DEVELOPMENT OF A MUSCULOSKELETAL MODEL

TO DETERMINE KNEE CONTACT FORCE

DURING WALKING ON BALLAST

USING OPENSIM SIMULATION

by

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ABSTRACT

Railroad workers experience a unique exposure to walking on ballast and uneven ground walking is a possible risk factor for knee osteoarthritis. However, the effect of ballast on workers is still not clear, especially for mechanical joint loads. Published research on walking on ballast principally examines temporal gait parameters and joint kinematics. The aim of this research is to investigate the change of knee contact force (KCF) during walking on ballast as surface condition, surface configuration, and uphill or downhill limbs by using an new OpenSim model.

There are two significant contributions of this research. First, a new OpenSim gait model with robust knee structures was developed, which included patella structures, a six degrees of freedom knee joint, and four main knee ligaments. Second, KCF was investigated when walking on ballast. Temporal gait parameters were found to be different between uphill and downhill limbs. A trend was observed that the second peak KCF decreased in ballast conditions compared with no ballast. The timing of the first peak KCF was different among no ballast, main ballast and walking ballast. Knee muscle cocontraction was higher in walking ballast compared with no ballast in both peak KCFs. Knee muscle cocontraction was also higher for the uphill limb than the downhill limb. Lateral collateral ligament force was larger and medial collateral ligament force was smaller for the downhill limb compared with the uphill limb in both peak KCFs. The effect of surface configuration was significant for some ligament bundles, including

anterior cruciate ligament and medial collateral ligament in the first peak KCF, and lateral collateral ligament in the second peak KCF.

There are two additional findings in this research. First, the ankle kinematics was found to be sensitive to toe marker placement error and muscle forces responded the residual variance of joint kinematics in various degrees based on the muscle function. Second, a method to combine ground reaction data from different trials was described, which can successfully simulate the gait cycle and obtain the results of joint moments and muscle forces in a certain acceptable range.

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CHAPTER 1

INTRODUCTION

Railroad workers who service trains experience a unique exposure to walking on ballast, the rock that is used to support the rails and provide drainage. They work in railroad yards or along tracks to make up trains, inspect cars, and pick up or drop off cars at industrial sites [1, 2]. According to the Federal Railroad Administration (FRA), walking contributed 13.9% to 16.5% of all railroad worker injuries and accounted for 16.7% to 20.3% of the days absent between 1998 and 2006 (FRA, 1999-2008). However, the effects on workers from walking on ballast are still not clear, especially regarding the mechanical joint loads. Given that walking on ballast is a significant part of some railroad workers' jobs and uneven ground walking is a possible risk factor for knee osteoarthritis (OA) [3], it is imperative to evaluate the knee contact force (KCF) in different ballast conditions for this population.

Lower limb biomechanics for gait on hard, level surfaces are widely investigated, including joint kinetics, muscle forces and joint contact forces [4-10]. Analyses need to be extended to irregular surface conditions since walking on ballast remains an important topic of concern for safety and health professionals working with railroad workers. Although KCF during walking on hard, level surfaces has been reported for many years, most musculoskeletal models are oversimplified and fail to account for many factors, including ligaments, complex knee joint articulations and other kinematic constraints. Also, no research in the published literature predicts KCF during walking on ballast. Therefore, the aim of this research is to develop a musculoskeletal model with robust knee structures to investigate KCF for different ballast conditions.

1.1 Knee Osteoarthritis

OA is the most common form of arthritis and is characterized by the degradation of articular cartilage [11]. The knee is the weight bearing joint most commonly affected by OA [12]. Knee OA can cause several severe function limitations, such as walking, running and stair climbing. This may ultimately result in a total knee replacement [13, 14]. Several factors have already been known to contribute to the development and progression of knee OA. Knee OA increases in prevalence with age [15] and female gender [16, 17]. Obesity, as described by body mass index, is also significantly associated with knee OA [18-21]. Other risk factors include, but are not limited to, knee injury history, heredity, high impact sports, occupational bending and lifting [16, 22-27].

1.2 Walking on Ballast

Railroad workers experience a unique exposure to walking and performing tasks on ballast. Two ballast types are defined as walking ballast (WB), which is smaller rocks used for walking, compared with main ballast (MB), which is used for tracks [1, 2, 28]. There is a paucity of research reporting the kinetic and kinematic characteristics of walking on ballast. A study performed by Jensen and Eenberg (1995) suggested that walking over uneven ground was one of the possible risk factors for developing knee OA

[3]. Research by Andre et al. (2005) indicated that greater rear foot range of motion during walking on MB compared with walking on either WB or no ballast (NB) [2]. A case study, led by Merryweather (2008), reported that mean medial ground reaction force (GRF) increased for the downhill limb and mean lateral GRF increased for the uphill limb in slope configuration compared with the level configuration [28]. A follow-up study performed by Quincy (2009) further reported that the downhill knee joint had a higher adduction moment compared with uphill knee joint during walking on sloped surfaces [29]. A recent study, conducted by Wade et al. (2010), suggested that temporal gait parameters were significantly different for MB than for either WB or NB, and cocontraction levels were significantly greater on ballast compared with NB [1].

1.3 Technology to Predict Muscle Force

Muscle force prediction is an important component in the study of injury biomechanics. Challis and Kerwin (1993) suggested the intersegmental forces and torques, calculated from inverse dynamics, were due to three contributors: muscles, ligaments and joint contact forces [30]. Research by Herzog (2003) indicated that muscle forces were the primary determinants of joint contact forces and that correctly predicted muscle forces should result in sensible estimates of joint contact loads [31]. However, to date, accurate measurement and prediction of individual muscle forces are still a major challenge.

Three different strategies are typically used to predict individual muscle force. The first method is to estimate muscle forces based on an objective function within an optimization routine [8, 32, 33]. Traditional optimization criteria include static optimization and dynamic optimization. Usually, more muscles can be included in the musculoskeletal model when using static optimization due to a lower computational cost than dynamic optimization. However, the results of predicted muscle forces using static optimization are easily influenced by the accuracy of the experimental data and reconstructed joint kinematics [34, 35]. Four common static criteria are generally applied to estimate individual muscle force, including minimization of total muscle forces, minimization of total muscle moments, minimization of total muscle stresses, and minimization of total muscle activations [34]. Dynamic optimization can pose a time-dependent performance criterion to reduce the influence of errors from experimentally derived data. However, the tremendous computational expense and correctness of performance criterion are two main disadvantages of dynamic optimization [8, 32, 36]. Recently a new approach, computed muscle control (CMC), can compute a set of muscle excitations to reasonably predict muscle force by combining proportional-derivative control and static optimization [36, 37].

The second method is to reduce the number of unknown muscle forces to make the number of equations equal to the number of unknowns, resulting in a determinate system [38, 39]. The underlying assumption of this method is that certain muscles do not influence the system significantly and can either be excluded in the analysis or grouped with other muscles to represent a good estimate of the force acting within each separate muscle [7].

The third method is to combine muscle electromyography (EMG) data with an appropriate musculoskeletal model to estimate muscle force [33, 40, 41]. It is assumed that the EMG signal can precisely represent the actual muscle activation. However, the assumption is limited because the EMG signal acquires noise while travelling through different tissues and surface EMG detectors usually record signals from multiple motor units instead of a single motor unit [42].

1.4 Technology to Predict KCF

The determination of KCF is quite valuable for clinicians, researchers and implant designers to evaluate new knee replacements, simulate orthopedic procedures, predict clinical outcomes and investigate loading mechanisms that may cause knee OA [7].

Two techniques have already been used to determine joint contact loads. Telemetry, which has been successfully used to estimate in vivo loads at the human hip joint [43-45], cannot accurately predict KCF [46, 47]. Recently, instrumented knee implants provide another direct way to measure KCF, but this method is limited by the expensive cost and small sample size [10, 48, 49]. The other technique is to create a mathematical model to estimate joint contact loads. The widely applied method to calculate KCF is the vector sum of the knee joint reaction force using inverse dynamics and the compressive forces from the muscles crossing the knee joint [8-10, 49]. To date, the range of the peak KCF for gait is reported between 1.7 to 7.1 body weight from different studies [7, 50, 51].

1.5 Prediction of Muscle Force During Gait

Many researchers have already predicted lower limb muscle forces during gait on hard, level surfaces using mathematical models. However, predicted values vary widely.

Sometimes even for the same muscle, the muscle force-time profiles being reported are quite different during the gait cycle [8, 34, 40, 52].

There are four main reasons that great variability exists in predicted muscle forces. First, different models contain different number of muscles, and no model contains all the muscles in the lower body. This requires some muscles in the model to be a combination of several anatomical muscles. Second, different optimization methods may result in different muscle forces predictions and verifying the methods can be quite challenging. Third, the accuracy of muscle parameter values has a significant influence on the predicted muscle force. These parameters included physiological cross-sectional area, maximum isometric force, muscle-fiber length and tendon rest length. Fourth, the diversity among individuals can cause different predicted muscle forces for the same activities [40, 53]

According to research by Anderson and Pandy (2001), static optimization and dynamic optimization were practically equivalent for predicted muscle forces during gait [8]. A study performed by Li et al. (1999) suggested that different static optimization criteria predicted nearly identical muscle forces. However, kinematic information involved in the optimization played an important role in prediction of muscle forces [34].

1.6 Knee Ligament Modeling

The knee ligaments, which attach the femur to the tibia or fibula, are very important in stabilizing the knee joint and preventing knee injuries. There are four main ligaments in the knee joint including: 1) anterior cruciate ligament (ACL), which is a primary restraint to anterior tibial translation and secondary restraint to tibia rotation, 2) posterior cruciate ligament (PCL), which mainly restrains posterior translation of the tibia, 3) medial collateral ligament (MCL), which counteracts valgus instability, and 4) lateral collateral ligament (LCL), which primarily restrains varus stress of the knee joint and resists tibial external rotation.

In the previous research, the knee ligaments are represented by either single line elements or as multiple bundles of fascicles, with the path as a straight line [54-57]. The ACL and PCL are commonly represented by an anterior and a posterior bundle respectively. The MCL is usually separated into two portions: the superficial layer, represented by an anterior bundle, an intermediate bundle, and a posterior bundle; and the deep layer, represented by an anterior bundle and a posterior bundle. The LCL is generally represented by one bundle [58-63]. The effect of ligament-bone contact was considered in research by Hefzy and Grood (1982) [64, 65], and by Blankevoot and Huiskes (1991) [66]. However, the sensitivity analysis by Blankevoot and Huiskes (1991) indicated that ligament-bone contact had practically no effect on the relative position of the bones during flexion [66].

In the literature, the ligament bundles were assumed to be nonlinear elastic which meant that the tension in a ligament bundle was only a function of its length L or strain ε. The ligament strain was defined by Equation 1-1[66].

$$\varepsilon = (L - L_0)/L_0 \qquad \text{(Eq. 1-1)}$$

in which L_0 was the zero-load length of a ligament.

The zero-load length of a ligament bundle was determined by Equation 1-2 if the reference length L_r and the reference strain ε_r of the bundle were available.

$$L_0 = L_r/(\varepsilon_r + 1) \tag{Eq. 1-2}$$

The force-strain relationship for ligaments bundle was described as quadratic for low strain and linear for strain higher than a certain level [55, 66]. Specific formulas were enumerated in Equations 1-3 through 1-5.

$$f = \frac{1}{4}k\varepsilon^2/\varepsilon_l \qquad 0 \leq \varepsilon \leq 2\varepsilon_l \tag{Eq. 1-3}$$

$$f = k(\varepsilon - \varepsilon_l) \qquad \varepsilon \geq 2\varepsilon_l \tag{Eq. 1-4}$$

$$f = 0 \qquad \varepsilon < 0 \tag{Eq. 1-5}$$

in which f was the tensile force in a line element, k was the ligament stiffness, ε_l was the linear strain limit, and ε was the strain in the ligament calculated from Equation 1-1.

1.7 OpenSim Simulation

OpenSim is open-source software used to study the musculoskeletal system and create dynamic simulation of movement. Six steps are available to obtain predicted muscle force, which is shown in Figure1.1.

Since each individual has different anthropometry, a scale function is used to alter the general model to match a participant. Each body segment is scaled by comparing the relative distances between pairs of markers obtained from a motion capture system and

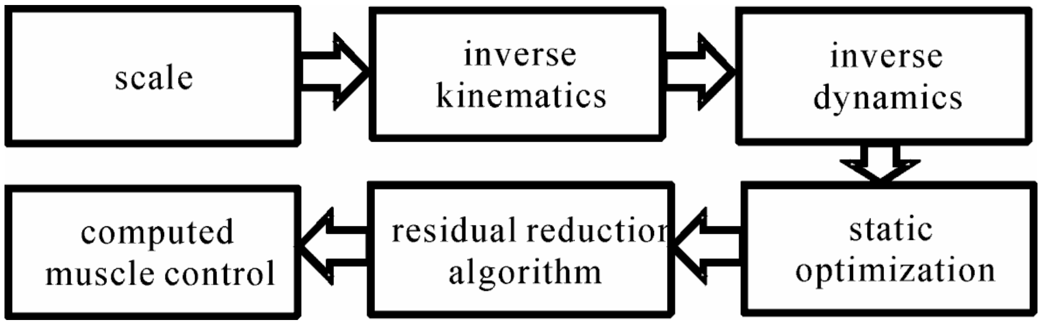

Figure 1.1: OpenSim Simulation Steps

the corresponding virtual marker located in the model. The inverse kinematics step, formulated as a weighted least squares problem, is used to reproduce the experimental kinematics recorded for a particular subject. Inverse dynamics and static optimization are optional steps for gait simulation. These steps can yield net moments and forces at each joint and distribute the net joint force to individual muscle forces at each instant in time. The residual reduction algorithm (RRA) step alters the torso mass center of the subject model and then slightly varies the kinematics of the model in order to make body kinematics more consistent with the dynamic GRF. The CMC step calculates muscle activation and muscle forces based on body kinematics and GRF from the previous steps [67].

OpenSim has several additional programs that help users to analyze a dynamic simulation. The body kinematics program can supply the position and orientation of each body reference frame in the global frame or a special local frame of the bodies. The point kinematics program can track any point’s position in any body-fixed coordinate by time series. The joint reaction program can report either joint reaction loads or joint contact loads, which are calculated as the forces and moments required to constrain the body motions based on the input information.

1.8 Research Purpose

The aim of this research is to investigate KCF during walking on ballast as surface conditions, surface configuration and uphill or downhill limbs change. The independent variables being controlled in this research are three surface conditions (MB, WB and NB); two surface configurations (smooth level surface and a slanted surface with a 7 °slope in the transverse plane), and the effect of the uphill or downhill limbs.

Following a general methods section for the fourth substudy of this dissertation in Chapter 2, four substudies will be discussed in Chapters 3-6, which are written as stand-alone manuscripts as follows:

- Chapter 3 (the first substudy) – Influence of toe marker placement error for lower limb joint kinematics and muscle force during gait
- Chapter 4 (the second substudy) – A method to combine force plate data together to simulate gait cycle and predict muscle force
- Chapter 5 (the third substudy) – Development of a new OpenSim model with robust knee structures
- Chapter 6 (the fourth substudy) – Investigation of knee contact force during walking on ballast

Marker error exists when recording data using motion capture systems. Misplacement of markers affects the accuracy of reconstruction and orientation body segments in a mathematical model. Some previous research focused on the effects of marker placement on different cases [68-70]. However, no research focused on the fluctuation of lower limb joint kinematics and muscle forces on toe marker placement error caused by footwear during gait. The hypothesis for the first substudy was that toe marker placement error caused by footwear affected lower limb joint kinematics and muscle forces during gait.

Successful trials are usually desired in order to predict muscle force in the lower limbs during gait. The criterion for successful trials is that both feet must be perfectly kept on two or more force plates during consecutive stance phases. Trials that do not meet this critical criterion will be rejected. This can significantly increase the total number of trials required to be collected [71]. A method to combine force plate data from different trials can effectively reduce the number of trials to be collected and help to predict lower limb muscle forces for a full gait cycle. The hypothesis for the second substudy was that the corresponding lower limb joint moments and muscle forces in the combined trial were not significantly different compared with the original, successful trial.

The KCF during walking on hard, level surfaces has been assessed in the literature [4-10]. However, most of the existing models include only muscles as force contributors and limit the knee joint to one degree of freedom (DOF) in the sagittal plane. Some previous research indicated that KCF was underestimated for gait on hard, level surfaces by excluding knee ligaments, especially the ACL [4, 6, 9]. Also previous efforts lacked body motions in the frontal plane and transverse plane, which could cause inaccurate muscle and joint reaction forces due to the different muscle excitation pattern [72, 73]. Therefore, a musculoskeletal model with robust knee structures was developed in the third substudy, which included four main knee ligaments, and multiple degrees of freedom for the knee joint, to provide more reasonable muscle forces and joint contact loads.

Railroad workers experienced a unique exposure to walking on ballast, which may be a possible risk factor for knee OA. A paucity of research reports the kinetic and kinematic characteristics during walking on ballast. However, the effect on workers is still not clear. Also, no research was found to evaluate KCF during walking on ballast. Therefore, the changes in KCF during walking on ballast were investigated as surface conditions, surface configuration and uphill or downhill limbs in the fourth substudy. It was hypothesized that KCF were significantly altered during walking on ballast compared with walking on hard, level surfaces. Additionally, it was hypothesized that walking on MB altered KCF more than walking on WB. The downhill limb was also hypothesized to have higher KCF than the uphill limb.

These chapters form a comprehensive body of research relative to modeling knee structure, simulating ballast gait and predicting KCF. The general conclusion of this dissertation, Chapter 7, consists of a discussion of the research as a whole. Common themes between chapters are addressed and directions for future work in this area are also outlined.

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CHAPTER 2

METHODS

2.1 Experimental Design

The independent variables being controlled for this research were: surface conditions, surface configurations, and uphill or downhill limbs. Surface conditions included MB, WB and NB (hard surface); surface configurations included a normal level surface and a slanted surface with a 7° slope in the transverse plane. The sloped surface represented the maximum slope of railroad yards [1]

Two tracks, 0.76 m wide and 7.3 m long, were built in the Ergonomics and Safety Laboratory in University of Utah, as shown in Figure 2.1. One track was filled with MB and the other with WB. Each track was filled 15-20 cm deep with aggregate, which was slightly compacted to minimize shifting during data collection. A hard surface made from structural plywood was placed over the walking ballast track to be used for NB trials.

The tracks were placed on the adjustable jacks so the same tracks could be used for both the level configuration trials and the sloped configuration trials. One force plate (model OR6-5-1000, AMTI, Watertown, MA) was embedded in the track. A custom force plate isolation fixture, shown in Figure 2.2, was developed to prevent significant dispersion of the surface force through the aggregate to the force plate. The fixture was found to effectively isolate the force plate and accurately record GRF [1].

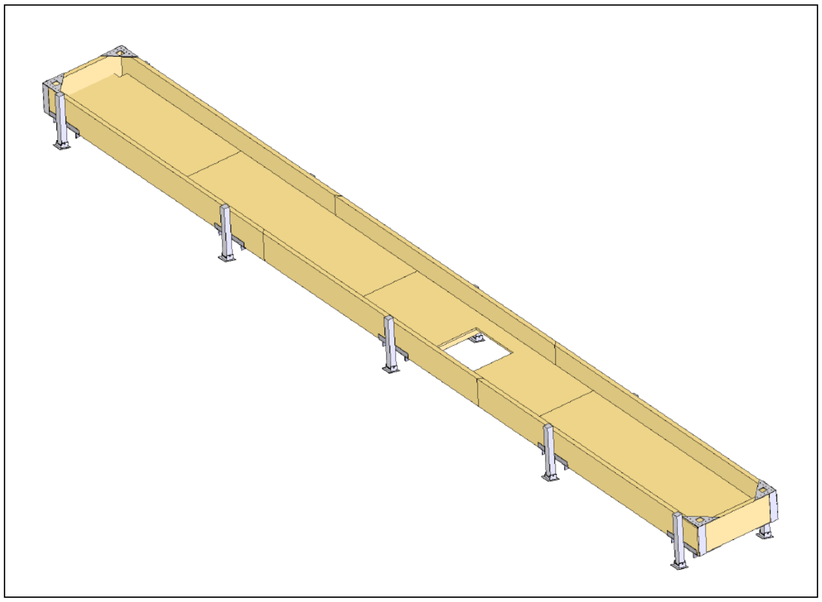

Figure 2.1: Track Design Model

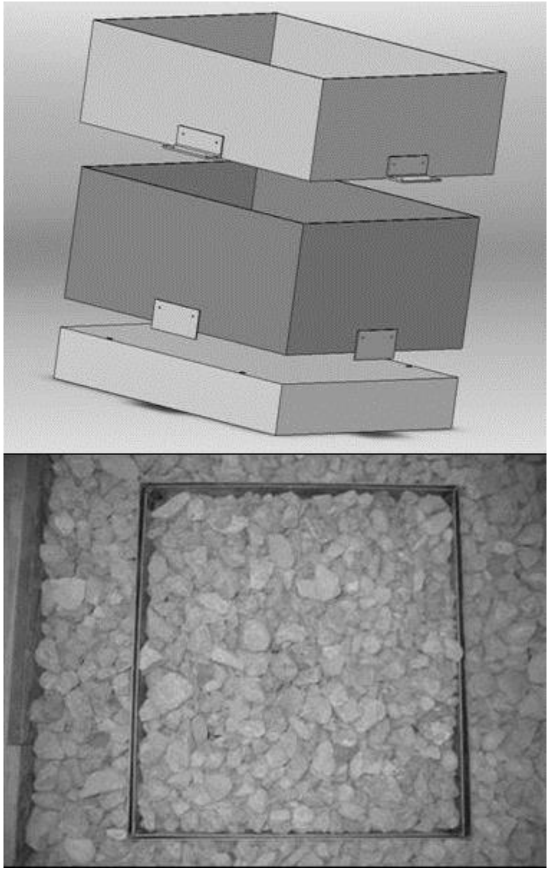

Figure 2.2: Force Plate Isolation Fixture

2.2 Data Collection

The study was approved by the University of Utah's Institutional Review Board (IRB). All the data were collected as part of a previous study [1]. Participants were brought to the Ergonomics and Safety Laboratory where they were interviewed to ensure they met all enrollment requirements. Then participants were outfitted with reflective markers. Marker locations were based on a modified Helen Hayes Marker Set [2]. Each participant was fitted with and given a new pair of model 2408 Red Wing work shoes for study participation. The markers on the foot and ankle were placed on the shoes bilaterally over the second metatarsal, heel, and lateral malleolus.

The combinations of surface conditions and configurations were randomized. Participants were allowed to walk on each surface to become familiar with each setup. This process also allowed researchers to find a suitable starting location on the track so that the foot was likely to have a clean strike on the force plate. For each experimental condition, five acceptable trials were collected for each limb. Acceptable trials had clean force plate strikes. The walking direction was kept the same for all trials. This meant that the right limb was always the uphill limb and the left limb was always the downhill limb for the sloped configuration. Each participant performed at least 60 trials (5 trials * 3 surface conditions * 2 surface configurations * 2 feet). An average of approximately 4 hours per session was needed to collect acceptable trials for each combination of conditions and configurations.

Motion data were collected at 60 Hz using a five camera Vicon Motus Video acquisition system (Vicon Motion Systems, Lake Forest, CA). Panasonic GS55 video cameras were used to capture the video. The force plate (model OR6-5-1000, AMTI, Watertown, MA) recorded GRF data at 600 Hz. A fourth order zero lag digital Butterworth filter with a cutoff frequency of 6 Hz was used to condition the raw marker position data. The global coordinate system was used for all trials with the positive X axis in the direction of motion, the positive Y axis right to left, and the positive Z axis upward. Calibration was done for each track condition prior to data collection.

2.3 Participant Inclusion Criteria

Eight railroad workers from Salt Lake City, Utah were selected to represent a healthy population of railroad workers. The participants consisted of conductors, switchmen, and other workers employed in positions involving walking on ballast in a train yard on a regular basis. Each participant read and signed an informed consent form approved by the IRB prior to participation. The study population demographics are shown in Table 2.1. The average participant was overweight as defined by BMI. More detailed information regarding data collection can be found in a publication of the previous study [1].

All participants met the following inclusion criteria:

- Age: 18-60
- BMI: Preferably between 18.5-24.9
- Railroad workers for minimum of 3 years
- Normal gait patterns
- No abnormal foot physical features
    - Club and flat feet
    - Extreme valgus or varus

Table 2.1: Study Population Demographics

Age (SD)	Years with Railroad (SD)	Height-m (SD)	Weight-kg (SD)	BMI (SD)
39.17 (8.80)	9.79 (8.30)	1.76 (0.09)	82.71 (14.14)	26.79 (4.01)

2.4 Statistical Analysis

The main variables of interest in this study include temporal gait parameters, the magnitude and timing of peak KCF, muscle cocontraction and ligament forces. Descriptive statistics were obtained for temporal gait parameters and peak KCF. Additional statistical tests were performed, specific to the data to be analyzed. These tests included t-test and analysis of variance (ANOVA). Results were considered statistically significant when $p < 0.05$ ($\alpha = 0.05$). Observed power was also computed. If the assumption of sphericity was violated, the Greenhouse-Geisser correction was used. Post hoc tests were performed using the Bonferroni adjustment to correct for multiple comparisons. All the statistics were performed using SPSS (IBM Corporation, Armonk, NY).

2.5 References

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CHAPTER 3

INFLUENCE OF TOE MARKER PLACEMENT ERROR FOR LOWER LIMB JOINT KINEMATICS AND MUSCLE FORCE DURING GAIT

3.1 Abstract

Marker placement and movement artifacts can be significant sources of error in biomechanics studies of human movement. Marker-based motion data is often collected where participants are shod during gait. The magnitude of toe (second metatarsal) marker placement error is amplified with footwear since the toe marker placement on the shoe only relies on an approximation of underlying anatomical landmarks. Limited research has been published regarding the fluctuation of lower limb joint kinematics and muscle force during gait resulting from toe marker placement error. The aim of the present study is to assess the influence of toe maker placement error caused by different footwear on lower limb joint kinematics and muscle force during gait.

The static trial combined with vertical height differences between heel marker and toe marker were used to generate a subject-specific model and determine the toe marker placement in four footwear conditions and a barefoot condition. A single dynamic gait trial was used to simulate these five conditions using OpenSim to obtain lower limb joint kinematics and muscle forces.

The results showed that ankle dorsi/plantarflexion had a statistically significant difference when comparing work shoe, sports shoe and leather shoe conditions with the barefoot condition. Statistically significant differences were found for hip flexion/extension, iliacus, psoas, rectus femoris, soleus, and tibialis posterior between the work shoe condition and the barefoot condition.

The present study suggested that ankle dorsi/plantarflexion was sensitive to toe marker placement error. The influence of toe marker placement error was relatively small for hip abduction/adduction and knee flexion/extension compared with hip flexion/extension and rotation. The lower limb muscle forces responded to the residual variance of joint kinematics in various degrees based on the muscle function for specific joint kinematics.

3.2 Introduction

Gait analysis is widely used to investigate normal and pathological gait to describe how humans walk, and has clinical value to rectify and refine treatment programs for abnormal gait [1-5]. The most commonly applied method of gait analysis is to structure around tracking clusters of reflective markers placed on the skin to identify various anatomical landmarks. These markers are used to reconstruct body segments and to define orientation of segments in space. However, some errors exist with this method and have been recognized on many occasions by previous researchers [6, 7]. Two of the largest sources of errors are marker misplacement and relative movement between the marker and the corresponding anatomical landmark during the period of marker capture [8-10]. The basic requirement for marker placement involves correct identification of a specific anatomical landmark on body segments. Then, markers can be used to represent anatomical landmarks to create a mathematical model or generate a subject- specific model.

Marker-based motion data are often collected where participants are shod during gait. Because of the obstruction from footwear, foot markers are usually placed on the footwear instead of a more accurate location on anatomical landmarks. The magnitude of marker placement error is amplified with footwear since the marker position on the shoes only relies on an approximation of underlying anatomical landmarks, as shown in Figure 3.1.

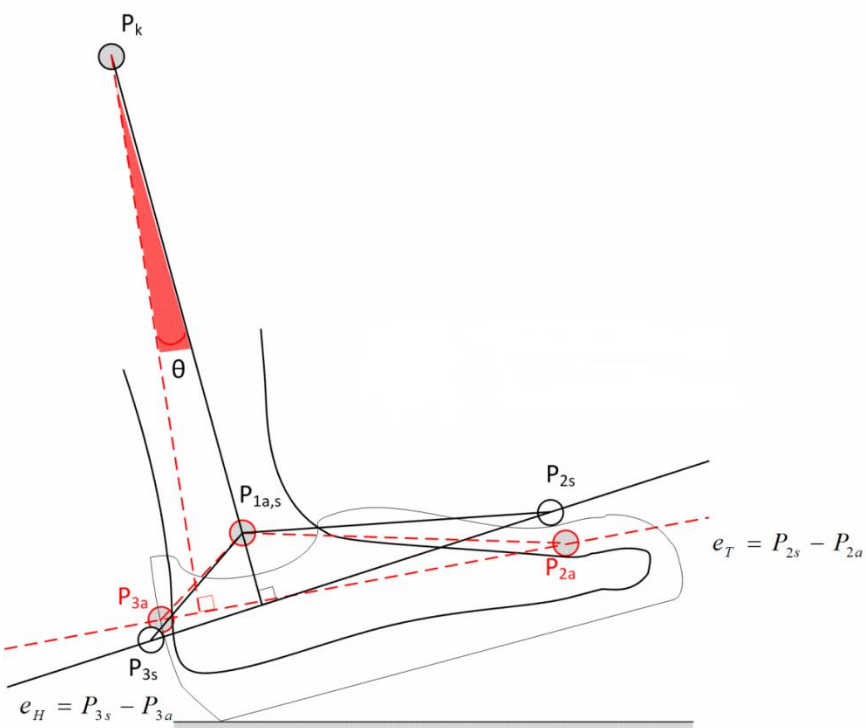

Figure 3.1: Foot Marker Placement Error

The darker markers (P_{1a}, P_{2a} and P_{3a}) and the light markers (P_{1s}, P_{2s} and P_{3s}) in Figure 3-1 represent anatomical positions and the approximate positions on a shoe, respectively. The ankle angle error in the sagittal plane is represented by θ. Toe and heel marker error between the anatomical position and the shod marker position are represented by e_T and e_H.

Some previous research has focused on the effects of marker placement on different cases. A case study performed by Szczerbik and Kalinowska (2011) evaluated the influence of knee marker placement error on gait kinematic parameters, their main finding was that kinematics for hip joint, knee joint and ankle joint was significant altered when knee marker position was changed in a systematical way [11]. O'Connor et al. (1993) carried out a study to investigate the effect of marker placement error on spinal motion. They found that marker placement had a significant effect for measuring the range motion of spinal flexion/extension and lateral side-bending [12]. A study, led by France and Nester (2001), evaluate the effect of error in the identification of anatomical landmarks for quadriceps angle. Their finding indicated that the quadriceps angle was highly sensitive to error in the definition of the center of the patella and tibial tuberosity [13]. However, no research was found to focus on the fluctuation of lower limb joint kinematics and muscle force on toe (second metatarsal) marker placement error caused by footwear during gait.

The purpose of this study was to assess the influence of toe maker placement error caused by footwear on lower limb joint kinematics and muscle forces during gait. It was hypothesized that toe marker placement error caused by footwear significantly affected lower limb joint kinematics and muscle forces during gait.

3.3 Methods

3.3.1 Experiment Data

The motion data, including marker-based video motion and GRF, were collected as part of a previous study [14]. An 83-year-old male, having a height of 166 cm and mass of 68 kg, was the subject for hard, level surface gait. Marker motion was collected at 120 Hz by using a 1camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA). Forty-five and 31 surface markers were attached to the subject in the static trial and the dynamic trial, respectively. The marker trajectory followed a modified Cleveland Clinic marker set. GRF data were collected at 3840Hz by using four force plates (AMTI Corporation, Watertown, MA).

Four types of footwear were chosen in this study to determine physically meaningful toe marker placement error as a function of common styles of footwear, which were shown in Figure 3.2. Toe marker placement for the barefoot condition was chosen as the reference position. The heel and toe markers were assumed to be at the same level in the sagittal plane in the barefoot condition. This meant that the ankle dorsiflexion angle was zero in the static barefoot condition. The heights of heel and toe markers in the sagittal plane were measured for the four pairs of shoes. The height difference h was calculated and is shown in Figure 3.3. The heights of heel and toe markers in the sagittal plane were represented by h_1 and h_2.

3.3.2 Data Process

Prior to running the OpenSim gait simulation, the motion data and GRF data were processed using MATLAB (The Mathworks, Inc., Natick, MA), including cubic spline

Figure 3.2: Four Types of Footwear
(A) work shoe (B) sports shoe (C) walking shoe (D) leather shoe

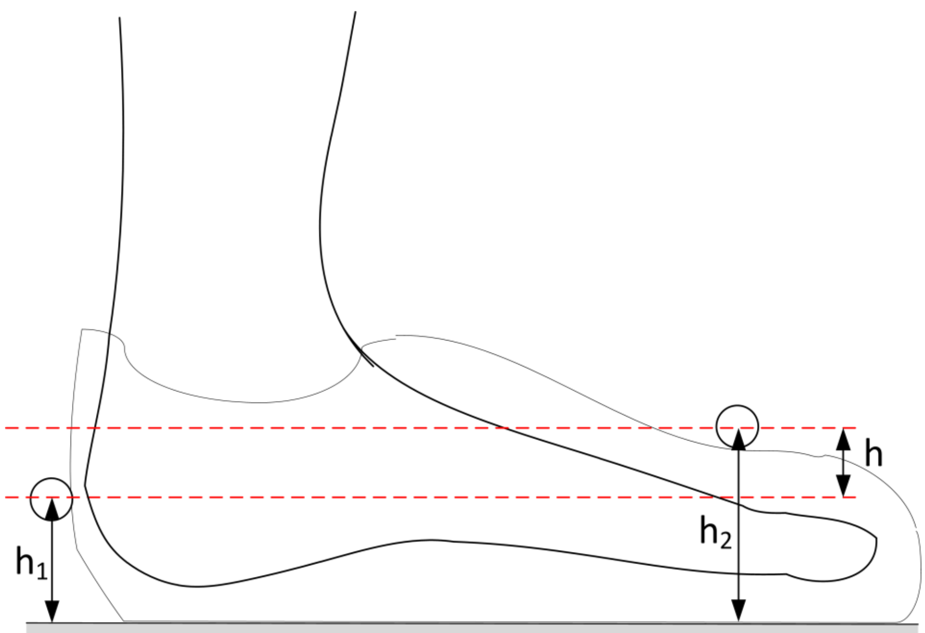

Figure 3.3: Height Difference Between Heel and Toe Markers

interpolation, cross-correlation and filtering (4th order Butterworth) with a low pass cut-off frequency of 15 Hz and 100 Hz, respectively. The synchronized frequency for the motion data and GRF data were 200 Hz and 1000 Hz, respectively.

3.3.3 Gait Simulation

OpenSim was used to generate 3D, subject-specific, muscle-actuated simulation for four footwear conditions and barefoot condition [15]. The 3D model used in this study consisted of 12 rigid segments, 23 DOFs, and 54 muscle actuators. The hip was represented as a 3 DOFs ball-and-socket joint, the knee was represented as a single DOF hinge joint and the ankle was represented as a single DOF universal joint. This model represented a simplified version of the lower extremity model proposed by Delp et al. (1990) [16], and was modified to include a torso and back joint based on the model of Anderson and Pandy (1999) [17].

The static trial was first used to generate a subject-specific model and locate markers in the model. In order to accurately locate toe markers and meet the assumption between toe and heel markers for the barefoot condition, the ankle dorsiflexion angle was set to zero during the scaling process. Once toe marker position was found in the subject-specific model for the barefoot condition, the position of the toe marker in the footwear conditions could be determined by the known the height difference combined with the assumption that heel marker had a fixed position in all five conditions. Then, a single dynamic trail and the subject-specific model were input into OpenSim for simulation of the five conditions. The inverse kinematics step was employed to determine joint kinematics by positioning the model as a “best match” pose, which was mathematically expressed as a weighted least squares problem. Lower limb muscle forces were reported after running the CMC step [15].

3.3.4 Statistical Analysis

The results of lower limb joint kinematics and muscle forces were compared between four footwear conditions and barefoot condition respectively. Descriptive statistics was obtained for joint kinematics and muscle force. Root mean square error (RMSE) and normalized root mean square error (NMSE) were used to describe the error magnitude and the residual variance respectively. The formulas for RMSE and NRMSE were shown in Equation 3-1 and Equation 3-2. Results were considered statistically significant when residual variance (NRMSE) above 10%. These statistics were performed using SPSS (IBM Corporation, Armonk, NY).

$$RMSE = \sqrt{\frac{\sum_{i=1}^{n}(x_{1,i}-x_{2,i})^2}{n}} \quad \text{(Eq. 3-1)}$$

$$N\text{RMSE} = \frac{RMSE}{x_{max}-x_{min}} \quad \text{(Eq. 3-2)}$$

3.4 Results

Height differences between heel and toe markers in the sagittal plane for different footwear and the barefoot conditions were reported in Table 3.1. The minimum and maximum height differences for footwear were 1.0 cm for the walking shoe and 5.5 cm for the work shoe, respectively.

Table 3.1: Height Difference Between Heel and Toe Markers

barefoot	work shoe	sports shoe	leather shoe	walking shoe
0cm	5.5cm	3.1cm	1.3cm	1.0cm

3.4.1 Lower Limb Joint Kinematics

All the corresponding lower limb joint kinematic curves in the four footwear conditions and the barefoot condition were visually similar except for ankle dorsi/plantarflexion, which was layered. The ankle dorsi/plantarflexion curves were significantly different when comparing the work shoe (NRMSE=43%), sports shoe (NRMSE=25%) and leather shoe (NRMSE=11%) to the barefoot condition. Work shoe condition had a statistically significant difference in hip flexion/extension compared with barefoot condition (NRMSE=14%). No statistically significant differences were found in hip abduction/adduction, hip rotation, and knee flexion/extension. These results are shown in Table 3.2, Figure 3.4 and Figure 3.5.

3.4.2 Lower Limb Muscle Forces

Sixteen lower limb muscles having maximum isometric forces above 500N, were chosen to compare between footwear conditions and the barefoot condition. Five lower limb muscle forces were significantly different in the work shoe condition compared with the barefoot condition. These muscles were iliacus (NRMSE=16%), psoas (NRMSE=16%), rectus femoris (NRMSE=13%), soleus (NRMSE=12%), and tibialis posterior (NRMSE=17%). No statistically significant differences were found for lower limb muscle forces in other footwear conditions compared with the barefoot condition. These results are shown in Table 3.3, Figure 3.6 and Figure 3.7.

Table 3.2: Kinematics Differences in Hip, Knee and Ankle Joints

		mean	RMSE	NRMSE
hip flex/extension	work shoe	9.42	5.63	14% *
	sports shoe	7.05	3.24	8%
	leather shoe	5.30	1.45	3%
	walking shoe	4.99	1.13	3%
	barefoot	3.87		
hip abd/adduction	work shoe	.66	0.50	2%
	sports shoe	.85	0.27	1%
	leather shoe	.94	0.11	1%
	walking shoe	.93	0.13	1%
	barefoot	-1.01		
hip rotation	work shoe	-5.16	1.94	10%
	sports shoe	-5.96	1.13	6%
	leather shoe	-6.55	0.55	3%
	walking shoe	-6.77	0.32	2%
	barefoot	-7.08		
knee flex/extension	work shoe	36.49	3.85	6%
	sports shoe	34.98	2.23	3%
	leather shoe	33.71	0.86	1%
	walking shoe	33.35	0.49	1%
	barefoot	32.93		
ankle dorsi/plantarflexion	work shoe	12.09	12.97	43% *
	sports shoe	6.47	7.34	25% *
	leather shoe	2.14	3.00	11% *
	walking shoe	1.36	2.22	8%
	barefoot	.85		

The units for mean and RMSE were degree
* Results were significant at the NRMSE >10% level

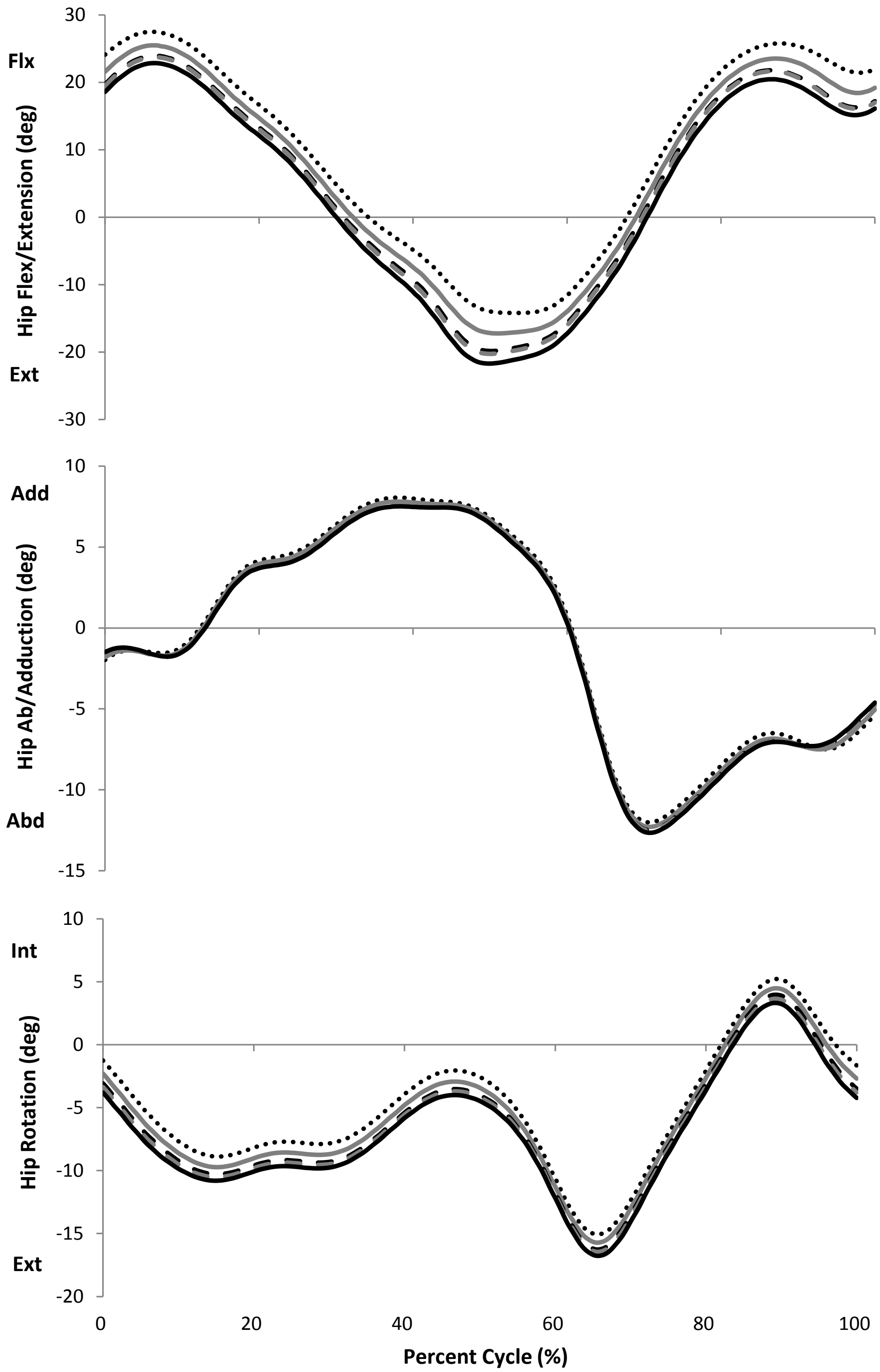

Figure 3.4: Comparison of Hip Joint Kinematics
Work shoe (point black), leather shoe (dashed black), walking shoe (dashed gray), sports shoes (solid gray) and barefoot (solid black)

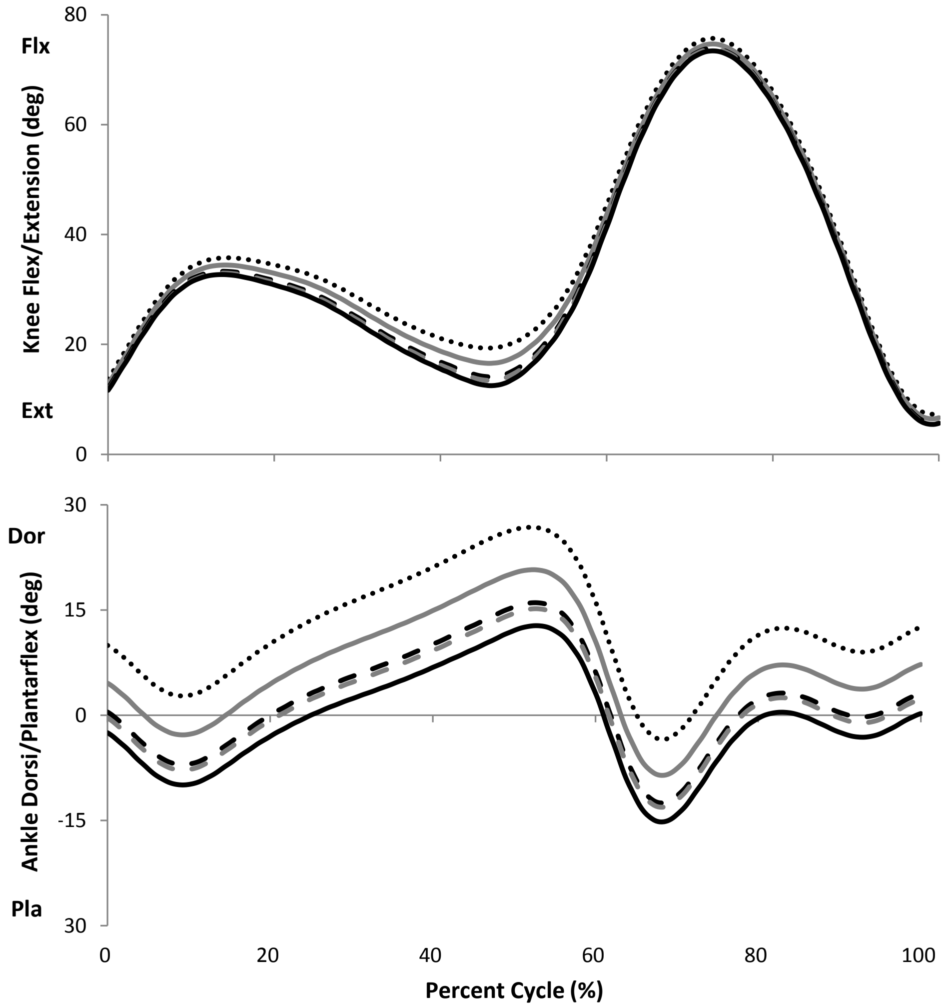

Figure 3.5: Comparison of Knee and Ankle Joint Kinematics
Work shoe (point black), leather shoe (dashed black), walking shoe (dashed gray), sports shoes (solid gray) and barefoot (solid black)

Table 3.3 Comparison of Lower Limb Muscle Force

Muscle	RMSE				NRMSE			
	work shoe	sports shoe	leather shoe	walking shoe	work shoe	sports shoe	leather shoe	walking shoe
gluteus maximus1	3.22	1.77	1.07	0.85	5%	3%	2%	2%
gluteus maximus2	5.79	3.55	2.14	1.82	5%	3%	2%	2%
gluteus medius1	28.66	17.91	7.70	5.82	4%	2%	1%	1%
gluteus medius2	9.11	6.58	3.92	3.82	3%	2%	1%	1%
gluteus medius3	18.77	11.62	6.28	6.16	6%	4%	2%	2%
iliacus	77.19	41.22	15.49	11.89	16% *	9%	3%	2%
psoas	83.46	44.47	17.16	13.25	16% *	8%	3%	2%
adductor magnus	6.17	4.34	1.61	1.45	9%	6%	2%	2%
biceps femoris long head	31.95	18.53	8.24	6.20	5%	3%	1%	1%
biceps femoris short head	23.75	14.02	6.64	4.97	6%	4%	2%	1%
rectus femoris	63.53	28.11	10.77	9.83	13% *	6%	3%	2%
vastus intermedius	98.00	53.78	19.31	12.09	9%	5%	2%	1%
soleus	197.73	140.21	58.58	48.38	12% *	9%	5%	4%
gastrocnemius	136.60	63.51	26.97	30.99	8%	5%	2%	3%
tibialis anterior	18.06	13.00	5.03	4.01	5%	4%	2%	1%
tibialis posterior	73.71	46.77	19.89	15.58	17% *	9%	4%	3%

The unit for RMSE was newton
*Results were significant at the NRMSE >10% level

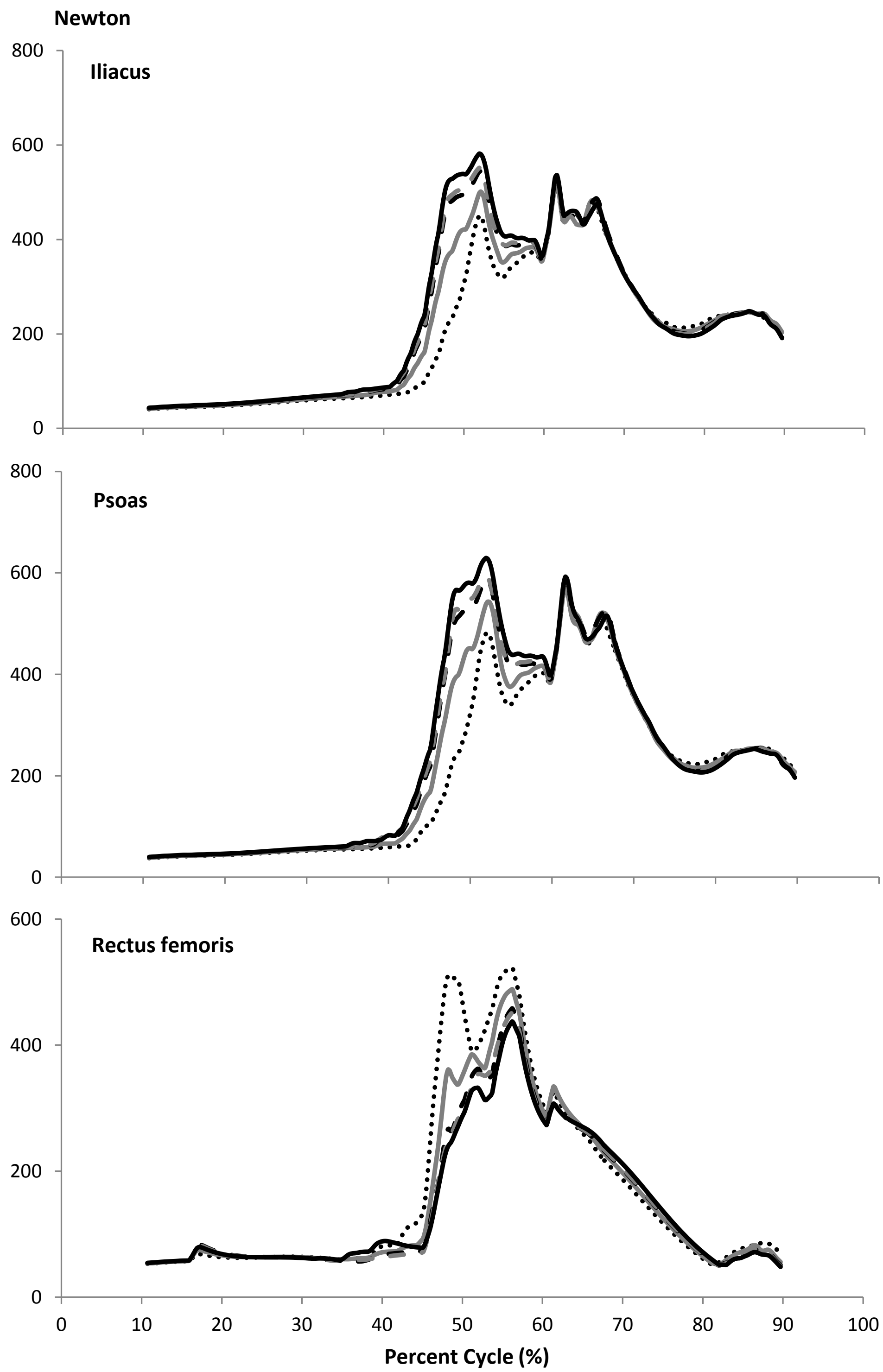

Figure 3.6: Comparison of Hip and Knee Joint Muscle Forces
Work shoe (point black), leather shoe (dashed black), walking shoe (dashed gray), sports shoes (solid gray) and barefoot (solid black)

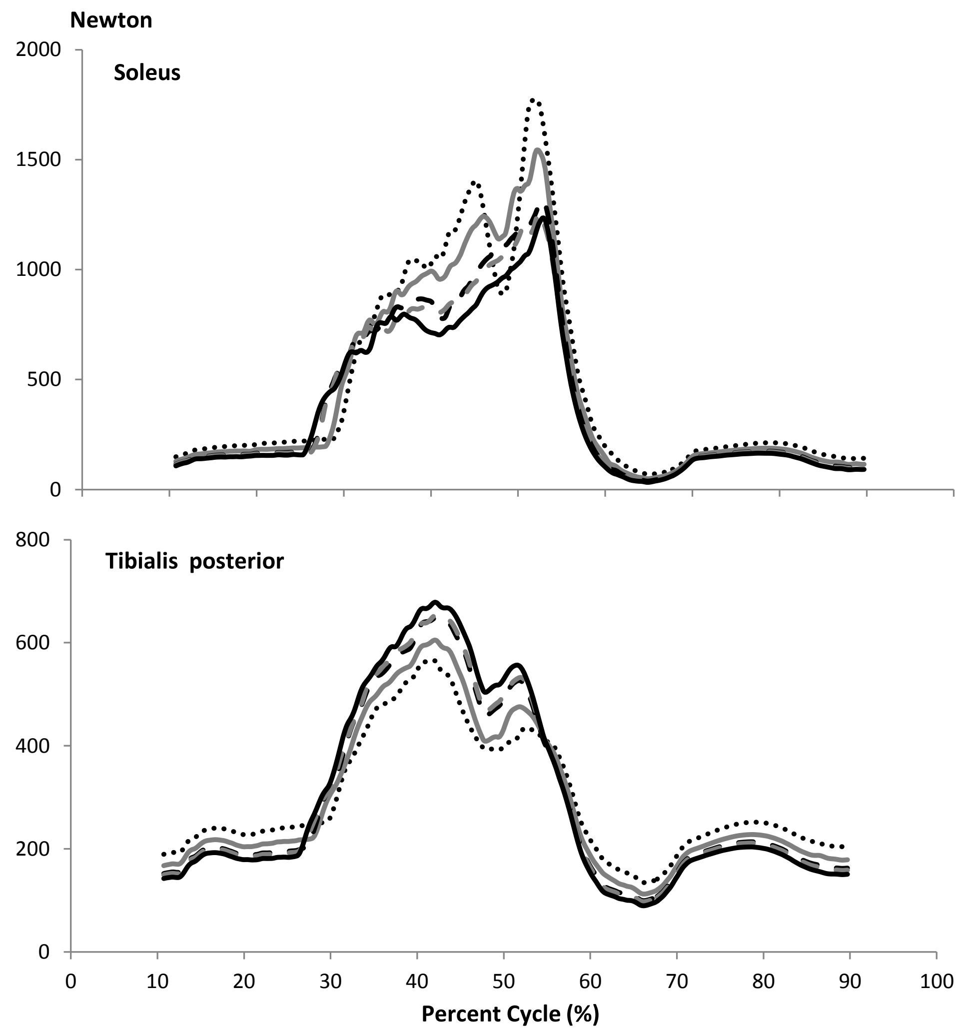

Figure 3.7: Comparison of Ankle Joint Muscle Forces
Work shoe (point black), leather shoe (dashed black), walking shoe (dashed gray), sports shoes (solid gray) and barefoot (solid black)

3.5 Discussion

Since one subject and a single dynamic gait trial were used in the present study, the inherent variability between individuals and the differences among gait trials were controlled. The same model incorporated identical muscle parameters and identical marker weight settings for all five conditions further removing other sources of error which may influence the computed joint kinematics and muscle forces. Therefore, the differences of lower limb joint kinematics and muscle forces between footwear conditions and barefoot condition were only a product of toe marker placement error.

One hypothesis of the present study was that toe marker placement error caused by footwear affected lower limb joint kinematics during gait. The hip joint and knee joint kinematics were not statistically different due to toe marker placement error except for hip flexion/extension for the work shoe condition. The hip joint and knee joint kinematics were mainly determined by the markers located on thigh and shank. However, other markers and the weight of markers also played a function in these joints because all joint kinematics were determined together as a marker weighted least square problem in OpenSim. Therefore, toe marker placement error theoretically affected all the joint kinematics in OpenSim simulation though the magnitudes were different. The residual variances of joint kinematics had a linear relationship with toe marker placement error for all lower limb joints. The ankle joint kinematics were more sensitive to the toe marker placement error than hip joint and knee joint kinematics since the toe and heel markers were main determinants for ankle dorsi/plantarflexion. This phenomenon was also indicated by the largest slope in residual variance for ankle dorsi/plantarflexion, which was shown in Figure 3.8. Toe maker placement error of 1.1cm would cause 10% residual

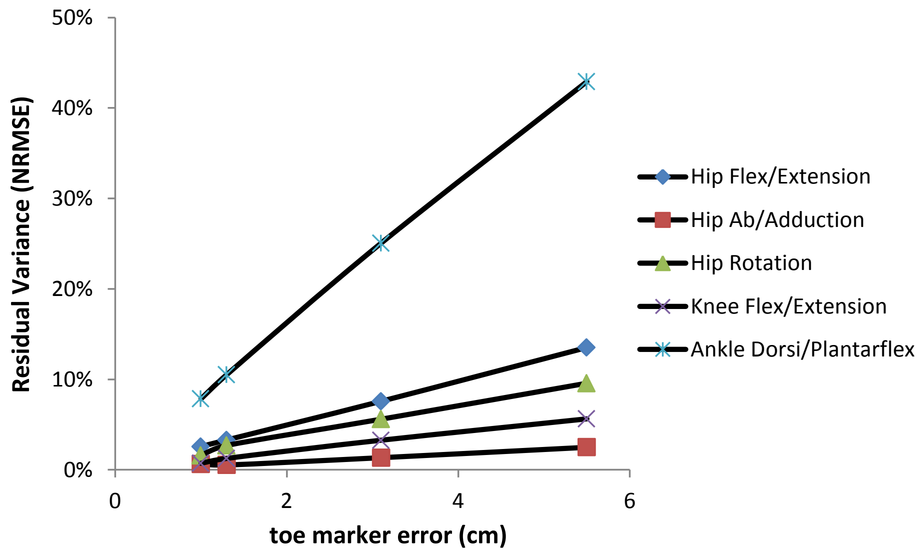

Figure 3.8: Joint Kinematics Residual Variances

variance for ankle joint kinematics and result in statistically significant differences in the present study. The influence of toe marker placement error was relatively small for hip abduction/adduction and knee flexion/extension compared to hip flexion/extension and hip rotation based on their residual variance slopes.

The other hypothesis of the present study was that toe marker placement error caused by footwear affected lower limb muscle forces during gait. Sixteen relatively large muscles were chosen in this study since they were the main force contributors for lower limb joints in this subject-specific model. Five lower limb muscle forces were significantly different between the work shoe condition and the barefoot condition. The significant differences for iliacus and psoas could be explained by their response to the residual variance for hip rotation in the work shoe condition since iliacus and psoas were the main functional muscles for hip external rotation. The significant differences for soleus and tibialis posterior resulted from the residual variance for ankle dorsiflexion/plantarflexion in the work shoe condition to prevent the body from falling forward and to keep body stabilization. The significant difference for rectus femoris was a compensation for the residual variance for knee flexion/extension and balanced the additional knee joint torque in the work shoe condition generated by gastrocnemius, which crosses both the knee and ankle joints in the model. The present study agreed with the conclusion from previous research that kinematic information played an important role in prediction of muscle force [18]. The toe marker placement error directly affected lower limb joint kinematics and indirectly altered muscle force in various degrees based on the muscle function for specific joint kinematics.

Toe marker placement error significantly affected joint kinematics (hip and ankle joints) and muscle forces (five muscles) in the work shoe condition compared with the barefoot condition. This error should be controlled for the work shoe condition in hard, level surface gait. A previous study performed by Merryweather reported that lower limb joint kinematics were similar when walking on ballast compared with NB [19]. Therefore, the effect of work shoes on predicted muscle forces in the present study can also be expected in ballast gait. Adjustment of the heel and toe markers to the same vertical height in the model during the static trial could effectively reduce toe marker placement error caused by footwear. This method could be used in ballast gait since subjects wore work shoe during walking on ballast in the fourth substudy.

3.5.1 Limitations

Some limitations existed in the present study. First, the change of gait pattern due to the footwear was neglected. Previous research by Cedirc et.al (2009) indicated that the shoes restricted the natural motion of the barefoot and imposed a specific foot motion pattern during the push-off phase [20]. A case study performed by Matthew et al. (2005) indicated the texture of footwear influenced ankle kinematics and muscle activities [21]. Second, the knee and ankle joints were both modeled as single DOF joints in the sagittal plane. Previous research indicated that the knee abduction/adduction, knee rotation and ankle rotation also existed during gait [22-25]. Lack of DOFs of knee and ankle joints would limit the ability to detect toe marker placement error for these two joints in the coronal and transverse planes and would further affect the corresponding functional muscles of the knee and ankle joints in these two planes. Third, the differences of gait patterns between elderly and young subjects were not considered in this study. Some previous studies reported that elderly people had different temporal gait parameters, decreased motion of the knee and hip joints compared with young subjects [26-29]. It is unclear if the predicted muscle forces and associated errors from marker placement had the same magnitude in young, healthy adults as was found with the 83-year-old male from this substudy. Finally, the mass of footwear was neglected in this study, which meant GRFs were the same for all footwear conditions and the barefoot condition.

3.5.2 Conclusion

In conclusion, the hypotheses that toe marker placement error caused by footwear affected lower limb joint kinematics and muscle forces during the gait cycle were partially supported. The ankle dorsi/plantarflexion was significantly different for the work shoe, sports shoe and leather shoe conditions compared to the barefoot condition, Also, hip flexion/extension and five muscle forces in the work shoe condition were different when compared to the barefoot condition. It was found that the ankle kinematics were very sensitive to toe marker placement error. The influences of the toe marker placement errors were relatively small for the hip abduction/adduction and knee flexion/extension compared with hip flexion/extension and hip rotation. The lower limb muscle forces responded to the joint kinematics residual variance to various degrees based on the muscle function for specific joint kinematics.

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CHAPTER 4

A METHOD TO COMBINE FORCE PLATE DATA TOGETHER TO SIMULATE GAIT CYCLE AND PREDICT MUSCLE FORCE

4.1 Abstract

Successful gait trials are important to clinical human walking research and related biomechanical studies. More representative data can typically be obtained as more trials are collected. However, due to the physical conditions of many clinical study subjects, the luxury of collecting many gait trials is uncommon. The ability to combine force plate data from different trials to obtain successful trials is meaningful and can significantly reduce the total number of trials to be collected. The aim of this study was to describe a method to combine force plate data from different trials to generate a combined trial to simulate full gait cycle biomechanics.

The most similar two trials from five successful trials, based on foot marker correlation, were chosen to generate a combined trial. GRF and center of pressure (COP) in the combined trial were generated by building a relationship between the chosen foot marker and GRF or COP in the two chosen trials. OpenSim was used to simulate the original trial and the combined trial. The results of lower limb joint moments and knee joint muscle forces were compared between the original trial and the combined trail to assess the method in this study.

The results indicated that GRF in the mediolateral direction and free torque in the vertical direction was significantly different in the combined trial compared to the original trial. Statistically significant differences were found for hip abduction/adduction moment, hip rotation moment, knee flexion/extension moment and ankle dorsi/plantarflexion moment. The muscle forces generated by the biceps femoris long head, gastrocnemius and rectus femoris were found to be significantly different between the original trial and the combined trial.

The method described in this study can be successfully used to combine GRF and COP from different trials to create a successful trial to simulate the gait cycle. Furthermore, joint moments and muscle forces are able to be obtained within a certain acceptable range. The findings of the present study depended on the repeatability of foot marker placement among the trials and the accepted level of residual variance in the specific research. This method could be applied to several situations with populations who were unable to complete a large number of trails, such as those impaired gait, the elderly, amputees and pediatrics. The proposed method could significantly reduce the total required number of trials to study lower limb biomechanics and movement disorders.

4.2 Introduction

GRF and COP are commonly recorded in gait analysis using force plates. These data allow the musculoskeletal model to calculate net joint moments using inverse dynamics and to obtain muscle forces using optimization methods [1, 2]. One of the major challenges of capturing data with force plates is that the subjects' feet may not both fall entirely on the force plate during the corresponding stance phases. This situation can significantly increase the number of rejected trials and total trials required to obtain the desired number of successful trials. Research by Bates et al. (1983) reported that a minimum of eight successful trials were necessary in order to achieve statistically stable data, which was based on a normal subject [3]. A case study performed by Hamill and McNiven (1990) examined the reliability of GRF time domain parameters over 20 trials, the main finding was that at least 10 successful trials were necessary for stable GRF data during walking [4]. Although the entire foot on the force plate is a critical criterion for the successful trial, subjects usually are not instructed to look at the force plate, or are not made aware of the presence of the force plate in order to prevent targeting. As a result, many trials are rejected which requires more repetition and incurs additional costs. A common solution is to adjust the starting point at a distance from the force plate to increase the possibility of an acceptable entire foot placement on the force plate. However, clinical populations often include those whose physical conditions may not tolerate numerous gait trials. The total number of trials is limited and any rejected trials, by reason of incomplete force plate data, represent the loss of a meaningful amount of data [5]. Therefore, the development of a method to combine force plate data from different trials has meaningful potential and could significantly reduce the total number of trials necessary to be collected.

The purpose of the present study was to describe a method to combine force plate data from different trials to create successful, sequential foot contact events in order to simulate full gait cycle biomechanics. The hypothesis of this study was that the corresponding lower limb joint moments and knee joint muscle forces in the combined trial were not significantly different compared with the original, successful trial.

4.3 Methods

4.3.1 Experimental Data

An 83-year-old male, having a height of 166 cm and mass of 68 kg, was the subject in this study. Five successful gait trials on a hard, level surface were collected as part of a previous study, which was same as described in Chapter 3 [6]. Marker-based motion data were collected by a 1camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA). Ground reaction data were recorded by four AMTI force plates (AMTI Corporation, Watertown, MA). The force plates were equally spaced except that the first force plate was adjacent to the second force plate. The criterion for successful trials in this study was that the subject had a clean right foot strike on the force plate 3 and a clean left foot strike on the force plate 2 and 4, which was shown in Figure 4.1. The global coordinate system was set as the X axis pointed forward from the subject, the Y axis pointed upward, and the Z axis pointed to the subjects' right.

4.3.2 Combination of Trails

This method comprised five steps to combine force plate data to generate a combined trial. 1) Gait Event Identification: the gait events in five successful trials were detected including heel-strike and toe-off. 2) Correlation Analysis: the correlation coefficients of 10 paired toe and heel markers from these five trials were calculated using SPSS (IBM Corporation, Armonk, NY). The paired trials with the highest correlation of

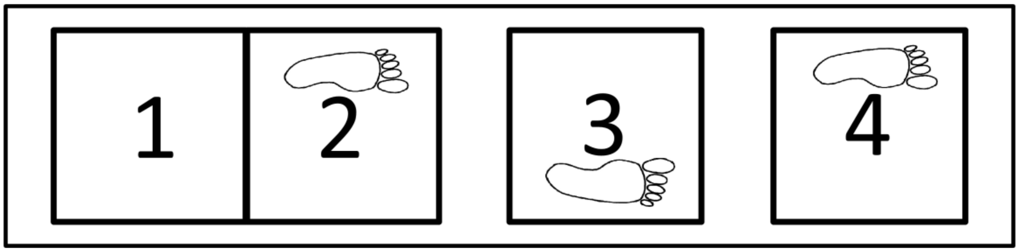

Figure 4.1: Layout of Force Plates

heel marker or toe marker were chosen and identified as No.1 and No.2. Then, the key marker in No.1 and No.2 was identified as the pair with the highest correlation of the toe marker or heel marker. 3) Motion Data Normalization: the marker data and force plate 3 data for the right foot stance phase in No.2 was normalized using MATLAB (The Mathworks, Inc., Natick, MA) to time match the corresponding motion data of the right foot stance phase in No.1. 4) Force Plate Data Combination: the new, combined force plate dataset for force plate 3 in No. 1 was generated based on the force plate 3 data in No.2 by utilizing Equation 4-1 through 4-3. 5) New Force Plate Data Normalization: the new, combined force plate dataset was normalized to match the original force plate 3 data in No. 1. The underlying assumptions for trials combination were that the GRF vector and the relative position between the key marker and the COP for No.1 and No.2 were equivalent..

$$COP_{x1} - X_{key1} = COP_{x2} - X_{key2} \quad F_{x1} = F_{x2} \quad Torque_{x1} = Torque_{x2} \qquad \text{(Eq.4-1)}$$

$$COP_{y1} - X_{key1} = COP_{y2} - X_{key2} \quad F_{y1} = F_{y2} \quad Torque_{y1} = Torque_{y2} \qquad \text{(Eq.4-2)}$$

$$COP_{z1} - X_{key1} = COP_{z2} - X_{key2} \quad F_{z1} = F_{z2} \quad Torque_{z1} = Torque_{z2} \qquad \text{(Eq.4-3)}$$

The variable COP_x, COP_y and COP_z were the coordinate values of the COP in the global frame; X, Y and Z were the corresponding coordinate of the key marker in the global frame; F_x, F_y and F_z were the GRF in three axes; $Torque_x$, $Torque_y$ and $Torque_z$ were the free moments in three axes.

4.3.3 Analysis Process

The 3D model used in the present study was the same model as described in Chapter 3, which included three DOFs for the hip joint and a single DOF for the knee and ankle joints, respectively. A seven muscles system was used in the knee joint in this model. This included five knee flexors: biceps femoris long head (BFLH), biceps femoris short head (BFSH), gracilis (GRAC), gastrocnemius (GAS), sartorius (SAR), and two knee extensors: rectus femoris (RF) and vastus intermedius (VAS). OpenSim was used to simulate the original No.1 trial and the new No.1 trial with combined force plate 3 data. Net joint moments were determined by the inverse dynamic step and muscle forces were determined by the CMC step.

Three statistical parameters were used in this study to compare GRF, COP, lower limb joint moment and knee joint muscle forces during the stance phase of the gait cycle. Root mean square error (RMSE) and normalized root mean square error (NRMSE) were used to describe the magnitude of differences and the residual variances, respectively. Correlation coefficients were also calculated to describe the linear relationship of the time series curve. Results were considered statistically significant when $p<0.01$ ($\alpha=0.01$) for the correlation coefficient and when the residual variance (NRMSE) was above 10%. The formulas for RMSE and NRMSE are the same as Equation 3-1 and Equation 3-2 in Chapter 3. Statistics were performed using SPSS (IBM Corporation, Armonk, NY).

4.4 Results

The heel marker was chosen as the key marker due to the higher anteroposterior and mediolateral correlation coefficient compared with the toe marker in the stance phase, which is shown in Table 4.1. The correlation coefficient in the vertical direction could be neglected because the value of COP_y was constant during walking on hard, level surface.

GRF and COP were found to have statistically significant correlation in all three directions (p<0.01). No statistically significant differences were found for GRF in the anteroposterior and vertical directions. However, the mediolateral GRF (NRMSE=15.31%) and free torque in the vertical direction (NRMSE=11.45%) were significantly different between the original and combined trials. The results are shown in Table 4.2, Figure 4.2 and Figure 4.3.

4.4.1 Lower Limb Joint Moments

The lower limb joint moments were found to have significant correlation for the hip, knee and ankle joints (p<0.01). No statistically significant differences were found for the hip flexion/extension moment. However, the significant differences were found for the hip rotation moment (NRMSE=11.49%), hip abduction/adduction moment (NRMSE=14.69%), knee flexion/extension moment (NRMSE=11.33%), and ankle dorsi/plantarflexion moment (NRMSE=13.90) between the original and the combined trials. The results are shown in Table 4.3, Figure 4.4 and Figure 4.5.

Table 4.1: Correlation Coefficient for Heel and Toe Markers

	Axis	Heel marker	Toe marker
Correlation coefficient	X (anteroposterior)	0.998	0.998
	Y (vertical)	0.996	0.998
	Z (mediolateral)	0.985	0.768

Table 4.2: Comparison of GRF, COP and Free Torque

	F_x	F_y	F_z	COP_x	COP_z	$Torque_y$
Correlation	0.988*	0.994*	0.866*	0.953*	0.583*	0.943*
RMSE	14.55N	21.33N	9.45N	3.1cm	2.1cm	0.45Nm
NRMSE	5.70%	2.94%	15.31%	8.12%	7.86%	11.45%

*correlation is significant at α=0.01 level (2-tailed)

Table 4.3: Comparison of Lower Limb Joint Moments

	Hip flexion	Hip adduction	Hip rotation	Knee extension	Ankle dorsiflexion
Correlation	0.987*	0.806*	0.934*	0.914*	0.938*
RMSE	8.62N.m	10.17N.m	2.68N.m	10.77N.m	15.80N.m
NRMSE	6.13%	14.69%	11.49%	11.33%	13.90%

*correlation is significant at α=0.01 level (2-tailed)

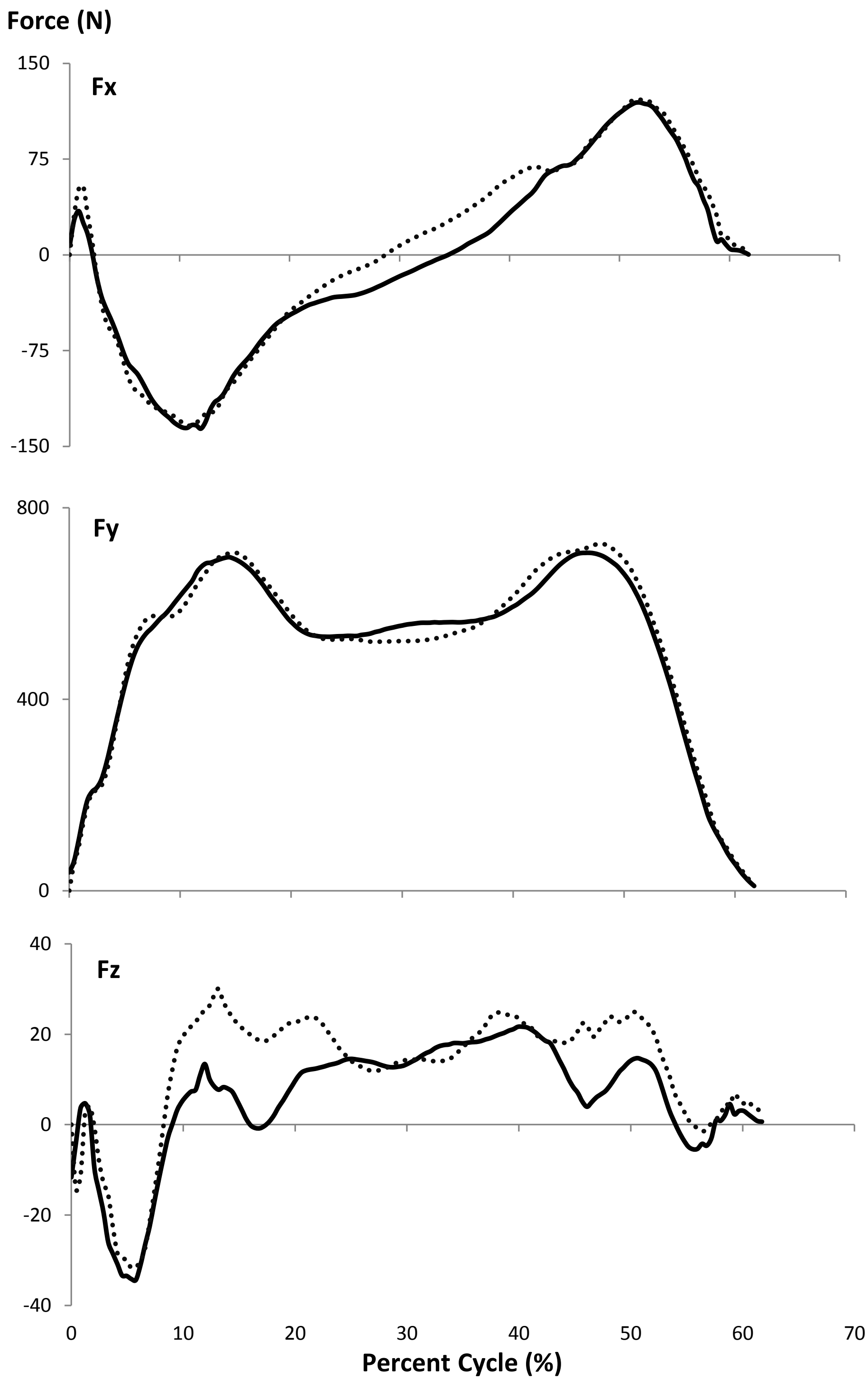

Figure 4.2: Comparison of GRF
Original trial (solid black), combined trial (point black)

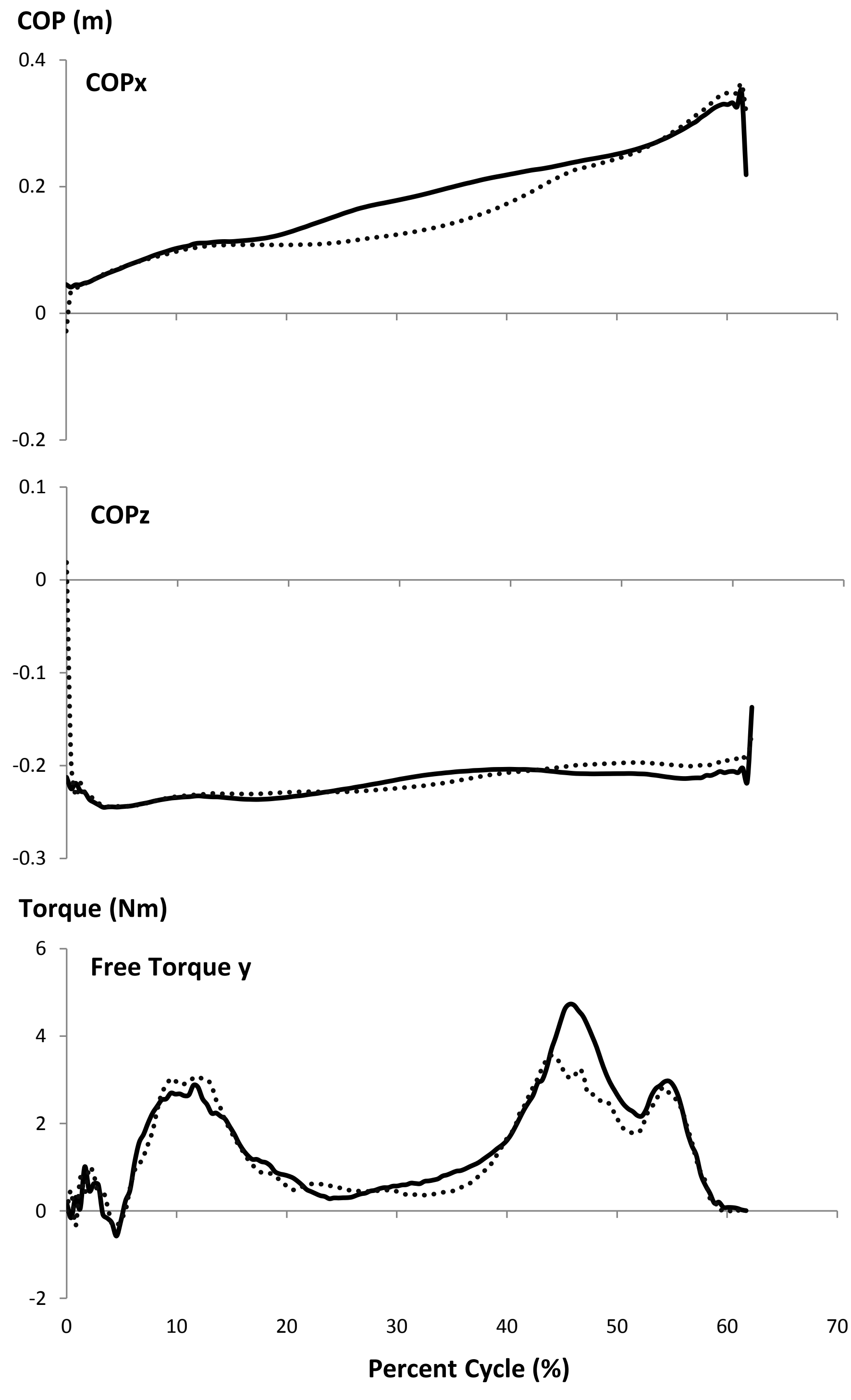

Figure 4.3: Comparison of COP and Free Torque
Original trial (solid black), combined trial (point black)

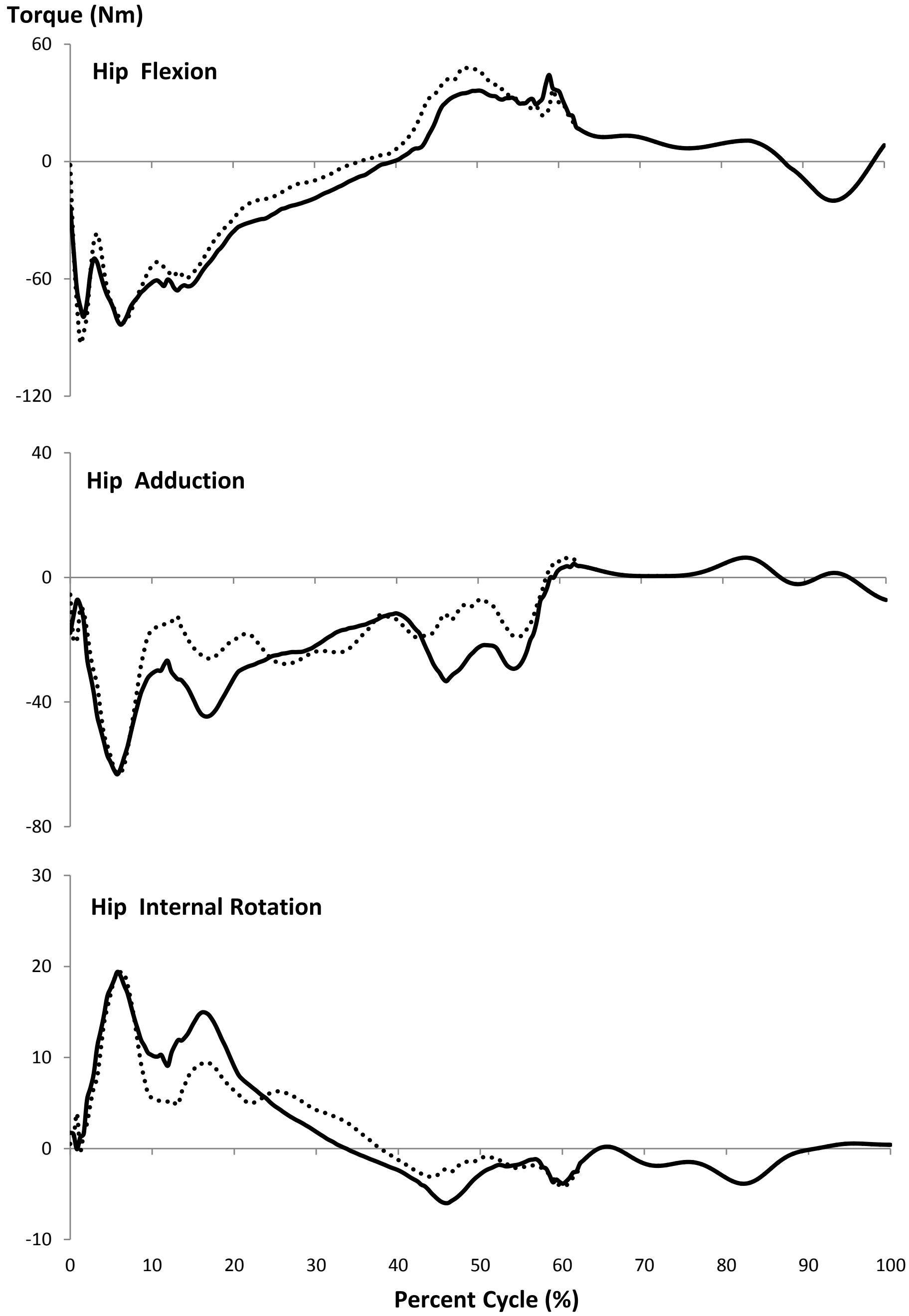

Figure 4.4: Comparison of Hip Joint Moments
Original trial (solid black), combined trial (point black)

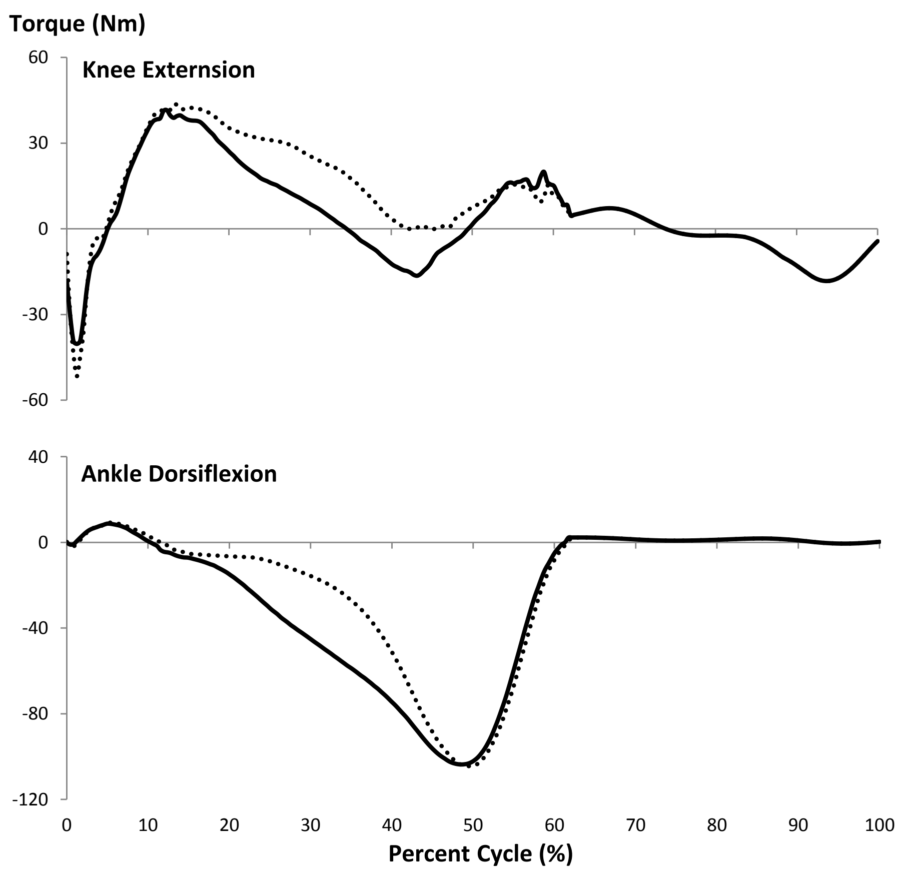

Figure 4.5: Comparison of Knee and Ankle Joint Moments
Original trial (solid black), combined trial (point black)

4.4.2 Knee Joint Muscle Forces

The knee joint muscle forces were found to have statistically significant correlation ($p<0.01$). No statistically significant differences were found for BFSH, GRAC, SAR and VAS. However, significant differences were found for BFLH (NRMSE=12.47%), GAS (NRMSE=13.37%) and RF (NRMSE=11.36%) between the original and the combined trials. The results are shown in Table 4.4, Figure 4.6 and Figure 4.7.

4.5 Discussion

4.5.1 Gait Symmetry and Variability

In the present study, the most similar two trials were chosen from five successful trials based on the highest correlation of heel markers among the pairs of trials. As the numbers of collected trials increased, there was a greater possibility that high correlation coefficients for key marker would be obtained. Although the number of successful trials (five trials) in this study was less than the necessary number mentioned by previous research (eight or ten trials) [3, 4], the correlation coefficient for key markers was still high for all three axes (0.998, 0.996 and 0.985). It was indicated that a mean value of GRF calculated from multiple trials, instead of a single trial, could reduce gait variability [3, 4, 7]. However, this method was difficult to use in conjunction with COP due to the variability of the contact position on the force plate in different trials. Some previous research suggested that the left-side stance and swing phase was the mirror of the right-side stance and swing phases, respectively, and that full gait cycle curves could be obtained by simulation of half a gait cycle [8-11]. However, research from Herzog (1989)

Table 4.4: Compare Knee Joint Muscle Forces

	BFLH	BFSH	GRAC	GAS	SAR	RF	VAS
Correlation	0.932*	0.994*	0.969*	0.954*	0.983*	0.945*	0.989*
RMSE	73.20N	14.69N	1.75N	171.38N	1.66N	61.12N	82.66N
NRMSE	12.47%	4.73%	5.85%	13.37%	6.32%	11.36%	7.65%

*correlation is significant at α=0.01 levels (2-tailed)

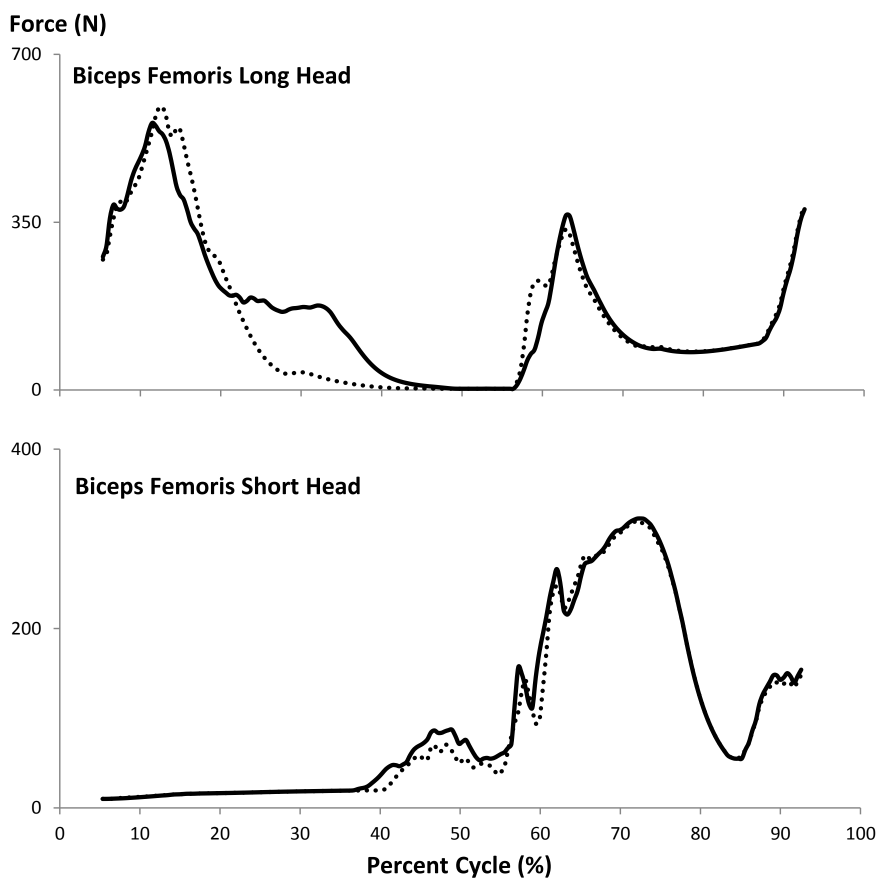

Figure 4.6: Knee Flexors Muscle Forces
Original trial (solid black), combined trial (point black)

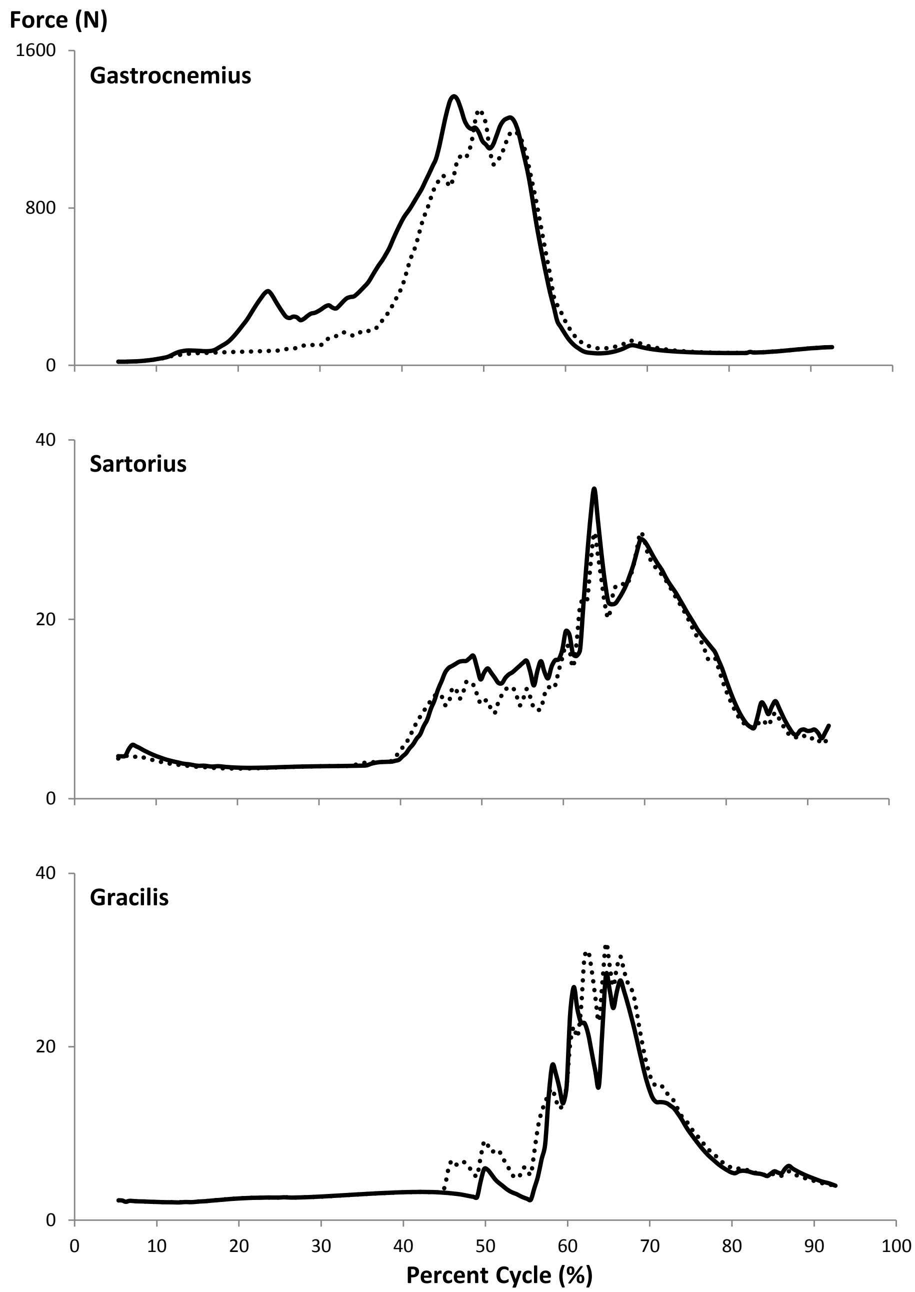

Figure 4.6: Continued
Original trial (solid black), combined trial (point black)

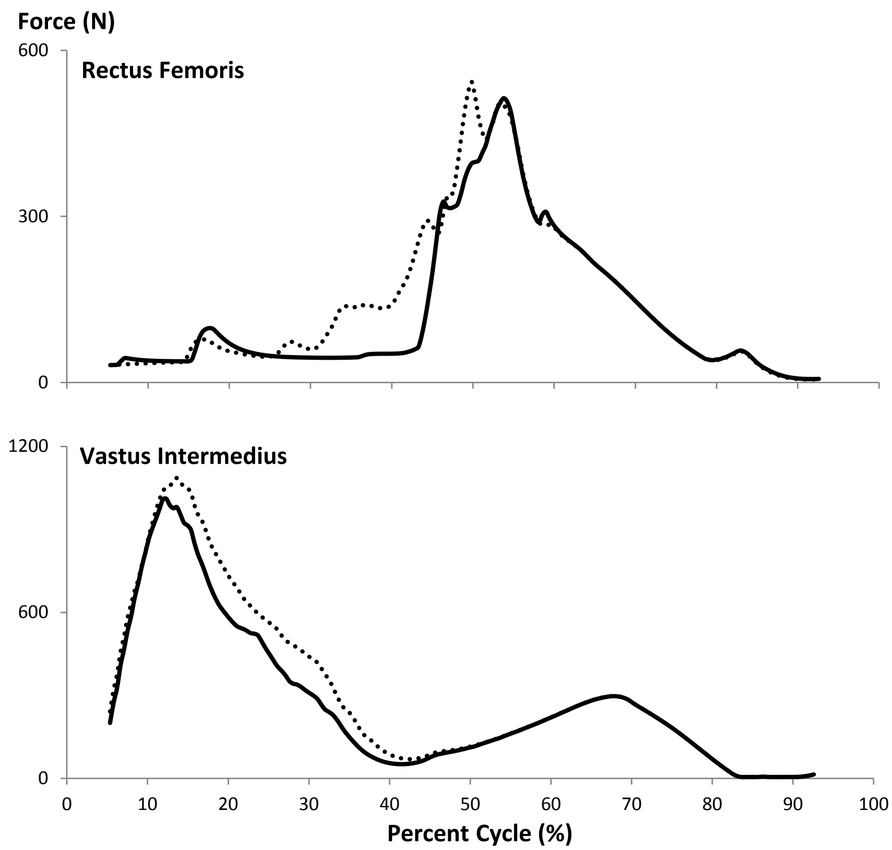

Figure 4.7: Knee Extensors Muscle Forces
Original trial (solid black), combined trial (point black)

indicated that asymmetry of GRF patterns for left and right side of human gait widely existed [12]. The assumption of gait symmetry may not be appropriate in the present study, especially for the subject who had a knee implant or lower limb injury history. The trial combination method used in this study could effectively evade the possible differences between the two legs by using the ipsilateral leg instead of the contralateral leg. It would also remove the differences of the corresponding foot marker positions between two feet, which may result from artificial error when marking the anatomical landmark.

GRF had the best repeatability in the vertical direction and worst repeatability in mediolateral direction based on the correlation and residual variance. The results in this study were in agreement with some previous research which showed that residual variance was below 5% for vertical GRF and below 10% for anteroposterior GRF [13-15]. The residual variance was about 15% for the mediolateral GRF which indicated that mediolateral GRF was not reliably measured and may be inappropriate for comparison purposes. This finding was also consistent with the results from previous studies [12, 14].

Anteroposterior and mediolateral COPs had high variability in the beginning and the end, which decreased the correlation and increased RMSE for COP_x and COP_z. This phenomenon could be explained in two ways. First, the noise included in GRF could propagate to the calculation of COP for most force plates. Second, since anteroposterior and mediolateral COPs were calculated by dividing vertical GRF, it was most sensitive at early and late stance where vertical GRF was low.

The most significant variability of free torque in the vertical direction occurred between 40%-60% of the gait cycle. This occurrence was most likely caused by mediolateral GRF, which had a digressive linear relationship with the vertical free moment.

4.5.2 Lower Limb Joint Moments and Knee Joint Muscle Forces

The hypothesis of the present study was that the corresponding joint moments and knee joint muscle forces in the combined trial were not significantly different compared with the original, successful trial. The results for the hip flexion/extension moment and four knee joint muscles supported this hypothesis, while the results for all other joint moments and three other knee joint muscles did not. The magnitude of differences and the residual variance for joint moments decreased from distal to superior in the sagittal plane. The differences in anteroposterior GRF first directly affect the ankle moment, and then indirectly affect the knee and hip moments in the sagittal plane. The high residual variance for the hip abduction/adduction moment was in response to the high residual variance of mediolateral GRF. The lack of DOF for the ankle and knee joints in the transverse and coronal planes likely accumulated as differences in GRF at the hip joint which caused high variability in these two planes. The large knee muscles were relatively more sensitive to the change of the knee flexion/extension moment than small muscles since they were the main force contributors.

4.5.3 Limitations

Some limitations existed in this study. First, the seven knee muscles system used in this model represented 13 muscles crossing the knee joint in the human body. This meant some muscles in the model represented the combination of multiple muscles in the human body that had similar functions. However, this may affect the evaluation of magnitude of differences and residual variance for knee muscle forces in this study. Second, some previous research indicates that the knee abduction/adduction and rotation occur during gait [16, 17]. Therefore, the lack of DOF for the knee joint in the transverse and coronal planes may lead to incorrectly predicted muscle forces, especially for muscles such as biceps femoris which plays a role in knee abduction/adduction and rotation,. Third, data for one subject were used for this study to make all the conclusions. This reduces the ability to confidently generalize study results. Possible future research may be to increase DOF for the knee joint and increase the sample size to reevaluate the conclusions of this study.

4.5.4 Conclusion

In conclusion, the hypothesis that the corresponding joint moments and knee joint muscle forces in the combined trial were not significantly different from the original, successful trial were partially supported by the results for the hip flexion/extension moment and three knee joint muscles. The method described in this study can successfully be employed to combine GRF and COP from different trials to create one successful trial to simulate the gait cycle. Furthermore, the joint moments and muscle forces can be obtained and can be within a certain acceptable range. The conclusions of the present study depend on the repeatability of foot markers among the trials and the accepted level of residual variance in specific research. The method described is not limited gait on hard, level surfaces gait and could be applied for several gait pattern trials where the subject’s physical condition does not allow for collection of numerous trials.

However, this method was not applied to the ballast trials in the fourth substudy because the threshold of acceptance for joint moments and muscle forces was not met.

4.6 References

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CHAPTER 5

DEVELOPMENT OF A NEW OPENSIM GAIT MODEL WITH ROBUST KNEE STRUCTURES

5.1 Abstract

A three-dimensional OpenSim gait model with robust knee structures was developed in this study based on an existing model. Three main contributions of the present model compared with the existing model were: 1) The patella and patella tendon including all the parameters were involved in the model, as well as patellotibial joint. 2) Six degrees of freedom knee joint was built in this model, which included three rotations and three translations. Three knee rotations and knee mediolateral translation were independent. The knee proximodistal and anteroposterior translations were defined as a function of passive knee flexion. 3) Knee cruciate ligaments and knee collateral ligaments were involved in the model, as well as the geometry and mechanical properties of the ligament.

The present model was used to simulate knee rotations in the three body planes to investigate the ligament function. Quantitative comparison of the results in this model with previous experimental data and knee models reported in the literature indicated that the geometry of the ligaments in the present model was similar to those evident in the physical knee and other existing knee models.

5.2 Introduction

Walking is a fundamental ability for humans. Kinetic and kinematic characteristics of human gait on hard, level surfaces have been investigated and well understood in the past decades. The knee joint is one of the most complex joints in human body due to complicated articular geometry, multibody contact, and multibody musculature that comprise the knee joint [1]. Determination of KCF is important and has major implications in at least three different areas: 1) prediction of the performance of new implant designs, 2) simulation of orthopedic surgery procedures and optimization of clinical outcomes based on proposed surgical parameters, and 3) investigation of loading mechanisms that contribute to degenerative joint disease, as well as clinical interventions to reduce these effects [2]. In recent years, much research has focus on prediction of knee joint muscle forces and KCF during gait owing to the vast increase of computer power and the availability of robust algorithms for predicting muscle force [3-9]. However, some potential limitations exist in previous musculoskeletal gait models.

First, muscle was considered as the only force generator and not all the muscles in the lower limb were included in most of previous gait models [5, 7-9]. Research by Anderson and Pandy (2001) indicates that a lack of some muscles would not significantly change the joint contact loads. They believed that increasing the number of muscles meant simply separating combined muscles into individual muscles if the new muscles had approximately the same moment arms as the combined muscles [5]. However, verification of the combined muscle would be quite challenging since electromyography, the most common method to confirm the predicted muscle force, was usually used to record single muscle activation levels.

Second, tissues, including ligaments, were not included in most previous gait models, so it was likely that KCF results were underestimations due to a lack of ligament forces [6, 9, 10]. Some previous studies suggested that knee ligaments, especially the anterior cruciate ligament (ACL), could influence knee joint loading depending on the movement task, [11-14]. The peak ACL force during gait on hard, level surfaces was reported to be range from 0.2 to 1.7 body weight [15-17]. Kevin et al. (2004) carried out a study to investigate the pattern of ACL loading during hard, level surface gait using two 3D models together. They found that ACL bore load throughout stance phase and a peak ACL force of about 300 N occurred at the beginning of single support phase. This was explained by the shear forces acting at the knee [13].

Third, most of previous gait models represented the knee joint as a single DOF hinge joint and reported similar knee flexion/extension curves during hard, level surface gait [7, 9, 18, 19]; However, it was indicated that knee adduction/abduction and knee rotation existed during hard, level surface gait based on marker-motion data, which are shown in Table 5.1. The average knee rotation was about 10 degrees in the frontal plane and 15 degrees in the transverse plane during the gait cycle [18, 224]. Some previous studies suggested that knee motions in the frontal plane and the transverse plane were usually restricted by knee flexion based on different human activity and heavily affected by knee ligaments [25-29]. Research from Xiao and Higginson (2008) and Glitsch and Baumann (1997) indicated that the lack of body motions in the frontal plane and the transverse plane may cause inaccurate muscle and joint forces due to the different muscle excitation pattern [30, 31].

Table 5.1: Mean Knee Joint Kinematics During Gait

	Kadaba et al. N=40	Sutherland N=15	Isacson et al. N=20	Chao et al. N=110
Knee (Degree)				
Flexion	56.7	58	60.6	68
Varus	13.4	N/A	9	10
Rotation	16	12	12.9	13

Most of the previous musculoskeletal models only included muscle as force contributors and limited the knee joint to a single DOF in the sagittal plane. The purpose of this study was to develop a new OpenSim gait model with robust knee structures including the patella, patellar attachments, knee ligaments and multiple DOFs for the knee joint. This model was to be calibrated and compared with previous research to assess the reasonableness of the ligament geometries.

5.3 Methods

The new OpenSim gait model in this study was based on the existing gait model (Gait2354 model) used in the first and second substudies which consisted of 12 rigid segments, 23 DOFs, and 54 muscle actuators. The hip was represented as a 3 DOF ball-and-socket joint, the knee was represented as a single DOF hinge joint, the ankle was represented as a single DOF universal joint. This existing model represents a simplified version of the lower extremity model proposed by Delp et al. (1990) [32], and was modified to include a torso and back joint based on the model of Anderson and Pandy (1999) [33].

5.3.1 Model of the Tibiofemoral Joint

The reference frame of the tibia was based on the transverse axis, which passed through the centers of the medial and lateral posterior femoral condyles. The origin of the tibial reference frame lied on the transverse axis at the midpoint between these centers. The transverse axis pointed laterally and was the z axis of the tibia. The y axis was perpendicular to the transverse axis and pointed proximally. The x axis pointed anteriorly and was formed by the taking the cross product of the y and z axes.

The reference frame of the femur was fixed at the center of the femoral head and had the same orientation as the reference frame of the tibia when the knee was fully extended. Six generalized coordinates described the position and orientation of the tibia relative to the femur: internal and external rotations about the y axis; abduction and adduction about the x axis; flexion and extension about the z axis. The position of the origin of the tibial reference frame was defined by translation along each of these axes: proximal and distal translations along the y axis; anterior and posterior translations along the x axis; and medial and lateral shifts along the z axis. The reference frames of the femur and tibia are shown in Figure 5.1.

5.3.2 Model of the Patellotibial Joint

The orientation of the patellar reference frame was the same as that of the tibial reference frame when the knee was fully extended; however, the origin of the patellar reference frame was located at the most distal point of the patella, which is shown in Figure 5.2. Rotation of the patellar with respect to the tibia in the sagittal plane was defined as a function of passive knee flexion. This was determined by experimental

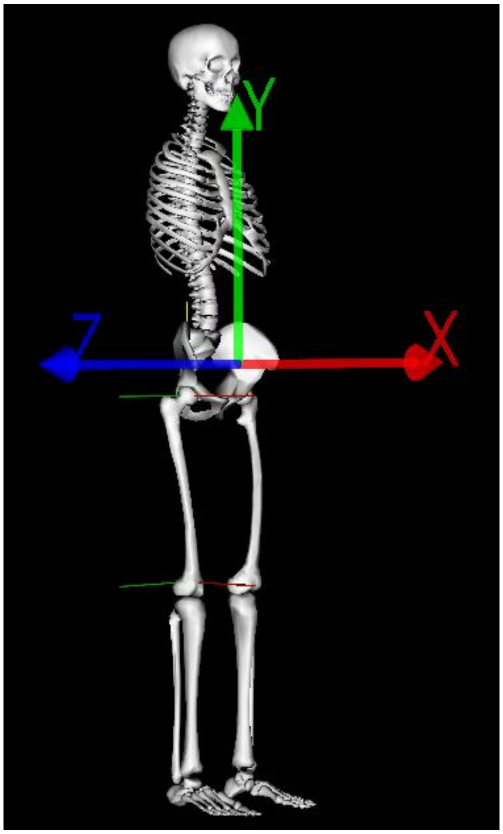


Figure 5.1: The Reference Frames of Femur and Tibia

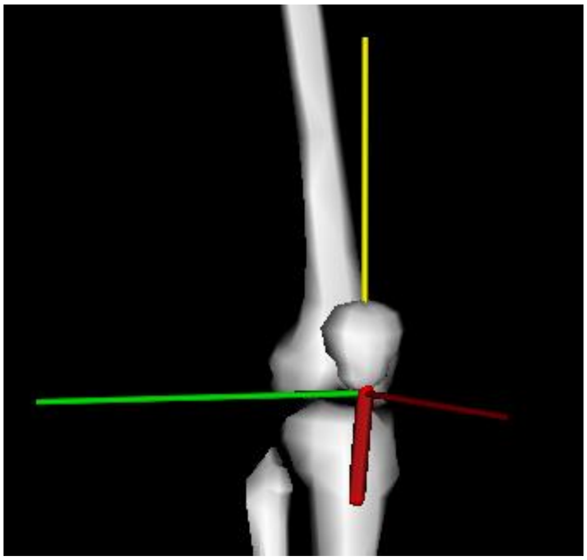


Figure 5.2: Patella Tendon and The Reference Frames of Patella

measures of patellar tendon and patellar rotation from Delp et al. [32] and van Eijden et al. [34]. Mediolateral translation, coronal plane rotation and transverse plane rotation were not defined between the patella and tibia in the present model. The patellar tendon was defined as muscle in this model and the parameters used for the patella and patellar tendon were from Arnold et al. [35].

5.3.3 Model of the Ligaments

Ten separate bundles were used to model the geometry and mechanical properties of knee cruciate ligaments and knee collateral ligaments, which are shown in Figure 5.3. The ACL and PCL were each represented by an anterior bundle and a posterior bundle; The MCL was separated into two portions: a superficial layer composed of an anterior bundle, an intermediate bundle, and a posterior bundle; and a deep layer, represented by an anterior bundle and a posterior bundle; The LCL was represented by one bundle [36-41]. The abbreviations of the bundles of ligament are shown in Table 5.2.

The attachment sites of ligament bundles in this study were based on the dataset reported by Blankevoot et al. [40]. The tibial insertion of pACL was assumed zero in mediolateral direction (z=0) in the reference frame of tibia in this model. The path of each ligament bundle was approximated as a straight line; the effect of ligament-bone contact was neglected. Each ligament bundle was assumed as elastic and its properties were described by a nonlinear, force-length curve [42]. The stiffness value and reference length of the ligament bundles in this model were based on the data reported by Pandy et al. [38], which is shown in Table 5.3.

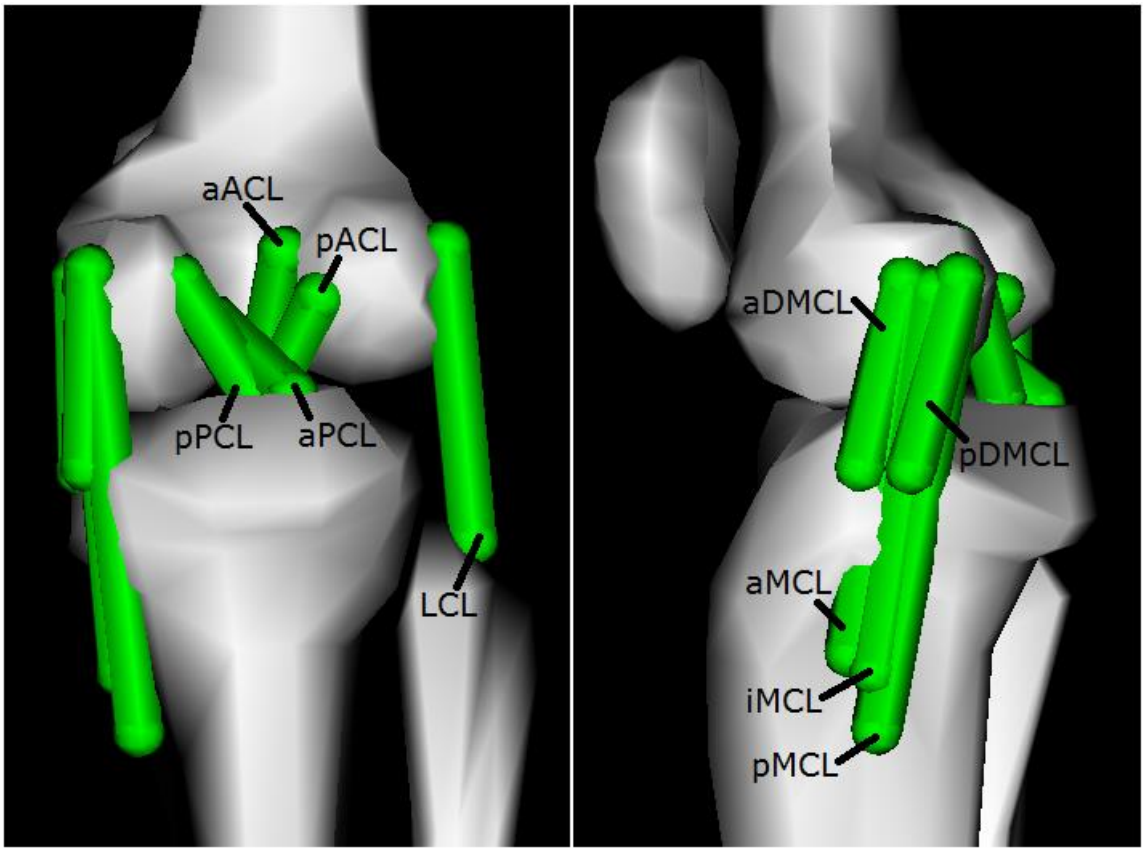


Figure 5.3: The Geometry of Knee Ligament Bundles

Table 5.2: Abbreviations of Ligament Bundles

ACL-Anterior Cruciate Ligament	aACL-anterior bundle
	pACL-posterior bundle
PCL-Posterior Cruciate Ligament	aPCL-anterior bundle
	pPCL-posterior bundle
MCL-Medial Collateral Ligament	aMCL-anterior bundle
	iMCL-inferior bundle
	pMCL-posterior bundle
DMCL-Deep Medial Collateral Ligament	aDMCL-anterior bundle
	pDMCL-posterior bundle
LCL-Lateral Collateral Ligament	

Table 5.3: The Parameters for Ligaments

Ligament bundle	Stiffness (N)	Reference Strain
aACL	1500	0.02
pACL	1600	0.01
aPCL	2600	0.23
pPCL	1900	0.02
LCL	2000	0.02
aMCL	2500	0.02
iMCL	3000	0.04
pMCL	2500	0.02
aDMCL	2000	0.08
pDMCL	4500	0.03

5.3.4 Calibration of the Attachment Sites of Ligaments

In order to calibrate the attachment sites of each ligament in the model, knee rotation, abduction/adduction and mediolateral translation were temporarily locked. Then, knee proximodistal translation and anteroposterior translation were defined as a function of passive knee flexion, which was based on data from Yamaguchi and Zajac (1989) [43] and Delp et al. (1990) [32]. Finally, the attachment sites of each ligament were adjusted in the interval of 2.5mm in all three directions to determine the best femoral and tibial insertions to match the data found in the literature [38-40]. The assumption was that the reference length and orientations of each bundle were constant during the process of calibrating the ligament attachment sites.

5.3.5 Model of the Muscles

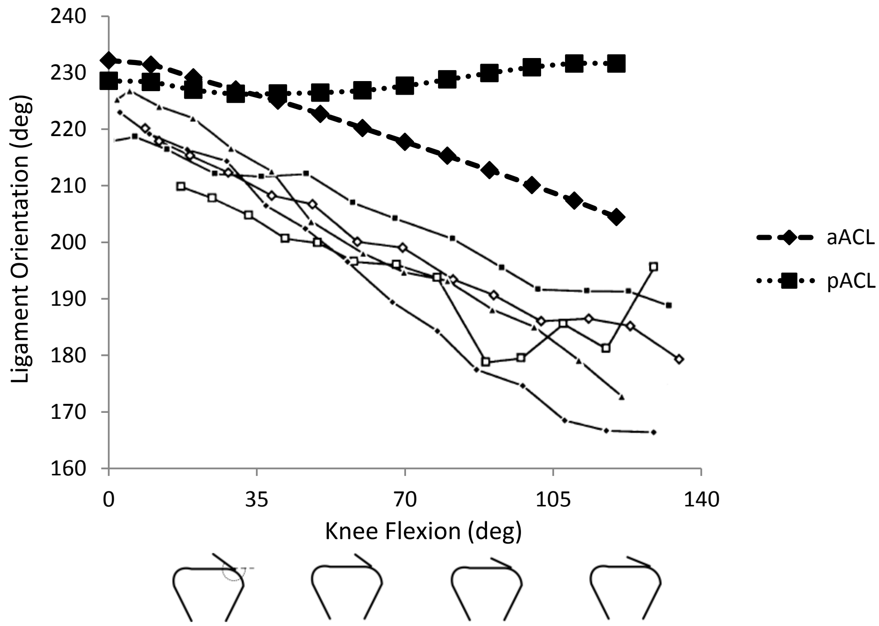
The muscles in the model were defined by musculotendinous units. Each unit represented as a three-element muscle in series with tendon, including a Hill-type contractile element, a series-elastic element, and a parallel-elastic element. Tendon was assumed to be elastic, and its properties were described by a linear force-length curve. The muscle path was defined by a series of attachment points in this model. Three different types of attachment points included fixed point, via point and moving point. Fixed point was the point whose XYZ offsets were constant in a body-fixed coordinate frame. Via point was the point fixed to a body frame, but they were only shown in the muscle path when a specified coordinate was in a certain range. Moving point was point whose XYZ offset in a body-fixed coordinate frame was a function of coordinates, rather than constants [44, 45]. The distal attachment sites of knee extensors in the present model were attached to the patella instead of using the tibial attachment from the existing model.

5.3.6 Model Simulation

The present model was used to simulate three conditions to assess the reasonableness of the ligament geometries: knee flexion (0 to 120 degrees), knee rotation (-40 to 30 degrees) and knee adduction (-15 to 15 degrees). In each condition, the knee rotations in the other two body planes were set to zero during simulation.

5.4 Results

The orientation of ligaments in the sagittal plane are shown in Figure 5-4. The length changes of each ligament bundle, relative to its zero-load length in knee flexion, knee rotation and knee adduction are shown in Figure 5.5, Figure 5.6 and Figure 5.7, respectively.



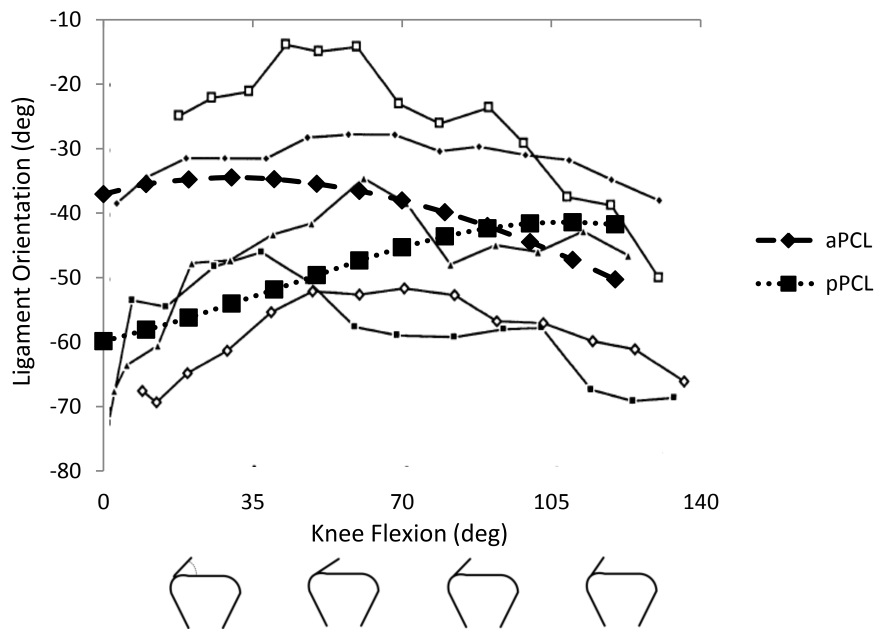
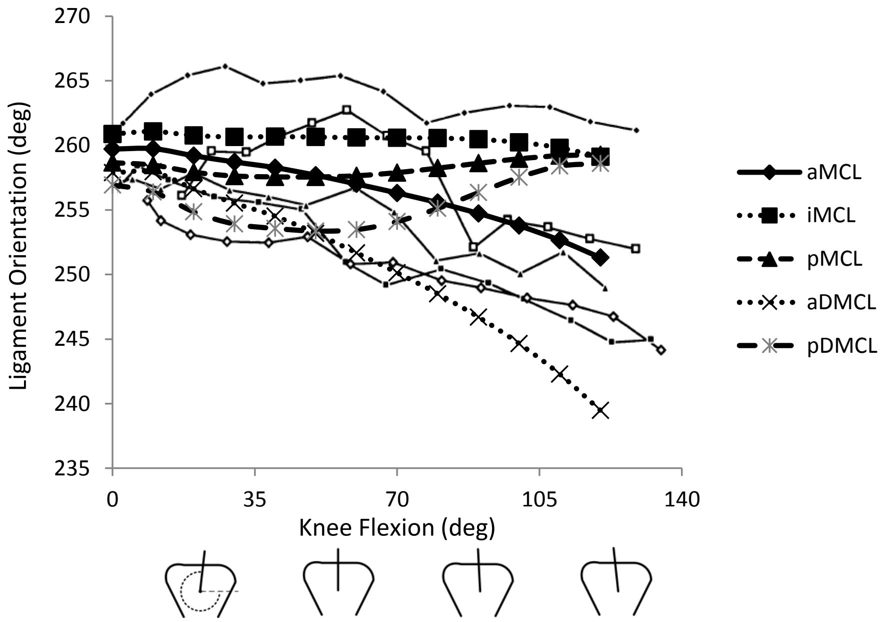


Figure 5.4: Orientation of Knee Ligaments in The Sagittal Plane
The orientation of a ligament bundle was defined as the angle formed between the ligament lines and the tibial plateau in the sagittal plane. Lines with small markers were experimental data from Herzog and Read (1993)



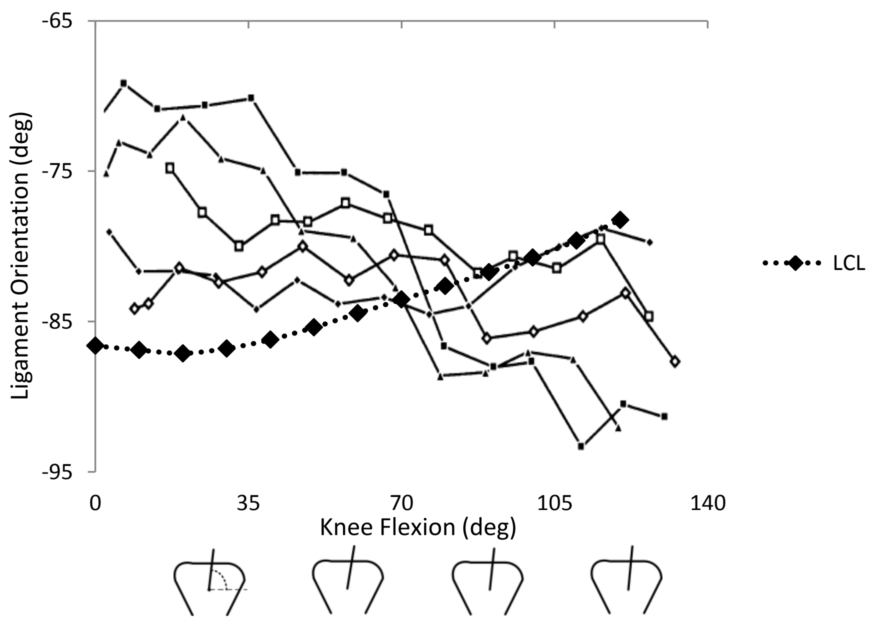
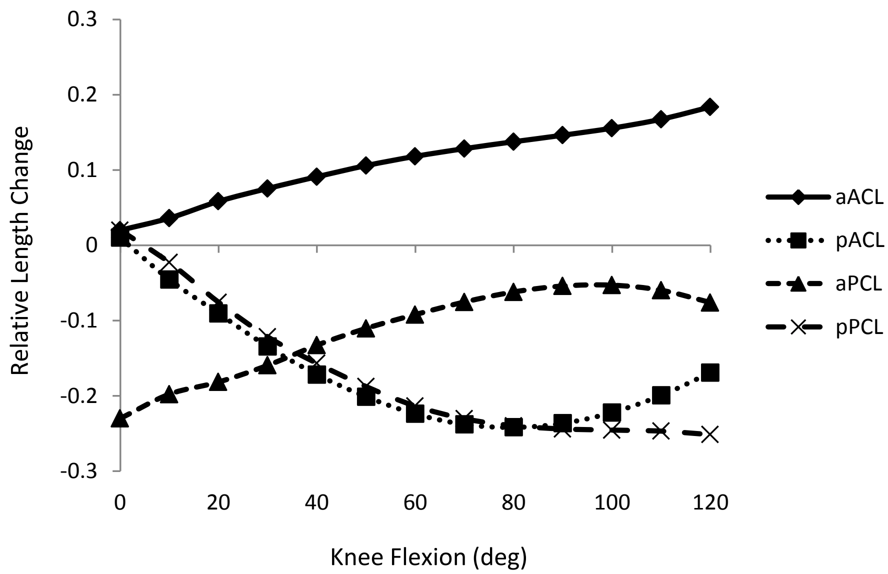


Figure 5.4: Continued



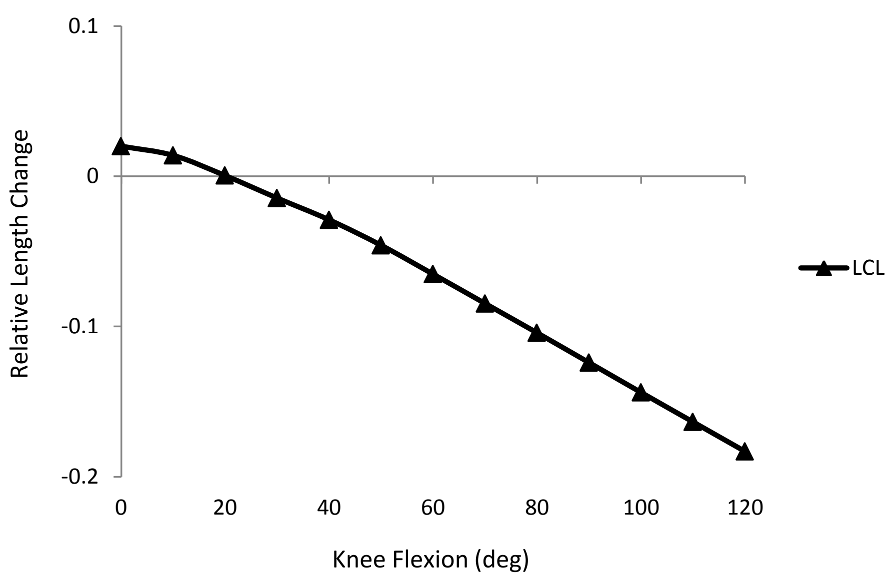


Figure 5.5: Ligament Length Change by Knee Flexion
Compared with the zero-load length (L/L1)

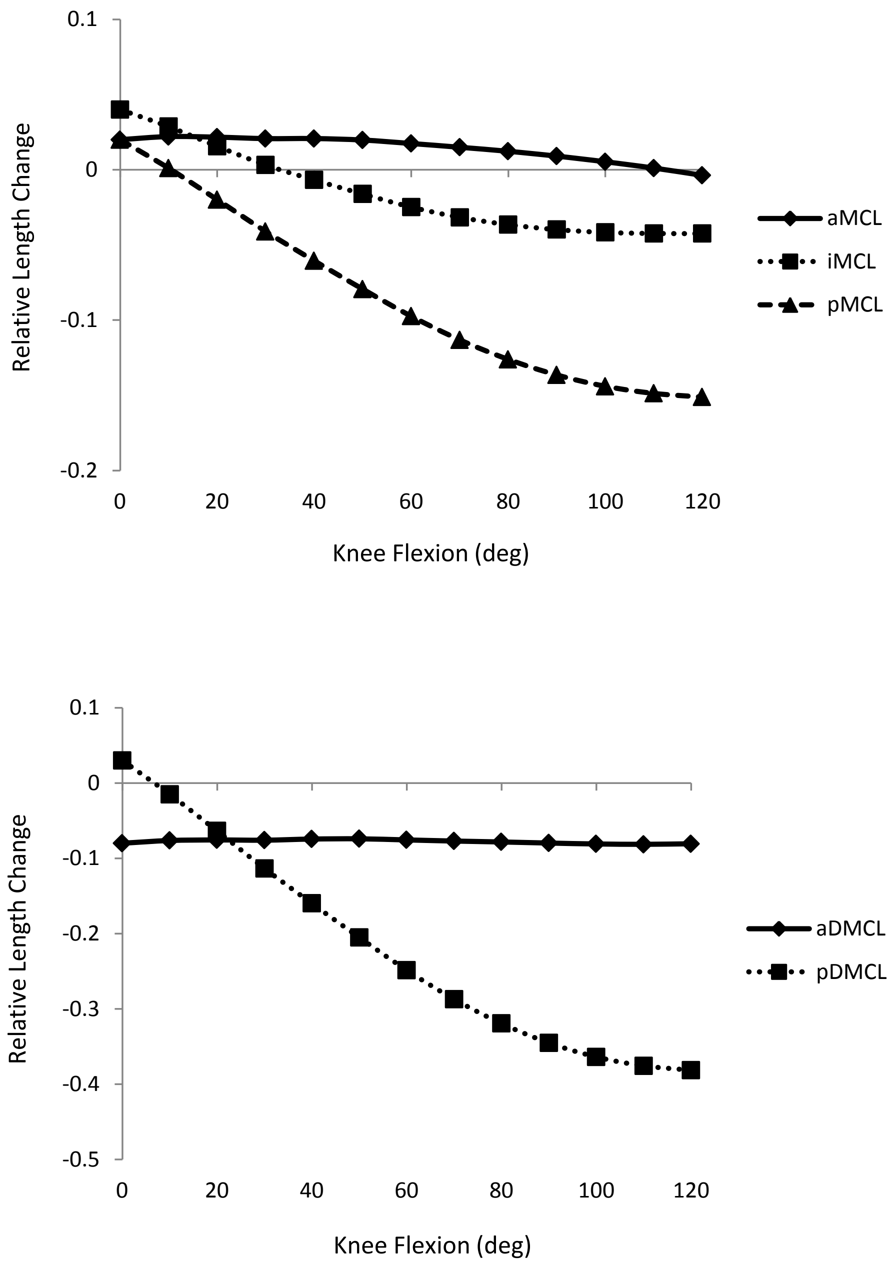
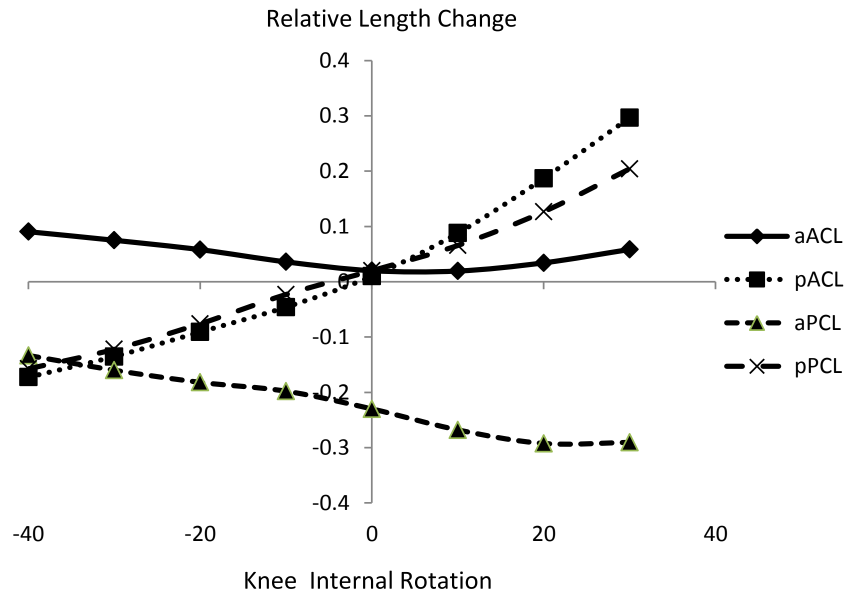


Figure 5.5: Continued



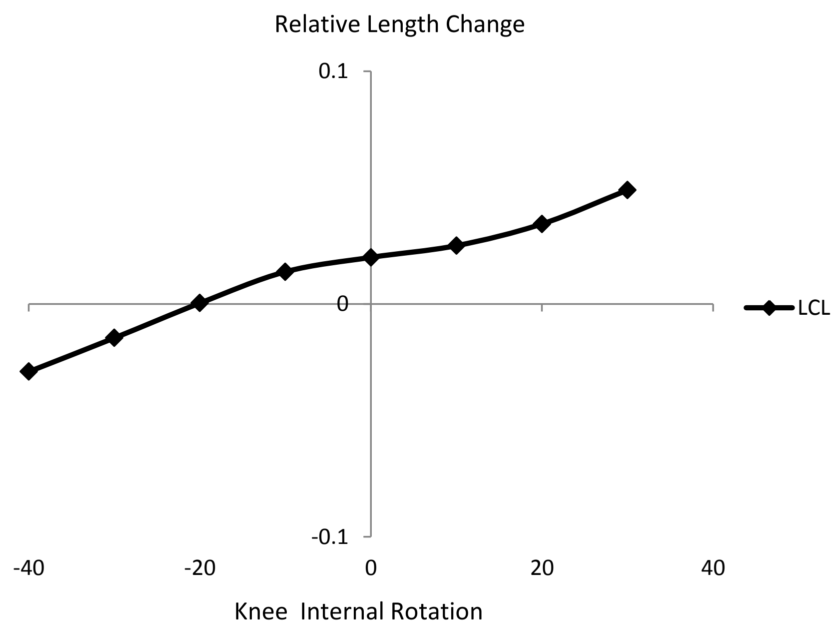
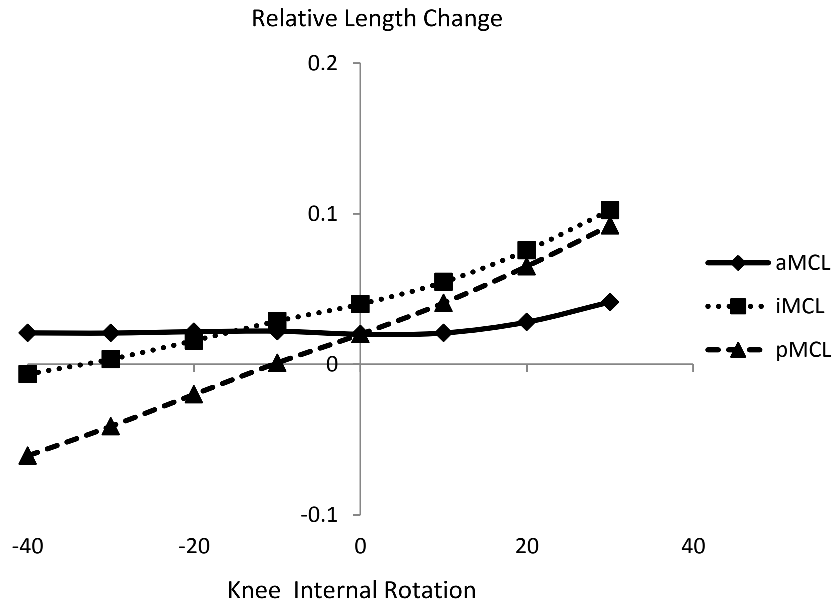


Figure 5.6: The Ligament Length Change by Knee Rotation
Compared with the zero-load length (L/L1)



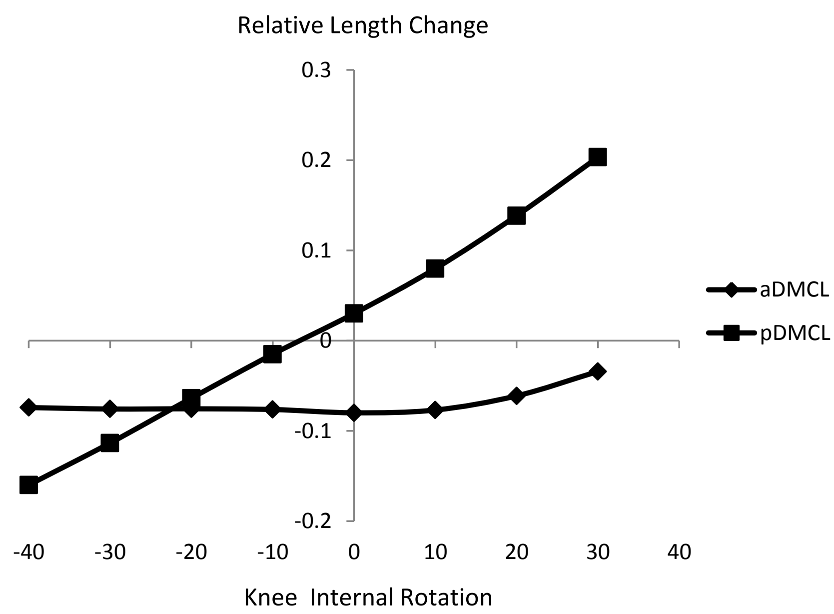
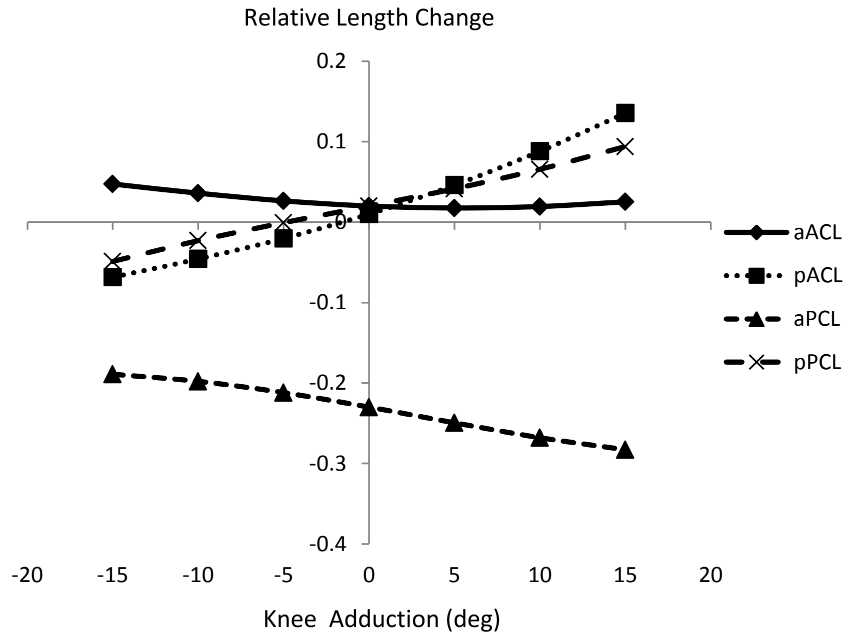


Figure 5.6: Continued



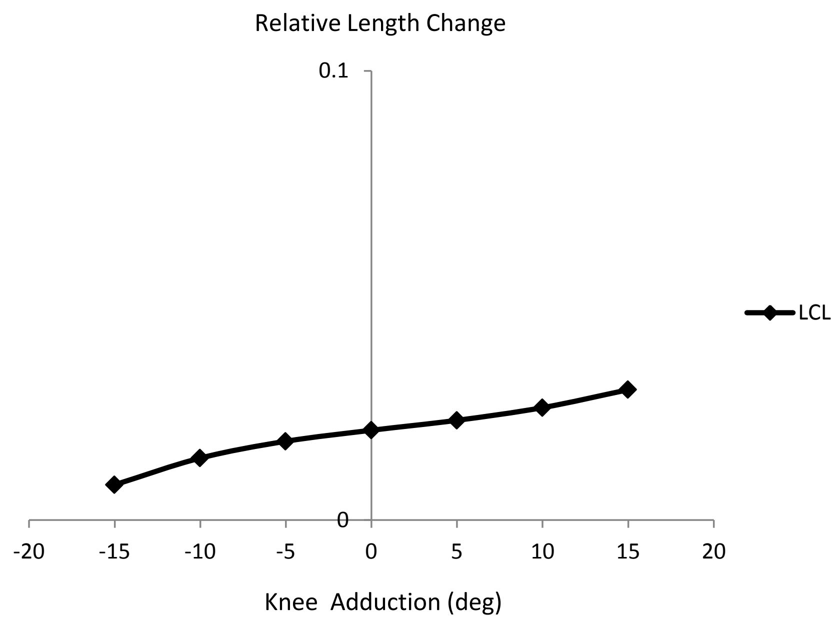
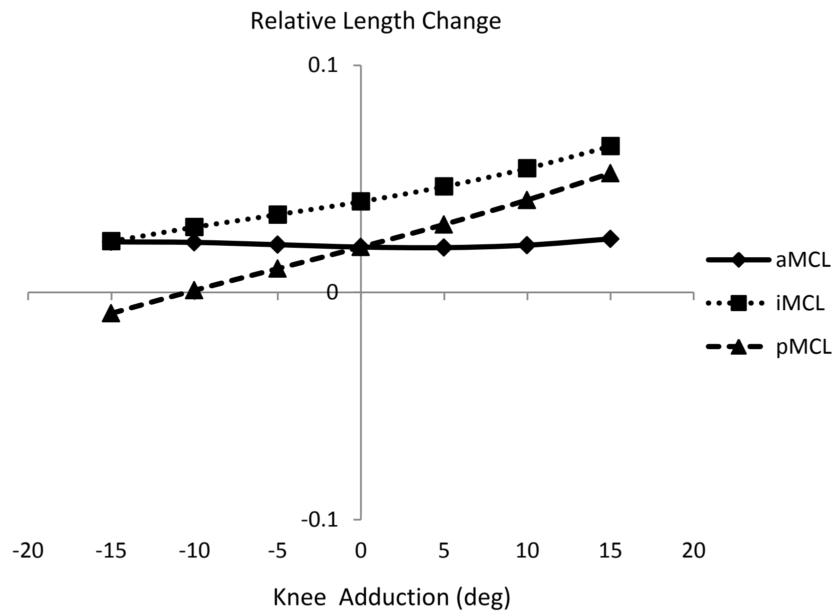


Figure 5.7: The Ligament Length Change by Knee Adduction
Compared with the zero-load length (L/L1)



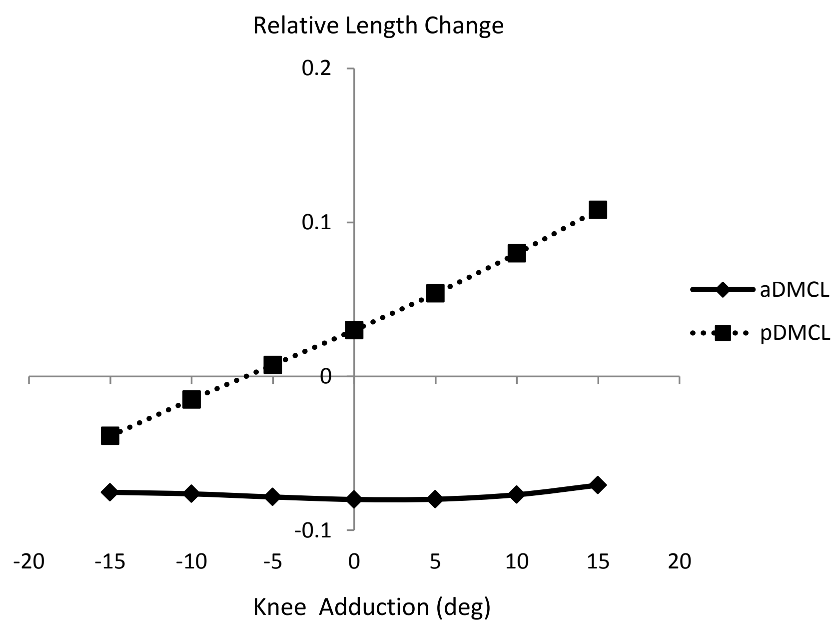


Figure 5.7: Continued

For knee cruciate ligaments, the anterior (aACL and aPCL) bundles and posterior bundles (pACL and pPCL) twisted during knee flexion. However, this intersection happened at approximately 40 degrees of knee flexion for the ACL, but nearly 90 degrees for the PCL. The changes in orientation of the aACL and pPCL were both about 25 degrees for the range of knee flexion. However, the angle decreased for the aACL, but increased for the pACL with increasing knee flexion. The aACL was recruited, but the aPCL was slack for the ranges of knee flexion, rotation and abduction/adduction. The pACL and pPCL were recruited for knee internal rotation and adduction, but not for knee flexion.

The changes in orientation of the LCL were less than 10 degrees throughout the range of knee flexion. The LCL was recruited in all the ranges of knee internal rotation and abduction/adduction, and was slack beyond 20 degrees of knee flexion and external rotation.

The orientations of the three MCL and two DMCL bundles changed only slightly with knee flexion except for the aDMCL. The aMCL was recruited and the aDMCL was slack throughout the range of knee flexion, rotation, and abduction/adduction. The iMCL was recruited in nearly all the range of knee rotation and knee abduction/adduction. The recruitment of the pMCL and pDMCL mostly happened for knee adduction and internal rotation.

5.5 Discussion

The change in the orientation of the aACL was similar (25 degrees) to experimental specimens reported by Herzog and Read (1993) [46] and other knee models reported by Lu and Connor (1996), and Pandy et al. (1997) [38, 47] when the knee flexion angle increased. The difference in the aACL orientation in knee flexion in the present model compared with the experimental specimens and other existing knee models was mostly due to the anatomical variances between study populations. Although the pACL experimental specimen data were not reported by Herzog and Read (1993) [46] due to the inaccessibility of the attachment point, the orientation of pACL was similar to the results from Lu and Connor (1996), and Pandy et al. (1997) [38, 47]. The function of the ACL was as a primary restraint to anterior tibial translation and a secondary restraint to tibial rotation and varus-valgus angulation in the intact knee [48-52]. The anterior tibial translation was about 1cm when knee flexion went from 0 to 120 degrees in a neutral motion pathway in the present model. The recruitment of the aACL throughout the range of knee flexion, rotation and abduction/adduction was in good agreement with the function of the ACL in the intact knee.

All of the PCL and MCL bundles were in the corresponding ranges reported by Herzog and Read (1993) [46]. Though the bundles of PCL were not really recruited in most of the passive knee motions due to the small reference strain (0.23), the tendencies for pPCL bundle recruitment in knee flexion and aPCL recruitment in knee external rotation were consistent with a previous study. It was reported that the PCL was mainly responsible for restraint to posterior tibial translation and that the secondary restraint was tibial external rotation in the intact knee [53].

It was found that the function of the MCL was primarily a restraint to valgus instability. However, contradictory results were found in the literature with respect to MCL function for knee rotation: Warren et al. (1974) and Jasty et al. (1982) reported external restraining function [54, 55], but Seering et al. (1980) and Markolf et al. (1976) reported internal restraining function [56, 57]. The recruitment of the aMCL and iMCL throughout the range of knee rotation supported both results in the previous research. The recruitment of the pMCL and pDMCL in knee internal rotation only indicated internal restraining function.

The orientation of the LCL in the present model was similar to the results from Pandy et al. (1997) [38], but the direction was different from experimental measurements [46] when knee flexion was increased. The LCL was recruited over the whole knee adduction range in this model. This result was in agreement with a previous study results which indicated that the function of the LCL was primarily a restraint to knee varus stress [58].

The gait model developed in the present study was expected to improve the ability to predict muscle forces and KCF. First, the patellar structures, including patella, patellar tendon and patellotibial joint, increased the integrity of the knee joint. This model considered the effect of patella mass and tendon force to improve the anatomical realism of the model. Second, the development of six DOFs knee joint in the present model could 1) improve the ability to restore the actual knee kinematics during gait, 2) facilitate the investigation of differences in knee abduction/adduction and rotation during various kinds of gait, 3) increase the ability to obtain more practical muscle excitation patterns by freeing the flexion constraint which existed in most gait models. Third, the inclusion of knee cruciate ligaments and knee collateral ligaments in the present model may reduce the likelihood that predicted muscle forces include components of passive ligament forces. This may lead to more realistic muscle force values.

The present model did not increase the number of muscles in order to minimize computational cost and increase the likelihood of successful simulation of gait. A hard, level surface gait simulation using another model (Gait2392 model), which included the same structures as Gait2354 model except more muscles (92 muscles), indicated consistent sums of the total knee muscle forces compared with gait simulation using Gait2354 model. This phenomenon suggested that increasing muscle numbers had little impact on calculated KCF, which was in agreement with a previous study [5].

5.5.1 Limitations

There were some limitations associated with the present model. First, the path of each ligament bundle was approximated as a straight line in this model. However, real ligaments attached over a finite area of bone and wrapped around the bones. This approximation affected the length of the ligament bundles and further altered the calculated ligament force. This approximation was accepted in the sagittal plane since the orientation of the model ligaments were similar to the intact knee measurement in passive knee flexion [46]. A ligament sensitivity analysis performed by Blankevoot and Huiskes (1991) indicated that the effect of ligament-bone contact can be neglected during knee flexion [42]. However, the observation that the bundles of the MCL and LCL fell into the tibia during knee rotation and knee abduction/adduction indicated that ligament-bone contact may redirect the knee collateral ligaments for knee rotation in the coronal and transverse planes.

Second, the zero-load length of the ligament bundle was a key parameter to determine whether the ligament bundle was recruited or not. However, the zero-load length was not directly known and had to be indirectly calculated from reference lengths and strains. Some previous research indicated that anatomical differences existed for the insertion locations and the reference lengths of the ligament [36, 59-64]. These differences were much larger than the error from identifying the attachment sites of ligaments. This may explain the variations for length patterns and strain change in different research. The previous sensitivity analysis indicated that knee ligament force was more sensitive to changes in ligament length than ligament stiffness [39, 42, 65]. Pandy and Sasaki (1997) reported that a 5% change in ligament reference length has an equal effect on ligament force as a 50% change in ligament stiffness [39]. Loch et al. (1992) reported that a 5% decrease in the reference length for the ACL would make the ACL force double [65]. The ligament strain can be measured with transducers directly attached to the ligaments in the cadavers [66-68], but data are limited for ligaments in the living body. Therefore, ligament reference strains were usually adapted to get close to the *in vitro* experimental data and were variable from different research [37, 38, 42].

5.5.2 Conclusion

The OpenSim model in the present study was a 3D musculoskeletal model with robust knee structures. There were three main contributions of this model compared with the existing model described in Chapter 3 and Chapter 4. First, the patella, patellar tendon and patellotibial joint were added to the model, including tendon parameters. Second, a three dimensional knee joint was built in this model which included three rotations and three translations. Knee rotations in three body planes and knee mediolateral translation were independent. Knee proximodistal translation and anteroposterior translation were defined as a function of passive knee flexion. Third, knee cruciate ligaments and knee collateral ligaments were involved in this model. This included the geometry and mechanical properties of the ligament. The reasonableness of the ligament geometries was verified by simulating knee motions in three body planes and comparing the results with previous research.

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CHAPTER 6

INVESTIGATION OF KNEE CONTACT FORCE DURING WALKING ON BALLAST

6.1 Abstract

Railroad workers who service trains experience a unique exposure to walking on ballast. The effect of this exposure on workers is still not clear, especially as it relates to mechanical joint loads during walking. Walking on uneven ground is a possible risk factor for knee OA. Published research for walking on ballast principally examines temporal gait parameters and joint kinematics. The aim of this study is to investigate the change of KCF during walking on ballast as surface conditions, surface configuration, and uphill or downhill limb.

Eight railroad workers were selected to represent a healthy population of railroad workers. Three-dimensional motion date was captured using the motion capture system. GRF data was recorded using a single force plate. The new OpenSim model described in Chapter 5 was utilized to simulate walking on ballast. Several biomechanical parameters of interest included temporal gait parameters, the magnitude and timing of peak KCF, muscle cocontraction and ligament forces.

The effect of surface conditions was found to be significant for the gait cycle (p=0.001) and double support (p=0.011). Several statistically significant differences were

found between the uphill and downhill limbs, including stance duration (*p*=0.025), swing duration (*p*=0.036), single support (*p*=0.009) and double support (*p*=0.003). The effect of surface conditions was found to be significant for the timing of the first peak KCF (*p*=0.001) and statistically significant differences were reported among NB, MB and WB (*p*=0.028, *p*=0.024 and *p*=0.001). A significant difference was also found for the timing of the second peak KCF between NB and WB (*p*=0.015). For muscle cocontraction in the first peak KCF, a statistically significant difference was indicated between NB and WB (*p*=0.025), and also between the uphill and downhill limbs (*p*=0.008). For muscle cocontraction in the second peak KCF, the effect of surface conditions was found to be significant (*p*=0.004) and statistically significant differences existed between NB and WB (*p*=0.041), and MB with WB (*p*=0.026). Several statistically significant differences were found between the uphill and downhill limbs, including the aACL (*p*=0.02), LCL (*p*=0.042) and iMCL (*p*=0.017) for the first peak KCF, and the LCL (*p*=0.011) and aMCL (*p*=0.017) for the second peak KCF. The effects of configurations were found to be significant for the aACL (*p*=0.042) and aMCL (*p*=0.022) for the first peak of KCF, and the LCL (*p*=0.017) for the second peak KCF.

Overall, the effects of surface conditions were reported for the gait cycle, the timing of peak KCF and muscle cocontraction. The effects of surface configuration changes were only found in some ligament forces. The effect of uphill and downhill limbs was observed in most of the parameters except for the magnitude and timing of peak KCFs.

6.2 Introduction

Railroad workers experience a unique exposure to walking and performing tasks on ballast. They work in railroad yards or along tracks to make-up trains, inspect cars, and pick up or drop off cars at industrial sites [1, 2]. Two ballast types are defined as WB which is small rocks used for walking, and MB which is large rocks used for tracks [1-3]. According to the Federal Railroad Administration (FRA), walking contributed 13.9% to 16.5% of all railroad worker injuries and accounted for 16.7% to 20.3% of the days absent from work between 1998 and 2006 (FRA, 1999-2008). However, the effects on workers from walking on ballast are still not clear, especially regarding the mechanical joint loads. Given that walking on ballast is a significant part of some railroad workers' jobs and the knee is the weight bearing joint most commonly affected by OA [4], it is imperative to evaluate KCF when walking on ballast as a possible risk factor for knee OA in this population.

A paucity of previous research has focused on kinetics and kinematics characteristic of walking on ballast. Andres et al. (2005) carried out a study to investigated rear foot motion when walking on ballast with five healthy male subjects. They found that walking on MB significantly increased rear foot range of motion, compared to walking on either WB or NB [2], which could cause increased stresses applied to the knee joint when walking on MB since rear foot eversion cause a coupled medial rotation of the tibia [5]. A study performed by Merryweather (2008) focused on lower limb biomechanics when walking on ballast with ten railroad workers. The main findings were that surface configuration had the greatest effect on mediolateral kinetics and knee flexion was greater when walking on ballast than NB [3]. A follow-up study performed by Quincy (2009) further reported that the downhill knee joint had a higher adduction moment compared with uphill knee joint during walking on the sloped configuration n [6]. A recent research, led by Wade et al (2010), examined the impact of ballast on gait biomechanics with 20 healthy adult males. The main findings were that walking on ballast increased muscle cocontraction levels compared with NB, based on EMG, and that the range of joint moments were smaller for MB and WB compared with NB [1].

So far, no research has reported KCF during walking on ballast. The aim of this study was to investigate KCF during walking on ballast by using OpenSim simulation. There were three hypotheses in the present study: first, KCF was significantly altered when walking on ballast compared with NB; second, walking on MB altered KCF response more than walking on WB; third, the downhill knee joint had a higher KCF than the uphill knee joint.

6.3 Methods

6.3.1 Experiment Data

Eight railroad workers from Salt Lake City, Utah were selected to represent a healthy population of railroad workers. The independent variables being controlled for this research were: surface conditions, surface configurations, and uphill or downhill limb. Surface conditions included MB, WB and NB; Surface configurations included a normal level surface and a slanted surface with a 7° slope in the transverse plane. The combinations of surface conditions and configurations were randomized. Each participant performed at least 60 trials (5 trials * 3 surface conditions * 2 surface configurations * 2 feet) to collect acceptable trials for each combination of conditions and configurations. Marker-based motion data were collected by a five camera motion capture system. GRF data was recorded by one force plate for each trial. The details of the experiment design and data collection can be found in Chapter 2 and in previously published research [3].

6.3.2 Temporal Gait Parameters

Major gait events (heel strike and toe off) were visually marked for each foot to identify the gait cycle. At least two complete gait cycles were digitized for each trial. To determine an appropriate trial for OpenSim simulation and statistical analysis, a representative trial instead of grouped averages was chosen from each group of trials for each condition and configuration combination. A total of 96 trials (8 subjects * 3 conditions * 2 configurations * 2 feet) were involved in the present study. The details of representative trial selection can be found in previous [3]. Temporal gait characteristics of interest in this study were the gait cycle (seconds), stance duration (percent cycle), swing duration (percent cycle), single support (percent cycle) and double support (percent cycle).

6.3.3 Gait Simulation

OpenSim was used to simulate the gait trials for each condition and configuration combination. A musculoskeletal model with robust knee structures was used in this study. The model consisted of 28 DOFs, 54 muscle actuators and 10 knee ligament bundles. The motion of the patella was defined as a function of knee flexion. The knee joint (tibiofemoral joint) was represented as a 6 DOF joint. Seven muscles crossing the knee joint were involved in each limb. The details of this model are described in Chapter 5.

In order to control the marker relative movement error and obtain reasonable knee joint kinematics, the knee mediolateral translation was set to zero during gait simulation. The functional curves of knee translation in the sagittal plane following by passive knee flexion from Yamaguchi and Zajac (1989) [7] were input to the model to represent knee proximodistal-flexion and anteroposterior-flexion relationships during walking on ballast and NB. Then, this model was scaled to a subject-specific model using three steps for hard, level surface trials: body segment scale, ligament attachment sites scale and marker location. First, the body segment of the model was scaled by the ratio of relative distances between chosen pairs of markers obtained from the motion capture system and the corresponding virtual marker located in the model [8]. The torso was scaled using the pairs of shoulder and sacrum markers; the pelvis was scaled using the pairs of left and right anterior superior iliac spine (ASIS) markers; the femur was scaled using the pairs of ASIS and knee lateral markers; the tibia was scaled using the pairs of knee lateral and ankle lateral markers; the foot was scaled using the pairs of heel and toe markers. Second, the attachment sites of ligament bundles in the femur and tibia were scaled using the pairs of markers for the femur and tibia, respectively. Third, an instant in time (heel strike) from the dynamic trial was chosen to locate the visual marker positions in the model since no static trials were available. All the lower body joint angles in the model were set to zero in heel strike moment except for hip flexion/extension, which were reported in many previous gait publications [9-13]. The markers, which were used for scaling the model in the first step, were given relatively large weights compared with all other markers. The visual markers in the model were then adjusted to match the corresponding experimental markers to finish the scaling step. Finally, the single support phase, where ground reaction data were available, was simulated by OpenSim to investigate ballast gait.

6.3.4 KCF and Ligament Forces

The joint reaction program in OpenSim was employed to calculate KCF in this study. This program computed the resultant forces that represented the internal loads carried by the joint structures. For this study, the KCF represented the contact force between the tibia and femoral cartilage and did not differentiate this force between medial and lateral compartments of the meniscus. This global KCF was calculated as the vector sum of the knee joint reaction forces, the compressive forces from knee joint muscles, and knee collateral and cruciate ligaments, which are shown in Equation 6-1, Equation 6-2 and Equation 6-3. The magnitude and the timing (percent cycle) for the first and second peak KCF were detected for each trial to verify the hypothesis of this study.

$$\vec{F}_{KCF} = \sum(\vec{F}_{KRF} + \vec{F}_{Muscle} + \vec{F}_{Ligament}) \qquad \text{(Eq. 6-1)}$$

$$\vec{F}_{muscle} = \sum(\vec{F}_{BFLH} + \vec{F}_{BFSH} + \vec{F}_{GRAC} + \vec{F}_{GAS} + \vec{F}_{SAR} + \vec{F}_{RF} + \vec{F}_{VAS} + \vec{F}_{PT}) \qquad \text{(Eq.6-2)}$$

$$\vec{F}_{Ligament} = \sum(\vec{F}_{ACL} + \vec{F}_{PCL} + \vec{F}_{MCL} + \vec{F}_{LCL}) \qquad \text{(Eq.6-3)}$$

The point kinematics program in OpenSim was recruited to track the attachment site of each ligament bundle during gait. Then, the ligament force in each instant time could be calculated by knowing the reference length and reference strain information.

The mechanical properties and relative parameters of the ligaments were described in Chapter 5. Finally, the force generated by each ligament bundle in peak KCFs were determined and used for examining the effect of three variables and their interactions.

6.3.5 Muscle Cocontraction

Muscle cocontraction was usually used to describe the simultaneous activity of various muscles acting around a joint. In the present study, knee muscle cocontraction was determined in the form of cocontraction index (CCI), which is shown in Equation 6-4 [14].

$$CCI = \frac{\sum F_{Total}^{M}}{\sum F_{Agonists}^{M}} - 1 \quad \text{(Eq. 6-4)}$$

where $\sum F_{Total}^{M}$ represented the total muscle force acting at the knee joint, $\sum F_{Agonist}^{M}$ represented the total muscle force of the agonist muscle groups in knee joint.

6.3.6 Model Sensitivity Analysis

Sixteen NB with the level configuration trials (two trials for each subject) were chosen for model sensitivity analysis. The mediolateral shear GRF and total GRF were increased and decreased by 10%, 20% and 50% separately to investigate KCF sensitivity for the changes in GRF.

6.3.7 Residual Forces and Moments in Peak KCF

The residual forces and moments for the first and second peak KCFs were evaluated for the RRA and CMC steps in OpenSim for 96 trials and root mean square residual values from these trials were calculated. The residual, general forces in the RRA step were used to evaluate the magnitude of dynamic inconsistency between ground reaction data and acceleration from measured marker kinematics. The residual, general force in CMC step was used to evaluate the robustness of the model when simulating ballast gait.

6.3.8 Statistical Analysis

Repeated analysis of variance was used for determining the effect of surface conditions, surface configurations, limbs and their interactions. Paired t-tests were used for comparison between uphill and downhill limbs. Results were considered statistically significant when $p<0.05$ ($\alpha=0.05$). If the assumption of sphericity was violated, the Greenhouse-Geisser correction was used. Post hoc tests were performed using the Bonferroni adjustment to correct for multiple comparisons. Observed power was also computed in this study. All the statistics were performed using SPSS (IBM Corporation, Armonk, NY).

6.4 Results

6.4.1 Temporal Gait Parameters

The effect of surface conditions was found to be significant for the gait cycle ($p=0.001$) and double support ($p=0.011$). Statistically significant differences were found for the gait cycle between walking on ballast and NB (*p*=0.017 and *p*=0.001). Several statistically significant differences were shown between the uphill and downhill limbs, including stance duration (*p*=0.025), swing duration (*p*=0.036), single support (*p*=0.009) and double support (*p*=0.003). The effect of surface configuration and four interactions were not statistically significant for temporal gait parameters. These results are shown in Table 6.1.

6.4.2 Magnitude and Timing of Peak KCF

The effect of surface conditions was significant for the timing of the first peak KCF (*p*=0.001) and statistically significant differences were reported among NB, MB and WB (*p*=0.028, *p*=0.024 and *p*=0.001). No significant effect of surface conditions was found for the timing of the second peak KCF; however, Bonferroni-adjusted pairwise comparison indicated a statistical difference between NB and WB (*p*=0.015) for the timing of the second peak KCF. No variables and interactions were found to have significant effects for the first and the second peak KCFs which are shown in Table 6.2.

6.4.3 Cocontraction Index in Peak KCF

In the first peak KCF, statistically significant knee muscle cocontraction differences were observed for walking on WB compared with NB (*p*=0.025), and also between the uphill and downhill limbs (*p*=0.008). The effect of surface conditions was significant (*p*=0.004) for knee muscle cocontraction in the second peak KCF. Statistically significant differences existed when walking on WB compared with NB (*p*=0.041) and MB (*p*=0.026). These results are shown in Figure 6.1, Figure 6.2 and Figure 6.3.

Table 6.1 Mean Temporal Gait Parameters

Variable		Gait Cycle (second)	Stance Duration (Gait Cycle)	Swing Duration (Gait Cycle)	Single Support (Gait Cycle)	Double Support (Gait Cycle)
Condition	no ballast	1.22 (0.03)	0.63 (0.01)	0.37 (0.01)	0.38 (0.02)	0.25 (0.02)
	main ballast	1.26 (0.04)*	0.62 (0.03)	0.38 (0.02)	0.38 (0.01)	0.24 (0.03)
	walking ballast	1.27 (0.04)*	0.63 (0.01)	0.37 (0.01)	0.37 (0.02)	0.26 (0.02)
Configuration	level	1.24 (0.04)	0.63 (0.01)	0.37 (0.01)	0.38 (0.01)	0.25 (0.02)
	slope	1.26 (0.03)	0.62 (0.02)	0.38 (0.02)	0.38 (0.01)	0.25 (0.02)
Leg	level left leg	1.25 (0.03)	0.63 (0.01)	0.37 (0.01)	0.37 (0.02)	0.26 (0.02)**
	level right leg	1.24 (0.04)	0.62 (0.01)	0.38 (0.01)	0.38 (0.01)	0.24 (0.02)
	slope left leg (downhill)	1.26 (0.03)	0.64 (0.01)**	0.37 (0.01)**	0.37 (0.01)**	0.26 (0.02)**
	slope right leg (uphill)	1.25 (0.03)	0.61 (0.03)	0.39 (0.03)	0.39 (0.01)	0.22 (0.03)

Standard deviations were shown in parentheses

* indicated a significant difference from no ballast in condition comparison

* * indicated a significant difference from the other leg in the same configuration

Table 6.2 Mean Peak KCF and Its Timing

Variable		1st Peak (Body Weight)	1st Peak Time (Gait Cycle)	2nd Peak (Body Weight)	2nd Peak Time (Gait Cycle)
Condition	no ballast	4.57 (0.52)	0.15 (0.02)**	6.41 (0.93)	0.46 (0.01)
	main ballast	4.63 (0.77)	0.13 (0.01)**	6.09 (0.97)	0.45 (0.03)
	walking ballast	4.51 (0.66)	0.13 (0.02)**	6.23 (0.75)	0.44 (0.03)*
Configuration	level	4.53 (0.60)	0.14 (0.01)	6.19 (0.79)	0.45 (0.02)
	slope	4.61 (0.61)	0.14 (0.01)	6.29 (0.81)	0.45 (0.02)
Leg	level left leg	4.53 (0.70)	0.13 (0.02)	6.14 (0.79)	0.45 (0.03)
	level right leg	4.53 (0.61)	0.14 (0.02)	6.23 (0.99)	0.45 (0.03)
	slope left leg (downhill)	4.52 (0.67)	0.14 (0.02)	6.33 (0.82)	0.45 (0.02)
	slope right leg (uphill)	4.71 (0.73)	0.14 (0.01)	6.25 (0.96)	0.45 (0.02)

Standard deviations were shown in parentheses
* indicated a significant difference from no ballast in condition comparison
** indicated a significant difference from other two conditions

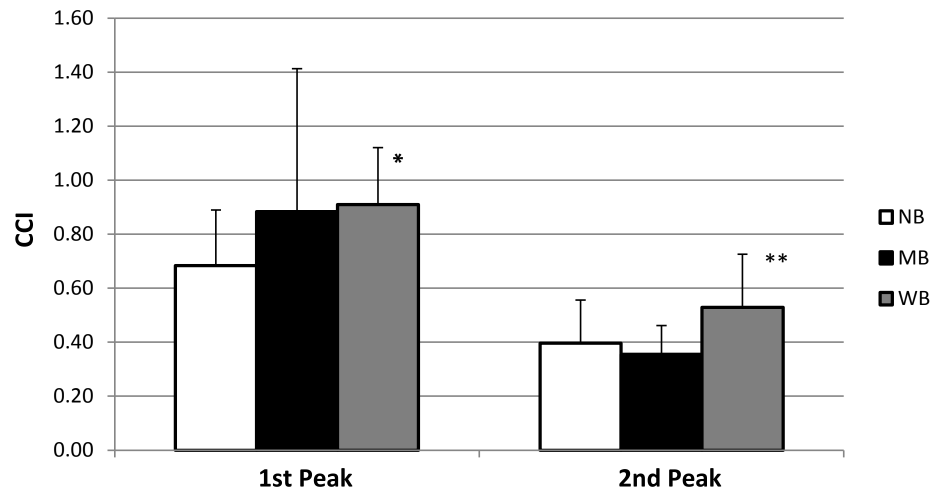

Figure 6.1: CCI by Surface Condition in Peak KCFs
* indicated a significant difference from NB
** indicated a significant difference from other two surface conditions

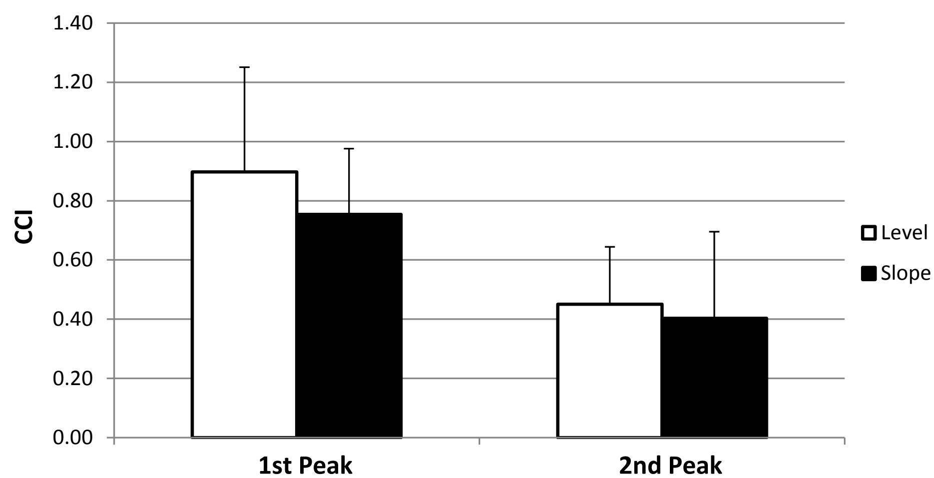

Figure 6.2: CCI by Surface Configuration in Peak KCFs

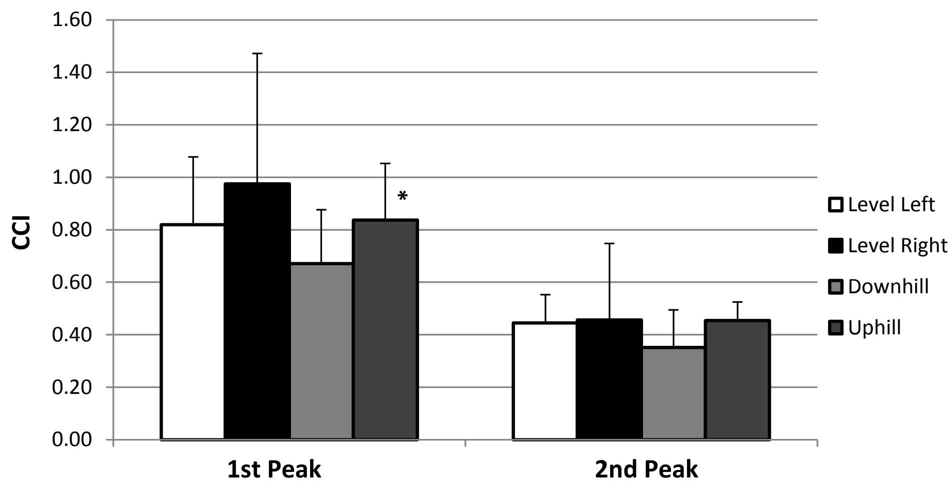

Figure 6.3: CCI by Limb in Peak KCFs
* indicated a significant difference with the other foot in the same surface configuration

6.4.4 Ligament Forces in Peak KCF

Several statistically significant differences were found between the uphill and downhill limbs, including the aACL (p=0.02), LCL (p=0.042) and iMCL (p=0.017) in the first peak KCF, and the LCL (p=0.011) and aMCL (p=0.017) in the second peak KCF. The configuration by limb interaction effect was statistically significant for the aACL (p=0.001), aMCL (p=0.01) and iMCL (p=0.016) in the first peak KCF, and the LCL (p=0.006) and aMCL (p=0.03) in the second peak KCF. No variables were found to have significantly effect for the pACL, PCL, pMCL and DMCL for either KCFs peak. These results are shown in Figure 6.4 to Figure 6.15. The observed powers for all parameters are shown in Table 6.3.

6.4.5 Model Sensitivity

KCF responded well to the change in GRF for the first peak KCF, but lagged in response to changes in GRF for the second peak KCF. The effect of changing

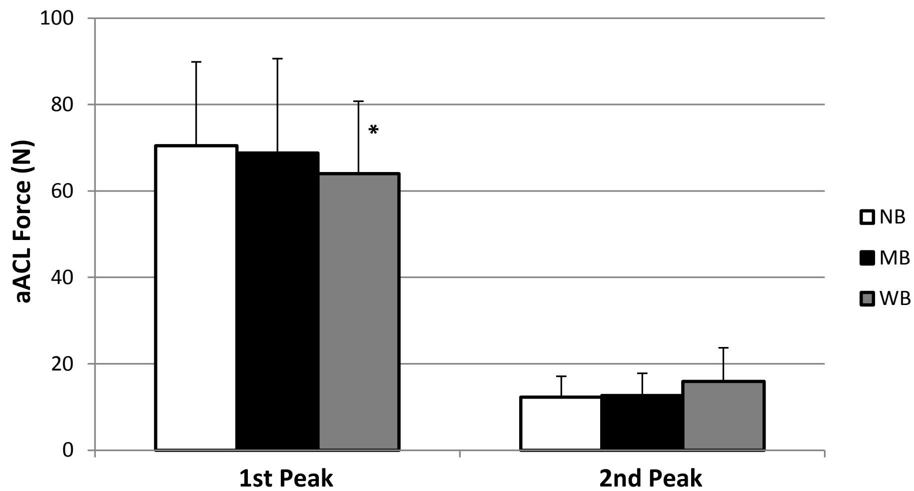

Figure 6.4: aACL by Surface Condition in Peak KCFs
* indicated a significant difference from NB

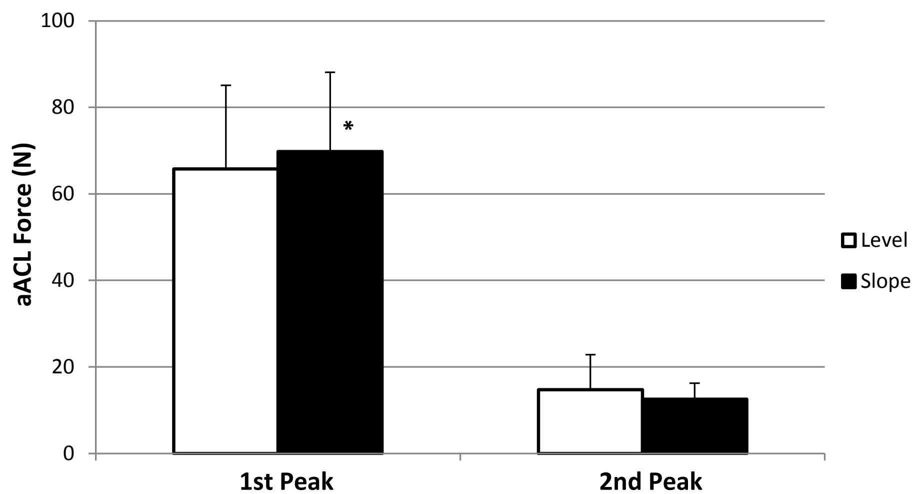

Figure 6.5: aACL by Surface Configuration in Peak KCFs
* indicated a significant difference from the level configuration

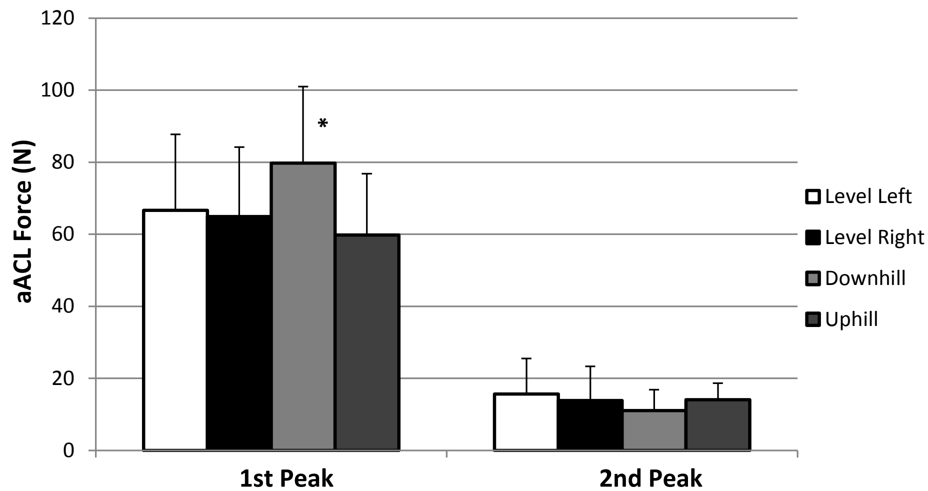

Figure 6.6: aACL by Limb in Peak KCFs
* indicated a significant difference with the other foot in the same surface configuration

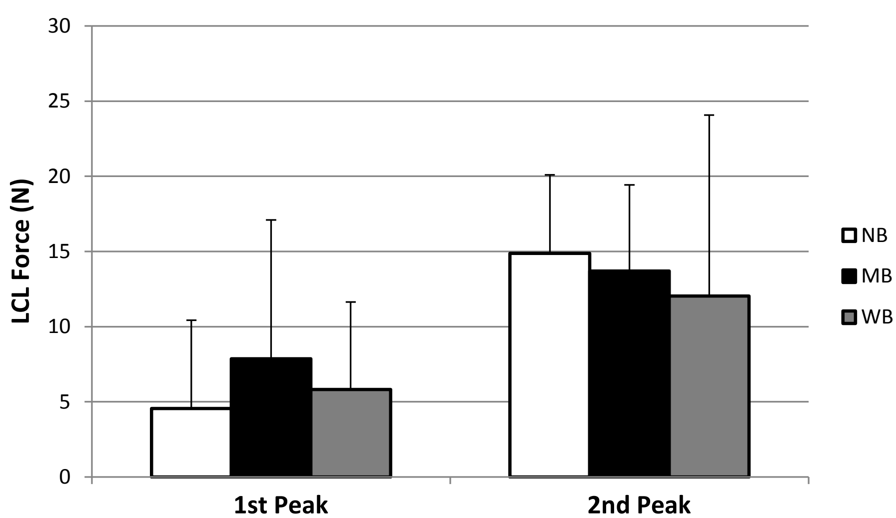

Figure 6.7: LCL by Surface Condition in Peak KCFs

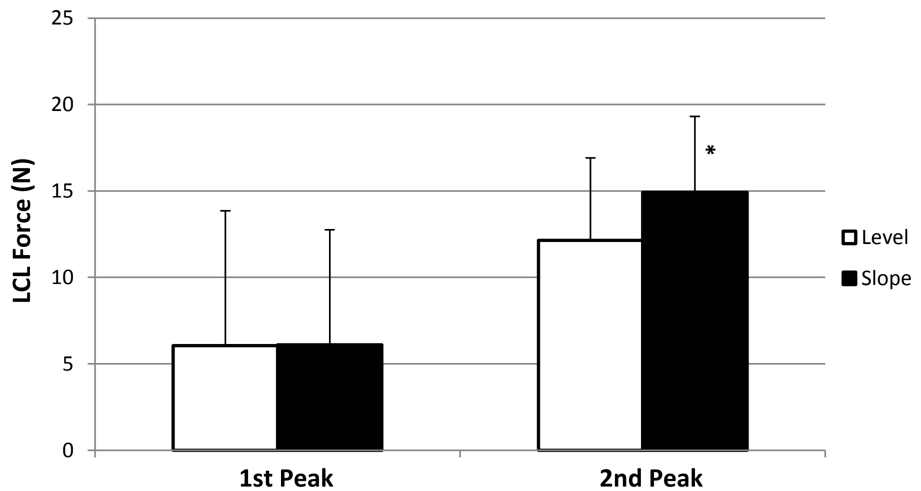

Figure 6.8: LCL by Surface Configuration in Peak KCFs
* indicated a significant difference from the level configuration

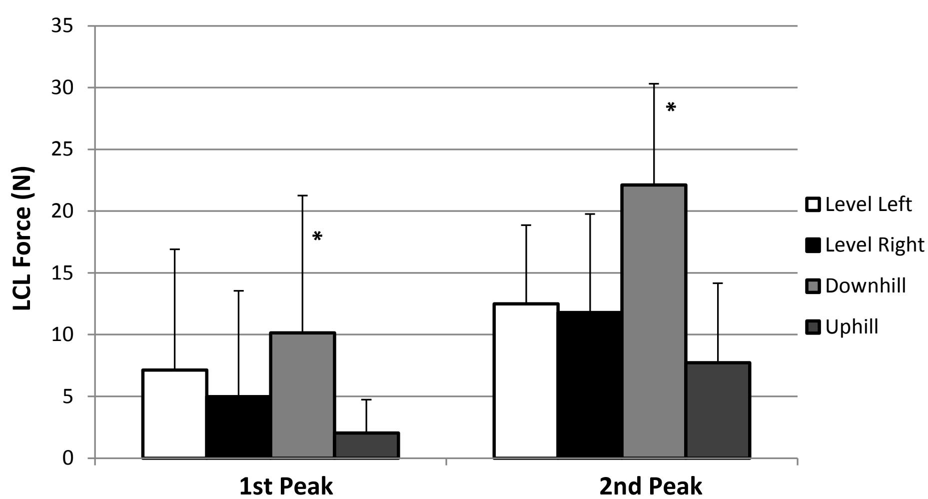

Figure 6.9: LCL by Limb in Peak KCFs
* indicated a significant difference with the other foot in the same surface configuration

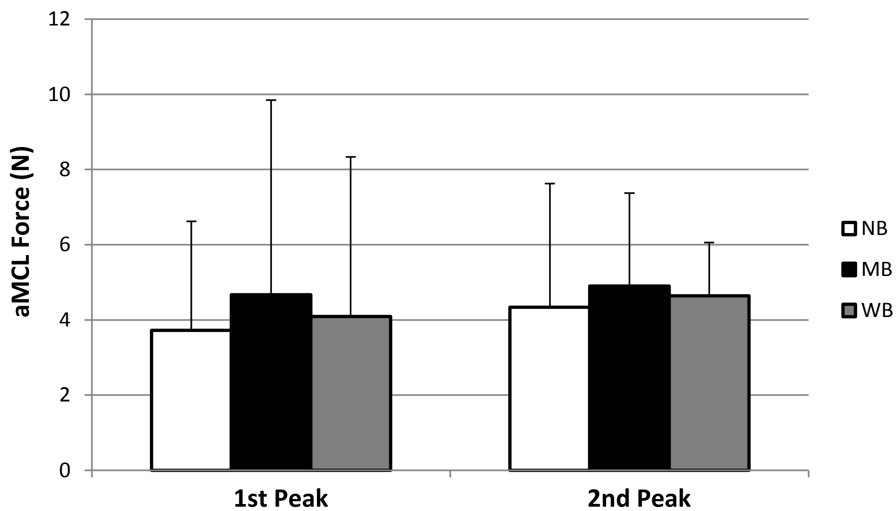

Figure 6.10: aMCL by Surface Condition in Peak KCFs

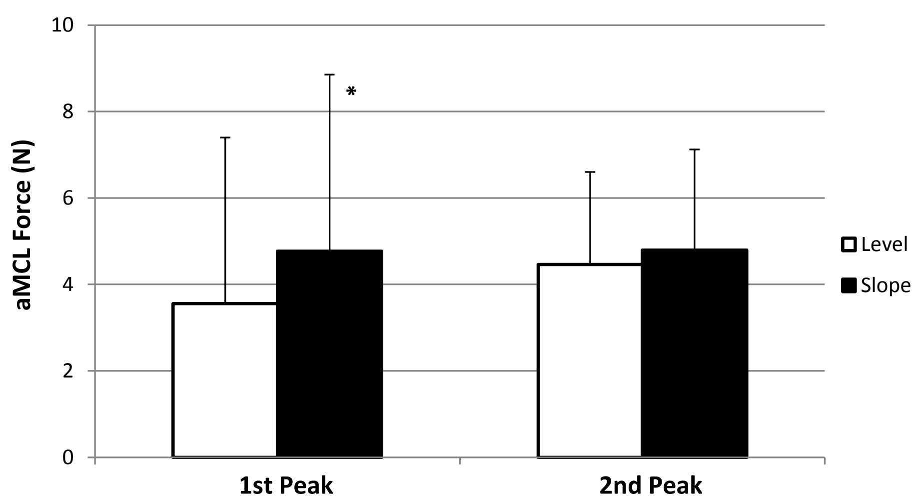

Figure 6.11: aMCL by Surface Configuration in Peak KCFs
* indicated a significant difference from the level configuration

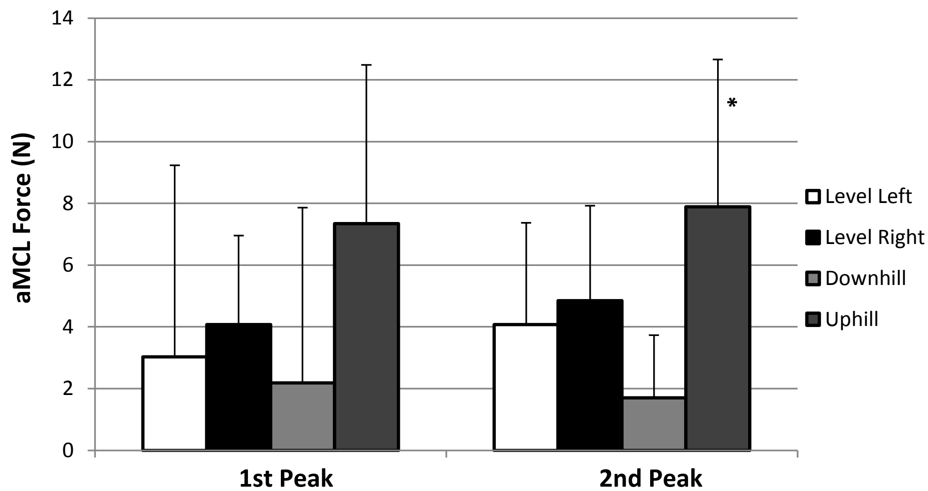

Figure 6.12: aMCL by Limb in Peak KCFs
* indicated a significant difference with the other foot in the same surface configuration

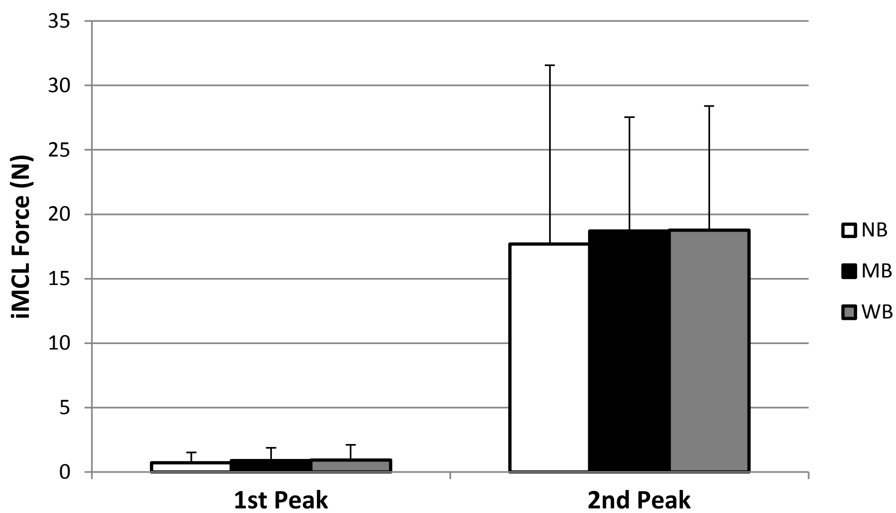

Figure 6.13: iMCL by Surface Condition in Peak KCFs

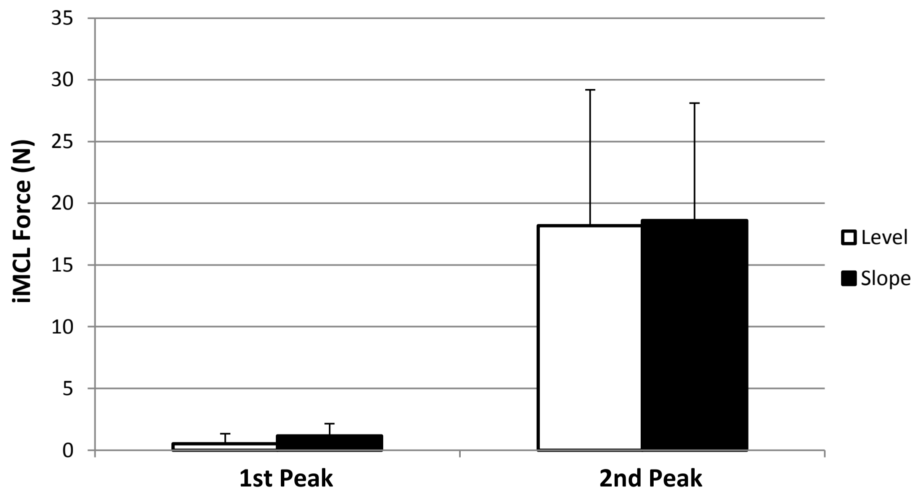

Figure 6.14: iMCL by Surface Configuration in Peak KCFs

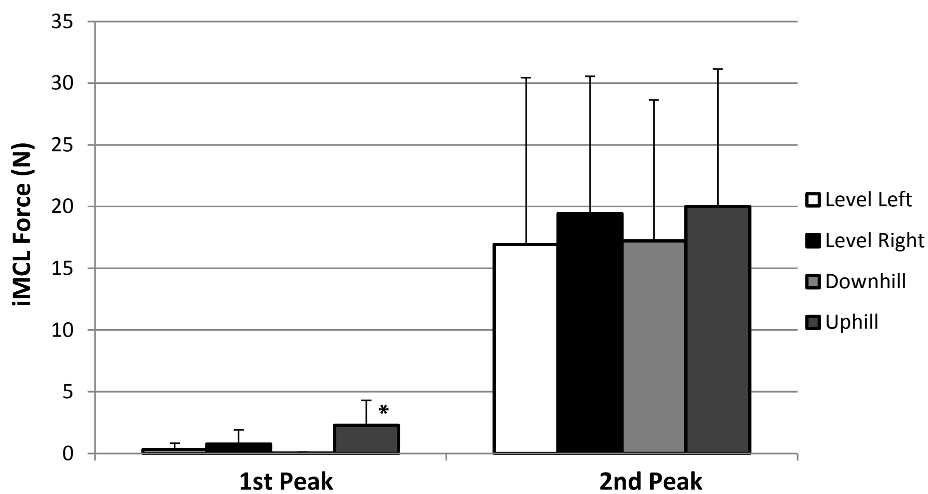

Figure 6.15: iMCL by Limb in Peak KCFs
* indicated a significant difference with the other foot in the same surface configuration

Table 6.3 Observed Power for All Parameters

	Condition	Configuration	Leg	Con*Conf	Con*Leg	Conf*Leg	Con*Conf*Leg
Gait Cycle	**0.999**	0.402	0.100	0.074	0.290	0.057	0.097
Stance Duration	0.263	0.116	**0.726**	0.172	0.223	0.466	0.147
Swing Duration	0.236	0.252	**0.813**	0.193	0.175	0.244	0.216
Single Support	0.409	0.250	**0.913**	0.055	0.231	0.051	0.153
Double Support	0.212	0.332	0.095	0.187	0.082	0.256	0.256
Timing of First Peak KCF	**1.000**	0.075	0.310	0.073	0.085	0.050	0.087
Timing of Second Peak KCF	0.406	0.055	0.069	0.137	0.335	0.068	0.064
First Peak KCF	0.216	0.290	**0.507**	0.116	0.184	0.050	0.186
Second Peak KCF	**0.925**	0.100	0.153	0.053	0.089	0.165	0.105
CCI_First Peak	0.495	0.062	0.402	0.068	0.069	0.238	0.188
CCI_Second Peak	0.168	0.050	**0.975**	0.053	0.404	0.051	0.188
aACL_First Peak	0.368	**0.572**	**0.675**	0.083	0.067	**0.999**	0.149
LCL_First Peak	0.423	0.050	0.317	0.071	0.584	0.396	**0.633**
aMCL_First Peak	0.099	**0.710**	0.227	0.198	0.236	**0.856**	0.312
iMCL_First Peak	0.074	**0.506**	**0.706**	0.065	0.077	**0.767**	0.054
aACL_Second Peak	**0.627**	0.154	0.054	0.388	0.050	0.242	0.098
pACL_Second Peak	0.095	0.353	0.438	0.142	0.080	0.451	0.098
pPCL_Second Peak	0.069	0.184	0.157	0.077	0.082	0.355	0.084
LCL_Second Peak	0.325	0.762	0.439	**0.606**	0.061	**0.917**	0.107
aMCL_Second Peak	0.076	0.076	0.408	0.052	0.072	**0.968**	0.138
iMCL_Second Peak	0.062	0.056	0.095	0.135	0.089	0.051	0.183
pMCL_Second Peak	0.058	0.121	0.108	0.200	0.135	0.301	0.136
pDMCL_Second Peak	0.061	0.105	0.116	0.343	0.136	0.320	0.134

Bold black indicated high and medium power mediolateral shear GRF was not significant for the peak KCFs. The maximum peak KCF change was less than 5% in the range of 50% change of mediolateral shear GRF. These results are shown in Table 6.4.

6.4.6 Residual Forces and Moments in Peak KCF

The largest residual forces in the RRA were shown in the vertical direction for both peak KCFs (13 N and 22 N). The residual moment for the RRA was small enough to be neglected. For the CMC step, the largest residual forces (75 N and 77 N) and moments (29 Nm and 69 Nm) were shown in the anteroposterior and mediolateral directions. For both peak KCFs. The residual force and moment results are shown in Table 6.5.

6.5 Discussion

6.5.1 Temporal Gait Parameters

No temporal gait parameters were significantly different for surface conditions and surface configurations except for the gait cycle in surface conditions. This indicates the similarity of gait. The gait cycle was significantly longer for the ballast conditions compared with NB suggesting a slower speed and a more cautious gait on the ballast due to the less stable surface. It was found that stance duration and double support phase were longer for the downhill limb than the uphill limb and that swing duration and single support phase were shorter for the downhill limb than the uphill limb. This finding indicated that the downhill limb may respond more to maintain the body balance and control for the gravity center of body than the uphill limb in the sloped configuration. Wade et al. (2010) reported that several temporal gait parameters were significantly

Table 6.4:The Change of Peak KCF by GRF

			First Peak	Second Peak
GRF	Increase	10%	9%	-1%
		20%	18%	10%
		50%	44%	24%
	Decrease	10%	-9%	3%
		20%	-16%	-25%
		50%	-39%	-52%
Mediolateral GRF	Increase	10%	0%	1%
		20%	1%	0%
		50%	3%	1%
	Decrease	10%	-1%	0%
		20%	-2%	-1%
		50%	-4%	-3%

Table 6.5: Residual Forces and Moments for RRA and CMC

			First Peak KCF	Second Peak KCF
RRA	Residual Force	Fx	2 N	7 N
		Fy	13 N	22 N
		Fz	3 N	18 N
	Residual Moment	Mx	0 Nm	0 Nm
		My	0 Nm	0 Nm
		Mz	0 Nm	0 Nm
CMC	Residual Force	Fx	75 N	77 N
		Fy	14 N	27 N
		Fz	7 N	32 N
	Residual Moment	Mx	24 Nm	26 Nm
		My	10 Nm	10 Nm
		Mz	29 Nm	69 Nm

different among walking on NB, MB and WB in the level configuration, including stance duration, swing duration, single support and double support [1]. These results were not observed in this study.

6.5.2 Peak Knee Contact Force

The first hypothesis of the present study was that KCF was significantly altered when walking on ballast compared with NB. This was not confirmed for either KCF peak. No statistically significant differences were found for either KCF peak when walking on ballast compared with NB. A case study performed by Kim et al. (2009) reported that the two peak KCFs both decreased followed by the reduction of walking speed [15]. However, a recent study, led by Richards and Higginson (2010) had a conflicting result that the reduction of walking speed only influenced the second peak KCF but not the first peak KCF [16]. Although no significant differences in second peak KCF were observed among surface conditions in the present study, a trend was observed that the second peak KCF decreased when walking on ballast compared with NB. Since gait cycles were significantly longer when walking on ballast compared with NB indicating a slow speed on ballast, the results in the present study were in agreement with the research performed by Richards and Higginson [16] .

The second hypothesis of the present study was that walking on MB altered KCF response more than walking on WB. This was not confirmed for either the first or the second peak KCF. The previous research performed by Wade et al. (2010) reported walking speed was significantly slower when walking on MB compared with WB [1]. Although no significant difference in walking speed was found in a previous study performed by Merryweather [3], a trend was observed that walking speed decreased when walking on MB compared with WB. Therefore, the possibility that this hypothesis was confirmed still exists.

The third hypothesis of the present study was that the downhill knee joint had a higher KCF than the uphill knee joint. This was not confirmed for either KCF peak. It was suggested that the mediolateral GRF increased laterally for the uphill limb and medially for the downhill limb to oppose the additional shear force acting down the slope [3], and the knee adduction moment was significantly greater for the downhill limb than the uphill limb [6]. These results indicated the possibility of increasing medial KCF for the downhill limb; however, the effect of the change in mediolateral GRF was still not clear for lateral KCF. The similar KCF presented in both downhill and uphill limbs indicates a symmetric compensatory strategy on both limbs for the sloped configuration.

The timing of both peak KCFs in the gait cycle was also investigated in the present study. The timing of the first peak KCF was significantly different when walking on three surface conditions. The timing difference was also found for the second peak KCF when walking on WB compared with NB. Surface configuration and limb had little effect on the timing of peak KCF in the gait cycle.

6.5.3 Cocontraction Index

The knee muscle cocontraction was found to be significantly higher when walking on WB compared with NB and MB in both peak KCFs, except for MB in the first peak KCF. These results were partly consistent with previous research led by Wade et al. (2010), which reported that vastus medialis-medial hamstring cocontraction was different among three surface conditions by using EMG measurement [1]. Muscle cocontraction was considered to be amplified in several situations. First, high muscle cocontraction was recruited to perform the activities which demanded high relative to their capability[17], and indicated the inefficiency of human movement[18]. Second, muscle cocontraction was shown to have important functions for a systematic distribution of compression forces across the articular surface [19]. In the present study, muscle cocontraction indicated that walking on WB resulted in a more cautious gait compared with walking on MB and NB. The other main finding for knee muscle cocontraction was that CCI was significantly higher for the uphill limb than the downhill limb in the sloped configuration. This trend was statistically significant for the first peak KCF, and trending towards statistical significance (p=0.055) for the second peak KCF. A significantly larger average value of knee flexion angle was found for the uphill limb than the downhill limb in previous research [6]. Therefore, higher CCI for the uphill limb indicated that more work was required for the uphill limb than the downhill limb to elevate the body and to prevent the toe from colliding with the ground.

6.5.4 Ligament Forces

Statistically significant differences in ligament forces existed between the uphill and downhill limbs. The main findings for ligament forces in both peak KCFs were that the LCL force was larger for the downhill limb than the uphill limb and MCL force was smaller for the downhill limb than the uphill limb. The previous research suggested that knee adduction moment, medial GRF and medial knee reaction force were larger for the downhill limb than the uphill limb [3, 6]. These previous findings can well explain the results in this study, which indicated that the LCL in the downhill limb and MCL in the uphill limb need to generate more force to restrain knee varus stress and knee valgus instability separately in the corresponding limbs. Significantly larger aACL force was found for the downhill limb than the uphill limb in this study, which can be explained by the larger adduction angle for the downhill limb [6].

Surface configuration was statistically significant for the aACL and aMCL in the first peak KCF, and the LCL in the second peak KCF. The significantly larger aACL, MCL and LCL force in the sloped configuration could be caused by the significantly larger knee flexion angle and knee adduction moment in the sloped configuration compared with the level configuration [6].

6.5.5 Model Sensitivity

Overall, the relative change of peak KCF responded well to the change in total GRF, but not for the mediolateral shear GRF. Although significantly different mediolateral GRF was found when walking on the level and the sloped configuration [3], this difference did not significantly alter peak KCFs, which were primarily determined by the muscles crossing the knee joint.

6.5.6 Residual Forces and Moments in Peak KCF

The residual values balanced the dynamic inconsistency between the ground reaction data and the acceleration estimated from measured marker kinematics due to the modeling assumptions, noise and other errors from motion capture process in RRA, and controlled the global position and orientation of the model in CMC. For the present study, the residual values in RRA and CMC indicated the agreement between the musculoskeletal model for simulation and the recorded ballast gait data. The residual moments and forces in RRA and CMC were in a high acceptance level (below 30 Nm) [20] and in a medium acceptance level (below 25 N) [20] for the first peak KCFs respectively when using full-body simulations of walking, except for anteroposterior residual force in CMC. For the second peak KCF, the residual values in RRA and residual moments in CMC were in a medium acceptance level (below 25 N and 30 Nm) [20], but the residual forces in CMC were in a low acceptance level (above 25 N) [20] when using full-body simulations of walking. These values indicate a greater level of confidence in the first peak KCF compared with the second peak when using this model for ballast gait simulation.

6.5.7 Comparison KCF with Previous Studies

Some instrumented tibiofemoral implant studies provided researchers with valuable opportunities for validation of predicted KCF using musculoskeletal models. They reported peak KCF ranging from 1.8 to 3.0 BW during overground gait and treadmill gait [21-30], which are shown in Table 6.6. On the other hand, most previous musculoskeletal modeling studies overestimate the peak KCF during gait, ranging from 1.8 to 8.1 BW [31-41]. In the present study, the average peak KCFs were 4.57 BW and 6.24 BW for the first and second peaks respectively. These were two or three times higher than the in vivo measurements, but in the range of most musculoskeletal model predictions. The higher predicted peak KCF that was observed in this study has several possible causes.

Table 6.6: Maximum in Vivo KCF During Gait

Study	Subject No.	Condition	Peak KCF	Subject Information
Taylor et al.[21]	1	Overground	2.5	one woman aged 41 with osteosarcoma in femur
Heinlein et al.[22]	2	Overground	2.1-2.8	two men aged 63 and 71 years with osteoarthritis
D'Lima et al.[23]	1	Overground	2.3	two men aged 83* and 81 years and one 67-year-old woman
D'Lima et al.[24]	3	Treadmill	1.8-2.5	
Kutzner et al.[25]	5	Overground	2.2-3.0	four men aged 60*,63*,70 and 71*, one woman aged 63 with osteoarthritis
Kutzner et al.[26]	3	Overground	2.1-2.5	
D'Lima et al.[27]	1	Overground	2.8	8year-old man with osteoarthritis
D'Lima et al.[27]	1	Treadmill	2	
D'Lima et al.[28]	1	Overground	2.4	
Fregly et al.[29]	1	Overground	2.3	
Zhao et al.[30]	1	Treadmill	2.2	

* indicated the subject was also included in the other study

First, the study population was very small for the published studies of in vivo measurement. Most of the previous studies only included one subject and no research had more than five subjects. Additionally, nearly all the participants were elderly subjects with osteoarthritis, which meant a relatively lower peak KCF due to the deceased walking speed compared with healthy, younger adults. So the results from in vivo measurement were difficult to be extrapolated to larger population other than elderly tibiofemoral implant patients.

Second, most previous musculoskeletal models used for predicting KCF only include muscles as force contributors and a single DOF knee joint in the sagittal plane [16, 38, 39, 42]. Although these models reported closer peak KCF to in vivo measurements than the models which included multiple DOFs knee joint and ligaments, they lacked the anatomical realism and may limit the ability to further investigate knee loading mechanisms. The overestimates of KCF in the present study may be indicative of inaccurate muscle parameters and ligament recruitment patterns. Validation of the magnitude of muscle and ligament forces were quite challenging since directly measuring in vivo muscle force and ligament force during gait was unavailable. To date, we could only verify the muscle activation level by experimental EMG data and ligament recruitment pattern by cadaver research.

Finally, a subject-specific muscle and ligament model is necessary to increase the likelihood of predicting more reasonable KCF. The present model used an oversimplified scaling method using general model parameters by marker pairs neglecting the anatomical variance that existed between study participants.

6.5.8 Limitations

There are some limitations in this study, most notably the small sample size. This limitation can be seen by examining the observed power for variables which were not shown to be significant or did not trend towards significance in Table 6-3. In most cases the observed power was well below 0.5. As a result, it is likely that some statistically significant effects may not have been detected due to the small sample size. Additionally, the ability to confidently generalize study results to the entire population of railroad workers is reduced.

The knee proximodistal translation and anteroposterior translation were defined as a function of passive knee flexion. Mediolateral translation was set to zero during the ballast gait simulations. Some previous musculoskeletal models which defined the knee translation in the sagittal plane as a function of passive knee flexion have already successfully simulated gait and predict muscle forces on hard, level surface [16, 42]. This demonstrates the possibility that the variance of knee translation between passive knee flexion and knee flexion in gait may be neglected for muscles. However, ligaments were not included in these models. The quantitative sensitivity analysis for the ligament in the present model suggested that a 10% increase of the ACL length (about 3.5mm) and PCL length (about 3.8mm) could increase ligament force about 200N and 300N, respectively when strained above 3%. Since marker error exists in all marker-based motion data and due to the sensitivity of the length change of ligaments, having independent knee proximodistal and anteroposterior translation was not practical in this study.

6.5.9 Conclusion

In conclusion, the three hypotheses in the present study were not supported: first, KCF was significantly altered when walking on ballast compared with NB; second, walking on MB altered KCF response more than walking on WB; third, the downhill knee joint had a higher KCF than the uphill knee joint. However, temporal gait parameters suggested the gait cycle was significant longer for ballast conditions compared with NB. The downhill limb was found to be longer in stance duration and double support and shorter in swing duration and single support than the uphill limb. No significantly different peak KCFs were found for surface condition and configuration. A trend was observed that the second peak KCF decreased for ballast conditions compared with NB. The timing of the first peak KCF was found to be significantly different among NB, WB and MB. Knee muscle cocontraction was significantly higher in WB compared with NB in the peak KCF, and was also higher for the uphill limb compared with the downhill limb in the sloped configuration. It was found that the LCL force was significantly larger and MCL forces were significantly smaller for the downhill limb compared with the uphill limb in the peak KCFs. The ligament force in the sloped configuration was significantly larger for the aACL and aMCL in the first peak KCF, and the LCL in the second peak KCF compared with the level configuration.

Overall, the effects of surface conditions were significant in the gait cycle, the time of peak knee contact loads and muscle cocontraction. The effects of surface configuration were only found in ligament forces. The effects of the uphill and downhill limbs were indicated by all the parameters except for the magnitude and the timing of peak KCFs

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CHAPTER 7

CONCLUSION

There are four substudies included in this dissertation. The significant contributions of this research and dissertation are the development of a musculoskeletal model with robust knee structures and an investigation into the changes in KCF during walking on ballast as surface condition, surface configuration and uphill or downhill limb. The first substudy evaluates the influence of toe marker placement error for lower limb joint kinematics and muscle forces during gait. This was necessary to understand the influence footwear has on predicted muscle forces in the gait model. The second substudy describes a method to combine GRF data from different trials to create a combined trial and further evaluates the accuracy of this method for joint moments and muscle forces. This was necessary to address the issue with the experimental data that only included a single force plate. The methods developed and described as part of this work can be used for any situation where an unacceptable force plate strike occurs or when a laboratory has data from multiple force plate strikes that may not occur sequentially.

Chapter 1 discusses the previous literature, regarding knee OA, ballast gait, prediction of muscle force and joint contact force, ligament modeling and OpenSim simulation. The objectives of this study were presented at the end of Chapter 1. Chapter 2 mainly describes the methods used in the fourth substudy to investigate KCF during walking on ballast. The four substudies were organized in Chapters 3-6.

Several findings from this research were statistically significant or otherwise relevant and are discussed by chapter herein.

7.1 Synopsis of Chapter 3

Ankle dorsi/plantarflexion was very sensitive to toe marker placement error, which indicated that the prediction of ankle joint kinematics was significantly affected in shod gait due to the toe marker placement on shoes. The effect of toe marker placement error was relatively small for hip abduction/adduction and knee flexion/extension compared with hip flexion/extension and hip rotation. These findings suggest that toe marker placement error affects all the joint kinematics though the magnitudes may be different. The lower limb muscle forces responded to residual variance of the joint kinematics to various degrees based on the muscle function for specific joint kinematics. Therefore, the effect of marker placement error for muscle forces is important to consider and should be evaluated individually by study design.

7.2 Synopsis of Chapter 4

A method to combine force plate data from different trials to create successful, sequential foot contact events was described. The combined, successful trials, can be used to reliably simulate a complete gait cycle and obtain the results of joint moments and muscle forces within a certain acceptable range. This method could be applied to for gait analysis situations for populations who may be unable to complete a large number of trails including impaired, elderly, amputee and pediatric gait. The proposed method could significantly reduce the total required number of trials to study lower limb biomechanics and movement disorders. Care should be taken to determine the validity of this method with an individual dataset because it is largely a function of the consistency of individual trials.

7.3 Synopsis of Chapter 5

A three-dimensional OpenSim gait model with robust knee structures was developed based on an existing model. Three main contributions of the present model compared with the existing model were: 1) The patella and patella tendon including all the parameters were involved in the model, as well as patellotibial joint. 2) Six degrees of freedom knee joint was built in this model, which included three rotations and three translations. Three knee rotations and knee mediolateral translation were independent. The knee proximodistal and anteroposterior translations were defined as a function of passive knee flexion. 3) Knee cruciate ligaments and knee collateral ligaments were involved in the model, as well as the geometry and mechanical properties of the ligament. The geometry of the ligaments in the present model was verified to be reasonable and similar to those evident in the physical knee and other existing knee models by simulation of knee motions in the three body planes. This musculoskeletal model offered the ability to investigate the effects of knee motion in the three body planes to predict more reasonable muscle forces and better understand knee biomechanics in different kinds of gait.

7.4 Synopsis of Chapter 6

Temporal gait parameters suggest that the gait cycle was significantly longer for ballast conditions compared with NB, indicating a more cautious gait when walking on ballast. Stance duration and double support duration were longer for the downhill limb than the uphill limb. Swing duration and single support duration were shorter for the downhill limb than the uphill limb.

Although no significantly different peak KCFs were found for surface conditions, surface configurations, or limbs, a trend was observed that the second peak KCF decreased for ballast conditions compared with NB due to the relatively slow walking speed. The timing of the first peak KCF was found to be significantly different among NB, MB and WB, which indicated that muscle forces were generated in advance for ballast conditions to prepare for the single support phase.

Knee muscle cocontraction was significantly higher in WB compared with NB in both peak KCFs, suggesting that walking on WB may be result in a more cautious gait compared to the other two surface conditions and requires more muscle cocontraction to account for gait variability and stability. CCI was significantly higher for the uphill limb compared with the downhill limb in the sloped configuration due to more work for uphill limb, such as elevating the body and preventing the toe from colliding with the aggregate.

LCL forces were significantly larger for the downhill limb and MCL forces were significantly larger for the uphill limb for both KCF peaks. This is necessary in order to resist knee varus and valgus instability for the corresponding lower limb. The ligament force in the sloped configuration was significantly larger for aACL and aMCL in the first

KCF peak, and LCL in the second KCF peak compared with the level configuration due to the significant difference in knee joint kinematics between surface configurations.

Overall, significant differences between surface conditions were observed in the gait cycle, the timing of peak KCF and muscle cocontraction. The effects of surface configurations were only found in ligament forces. The significant differences of uphill and downhill limbs were witnessed in all the parameters except for the magnitude and timing of KCF peaks.

7.5 Future Work

Four substudies were conducted to fulfill the objectives of the present research and provide a beneficial understanding of KCF when walking on ballast. During the course of this research three specific avenues for future work have been recognized. First, increased sample size might yield more significant results with greater ability to generalize. Due to the limited sample size, there may have been some significant effects which went undetected. This is evidenced by the typically low power levels for parameters which were not found to be statistically significant. Second, a new tracking method may be needed to better track the knee kinematics during different walking conditions, especially for knee translations. This type of tracking method can liberate the knee translation from knee flexion and result in more accurate ligament forces and predicted muscle forces. Finally, more reasonable KCF values using model simulations should be explored in the future.