral direction with 5-degree valgus modified orthoses. The magnitude of the shift in the current study was similar from Hinman et al. (2008) and Maly et al. (2002). Kakihana et al. (2005) reported a lateral shift in the average center of pressure trajectory in OA patients using lateral wedges that parallels the subtalar joint axis. One explanation could be that a greater center of pressure shift would reduce the knee adduction moment. Hinman et al. (2008) hypothesized that a full-length wedge was more effective compared with rearfoot wedge by comparing the COP shifts. The lateral wedges increased the inversion at the subtalar joint, causing a lateral shift in COP, while a lateral shift decreased the knee joint moment arm and results in a reduction in knee adduction moment. However, inconsistencies were also found, possibly due to compensatory pronation or supination and the orientation of the subtalar joint axis. The interaction test supports that forefoot movement has a much higher effect on the COP. This change in loading pattern is a potential mechanism of gait compensation used to reduce the mediolateral distance between the center of mass and the knee joint center, thereby reducing the moment arm of the ground reaction force and supposedly reducing the knee adduction moment at a later point in the stance phase.

The internal rotation moments and flexion moment of the knee were less reported in previous literature. LR decreases the knee internal rotation moment by 8.92%, though both LW and LF (8.85% and 4.07%) increased this moment. The study could not demonstrate significant differences between conditions. The LR condition shows the greatest increases in knee flexion. However, significance could not be demonstrated. Reducing the knee adduction moment may have been offset by an increase in the knee flexion moment. In fact, the increase in knee flexion moment could be well accommodated by the patellofemoral dynamics in the knee joint.

LW resulted in about a 36.90% increase in ankle eversion moment, while LR showed a 0.85% reduction. According to Hinman et al.'s hypothesis (2008), the COP shift resulting from a lateral wedge could lead to an increased ankle inversion moment. However, the difference caused by full-length wedges on other locations could be due to compensatory actions at the subtalar joint (Schmalz et al., 2006), who identified the effect of compensatory action by applying rigid and semi-rigid ankle-foot-orthoses while evaluating wedges. They discovered that wedge application might lead to movement in either joint. However, additional verification is required. Current study found the differences in the ankle inversion moment between groups were larger than differences in the knee adduction moments, which suggest that the potential mechanism of gait compensation is more likely driven by changes in ankle-joint kinetics.

Most of the research focuses on alleviating medial compartment osteoarthritis, whereas lateral wedges could reduce the knee moment and hence provide pain relief. Medial wedges were rarely considered for lateral compartment osteoarthritis. Mundermannet al. (2005) evaluated orthoses for runners. Medial wedges reduced the maximum foot eversion angle and the ankle inversion moment. That agreed with Mundermann et al. (2005), who showed that the first peak ankle inversion moment was 2-fold lower than the control group, though they did not achieve significance ($p=0.362$). Current studies investigated the medial wedge conditions reduced the

KADM1, except the MR condition. MF insoles shifted the COP laterally, while medially shifted approximately 0.24FtW in the MW condition. However, MW insoles reduced the peaks of knee adduction moment. The medial wedges did not increase the peak knee adduction moment as a reverse of the lateral condition.

The second objective of this research was to investigate the transfer mechanism of the wedges from the foot segment to the knee segment. The scope was achieved by correlation studies between different kinematic and kinetic variables. Existing research considered the transfer mechanism both by hypothesis and postulation of kinematic variables. Hinman et al. (2008) explained that the COP lateral shift induces increased ankle inversion or eversion moments and decreases the knee joint moment arm. Current studies found that a fair correlation between the COPL and the second peak knee adduction moment (r=0.304). The correlation between COP and ankle inversion moment in this study supported the hypothesis that a lateral shift in COP could alter the ankle inversion moment (r=0.385). Hurwitz et al. (2002) suggested that the toe-out angle could predict the peak adduction moment during the late stance phase. Current correlation studies suggested that the toe-out angle correlated with the COPL (r=-0.405), while secondary correlating with the second peak adduction moment.

Though significant differences were demonstrated between conditions, wedges affected the COP of subjects. A shifted COP changed the forefoot adduction angle (r=0.265) and ankle inversion moment (r=0.385), of which the forefoot adduction angle and ankle eversion moment themselves were also interrelated (r=-0.307). Interestingly, the ankle inversion moment was little correlated and the first peak knee adduction moment, while the first knee adduction moment correlated with ankle inversion angle. This correlation may be due to the relationship between the ankle eversion angle was strong correlated with the ankle inversion moment (r=0.730).

In this study, all wedged insoles reduced the knee adduction moments except the MR increased. However, just LW and MF shifted the COP lateral compared with FF. For the correlations, the COP shift laterally the magnitude of knee adduction moment was reduced. The ankle inversion angle, ankle adduction angle, ankle abduction moment and knee extension moment could affect the knee adduction moments. The magnitude of AABM increased, knee adduction moment decreased for all lateral wedged and MF insoles. The ankle inversion angles were also affected by the AABM. The COPL increased and the magnitude of AABM increased, the FADA decreased by all wedged insoles. That indicated the ankle joint motion and forefoot rearfoot motion make complex compensative for machines of lower extremities.

In summary, the current correlation study supported the hypothesis that compensatory action influenced the knee adduction moment, while the COP had more of an influence. These results agree the opinion that a lateral shift in COP increased the moment arm. The correlation analysis revealed that changing the ankle angles and moments was more effective than shifting the COP alone in reducing the knee adduction moment. Although MacLean et al. (2006) made use of custom-made orthoses for purposes other than OA, they discovered significant decreases in ankle angle and moments.

The variance in gait analysis was quite large. Previous research reported similar problems with inconsistent results that made it difficult to reach statistical significance. Kaihana et al. (2005) demonstrated that subjects had inconsistent responses when using lateral wedges, while Maly et al. (2002) doubted that wedges were the primary factor leading to inconsistent results. Schmalz et al. (2006) discussed the contradictory results of Crenshaw et al. (2000), Kerrigan et al. (2002) and Nester et al. (2003), and reported that this discrepancy could be due to the compensatory action of the ankle-foot complex. Maly et al. (2002) explained that the variance could be due to neuromuscular adaptations, while Kuroyanagi et al. (2007) argued about the methodology of the experiment.

Current studies supported the argument of Schmalz et al. (2006). Schmalz hypothesized that different people have different responsiveness to the wedges, which include factors such as the forefoot-rearfoot angle, ankle and subtalar joint flexibility, frontal plane knee alignment and baseline knee adduction moment. The correlation analysis between kinematic variables suggests that there is a higher correlation between consecutive segments, such as COP and forefoot-rearfoot angle or forefoot-rearfoot angle and ankle moments. Despite the correlation between segments, the correlation across segments was relatively low. For example, the first peak knee adduction moment was weakly correlated with the forefoot angles, providing evidence of a compensatory action across subtalar and ankle joints.

In terms of the simulation, the peak knee resultant forces obtained by the simulation model was found during walking (3.97 to 4.23BW) and decreased with the use of all wedged insoles. Muscles contributed substantially to the knee resultant forces, adding forces up to 4.00BW. Others have also concluded that muscles play a significant role in knee joint loading in healthy adults (Winby et al., 2009). The knee implant studies have provided the invaluable data regarding the dynamic loading conditions for the values comparison study (Zhao et al., 2007, Mundermann et al., 2008, Kim et al., 2009). The knee resultant forces ranged from 1.6 to 3.5BW during level walking, although values as high as 4.45BW were reported, using OpenSim software and a similar model(Richards and Higginson, 2010).

The musculoskeletal models used in this study have several limitations. The knee joint only has a single DOF and does not include the patella or ligaments. Although this simplified the model with a representation of the knee and a net knee reaction forces, the frontal plane effect was not addressed.

There are certain limitations associated with current studies, including subject variance, instrumentation error and experimental design. Current studies proposed that this discrepancy resulted from the foot-ankle compensatory action and differences in response mediation (Schmalz et al., 2006). Anthropometric differences, alignment, neuromuscular factors and joint laxity could also cause these deviations. The study could be improved by further dividing the subject population, according to their alignment, such as Q-angle and joint flexibility test.

The motion capture system induces instrumental errors when markers are placed on the anatomical landmarks. The repeatability and variability of the marker placement on different subjects and trials should be further assessed. Moreover, the vibration of markers during walking trials could induce motion artifacts during data acquisition. Suitable filters are needed before data processing. The kinematic calculations from the motion capture data relied on the musculoskeletal model and its corresponding parameters. EMG and cadaveric experiments could help optimize these parameters. EMG data could also provide more information about the role of neuromuscular action in various wedge applications. The calculation of COP also depends on the marker with the associated algorithm available in the package software. Validation, such as control segment or pedobarography, would be useful in confirming the results of COP.

Current results showed a relatively high reduction in the first peak of knee adduction moment. The insole material stiffness was hypothesized to be one of the influencing factors in this reduction. Footwear and corresponding accessories had a large impact on the experimental design, while current research only considers one type of footwear. Toda et al. (2001; 2004; 2006) investigated the use of elastic strapping together with wedges, while different types of footwear and footwear stiffness influenced the gait analysis (Shakoor et al., 2008, Shakoor and Block, 2006, Nigg et al., 2006, Kerrigan et al., 2003, Kerrigan et al., 2005, Kemp et al., 2008, Fisher et al., 2007, Erhart et al., 2008). Future studies could consider differences in footwear and insole material. The current study focused on the use of 5-degree lateral wedges, which were commonly assessed in previous literature, though 8-degree and 16-degree wedges were also commonly applied in clinical practice. A more comprehensive mixed design of these parameters and independent variables could provide more insight on orthotic interventions for osteoarthritis patients.

CHAPTER VI CONCLUSIONS AND FURTHER STUDIES

6.1 Conclusions

Wedged insoles are commonly prescribed for pain-relief for OA patients. Experimental gait analysis provides a platform to systematically understand the effectiveness of wedged insoles on lower-limb alignment and knee loading. In this study, ten healthy female subjects participated in this experiment. Their gait patterns were studied with six different wedged conditions along with a controlled flat insole. They walked with a controlled cadence of 120 per minute assigned with random insoles. Forty-four reflective marker clusters were attached to the pelvis, thigh, shank, forefoot and rearfoot. Motion analysis system and force platforms were used to capture the subject motion and force. The kinetic and kinematic model was developed to investigate the effects of different wedges on the center of pressure, joint angle and joint moment. The result of the ICC test demonstrated that the experimental design was reliable, reliability within a subject was 0.946, and while between subjects was 0.898.

Significances were demonstrated for KADM1 (p=0.039) and KADM2 (p=0.008). The second peak knee adduction moment was greater in magnitude of reduction compared with that of the first peak. All lateral wedge conditions showed a reduction of the first and second peak knee adduction moment. The full-length lateral wedge insole was more efficiency for reduction approximately 27.86%. Comparing with FF conditions, the differences were significant at LF and MF for second peak knee adduction moment. Forefoot wedge insoles were more efficiency for reducing the knee adduction moment. Despite interaction study between different insoles conditions was conducted, the sole factor of LW has significant differences among the interactions.

Current studies demonstrated the effect of the insoles for lateral shift COP was significance ($p=0.047$). The correlation test supported the knee adduction moment reduced the COP shift more laterally. It was suggested that the lateral wedge could increase the eversion at the subtalar joint, a lateral shift in COP, while the knee joint moment arm decreased and eventually the knee adduction moment reduced. In studying ankle joint motion, the wedged insoles could affect the ankle inversion angle, adduction angle and inversion moment.

The correlation studies between different kinematic and kinetic parameters investigated the transfer mechanism of the wedges from the foot segment to the knee segment. Though the wedge insole could not alter the forefoot motion, the forefoot motion correlated to knee extension moment. Reducing the knee adduction moment was compromised by an increase in knee flexion moment. The increase in knee flexion moment could be well accommodated by the patellofemoral dynamics in the knee joint. Shifting the COP could change the rearfoot-forefoot angle and ankle inversion moment, of which the forefoot-rearfoot and ankle inversion moment themselves were also inter-related. The correlation analysis revealed that changing ankle angles and moments were more effective than shifting the COP alone at reducing the knee adduction moment. Regarding to the relationship between the progression of knee OA and the knee moments, interventions for knee OA should therefore be assessed for their effects not only on the knee joint mechanics, but also on the ankle joint and forefoot motion.

.The model simulated the experimental conditions. The knee resultant forces decreased by the wedged insoles caused of the effect of lateral wedge insole on reducing the total muscle forces. The knee-foot-ankle structure changed with the addition of lateral wedged insoles to investigate the support effects on load transfer. The computational model helps us understand the effects of both external and internal influences on the mechanical properties of knee joint loading. The muscle forces played a major role in the knee joint force. This could help giving more advices for the knee OA treatments, such as the muscle training.

6.2 Future Work

Although these experiments demonstrated good repeatability for different trials within a subject, the between-subject variation was remarkable. Recruiting more subjects and categorizing them into groups to take individual differences, such as anthropometry, alignment and joint flexibility, could improve this repeatability. The relationship between the Q-angle, foot-type and resulting knee joint loading should be addressed.

As non-level walking is the most challenging for OA patients, studies on non-level walking, such as stair climbing, should be conducted. The stair-climbing experiment has been set up. A four-step stair with an adjustable riser and tread was mounted on two platforms. The subject's climbing action was captured by a motion analysis system. More information on knee joint loading will be tested out when this stair-climbing test is executed with different wedges in place.

In the future, simulations should be continued to compute the knee joint force and muscle activity for all ten subjects and seven conditions. Additionally, the interventional parameter should be analyzed to obtain individual data. The model data should be analyzed and compared with the experimental results. The effect of muscle activity on the lower extremities would be conducted.

The EMG of the lower extremities should be measured to provide more information about the joint movement and to validate the model. Bipolar surface electrodes (Ag-AgCI) could be placed on the vastus lateralis and medialis, rectus femoris, biceps femoris (long head), tibialis anterior, peroneus longus and gastrocnemuius medialis muscles after removing the hair and cleaning using isopropyl wipes, securing with alcohol stretch tape. A ground electrode would be placed on the tibial tuberosity. The placement of the electrodes is marked to ensure similar placement for all sessions. The EMG signal could be preamplified at the source and the wireless EMG recording system would be better to use. The timing of the heel-strike and toe-off for one step per trial should take from the GRF data. Using these two events, the EMG data can be related to different phases of ground contact during walking.

The data from the experiment and the musculoskeletal models could be used as boundary conditions and material properties in finite element models (Fig. 6-3). Ultimately, the biomechanical consequence of internal structure, such as bone stress and joint inter-facial pressure could be addressed.

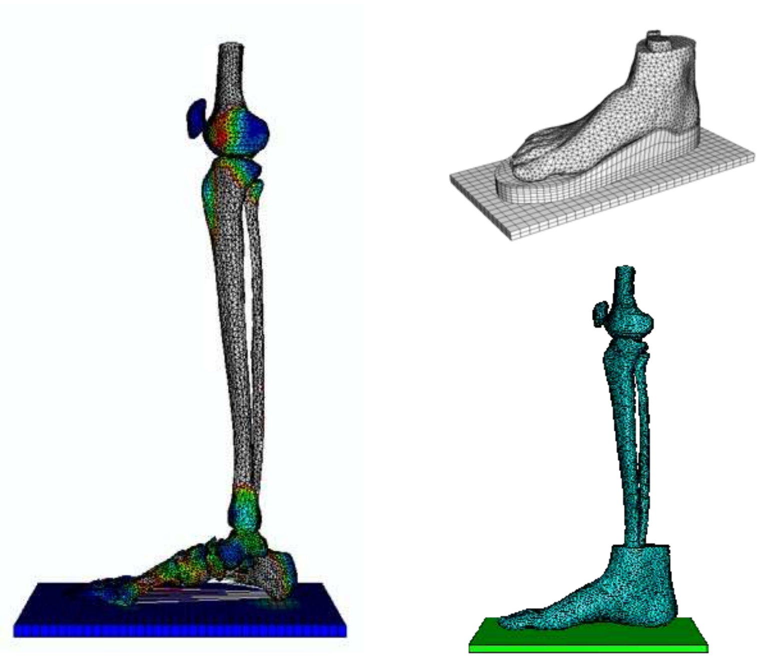

Figure 6-1. FE model of foot-ankle-knee and insoles (Yu 2009)

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Remedy health media, LLC, http://www.podiatrychannel.com

APPENDICES

Appendices-A Information Sheet

INFORMATION SHEET

Effect of Foot Supports on Knee Joint Loading

You are invited to participate on a study conducted by Meng HUANG, who is a post-graduate student of the Health Technology and Informatics Department in The Hong Kong Polytechnic University.

The aim of this study is researching mechanism of both lower limb and foot supports. The study will involve gait analysis while walking. It is hoped that this information will help to understand the knee pain related to lower limb mechanism in order to develop better joint posture and foot support insoles.

The testing should not result in any undue discomfort, but you will need to (anything the subject is required to do, e.g. disrobe, be photographed, videotaped, etc.). All information related to you will remain confidential, and will be identifiable by codes only known to the researcher. You have every right to withdrawn from the study before or during the measurement without penalty of any kind. The whole investigation will take about two-hour.

If you have any complaints about the conduct of this research study, please do not hesitate to contact Mr. Eric Chan, Secretary of the Human Subjects Ethics Sub-Committee of The Hong Kong Polytechnic University in person or in writing (c/o Human Resources Office of the University).If you would like more information about this study, please contact Ms. Meng HUANG on tel. no. 34003208 or Dr. Zhang on tel. no. 27664939.

Thank you for your interest in participating in this study.
Meng HUANG

MPhil Student

Dr. ZHANG Ming
Chief Investigator

Appendices-B Consent Form

THE HONG KONG
POLYTECHNIC UNIVERSITY
香港理工大學

CONSENT FORM

I, _________________________ (name), hereby consent to participate in as a subject for the research entitled "Effect of Foot Supports on Knee Joint Loading". I understand the effect and details of the experimental procedures that have been explained to me.

I understand I have the right to discontinue, with no reasons given, my participation anytime, even during the experiment. I realize that any findings of the study will only be used for research purpose and will be the properties of the Rehabilitation Engineering Centre, The Hong Kong Polytechnic University

Signed____________________
_

Date ____________________

Appendices-C Transportation Allowance

Transportation Allowance

This is to acknowledge that the following persons received from the Department of Health Technology & Informatics the list amount of transportation allowance (or swimming suit) on following date (Participating in the gait analysis experiments as subjects).

No.	Name	HKID No.	Allowance	Date	Signature
1					
2					
3					
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Appendices-D Subject information

Subject Information

<table>
<tr><td colspan="2">Participant:</td><td colspan="3">Contact number:</td></tr>
<tr><td colspan="2">Age:</td><td colspan="3">Height (cm):</td></tr>
<tr><td colspan="2">Weight(kg):</td><td colspan="3"></td></tr>
<tr><td>Brief Medical History:</td><td colspan="4">Sports Injury:
Anthracites:
Foot Pain
Knee Pain
Others:</td></tr>
<tr><td></td><td colspan="2">Left</td><td colspan="2">Right</td></tr>
<tr><td>Leg Length (cm):</td><td colspan="2"></td><td colspan="2"></td></tr>
<tr><td>Ankle Range of Motion (ROM)</td><td colspan="2">Normal/</td><td colspan="2">Normal/</td></tr>
<tr><td>Knee ROM</td><td colspan="2">Normal/</td><td colspan="2">Normal/</td></tr>
<tr><td>Q-angle</td><td colspan="2"></td><td colspan="2"></td></tr>
<tr><td rowspan="2">RSCA</td><td colspan="2">Valgus/Varus</td><td colspan="2">Valgus/Varus</td></tr>
<tr><td>0-2°</td><td></td><td>0-2°</td><td></td></tr>
<tr><td rowspan="2">Inter-condylar/Inter-malleolar gap (cm)</td><td colspan="4"></td></tr>
<tr><td colspan="4">Inter-condylar/Inter-malleolar</td></tr>
</table>

Appendices-E Statistic tables

One-way repeated measurement ANOVA in comparing each mutual condition for variables

Pairwise Comparisons

Measure:MEASURE_1

(I) KABM	(J) KABM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.028	.026	.305	-.031	.087
	3	.030	.017	.107	-.008	.067
	4	.024	.018	.224	-.017	.065
	5	.020	.009	.053	.000	.040
	6	.014	.023	.550	-.038	.066
	7	.040	.019	.066	-.003	.082
2	1	-.028	.026	.305	-.087	.031
	3	.001	.014	.928	-.031	.034
	4	-.005	.014	.752	-.037	.027
	5	-.009	.019	.669	-.052	.035
	6	-.014	.022	.532	-.063	.035
	7	.011	.013	.412	-.018	.041
3	1	-.030	.017	.107	-.067	.008
	2	-.001	.014	.928	-.034	.031
	4	-.006	.009	.522	-.026	.014
	5	-.010	.010	.334	-.032	.012
	6	-.015	.017	.388	-.054	.023
	7	.010	.012	.423	-.017	.036
4	1	-.024	.018	.224	-.065	.017
	2	.005	.014	.752	-.027	.037
	3	.006	.009	.522	-.014	.026
	5	-.004	.011	.728	-.029	.021
	6	-.009	.014	.501	-.040	.021
	7	.016	.008	.066	-.001	.033
5	1	-.020	.009	.053	-.040	.000
	2	.009	.019	.669	-.035	.052
	3	.010	.010	.334	-.012	.032
	4	.004	.011	.728	-.021	.029
	6	-.006	.016	.739	-.042	.031
	7	.020	.012	.128	-.007	.046
6	1	-.014	.023	.550	-.066	.038
	2	.014	.022	.532	-.035	.063
	3	.015	.017	.388	-.023	.054
	4	.009	.014	.501	-.021	.040
	5	.006	.016	.739	-.031	.042
	7	.025	.018	.184	-.014	.065
7	1	-.040	.019	.066	-.082	.003
	2	-.011	.013	.412	-.041	.018
	3	-.010	.012	.423	-.036	.017
	4	-.016	.008	.066	-.033	.001
	5	-.020	.012	.128	-.046	.007
	6	-.025	.018	.184	-.065	.014

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

Pairwise Comparisons

Measure:MEASURE_1

(I) KADM1	(J) KADM1	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.009	.010	.396	-.013	.031
	3	-.040	.020	.071	-.084	.004

	4	.032	.026	.246	-.026	.090
	5	-.016	.019	.427	-.059	.027
	6	.004	.022	.861	-.046	.054
	7	.032	.026	.243	-.026	.090
2	1	-.009	.010	.396	-.031	.013
	3	-.049*	.018	.023	-.089	-.009
	4	.023	.026	.402	-.037	.083
	5	-.025	.020	.255	-.071	.021
	6	-.005	.025	.855	-.061	.052
	7	.023	.026	.398	-.036	.083
3	1	.040	.020	.071	-.004	.084
	2	.049*	.018	.023	.009	.089
	4	.072*	.026	.021	.014	.131
	5	.024*	.007	.008	.008	.040
	6	.044	.024	.094	-.009	.097
	7	.072*	.026	.023	.012	.132
4	1	-.032	.026	.246	-.090	.026
	2	-.023	.026	.402	-.083	.037
	3	-.072*	.026	.021	-.131	-.014
	5	-.048	.023	.065	-.100	.004
	6	-.028*	.007	.003	-.044	-.012
	7	5.551E-17	.018	1.000	-.041	.041
5	1	.016	.019	.427	-.027	.059
	2	.025	.020	.255	-.021	.071
	3	-.024*	.007	.008	-.040	-.008
	4	.048	.023	.065	-.004	.100
	6	.020	.019	.324	-.023	.063
	7	.048	.024	.082	-.007	.103
6	1	-.004	.022	.861	-.054	.046
	2	.005	.025	.855	-.052	.061
	3	-.044	.024	.094	-.097	.009
	4	.028*	.007	.003	.012	.044
	5	-.020	.019	.324	-.063	.023
	7	.028	.019	.165	-.014	.070
7	1	-.032	.026	.243	-.090	.026
	2	-.023	.026	.398	-.083	.036
	3	-.072*	.026	.023	-.132	-.012
	4	-5.551E-17	.018	1.000	-.041	.041
	5	-.048	.024	.082	-.103	.007
	6	-.028	.019	.165	-.070	.014

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

(I) KADM2	(J) KADM2	Mean Difference (I-J)	Std. Error	Sig.a	95% Confidence Interval for Differencea Lower Bound	Upper Bound
1	2	-.032*	.012	.027	-.059	-.004
	3	-.061*	.014	.002	-.093	-.029
	4	.031	.019	.136	-.012	.074
	5	-.003	.015	.847	-.037	.031
	6	.001	.015	.949	-.033	.035
	7	.026	.017	.156	-.012	.064
2	1	.032*	.012	.027	.004	.059
	3	-.029*	.010	.019	-.052	-.006
	4	.063*	.022	.017	.014	.111
	5	.029*	.013	.049	.000	.057
	6	.033	.017	.083	-.005	.070
	7	.058*	.012	.001	.031	.085

3	1	.061*	.014	.002	.029	.093
	2	.029*	.010	.019	.006	.052
	4	.092*	.022	.002	.042	.141
	5	.058*	.011	.001	.032	.083
	6	.062*	.018	.008	.021	.103
	7	.087*	.013	.000	.058	.115
4	1	-.031	.019	.136	-.074	.012
	2	-.063*	.022	.017	-.111	-.014
	3	-.092*	.022	.002	-.141	-.042
	5	-.034	.015	.051	-.068	.000
	6	-.030*	.009	.011	-.051	-.009
	7	-.005	.020	.805	-.050	.040
5	1	.003	.015	.847	-.031	.037
	2	-.029*	.013	.049	-.057	.000
	3	-.058*	.011	.001	-.083	-.032
	4	.034	.015	.051	.000	.068
	6	.004	.014	.774	-.027	.035
	7	.029*	.011	.026	.004	.054
6	1	.000	.015	.949	-.035	.033
	2	-.033	.017	.083	-.070	.005
	3	-.062*	.018	.008	-.103	-.021
	4	.030*	.009	.011	.009	.051
	5	-.004	.014	.774	-.035	.027
	7	.025	.015	.125	-.008	.058
7	1	-.026	.017	.156	-.064	.012
	2	-.058*	.012	.001	-.085	-.031
	3	-.087*	.013	.000	-.115	-.058
	4	.005	.020	.805	-.040	.050
	5	-.029*	.011	.026	-.054	-.004
	6	-.025	.015	.125	-.058	.008

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

Pairwise Comparisons

(I) knf	(J) knf	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.078*	.028	.020	-.141	-.016
	3	-.049*	.014	.006	-.079	-.018
	4	-.053	.029	.096	-.118	.012
	5	-.045	.036	.248	-.127	.037
	6	-.069*	.023	.016	-.122	-.017
	7	-.043	.020	.059	-.089	.002
2	1	.078*	.028	.020	.016	.141
	3	.029	.026	.280	-.029	.088
	4	.025	.015	.133	-.009	.059
	5	.033	.018	.098	-.007	.074
	6	.009	.021	.690	-.039	.057
	7	.035	.033	.324	-.041	.111
3	1	.049*	.014	.006	.018	.079
	2	-.029	.026	.280	-.088	.029
	4	-.005	.025	.858	-.061	.052
	5	.004	.031	.903	-.067	.075
	6	-.021	.021	.342	-.067	.026
	7	.005	.024	.824	-.048	.059
4	1	.053	.029	.096	-.012	.118
	2	-.025	.015	.133	-.059	.009
	3	.005	.025	.858	-.052	.061
	5	.009	.019	.669	-.035	.052
	6	-.016	.018	.401	-.058	.025

	7	.010	.039	.806	-.079	.099
5	1	.045	.036	.248	-.037	.127
	2	-.033	.018	.098	-.074	.007
	3	-.004	.031	.903	-.075	.067
	4	-.009	.019	.669	-.052	.035
	6	-.025	.022	.293	-.075	.025
	7	.001	.041	.972	-.090	.093
6	1	.069*	.023	.016	.017	.122
	2	-.009	.021	.690	-.057	.039
	3	.021	.021	.342	-.026	.067
	4	.016	.018	.401	-.025	.058
	5	.025	.022	.293	-.025	.075
	7	.026	.030	.400	-.041	.093
7	1	.043	.020	.059	-.002	.089
	2	-.035	.033	.324	-.111	.041
	3	-.005	.024	.824	-.059	.048
	4	-.010	.039	.806	-.099	.079
	5	-.001	.041	.972	-.093	.090
	6	-.026	.030	.400	-.093	.041

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

Pairwise Comparisons

Measure:KEM1

(I) KF	(J) KF	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a] Lower Bound	Upper Bound
1	2	.005	.028	.849	-.058	.069
	3	-.079	.085	.379	-.272	.114
	4	-.017	.046	.715	-.122	.087
	5	.120	.125	.362	-.162	.402
	6	-.140*	.048	.017	-.249	-.032
	7	-.075	.075	.342	-.245	.095
2	1	-.005	.028	.849	-.069	.058
	3	-.084	.087	.357	-.281	.112
	4	-.023	.051	.664	-.138	.092
	5	.114	.127	.392	-.173	.401
	6	-.146*	.045	.010	-.247	-.045
	7	-.081	.064	.238	-.226	.064
3	1	.079	.085	.379	-.114	.272
	2	.084	.087	.357	-.112	.281
	4	.061	.066	.375	-.087	.210
	5	.199	.103	.086	-.034	.431
	6	-.061	.063	.352	-.203	.080
	7	.003	.058	.954	-.127	.134
4	1	.017	.046	.715	-.087	.122
	2	.023	.051	.664	-.092	.138
	3	-.061	.066	.375	-.210	.087
	5	.137	.106	.229	-.103	.377
	6	-.123*	.041	.015	-.215	-.030
	7	-.058	.077	.472	-.233	.117
5	1	-.120	.125	.362	-.402	.162
	2	-.114	.127	.392	-.401	.173
	3	-.199	.103	.086	-.431	.034
	4	-.137	.106	.229	-.377	.103
	6	-.260	.124	.065	-.540	.020
	7	-.195	.131	.170	-.491	.101
6	1	.140*	.048	.017	.032	.249
	2	.146*	.045	.010	.045	.247

(I) KEM	(J) KEM	Mean Difference (I-J)	Std. Error	Sig.[a]	Lower Bound	Upper Bound
	3	.061	.063	.352	-.080	.203
	4	.123*	.041	.015	.030	.215
	5	.260	.124	.065	-.020	.540
	7	.065	.052	.246	-.053	.183
7	1	.075	.075	.342	-.095	.245
	2	.081	.064	.238	-.064	.226
	3	-.003	.058	.954	-.134	.127
	4	.058	.077	.472	-.117	.233
	5	.195	.131	.170	-.101	.491
	6	-.065	.052	.246	-.183	.053

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:KEM2

(I) KEM	(J) KEM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.055	.027	.078	-.117	.008
	3	-.050*	.019	.030	-.093	-.006
	4	-.052*	.018	.020	-.094	-.010
	5	-.041	.035	.267	-.119	.037
	6	.004	.050	.932	-.109	.118
	7	.018	.018	.342	-.023	.059
2	1	.055	.027	.078	-.008	.117
	3	.005	.020	.811	-.040	.050
	4	.003	.025	.920	-.053	.058
	5	.013	.041	.748	-.078	.105
	6	.059	.057	.327	-.070	.188
	7	.073	.036	.072	-.008	.153
3	1	.050*	.019	.030	.006	.093
	2	-.005	.020	.811	-.050	.040
	4	-.002	.019	.902	-.045	.040
	5	.009	.029	.772	-.056	.073
	6	.054	.046	.269	-.050	.158
	7	.068*	.024	.022	.012	.123
4	1	.052*	.018	.020	.010	.094
	2	-.003	.025	.920	-.058	.053
	3	.002	.019	.902	-.040	.045
	5	.011	.029	.717	-.055	.077
	6	.056	.046	.250	-.047	.160
	7	.070*	.028	.032	.007	.133
5	1	.041	.035	.267	-.037	.119
	2	-.013	.041	.748	-.105	.078
	3	-.009	.029	.772	-.073	.056
	4	-.011	.029	.717	-.077	.055
	6	.045	.048	.367	-.063	.154
	7	.059	.036	.134	-.022	.141
6	1	-.004	.050	.932	-.118	.109
	2	-.059	.057	.327	-.188	.070
	3	-.054	.046	.269	-.158	.050
	4	-.056	.046	.250	-.160	.047
	5	-.045	.048	.367	-.154	.063
	7	.014	.053	.801	-.107	.134
7	1	-.018	.018	.342	-.059	.023
	2	-.073	.036	.072	-.153	.008
	3	-.068*	.024	.022	-.123	-.012
	4	-.070*	.028	.032	-.133	-.007
	5	-.059	.036	.134	-.141	.022

	6	-.014	.053	.801	-.134	.107

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:KIRM1

(I) KIRM	(J) KIRM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.018	.015	.267	-.016	.051
	3	-.033	.018	.102	-.074	.008
	4	-.008	.011	.469	-.032	.016
	5	.005	.024	.832	-.048	.059
	6	-.018	.021	.421	-.065	.030
	7	-.010	.021	.643	-.058	.038
2	1	-.018	.015	.267	-.051	.016
	3	-.051	.024	.065	-.105	.004
	4	-.026	.014	.097	-.057	.006
	5	-.012	.026	.637	-.070	.045
	6	-.035	.019	.104	-.079	.009
	7	-.028	.020	.189	-.072	.016
3	1	.033	.018	.102	-.008	.074
	2	.051	.024	.065	-.004	.105
	4	.025	.016	.154	-.011	.061
	5	.038	.017	.054	.000	.077
	6	.015	.017	.377	-.022	.053
	7	.023	.015	.168	-.012	.057
4	1	.008	.011	.469	-.016	.032
	2	.026	.014	.097	-.006	.057
	3	-.025	.016	.154	-.061	.011
	5	.013	.018	.483	-.028	.054
	6	-.009	.013	.468	-.038	.019
	7	-.002	.019	.915	-.046	.042
5	1	-.005	.024	.832	-.059	.048
	2	.012	.026	.637	-.045	.070
	3	-.038	.017	.054	-.077	.001
	4	-.013	.018	.483	-.054	.028
	6	-.023*	.010	.041	-.044	-.001
	7	-.015	.017	.380	-.053	.022
6	1	.018	.021	.421	-.030	.065
	2	.035	.019	.104	-.009	.079
	3	-.015	.017	.377	-.053	.022
	4	.009	.013	.468	-.019	.038
	5	.023*	.010	.041	.001	.044
	7	.007	.015	.641	-.027	.042
7	1	.010	.021	.643	-.038	.058
	2	.028	.020	.189	-.016	.072
	3	-.023	.015	.168	-.057	.012
	4	.002	.019	.915	-.042	.046
	5	.015	.017	.380	-.022	.053
	6	-.007	.015	.641	-.042	.027

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:KIRM2

(I) KIRM	(J) KIRM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]

					Lower Bound	Upper Bound
1	2	.001	.011	.895	-.023	.026
	3	.001	.008	.853	-.016	.018
	4	.002	.008	.818	-.015	.019
	5	-.004	.007	.555	-.019	.011
	6	-.003	.012	.808	-.030	.024
	7	.014	.007	.083	-.002	.030
2	1	-.001	.011	.895	-.026	.023
	3	-5.563E-5	.005	.991	-.011	.011
	4	.000	.010	.978	-.023	.024
	5	-.006	.006	.385	-.019	.008
	6	-.004	.007	.564	-.021	.012
	7	.012	.007	.137	-.005	.029
3	1	-.001	.008	.853	-.018	.016
	2	5.563E-5	.005	.991	-.011	.011
	4	.000	.008	.965	-.017	.018
	5	-.006	.005	.284	-.017	.005
	6	-.004	.008	.596	-.023	.014
	7	.012*	.005	.029	.002	.023
4	1	-.002	.008	.818	-.019	.015
	2	.000	.010	.978	-.024	.023
	3	.000	.008	.965	-.018	.017
	5	-.006	.007	.403	-.021	.009
	6	-.005	.010	.628	-.026	.017
	7	.012	.010	.263	-.011	.034
5	1	.004	.007	.555	-.011	.019
	2	.006	.006	.385	-.008	.019
	3	.006	.005	.284	-.005	.017
	4	.006	.007	.403	-.009	.021
	6	.001	.007	.881	-.015	.017
	7	.018*	.006	.021	.003	.032
6	1	.003	.012	.808	-.024	.030
	2	.004	.007	.564	-.012	.021
	3	.004	.008	.596	-.014	.023
	4	.005	.010	.628	-.017	.026
	5	-.001	.007	.881	-.017	.015
	7	.017	.010	.144	-.007	.040
7	1	-.014	.007	.083	-.030	.002
	2	-.012	.007	.137	-.029	.005
	3	-.012*	.005	.029	-.023	-.002
	4	-.012	.010	.263	-.034	.011
	5	-.018*	.006	.021	-.032	-.003
	6	-.017	.010	.144	-.040	.007

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:KERM

(I) KIRM	(J) KIRM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.024	.012	.070	-.050	.002
	3	-.015	.010	.147	-.037	.007
	4	-.018	.011	.143	-.044	.008
	5	-.012	.009	.216	-.032	.008
	6	-.026*	.011	.049	-.052	.000
	7	-.020	.009	.069	-.041	.002
2	1	.024	.012	.070	-.002	.050
	3	.009	.005	.117	-.003	.020

	4	.006	.004	.232	-.004	.015
	5	.012	.007	.124	-.004	.028
	6	-.002	.005	.727	-.014	.010
	7	.004	.006	.491	-.010	.019
3	1	.015	.010	.147	-.007	.037
	2	-.009	.005	.117	-.020	.003
	4	-.003	.007	.654	-.018	.012
	5	.003	.007	.648	-.013	.020
	6	-.010	.005	.067	-.022	.001
	7	-.004	.003	.221	-.011	.003
4	1	.018	.011	.143	-.008	.044
	2	-.006	.004	.232	-.015	.004
	3	.003	.007	.654	-.012	.018
	5	.007	.007	.353	-.009	.022
	6	-.007	.004	.093	-.016	.002
	7	-.001	.008	.895	-.019	.017
5	1	.012	.009	.216	-.008	.032
	2	-.012	.007	.124	-.028	.004
	3	-.003	.007	.648	-.020	.013
	4	-.007	.007	.353	-.022	.009
	6	-.014	.007	.084	-.030	.002
	7	-.008	.009	.421	-.028	.013
6	1	.026*	.011	.049	.000	.052
	2	.002	.005	.727	-.010	.014
	3	.010	.005	.067	.000	.022
	4	.007	.004	.093	-.002	.016
	5	.014	.007	.084	-.002	.030
	7	.006	.006	.339	-.008	.021
7	1	.020	.009	.069	-.002	.041
	2	-.004	.006	.491	-.019	.010
	3	.004	.003	.221	-.003	.011
	4	.001	.008	.895	-.017	.019
	5	.008	.009	.421	-.013	.028
	6	-.006	.006	.339	-.021	.008

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:ADFM

(I) AM	(J) AM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.012	.008	.166	-.029	.006
	3	-.054	.045	.260	-.156	.048
	4	-.036*	.010	.006	-.058	-.013
	5	.053	.077	.512	-.122	.228
	6	-.049*	.015	.010	-.083	-.015
	7	-.064	.051	.238	-.178	.050
2	1	.012	.008	.166	-.006	.029
	3	-.042	.050	.418	-.156	.071
	4	-.024	.011	.062	-.049	.001
	5	.065	.077	.421	-.109	.238
	6	-.037*	.015	.033	-.071	-.004
	7	-.052	.049	.311	-.162	.058
3	1	.054	.045	.260	-.048	.156
	2	.042	.050	.418	-.071	.156
	4	.018	.049	.713	-.091	.128
	5	.107	.083	.229	-.081	.295
	6	.005	.051	.920	-.110	.120

	7	-.010	.075	.899	-.180	.161
4	1	.036*	.010	.006	.013	.058
	2	.024	.011	.062	-.001	.049
	3	-.018	.049	.713	-.128	.091
	5	.089	.077	.277	-.085	.262
	6	-.013	.011	.262	-.038	.012
	7	-.028	.049	.577	-.139	.082
5	1	-.053	.077	.512	-.228	.122
	2	-.065	.077	.421	-.238	.109
	3	-.107	.083	.229	-.295	.081
	4	-.089	.077	.277	-.262	.085
	6	-.102	.068	.170	-.256	.052
	7	-.117	.082	.187	-.302	.068
6	1	.049*	.015	.010	.015	.083
	2	.037*	.015	.033	.004	.071
	3	-.005	.051	.920	-.120	.110
	4	.013	.011	.262	-.012	.038
	5	.102	.068	.170	-.052	.256
	7	-.015	.043	.734	-.112	.082
7	1	.064	.051	.238	-.050	.178
	2	.052	.049	.311	-.058	.162
	3	.010	.075	.899	-.161	.180
	4	.028	.049	.577	-.082	.139
	5	.117	.082	.187	-.068	.302
	6	.015	.043	.734	-.082	.112

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:APFM

(I) AM	(J) AM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.012	.008	.166	-.029	.006
	3	-.054	.045	.260	-.156	.048
	4	-.036*	.010	.006	-.058	-.013
	5	.053	.077	.512	-.122	.228
	6	-.049*	.015	.010	-.083	-.015
	7	-.064	.051	.238	-.178	.050
2	1	.012	.008	.166	-.006	.029
	3	-.042	.050	.418	-.156	.071
	4	-.024	.011	.062	-.049	.001
	5	.065	.077	.421	-.109	.238
	6	-.037*	.015	.033	-.071	-.004
	7	-.052	.049	.311	-.162	.058
3	1	.054	.045	.260	-.048	.156
	2	.042	.050	.418	-.071	.156
	4	.018	.049	.713	-.091	.128
	5	.107	.083	.229	-.081	.295
	6	.005	.051	.920	-.110	.120
	7	-.010	.075	.899	-.180	.161
4	1	.036*	.010	.006	.013	.058
	2	.024	.011	.062	-.001	.049
	3	-.018	.049	.713	-.128	.091
	5	.089	.077	.277	-.085	.262
	6	-.013	.011	.262	-.038	.012
	7	-.028	.049	.577	-.139	.082
5	1	-.053	.077	.512	-.228	.122
	2	-.065	.077	.421	-.238	.109
	3	-.107	.083	.229	-.295	.081

	4	-.089	.077	.277	-.262	.085
	6	-.102	.068	.170	-.256	.052
	7	-.117	.082	.187	-.302	.068
6	1	.049*	.015	.010	.015	.083
	2	.037*	.015	.033	.004	.071
	3	-.005	.051	.920	-.120	.110
	4	.013	.011	.262	-.012	.038
	5	.102	.068	.170	-.052	.256
	7	-.015	.043	.734	-.112	.082
7	1	.064	.051	.238	-.050	.178
	2	.052	.049	.311	-.058	.162
	3	.010	.075	.899	-.161	.180
	4	.028	.049	.577	-.082	.139
	5	.117	.082	.187	-.068	.302
	6	.015	.043	.734	-.082	.112

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:AIVM

(I) AM	(J) AM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a] Lower Bound	Upper Bound
1	2	.001	.010	.885	-.020	.023
	3	.016	.029	.580	-.048	.081
	4	-.028	.023	.241	-.079	.023
	5	.001	.020	.977	-.045	.046
	6	-.061*	.014	.002	-.093	-.030
	7	-.024	.023	.323	-.076	.028
2	1	-.001	.010	.885	-.023	.020
	3	.015	.028	.604	-.048	.078
	4	-.030	.020	.172	-.075	.016
	5	.000	.019	.967	-.044	.042
	6	-.063*	.016	.003	-.098	-.028
	7	-.025	.025	.326	-.081	.030
3	1	-.016	.029	.580	-.081	.048
	2	-.015	.028	.604	-.078	.048
	4	-.045*	.015	.016	-.079	-.011
	5	-.016	.016	.336	-.051	.019
	6	-.078*	.025	.012	-.134	-.022
	7	-.040	.026	.154	-.099	.018
4	1	.028	.023	.241	-.023	.079
	2	.030	.020	.172	-.016	.075
	3	.045*	.015	.016	.011	.079
	5	.029*	.008	.007	.010	.048
	6	-.033	.020	.127	-.078	.011
	7	.004	.019	.830	-.039	.048
5	1	.000	.020	.977	-.046	.045
	2	.001	.019	.967	-.042	.044
	3	.016	.016	.336	-.019	.051
	4	-.029*	.008	.007	-.048	-.010
	6	-.062*	.015	.003	-.097	-.027
	7	-.025	.020	.247	-.070	.020
6	1	.061*	.014	.002	.030	.093
	2	.063*	.016	.003	.028	.098
	3	.078*	.025	.012	.022	.134
	4	.033	.020	.127	-.011	.078

(I) AM	(J) AM	Mean Difference (I-J)	Std. Error	Sig.[a]	Lower Bound	Upper Bound
	5	.062*	.015	.003	.027	.097
	7	.037	.025	.174	-.020	.094
7	1	.024	.023	.323	-.028	.076
	2	.025	.025	.326	-.030	.081
	3	.040	.026	.154	-.018	.099
	4	-.004	.019	.830	-.048	.039
	5	.025	.020	.247	-.020	.070
	6	-.037	.025	.174	-.094	.020

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:AEVM

(I) AM	(J) AM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.010	.008	.233	-.028	.008
	3	-.011	.016	.524	-.048	.026
	4	-.010	.021	.662	-.058	.039
	5	-.002	.017	.904	-.041	.036
	6	-.036*	.015	.039	-.071	-.002
	7	-.023	.021	.307	-.071	.025
2	1	.010	.008	.233	-.008	.028
	3	.000	.017	.974	-.039	.037
	4	.001	.016	.971	-.036	.037
	5	.008	.012	.497	-.018	.034
	6	-.026*	.010	.026	-.048	-.004
	7	-.013	.019	.532	-.057	.031
3	1	.011	.016	.524	-.026	.048
	2	.001	.017	.974	-.037	.039
	4	.001	.023	.960	-.051	.053
	5	.009	.018	.641	-.032	.050
	6	-.026	.015	.128	-.060	.009
	7	-.012	.019	.549	-.056	.032
4	1	.010	.021	.662	-.039	.058
	2	.000	.016	.971	-.037	.036
	3	-.001	.023	.960	-.053	.051
	5	.008	.011	.527	-.018	.034
	6	-.027	.014	.089	-.058	.005
	7	-.013	.022	.566	-.064	.037
5	1	.002	.017	.904	-.036	.041
	2	-.008	.012	.497	-.034	.018
	3	-.009	.018	.641	-.050	.032
	4	-.008	.011	.527	-.034	.018
	6	-.034*	.011	.010	-.058	-.010
	7	-.021	.020	.332	-.067	.025
6	1	.036*	.015	.039	.002	.071
	2	.026*	.010	.026	.004	.048
	3	.026	.015	.128	-.009	.060
	4	.027	.014	.089	-.005	.058
	5	.034*	.011	.010	.010	.058
	7	.013	.015	.391	-.020	.047
7	1	.023	.021	.307	-.025	.071
	2	.013	.019	.532	-.031	.057
	3	.012	.019	.549	-.032	.056
	4	.013	.022	.566	-.037	.064
	5	.021	.020	.332	-.025	.067
	6	-.013	.015	.391	-.047	.020

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:AADM

(I) AM	(J) AM	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.000	.006	.909	-.014	.012
	3	-.015	.015	.348	-.049	.019
	4	-.005	.006	.492	-.019	.010
	5	.009	.011	.438	-.016	.034
	6	-.015*	.006	.033	-.028	-.001
	7	.001	.006	.862	-.012	.014
2	1	.001	.006	.909	-.012	.014
	3	-.014	.015	.359	-.047	.019
	4	-.004	.003	.256	-.011	.003
	5	.010	.011	.383	-.014	.034
	6	-.014*	.003	.002	-.022	-.007
	7	.002	.003	.581	-.005	.009
3	1	.015	.015	.348	-.019	.049
	2	.014	.015	.359	-.019	.047
	4	.010	.014	.492	-.022	.043
	5	.024	.016	.170	-.012	.060
	6	-3.800E-5	.016	.998	-.036	.036
	7	.016	.015	.301	-.017	.049
4	1	.005	.006	.492	-.010	.019
	2	.004	.003	.256	-.003	.011
	3	-.010	.014	.492	-.043	.022
	5	.014	.012	.280	-.013	.040
	6	-.010*	.004	.032	-.020	-.001
	7	.006	.005	.290	-.006	.017
5	1	-.009	.011	.438	-.034	.016
	2	-.010	.011	.383	-.034	.014
	3	-.024	.016	.170	-.060	.012
	4	-.014	.012	.280	-.040	.013
	6	-.024	.011	.063	-.049	.002
	7	-.008	.008	.370	-.027	.011
6	1	.015*	.006	.033	.001	.028
	2	.014*	.003	.002	.007	.022
	3	3.800E-5	.016	.998	-.036	.036
	4	.010*	.004	.032	.001	.020
	5	.024	.011	.063	-.002	.049
	7	.016*	.006	.018	.004	.028
7	1	-.001	.006	.862	-.014	.012
	2	-.002	.003	.581	-.009	.005
	3	-.016	.015	.301	-.049	.017
	4	-.006	.005	.290	-.017	.006
	5	.008	.008	.370	-.011	.027
	6	-.016*	.006	.018	-.028	-.004

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:FPFA

(I) FA	(J) FA	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-1.025	2.567	.699	-6.832	4.782

	3	-1.653	1.738	.367	-5.585	2.280
	4	.770	3.499	.831	-7.145	8.685
	5	3.790	4.255	.396	-5.835	13.414
	6	2.216	2.966	.474	-4.494	8.926
	7	.023	2.843	.994	-6.407	6.454
2	1	1.025	2.567	.699	-4.782	6.832
	3	-.628	2.139	.776	-5.467	4.212
	4	1.795	1.589	.288	-1.799	5.389
	5	4.815	2.930	.135	-1.814	11.443
	6	3.241	1.475	.056	-.094	6.577
	7	1.048	1.346	.456	-1.997	4.093
3	1	1.653	1.738	.367	-2.280	5.585
	2	.628	2.139	.776	-4.212	5.467
	4	2.423	3.170	.464	-4.749	9.594
	5	5.442	3.452	.149	-2.367	13.251
	6	3.869	2.164	.107	-1.026	8.763
	7	1.676	2.153	.456	-3.196	6.547
4	1	-.770	3.499	.831	-8.685	7.145
	2	-1.795	1.589	.288	-5.389	1.799
	3	-2.423	3.170	.464	-9.594	4.749
	5	3.020	2.283	.218	-2.144	8.183
	6	1.446	1.385	.324	-1.688	4.580
	7	-.747	1.535	.638	-4.219	2.725
5	1	-3.790	4.255	.396	-13.414	5.835
	2	-4.815	2.930	.135	-11.443	1.814
	3	-5.442	3.452	.149	-13.251	2.367
	4	-3.020	2.283	.218	-8.183	2.144
	6	-1.573	2.008	.453	-6.116	2.969
	7	-3.767	1.865	.074	-7.984	.451
6	1	-2.216	2.966	.474	-8.926	4.494
	2	-3.241	1.475	.056	-6.577	.094
	3	-3.869	2.164	.107	-8.763	1.026
	4	-1.446	1.385	.324	-4.580	1.688
	5	1.573	2.008	.453	-2.969	6.116
	7	-2.193*	.912	.040	-4.255	-.131
7	1	-.023	2.843	.994	-6.454	6.407
	2	-1.048	1.346	.456	-4.093	1.997
	3	-1.676	2.153	.456	-6.547	3.196
	4	.747	1.535	.638	-2.725	4.219
	5	3.767	1.865	.074	-.451	7.984
	6	2.193*	.912	.040	.131	4.255

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:FDFA

(I) FDFA	(J) FDFA	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a] Lower Bound	Upper Bound
1	2	1.658	2.373	.503	-3.711	7.026
	3	.347	2.453	.891	-5.201	5.896
	4	3.260	3.374	.359	-4.373	10.894
	5	1.110	2.756	.696	-5.124	7.344
	6	1.175	1.717	.511	-2.709	5.059
	7	-4.087	2.739	.170	-10.283	2.109
2	1	-1.658	2.373	.503	-7.026	3.711
	3	-1.310	2.932	.666	-7.943	5.323
	4	1.603	3.433	.652	-6.163	9.369
	5	-.547	2.006	.791	-5.084	3.990

	6	-.483	2.086	.822	-5.200	4.235
	7	-5.744	2.544	.050	-11.500	.011
3	1	-.347	2.453	.891	-5.896	5.201
	2	1.310	2.932	.666	-5.323	7.943
	4	2.913	2.704	.309	-3.204	9.031
	5	.763	2.252	.743	-4.331	5.857
	6	.827	2.880	.780	-5.687	7.342
	7	-4.434	3.836	.277	-13.111	4.243
4	1	-3.260	3.374	.359	-10.894	4.373
	2	-1.603	3.433	.652	-9.369	6.163
	3	-2.913	2.704	.309	-9.031	3.204
	5	-2.150	1.768	.255	-6.149	1.848
	6	-2.086	3.387	.553	-9.748	5.577
	7	-7.347	4.100	.107	-16.622	1.927
5	1	-1.110	2.756	.696	-7.344	5.124
	2	.547	2.006	.791	-3.990	5.084
	3	-.763	2.252	.743	-5.857	4.331
	4	2.150	1.768	.255	-1.848	6.149
	6	.065	2.548	.980	-5.698	5.828
	7	-5.197	3.484	.170	-13.079	2.685
6	1	-1.175	1.717	.511	-5.059	2.709
	2	.483	2.086	.822	-4.235	5.200
	3	-.827	2.880	.780	-7.342	5.687
	4	2.086	3.387	.553	-5.577	9.748
	5	-.065	2.548	.980	-5.828	5.698
	7	-5.262	3.111	.125	-12.300	1.776
7	1	4.087	2.739	.170	-2.109	10.283
	2	5.744	2.544	.050	-.011	11.500
	3	4.434	3.836	.277	-4.243	13.111
	4	7.347	4.100	.107	-1.927	16.622
	5	5.197	3.484	.170	-2.685	13.079
	6	5.262	3.111	.125	-1.776	12.300

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

Pairwise Comparisons

Measure:FIVA

(I) factor1	(J) factor1	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	4.415	2.239	.080	-.650	9.480
	3	4.855*	1.909	.032	.537	9.173
	4	2.672	1.609	.131	-.969	6.312
	5	2.205	2.475	.396	-3.394	7.804
	6	6.014*	2.242	.025	.943	11.085
	7	4.277	1.993	.060	-.231	8.786
2	1	-4.415	2.239	.080	-9.480	.650
	3	.441	1.996	.830	-4.076	4.957
	4	-1.743	1.796	.357	-5.805	2.319
	5	-2.210	2.350	.372	-7.526	3.107
	6	1.599	1.951	.434	-2.815	6.013
	7	-.137	1.210	.912	-2.875	2.600
3	1	-4.855*	1.909	.032	-9.173	-.537
	2	-.441	1.996	.830	-4.957	4.076
	4	-2.184	1.358	.142	-5.256	.888
	5	-2.650	2.264	.272	-7.772	2.471
	6	1.159	1.136	.335	-1.412	3.729
	7	-.578	1.230	.650	-3.361	2.205

4	1	-2.672	1.609	.131	-6.312	.969
	2	1.743	1.796	.357	-2.319	5.805
	3	2.184	1.358	.142	-.888	5.256
	5	-.466	2.735	.868	-6.653	5.720
	6	3.342	1.608	.067	-.295	6.980
	7	1.606	1.343	.262	-1.432	4.643
5	1	-2.205	2.475	.396	-7.804	3.394
	2	2.210	2.350	.372	-3.107	7.526
	3	2.650	2.264	.272	-2.471	7.772
	4	.466	2.735	.868	-5.720	6.653
	6	3.809	1.985	.087	-.681	8.299
	7	2.072	2.172	.365	-2.841	6.985
6	1	-6.014*	2.242	.025	-11.085	-.943
	2	-1.599	1.951	.434	-6.013	2.815
	3	-1.159	1.136	.335	-3.729	1.412
	4	-3.342	1.608	.067	-6.980	.295
	5	-3.809	1.985	.087	-8.299	.681
	7	-1.737	1.257	.200	-4.580	1.107
7	1	-4.277	1.993	.060	-8.786	.231
	2	.137	1.210	.912	-2.600	2.875
	3	.578	1.230	.650	-2.205	3.361
	4	-1.606	1.343	.262	-4.643	1.432
	5	-2.072	2.172	.365	-6.985	2.841
	6	1.737	1.257	.200	-1.107	4.580

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:FEVA

(I) factor1	(J) factor1	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	2.144	1.170	.100	-.504	4.791
	3	2.899	1.388	.066	-.241	6.039
	4	.456	1.635	.787	-3.243	4.154
	5	-.715	1.700	.684	-4.561	3.131
	6	2.991*	1.266	.042	.127	5.855
	7	.071	.811	.932	-1.764	1.907
2	1	-2.144	1.170	.100	-4.791	.504
	3	.755	1.471	.620	-2.574	4.083
	4	-1.688	1.924	.403	-6.041	2.665
	5	-2.859	1.971	.181	-7.317	1.599
	6	.848	1.541	.596	-2.637	4.332
	7	-2.073	1.143	.103	-4.659	.514
3	1	-2.899	1.388	.066	-6.039	.241
	2	-.755	1.471	.620	-4.083	2.574
	4	-2.443	1.472	.131	-5.772	.886
	5	-3.614*	1.337	.024	-6.639	-.588
	6	.093	1.559	.954	-3.435	3.621
	7	-2.827	1.373	.070	-5.934	.279
4	1	-.456	1.635	.787	-4.154	3.243
	2	1.688	1.924	.403	-2.665	6.041
	3	2.443	1.472	.131	-.886	5.772
	5	-1.171	2.048	.582	-5.804	3.463
	6	2.536	1.517	.129	-.895	5.967
	7	-.384	1.827	.838	-4.517	3.749
5	1	.715	1.700	.684	-3.131	4.561
	2	2.859	1.971	.181	-1.599	7.317
	3	3.614*	1.337	.024	.588	6.639

	4	1.171	2.048	.582	-3.463	5.804
	6	3.706	1.783	.067	-.328	7.741
	7	.786	1.813	.675	-3.314	4.887
6	1	-2.991*	1.266	.042	-5.855	-.127
	2	-.848	1.541	.596	-4.332	2.637
	3	-.093	1.559	.954	-3.621	3.435
	4	-2.536	1.517	.129	-5.967	.895
	5	-3.706	1.783	.067	-7.741	.328
	7	-2.920	1.528	.088	-6.377	.537
7	1	-.071	.811	.932	-1.907	1.764
	2	2.073	1.143	.103	-.514	4.659
	3	2.827	1.373	.070	-.279	5.934
	4	.384	1.827	.838	-3.749	4.517
	5	-.786	1.813	.675	-4.887	3.314
	6	2.920	1.528	.088	-.537	6.377

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:FADA

(I) factor1	(J) factor1	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	2.952	2.053	.184	-1.693	7.596
	3	6.205*	2.030	.014	1.613	10.797
	4	1.692	1.954	.409	-2.727	6.112
	5	.948	2.130	.667	-3.871	5.767
	6	3.070	2.316	.218	-2.170	8.310
	7	3.121	3.003	.326	-3.671	9.914
2	1	-2.952	2.053	.184	-7.596	1.693
	3	3.253*	.927	.007	1.155	5.350
	4	-1.259	1.432	.402	-4.498	1.979
	5	-2.004	2.346	.415	-7.312	3.304
	6	.118	2.298	.960	-5.081	5.317
	7	.170	1.988	.934	-4.328	4.667
3	1	-6.205*	2.030	.014	-10.797	-1.613
	2	-3.253*	.927	.007	-5.350	-1.155
	4	-4.512*	1.697	.026	-8.350	-.674
	5	-5.257*	2.251	.044	-10.349	-.164
	6	-3.135	1.688	.096	-6.952	.683
	7	-3.083	1.561	.080	-6.614	.447
4	1	-1.692	1.954	.409	-6.112	2.727
	2	1.259	1.432	.402	-1.979	4.498
	3	4.512*	1.697	.026	.674	8.350
	5	-.745	1.634	.659	-4.442	2.952
	6	1.378	2.506	.596	-4.292	7.047
	7	1.429	2.843	.627	-5.003	7.861
5	1	-.948	2.130	.667	-5.767	3.871
	2	2.004	2.346	.415	-3.304	7.312
	3	5.257*	2.251	.044	.164	10.349
	4	.745	1.634	.659	-2.952	4.442
	6	2.122	2.664	.446	-3.904	8.148
	7	2.174	2.949	.480	-4.497	8.845
6	1	-3.070	2.316	.218	-8.310	2.170
	2	-.118	2.298	.960	-5.317	5.081
	3	3.135	1.688	.096	-.683	6.952
	4	-1.378	2.506	.596	-7.047	4.292
	5	-2.122	2.664	.446	-8.148	3.904
	7	.051	2.464	.984	-5.523	5.626

7	1	-3.121	3.003	.326	-9.914	3.671
	2	-.170	1.988	.934	-4.667	4.328
	3	3.083	1.561	.080	-.447	6.614
	4	-1.429	2.843	.627	-7.861	5.003
	5	-2.174	2.949	.480	-8.845	4.497
	6	-.051	2.464	.984	-5.626	5.523

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:FABA

(I) factor1	(J) factor1	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.982	1.725	.583	-2.920	4.884
	3	-1.728	1.603	.309	-5.354	1.899
	4	.039	2.034	.985	-4.562	4.640
	5	-1.858	2.547	.484	-7.620	3.903
	6	-1.986	1.828	.305	-6.121	2.149
	7	.679	2.340	.778	-4.614	5.972
2	1	-.982	1.725	.583	-4.884	2.920
	3	-2.709	1.450	.095	-5.990	.572
	4	-.942	1.833	.620	-5.089	3.204
	5	-2.840	2.536	.292	-8.577	2.897
	6	-2.968	2.402	.248	-8.401	2.466
	7	-.303	2.664	.912	-6.329	5.724
3	1	1.728	1.603	.309	-1.899	5.354
	2	2.709	1.450	.095	-.572	5.990
	4	1.767	1.468	.259	-1.553	5.087
	5	-.131	1.822	.944	-4.252	3.990
	6	-.259	1.558	.872	-3.783	3.266
	7	2.406	2.895	.427	-4.143	8.956
4	1	-.039	2.034	.985	-4.640	4.562
	2	.942	1.833	.620	-3.204	5.089
	3	-1.767	1.468	.259	-5.087	1.553
	5	-1.898	1.517	.242	-5.328	1.533
	6	-2.025	1.886	.311	-6.291	2.241
	7	.640	2.543	.807	-5.113	6.393
5	1	1.858	2.547	.484	-3.903	7.620
	2	2.840	2.536	.292	-2.897	8.577
	3	.131	1.822	.944	-3.990	4.252
	4	1.898	1.517	.242	-1.533	5.328
	6	-.128	2.651	.963	-6.124	5.869
	7	2.537	3.714	.512	-5.865	10.940
6	1	1.986	1.828	.305	-2.149	6.121
	2	2.968	2.402	.248	-2.466	8.401
	3	.259	1.558	.872	-3.266	3.783
	4	2.025	1.886	.311	-2.241	6.291
	5	.128	2.651	.963	-5.869	6.124
	7	2.665	2.398	.295	-2.759	8.089
7	1	-.679	2.340	.778	-5.972	4.614
	2	.303	2.664	.912	-5.724	6.329
	3	-2.406	2.895	.427	-8.956	4.143
	4	-.640	2.543	.807	-6.393	5.113
	5	-2.537	3.714	.512	-10.940	5.865
	6	-2.665	2.398	.295	-8.089	2.759

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

Pairwise Comparisons

Measure: COPL

(I) factor1	(J) factor1	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.053	.025	.060	-.003	.109
	3	.043	.021	.068	-.004	.089
	4	-.001	.015	.948	-.035	.033
	5	.006	.011	.592	-.019	.032
	6	-.021	.024	.406	-.074	.033
	7	.042	.027	.146	-.018	.102
2	1	-.053	.025	.060	-.109	.003
	3	-.011	.012	.402	-.038	.017
	4	-.054*	.016	.009	-.091	-.017
	5	-.047	.023	.067	-.098	.004
	6	-.074*	.026	.018	-.132	-.016
	7	-.011	.029	.716	-.077	.055
3	1	-.043	.021	.068	-.089	.004
	2	.011	.012	.402	-.017	.038
	4	-.044*	.014	.014	-.076	-.011
	5	-.036	.023	.145	-.088	.015
	6	-.063*	.026	.035	-.121	-.005
	7	.000	.023	.988	-.052	.051
4	1	.001	.015	.948	-.033	.035
	2	.054*	.016	.009	.017	.091
	3	.044*	.014	.014	.011	.076
	5	.007	.014	.622	-.025	.040
	6	-.020	.019	.323	-.062	.023
	7	.043*	.018	.040	.002	.084
5	1	-.006	.011	.592	-.032	.019
	2	.047	.023	.067	-.004	.098
	3	.036	.023	.145	-.015	.088
	4	-.007	.014	.622	-.040	.025
	6	-.027	.020	.207	-.072	.018
	7	.036	.030	.255	-.031	.103
6	1	.021	.024	.406	-.033	.074
	2	.074*	.026	.018	.016	.132
	3	.063*	.026	.035	.005	.121
	4	.020	.019	.323	-.023	.062
	5	.027	.020	.207	-.018	.072
	7	.063*	.024	.028	.008	.117
7	1	-.042	.027	.146	-.102	.018
	2	.011	.029	.716	-.055	.077
	3	.000	.023	.988	-.051	.052
	4	-.043*	.018	.040	-.084	-.002
	5	-.036	.030	.255	-.103	.031
	6	-.063*	.024	.028	-.117	-.008

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:COPM

(I) factor1	(J) factor1	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.091	.060	.166	-.045	.227
	3	.082*	.033	.035	.007	.156

(I)	(J)	Mean Difference (I-J)	Std. Error	Sig.[a]	Lower Bound	Upper Bound
	4	.006	.048	.899	-.103	.116
	5	.013	.024	.585	-.040	.067
	6	.021	.050	.688	-.092	.133
	7	.096*	.039	.038	.007	.185
2	1	-.091	.060	.166	-.227	.045
	3	-.009	.040	.826	-.099	.081
	4	-.084	.046	.099	-.188	.019
	5	-.077	.043	.107	-.175	.020
	6	-.070	.061	.283	-.209	.069
	7	.005	.059	.931	-.129	.139
3	1	-.082*	.033	.035	-.156	-.007
	2	.009	.040	.826	-.081	.099
	4	-.075	.049	.159	-.186	.036
	5	-.068*	.027	.034	-.130	-.006
	6	-.061	.054	.287	-.183	.061
	7	.014	.042	.740	-.081	.109
4	1	-.006	.048	.899	-.116	.103
	2	.084	.046	.099	-.019	.188
	3	.075	.049	.159	-.036	.186
	5	.007	.038	.854	-.078	.093
	6	.014	.037	.705	-.069	.097
	7	.090	.040	.051	.000	.180
5	1	-.013	.024	.585	-.067	.040
	2	.077	.043	.107	-.020	.175
	3	.068*	.027	.034	.006	.130
	4	-.007	.038	.854	-.093	.078
	6	.007	.040	.863	-.084	.098
	7	.082	.038	.056	-.002	.167
6	1	-.021	.050	.688	-.133	.092
	2	.070	.061	.283	-.069	.209
	3	.061	.054	.287	-.061	.183
	4	-.014	.037	.705	-.097	.069
	5	-.007	.040	.863	-.098	.084
	7	.075*	.026	.018	.016	.134
7	1	-.096*	.039	.038	-.185	-.007
	2	-.005	.059	.931	-.139	.129
	3	-.014	.042	.740	-.109	.081
	4	-.090	.040	.051	-.180	.000
	5	-.082	.038	.056	-.167	.002
	6	-.075*	.026	.018	-.134	-.016

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

*. The mean difference is significant at the .05 level.

Pairwise Comparisons

Measure:COPA

(I) COPAP	(J) COPAP	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.001	.122	.991	-.274	.277
	3	.180	.135	.213	-.124	.485
	4	.110	.089	.250	-.092	.312
	5	-.102	.063	.141	-.244	.041
	6	-.048	.081	.570	-.231	.136
	7	-.106	.104	.334	-.342	.129
2	1	-.001	.122	.991	-.277	.274
	3	.179	.175	.333	-.217	.575
	4	.108	.113	.363	-.148	.364
	5	-.103	.103	.341	-.336	.129

	6	-.049	.102	.640	-.279	.181
	7	-.108	.121	.397	-.382	.167
3	1	-.180	.135	.213	-.485	.124
	2	-.179	.175	.333	-.575	.217
	4	-.071	.070	.340	-.229	.088
	5	-.282	.163	.118	-.652	.087
	6	-.228	.162	.194	-.596	.139
	7	-.287	.180	.146	-.695	.121
4	1	-.110	.089	.250	-.312	.092
	2	-.108	.113	.363	-.364	.148
	3	.071	.070	.340	-.088	.229
	5	-.212	.110	.086	-.460	.037
	6	-.158	.111	.190	-.409	.094
	7	-.216	.134	.141	-.519	.086
5	1	.102	.063	.141	-.041	.244
	2	.103	.103	.341	-.129	.336
	3	.282	.163	.118	-.087	.652
	4	.212	.110	.086	-.037	.460
	6	.054	.037	.179	-.030	.138
	7	-.005	.056	.938	-.131	.122
6	1	.048	.081	.570	-.136	.231
	2	.049	.102	.640	-.181	.279
	3	.228	.162	.194	-.139	.596
	4	.158	.111	.190	-.094	.409
	5	-.054	.037	.179	-.138	.030
	7	-.059	.035	.129	-.138	.021
7	1	.106	.104	.334	-.129	.342
	2	.108	.121	.397	-.167	.382
	3	.287	.180	.146	-.121	.695
	4	.216	.134	.141	-.086	.519
	5	.005	.056	.938	-.122	.131
	6	.059	.035	.129	-.021	.138

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).

Pairwise Comparisons

Measure:COPP

(I) COPAP	(J) COPAP	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.019	.056	.737	-.146	.107
	3	-.125	.089	.194	-.327	.077
	4	-.096	.090	.312	-.300	.107
	5	.042	.058	.485	-.089	.173
	6	-.071	.055	.228	-.195	.053
	7	-.055	.066	.429	-.203	.094
2	1	.019	.056	.737	-.107	.146
	3	-.106	.116	.384	-.368	.156
	4	-.077	.134	.578	-.379	.225
	5	.061	.082	.473	-.124	.247
	6	-.051	.086	.566	-.247	.144
	7	-.035	.077	.659	-.209	.139
3	1	.125	.089	.194	-.077	.327
	2	.106	.116	.384	-.156	.368
	4	.029	.076	.714	-.144	.202
	5	.167	.106	.150	-.073	.408
	6	.055	.082	.525	-.132	.241
	7	.071	.103	.509	-.162	.304
4	1	.096	.090	.312	-.107	.300
	2	.077	.134	.578	-.225	.379

	3	-.029	.076	.714	-.202	.144
	5	.139	.089	.155	-.063	.340
	6	.026	.081	.759	-.157	.209
	7	.042	.118	.730	-.225	.308
5	1	-.042	.058	.485	-.173	.089
	2	-.061	.082	.473	-.247	.124
	3	-.167	.106	.150	-.408	.073
	4	-.139	.089	.155	-.340	.063
	6	-.113	.059	.090	-.247	.021
	7	-.097	.074	.225	-.264	.071
6	1	.071	.055	.228	-.053	.195
	2	.051	.086	.566	-.144	.247
	3	-.055	.082	.525	-.241	.132
	4	-.026	.081	.759	-.209	.157
	5	.113	.059	.090	-.021	.247
	7	.016	.040	.693	-.074	.106
7	1	.055	.066	.429	-.094	.203
	2	.035	.077	.659	-.139	.209
	3	-.071	.103	.509	-.304	.162
	4	-.042	.118	.730	-.308	.225
	5	.097	.074	.225	-.071	.264
	6	-.016	.040	.693	-.106	.074

Based on estimated marginal means

a. Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).